A Thesis

entitled

Knee Response during Squats with Heels Up and Down

by

Francis G. Metelues

Submitted to the Graduate Faculty as partial fulfillment of the requirements for the
Masters of Science Degree in Bioengineering

Dr. Mohamed Samir Hefzy, Committee Chair

Dr. Charles Armstrong, Committee Member

Dr. Vijay Goel, Committee Member

Dr. Patricia R. Komuniecki, Dean
College of Graduate Studies

The University of Toledo
May 2014
The higher incident rate of osteoarthritis in the Asian countries compared to Western countries spurred research in the biomechanics field [1,12,48,40,51,181,193]. Genetics and diet cannot be completely ruled out; however, viewing the Asian lifestyle in terms of biomechanics brings an entirely new perspective. Researchers suggested that the prolonged squatting, which is prevalent in the daily lives of most of the Asian cultures, may provide clues to the higher rate of osteoarthritis [1,54,11,175]. This research investigated the loads associated with daily prolonged squatting. In particular, the potential cyclical loading exhibited during this activity could begin to explain the premature deterioration of the knee joint. Other studies did not conduct trials that compared various types of squatting and potential biomechanical significance. A core of this research was to extend the investigation into several squatting variants. This study investigated the influence on the biomechanical loads when the heels are down or up. Additionally, the research focused on the effects on the knee if a squat produced at the end of gait and from a standing position. Typically the kinetics and kinematics analysis focus on the transition of going into and out of a squat. This research will take the first glimpse of the force associated while shifting the body during the squat. The various types of squats would be exposed to the body weight shifting from one foot to next in the mediolateral direction. This movement occurs naturally when someone in a prolong squat attempting to achieve comfort. I hypothesized that
the moments and the shear loads from prolong shifting squats will generate a cyclical loading at the knee joint. Biomechanical loads such as the flexion-extension moment, abduction-adduction moment, bone on bone compressive and shear loads estimations for gait and various squats were analyzed to investigate any cyclical behavior. The research indicated that despite the higher moments at the knees generated during squatting, the compressive and shear forces at the knee joint for a majority of the squatting were slightly lower than the gait trials due to the reduced acceleration or ground reaction forces. This also confirmed that lower ground reaction force played a critical factor in the reduced contact shear force. However, despite comparable compressive and shear loads, the shifting squats not only produced significantly higher moments but additionally produced cyclical loading in all the biomechanical loads. A shifting squat generated cyclical moment loading in the frontal and sagittal planes while the knee joint is exposed to loading from compressive and shear forces. The relevance of this research is the squatting exhibited in the Asian culture is not just for a few seconds. Many of the squats performed daily are prolonged due to the shifting of the body for a comfortable position. The cyclical loading of the knee joint during a squat makes this a major risk factor for osteoarthritis.
I would like to dedicate this body of research to my wife Kim and children Malia and Dalen who provided support and love despite taking time away from them.
Acknowledgments

I would like to acknowledge Dr. Mohamed Samir Hefzy my advisor for taking a leap of faith on working with me on this project. In addition, I would like to acknowledge Dr. Charles Armstrong for providing guidance throughout my graduate program. Last but not least Dr. Vijay Goel for showing me how to connect the dots between theory and application. More importantly I would like to thank all of you for kindling the passion for research and applying it in a way that may potentially touch many lives.
Contents

Abstract iii

Acknowledgments vi

Contents vii

List of Tables x

List of Figures xii

List of Abbreviations xix

List of Symbols xx

1 Introduction 1

1.1 Knee Function ........................................... 1
1.2 Osteoarthritis ........................................... 7
1.3 Knee Implants & Treatment ............................. 11
1.4 Purpose of Research ................................. 13

2 Literature Review 14

2.1 Knee Osteoarthritis Risk Factors ....................... 17
  2.1.1 Excessive Weight or Obesity ....................... 18
  2.1.2 Age Impact ........................................ 20
  2.1.3 Genetics ........................................ 21
4.1.3 Comparison of Knee Joint Flexion Moment of the Various Squats 90
4.1.4 Knee Joint Abduction and Adduction Moment 94
4.1.5 Abduction-Adduction Moment for Shifting Squats 100
4.1.6 Comparison of Abduction-Adduction Knee Moment for Various Squats 104
4.1.7 Knee Joint Bone on Bone Compressive Force 108
4.1.8 Compressive Forces for Shifting Squat 116
4.1.9 Comparison of Compressive Forces for Squats 122
4.1.10 Shear Forces for Squats 127
4.1.11 Shear Forces for Shifting Squats 134

5 Discussion 138

6 Conclusion 172

References 175

A Biomechanics Derivation 194

B BCP Program 197
# List of Tables

2.1 Compressive Knee Joint Force During Gait ................................................. 38  
2.2 Compressive Knee Joint Force During Squat ................................................. 39  
3.1 Anthropometric Data of Subjects ................................................................. 43  
3.2 Reflective marker definition and anatomical locations .................................... 64  
4.1 Weight shift on the right leg for squats with heels up from a standing position ................................................................. 82  
4.2 Weight shift on the right leg for squats with heels down from a standing position ................................................................. 85  
4.3 Weight shift on the right leg for squats with heels up after performing a gait ................................................................. 87  
4.4 Weight shift on the right leg for squats with heels down after performing a gait ................................................................. 88  
4.5 Compressive knee joint force during gait ....................................................... 109  
4.6 Compressive knee joint force for squats with heels up and down from standing position ................................................................. 111  
4.7 Compressive knee joint force for squats with heels up and down at the end of a gait ................................................................. 115  
4.8 Compressive knee joint force for shifting squats with heels up from standing position ................................................................. 117
4.9 Compressive knee joint force for shifting squats with heels down from standing position .............................................................. 117

4.10 Compressive knee joint force for shifting squats with heels up executed after gait ................................................................. 120

4.11 Compressive knee joint force for shifting squats with heels down executed after gait ................................................................. 120

4.12 Shear knee joint force for squats with heels up and down from standing position ......................................................................... 129

4.13 Shear knee joint force for squats with heels up and down executed at end of gait ................................................................. 133

5.1 Compressive knee joint force for shifting squats with heels up executed after gait ................................................................. 171

A.1 Reflective marker definition and anatomical locations ................................................. 196
List of Figures

1-1 Knee Anatomy .......................................................... 2
1-2 Knee Flexion Arc Definition ........................................ 4
1-3 Medial Condyle Facet Description ................................. 6
1-4 Lateral Condyle Facet Description ............................... 8
1-5 Knee Replacement Components ................................... 12

2-1 Knee Osteoarthritis ................................................... 16
2-2 Knee Alignment ....................................................... 24
2-3 ACL Injury ............................................................. 26
2-4 Tendon Harvest ........................................................ 28
2-5 Tibial Spine and Femoral Cam Contact ........................... 34
2-6 TKA Implant ............................................................. 36

3-1 Visual 3D Skeletal Generation during Trial ...................... 42
3-2 Reflective Marker Locations ....................................... 45
3-3 Global lab and the local force plate coordinate systems ....... 46
3-4 Free body diagram of the knee joint .............................. 52
3-5 Free body diagram of the ankle joint .............................. 56
3-6 Knee joint FBD and tibial plateau angle ......................... 59
3-7 Knee joint FBD and tibial plateau angle ......................... 62

4-1 A comparison of knee joint moment when performing a gait between calculated and Visual 3D simulation .................. 66
4-2 Lab simulation of the knee joint when performing a gait .................. 67
4-3 Average knee joint moment when performing a gait ....................... 68
4-4 A comparison of BCP and V3D for knee joint moment of a squat from standing position with heels up simulation ................................. 69
4-5 Knee joint moment of a squat from a standing position with heels up . 70
4-6 Average knee joint moment when for a squat with heels up from standing position ................................................................. 71
4-7 A comparison of BCP and V3D for knee joint moment of a squat from standing position with heels down simulation ......................... 72
4-8 Knee joint moment of a squat from a standing position with heels down . 73
4-9 Average knee joint moment when a squat with heels down from standing position ................................................................. 74
4-10 A comparison of knee joint moment of walking then squatting with heels up off ground between BCP and V3D ................................. 75
4-11 Knee joint angle graphic illustration ........................................... 76
4-12 Average knee joint moment after a walk then squat with heels up ..... 77
4-13 Biomechanical lab knee joint moment of walking then squatting with heels up off the ground ......................................................... 78
4-14 Average knee joint moment after a walk then squat with heels down ... 79
4-15 A knee joint moment of a shifting squat from a standing position with the heels off the ground. ..................................................... 80
4-16 Average knee moments of subjects squatting with heels up and shifting from a standing position .................................................... 81
4-17 The knee joint moment of a shifting squat with heels up on ground ... 83
4-18 The average knee joint moment of walking then squatting with heels up on ground ................................................................. 84
4-19 Calculated knee joint moment of squatting while shifting with heels off
the ground compared to V3D 86
4-20 Knee joint moment of walking then perform a shifting squat with heels up
off ground 86
4-21 The average knee joint moment of a shifting squat with heels up off ground
at the end of a gait 87
4-22 The knee joint moment of a shifting squat with heels up on ground 88
4-23 The average knee joint moment of walking then squatting with heels up
on ground 89
4-24 The comparison of all the squat with heels down 90
4-25 The comparison of all the squat with heels up 91
4-26 The comparison of all the squat shifting with heels down and up 92
4-27 The comparison of all the squat with heels down and up 93
4-28 Frontal knee moment during gait 94
4-29 Frontal moment during a squat with heels up from standing 96
4-30 Frontal moment during a squat with heels down from standing 97
4-31 Frontal moment during a squat with heels up at end of gait 98
4-32 Frontal moment during a squat with heels down at end of gait 99
4-33 Frontal moment during a shifting squat with heels up from standing 100
4-34 Frontal moment during a shifting squat with heels down from standing 101
4-35 Frontal moment during a shifting squat with heels up at end of gait 102
4-36 Frontal moment during a shifting squat with heels down at end of gait 103
4-37 Frontal moment for squats with heels down 104
4-38 Frontal moment for squats with heels up 105
4-39 Frontal moment for stand-squat-stand 106
4-40 Frontal moment for walk-squat-walk 107
4-41 Compressive force of the knee joint when performing a gait 108
4-42 Knee joint compressive force from standing position to squatting with heels up .......................................................... 110
4-43 Knee joint compressive force from standing position to squatting with heels down .................................................. 112
4-44 Knee joint compressive force for a squat with heels up at end of gait ................................................................. 113
4-45 Knee joint compressive force for a squat with heels down at end of gait ............................................................... 114
4-46 Knee joint compressive force from standing position to a shifting squatting with heels up .................................. 116
4-47 Knee joint compressive force from standing position to a shifting squatting with heels up .................................. 118
4-48 Knee joint compressive force for walk then squat and shift with heels up ........................................................... 119
4-49 Knee joint compressive force for walk then squat and shift with heels down ......................................................... 121
4-50 Knee joint compressive force for squats with heels up ......................................................................................... 123
4-51 Knee joint compressive force for squats with heels down .................................................................................... 124
4-52 Knee joint compressive force for squats & shift with heels down ......................................................................... 125
4-53 Knee joint compressive force for squats & shift with heels up ............................................................................. 126
4-54 Knee joint compressive force from standing position to squatting .................................................................. 127
4-55 Knee joint shear force from standing position to squatting with heels up ............................................................ 128
4-56 Knee joint shear force from standing position to squatting with heels down ....................................................... 130
4-57 Knee joint shear force for a squat with heels up at end of gait .............................................................................. 131
4-58 Knee joint shear force for a squat with heels down at end of gait ........................................................................ 132
4-59 Knee joint shear force from standing position to a shifting squatting with heels up ........................................ 134
4-60 Knee joint shear force from standing position to a shifting squatting with heels down .................................... 135
4-61 Knee joint shear force for a shifting squat with heels up at end of gait ............................................................... 136
4-62 Knee joint shear force for a shifting squat with heels down at end of gait ............................................................ 137
5-1 Four Types of Non shifting squats for subject 1 ........................................ 141
5-2 Knee joint vertical reaction force and moment arm length for non shifting
   squats from standing position ............................................................... 142
5-3 Knee joint anterior-posterior force for non shifting squats from standing
   position .................................................................................................... 143
5-4 Knee joint vertical reaction force for non shifting squats performed at end
   of gait ....................................................................................................... 144
5-5 Knee joint anterior-posterior force for non shifting squats performed at
   end of gait ............................................................................................... 145
5-6 Four Types of Non shifting squats for subject 2 ........................................ 146
5-7 Knee joint vertical reaction force and moment arm length for non shifting
   squats from standing position ............................................................... 147
5-8 Knee joint anterior-posterior force for non shifting squats from standing
   position .................................................................................................... 148
5-9 Knee joint vertical reaction force for non shifting squats performed at end
   of gait ....................................................................................................... 149
5-10 Knee joint anterior-posterior force for non shifting squats performed at
    end of gait ............................................................................................. 150
5-11 Four Types of Non shifting squats for subject 1 ........................................ 151
5-12 Knee joint vertical reaction force and moment arm length for non shifting
   squats from standing position ............................................................... 152
5-13 Knee joint anterior-posterior force for non shifting squats from standing
    position .................................................................................................... 153
5-14 Knee joint vertical reaction force for non shifting squats performed at end
    of gait ..................................................................................................... 154
5-15 Knee joint anterior-posterior force for non shifting squats performed at
    end of gait ............................................................................................. 155
5-16 Four Types of Non shifting squats for subject 4 .......................... 156
5-17 Knee joint vertical reaction force and moment arm length for non shifting
squats from standing position ........................................... 157
5-18 Knee joint anterior-posterior force for non shifting squats from standing
position ................................................................. 158
5-19 Knee joint vertical reaction force for non shifting squats performed at end
of gait ................................................................. 159
5-20 Knee joint anterior-posterior force for non shifting squats performed at end of gait ........................................... 160
5-21 Four Types of Non shifting squats for subject 5 .......................... 161
5-22 Knee joint vertical reaction force and moment arm length for non shifting
squats from standing position ........................................... 162
5-23 Knee joint anterior-posterior force for non shifting squats from standing
position ................................................................. 163
5-24 Knee joint vertical reaction force for non shifting squats performed at end
of gait ................................................................. 164
5-25 Knee joint anterior-posterior force for non shifting squats performed at end of gait ........................................... 165
5-26 Four Types of Non shifting squats for subject 1 .......................... 166
5-27 Knee joint vertical reaction force and moment arm length for non shifting
squats from standing position ........................................... 167
5-28 Knee joint anterior-posterior force for non shifting squats from standing
position ................................................................. 168
5-29 Knee joint vertical reaction force for non shifting squats performed at end
of gait ................................................................. 169
5-30 Knee joint anterior-posterior force for non shifting squats performed at end of gait ........................................... 170
B-1 Raw data input .................................................. 198
B-2 Continued raw data input array ............................... 199
B-3 Continued raw data input array ............................... 200
B-4 Knee flexion angle calculations .............................. 201
B-5 Center of gravity location for the leg ....................... 202
B-6 Knee reaction force calculations ............................ 203
B-7 Flexion moment calculations ............................... 204
B-8 Flexion moment curve comparison ......................... 205
B-9 Patella tendon line of action and moment arm .......... 206
B-10 Patella tendon line of action and moment arm for rest of subjects .......... 207
B-11 Patella force ........................................... 208
B-12 Anterior-Posterior force .................................. 209
B-13 Compressive force equations ............................. 210
B-14 Shear and compressive force graphs .................... 211
B-15 Shear and compressive force graphs .................... 212
B-16 Shear and compressive force graphs .................... 213
B-17 Shear and compressive force graph of all subjects ..... 214
List of Abbreviations

V3D ........................ Visual 3D Biomechanics Simulation System
COM ........................ Center of mass of
COP ........................ Center of Pressure
KJC ........................ Knee Joint Center
OA .......................... Osteoarthritis
GCS ........................ Global Coordinate System
BW .......................... Body Weight
BMI .......................... Body Mass Index
TKA ........................ Total Knee Arthroplasty
MCL ........................ Medial Collateral Ligament
PCL ........................ Patella Collateral Ligament
UHMWPE .................... Ultra High Molecular Weight Polyethylene
HKA ........................ Hip Knee Angle
UKA ........................ Unicompartmental Hip Arthroplasty
FBD .......................... Free Body Diagram
LCL ........................ Lateral Collateral Ligament
ACL ........................ Anterior Cruciate Ligament
PHF ........................ Posterior
AHF ........................ Anterior
EF ............................ Effective Facet
MMP ........................ Matrix Metalloproteinases
MMP-1 ...................... Matrix Metalloproteinases Collagenase-1
MMP-13 ..................... Matrix Metalloproteinases Collagenase-3
ECM ........................ Extracellular Matrix
NSAID ........................ Non Steroidal Anti Inflammatory Drugs
ACI .......................... Autologous Chondrocyte Implantation
BMD ........................ Bone Mass Density
GRF ........................ Ground Reaction Force
BCP ........................ Biomechanical Computer Program
FF ............................ Flexion Facet
AP .......................... Anterior-Posterior
ANOVA .................... Analysis of Variance
List of Symbols

\( \alpha \) ........... Angle of rotation around internal rotation axis
\( \beta \) ........... the patella line of action
\( \gamma \) ........... The angle between the patella force and the longitudinal axis of the tibia
\( \omega \) ........... The angle between the tibial plateau and the longitudinal axis of the tibia
GRFy ....... Ground reaction force in anterior-posterior direction
GRFz ....... Ground reaction force in superior-inferior direction
\( \delta 1 \) ........... Moment arm vertical distance from the tibial plateau to the ground
\( \delta 2 \) ........... Moment arm vertical distance from the knee contact point to the center of gravity of the shank
\( \delta 3 \) ........... Moment arm horizontal distance from the knee contact point to the center of gravity of the shank
\( \delta 4 \) ........... Moment arm horizontal distance from the knee contact point to the ground reaction force coordinate
\( \rho o \) ........... Radius of gyration
mfoot ........ Mass of the foot
Xcop ........ Center of pressure along the X axis
Ycop ........ Center of pressure along the Y axis
\( \xi 1 \) ........... Moment arm horizontal distance from the center of the malleolus to the center of mass of the foot location
\( xi2 \) ........... Moment arm vertical distance from the center of the malleolus to the center of mass of the foot location
g ........... Gravity acceleration
ay ........... Acceleration in the anterior-posterior direction
az ........... Acceleration in the superior-inferior direction
Fy ........... Summation force in anterior-posterior direction
Fz ........... Summation force in superior-inferior direction
mshank ..... Mass of the shank
\( \delta \)patella ...... Moment arm length from knee contact to the patella
\( \delta \)bicepfoemoris Moment arm length from knee contact to the bicep femoris
\( \delta \)patella ...... Line of action of the patella force
\( \delta \)bicepfemoris Line of action of the bicep femoris force
IG ........ Inertia value for the shank
Mankle ..... Moment about the ankle joint in the sagittal plane
Mknee ..... Flexion/extension moment about the knee joint in the sagittal
Fpatella ..... The estimated consolidation force of the patella
FYpatella ... The estimated consolidation force of the patella in the anterior-posterior
direction
FZpatella ... The estimated consolidation force of the patella in the superior-inferior
direction
Fbicepfemoris The estimated consolidation force of the bicep femoris
FYshear .... Shear force from the equilibrium flexion moment
FZcomp ..... Compressive force from the equilibrium flexion moment
IG ........ Inertia value for the shank
Chapter 1

Introduction

The degeneration of the cartilage in knees has spurred many patients to find relief in pain medications prior to relenting to invasive surgery and eventually artificial knee replacements. One cause of cartilage degeneration involves the mechanical loading generated during various movements. This study delves into understanding the biomechanical loading of various squats on the knee and the potential reasons that make this risk factor lead to precipitation onset of osteoarthritis. A comprehensive look at how the knee function was necessary prior to analyzing the effects of the biomechanical loads from the various squats.

1.1 Knee Function

The complexity of the knee joint is quickly appreciated because of the various functions it provides such as the ability to absorb force, articulation, force transmission, body positioning, and long term protection from degeneration [2,3,7,35-37,174]. Weight bearing structures such as the femur (thigh bone), tibia (shin bone), and the fibula (outer shin bone) plays the most important role in the complex knee anatomy. Although the patella (knee cap) is another bone in the knee anatomy, the function is a conduit for transmitting muscle force rather than weight bearing [2,3,7,35-37,174]. A patellar tendon seen in Figure 1-1 connects from the tibial tubercle to the distal...
edge of the patella. This strong, flat tendon is also connected from the proximal edge of the patella to the femur and positioned in the center of the common tendons of the quadriceps. The rectus femoris, vastus lateralis, vastus medialis, and vastus intermedius are four large muscles of the quadriceps that curves around the patella both medially and laterally to insert into the tuberosity. The proximal end of the quadricep tendons connects to both the pelvis and same location as the patella tendon on the femur for complete control of leg movement. The patella rides in a concave groove shape called the patellofemoral or articulating cartilage. The purpose of this groove is to ensure that the force from the patella tendon and the quadriceps tendon does not shift to either side of the knee during flexion [2,3,7,35-37,174]. This configuration of the tendons is more analogous to that of a pulley for the extensors muscles [2]. A hyaline cartilage covers the distal surface of the femur cartilage to provide a low friction articulation surface for the patello-femoral groove and the mating crescent shape patella [2,3,7,35-37,174]. Wear resistant and mechanical properties of the hyaline produce an effective mechanism to distribute and transmit loads. The menisci which rest upon a tibial plateau that slopes downward from anterior to posterior supplies two significant functions [2,3,7,35-37,174]. The crescent shaped menisci absorb

![Knee Anatomy Structure](http://www.scoi.com/knee.php)

**Figure 1-1: Knee Anatomy Structure**

impact loads and provide a load distribution platform for the femur and tibia. A majority of the loads transmitted in the medial and lateral compartments is due to the menisci \[1,32,62,180,190,192\]. The connective tissues that make up the meniscus have limited movement relative to both the tibia and femur. This limitation provides an antero-posterior restraint in knees with compromised cruciate ligaments \[2,3,7,35-37,174\]. The surface of the menisci in conjunction with the synovial fluid supplies a very low friction interaction between the tibia and femur during motion. There are four ligaments that control the relative displacement between the tibial and femoral \[2,3,7,35-37,174\]. Ligaments such as the anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), medial collateral ligament (MCL) and Lateral collateral ligament (LCL) provide stability to the joint in addition, to resisting the tibiofemoral relative movement. The medial and lateral ligaments are taut when the knee extends in order to resist hyperextension and also limit the abduction and adduction movement. These ligaments also provides a lateral rotational control relative to the femur \[2,3,7,35-37,174\]. Positioning of the ACL prevents the femur sliding anteriorly relative to the tibia while the PCL controls the posterior sliding. The primary major muscle group (quadriceps and hamstrings) coordinate the movement of leg and knee joint. Location and understanding of the knee anatomic components during different activities will enlighten on how the knee joint plays a crucial role \[2,3,7,35-37,174\].

A sagittal view of the knee joint allows the flexion to be divided into three stages that cover all human activities. The figure 1-2 illustrates the three divided flexion arcs which are full extension, active functional, and full flexion \[1,13,35,36\]. The knee may be subjected on a daily basis to the three flexion ranges \[1,13,35,36\]. Most daily tasks are done in the active functional flexion range. Full flexion range is associated with tasks such as kneeling and squatting. Specifically, gait and squatting are two activities that provide a good understanding how the knee functions under those conditions. The geometrical shape of the femur and how movement is initiated in
the different flexion stages must be covered first. Two semi circular shape condyles covered in cartilage complete the distal end of the femur. The sagittal plane reveals that the radius of curvature of both the medial and lateral condyles varies. It is more conducive to subdivide the various radius of curvature so that the interaction with the tibial plateau can be clearly observed. The subdivision of the radius of curvature for the medial condyle is illustrated in Figure 1-3. The names of the subdivision or corresponding radius of curvatures are anterior horn facet (AHF), extension facet (EF), flexion facet (FF) and posterior horn facet (PHF) labeled in Figure 1-3 and Figure 1-4 [35,36]. Radius of curvature for the medial femoral condyle for the PHF, FF, EF and AHF are approximately 17 mm, 25 mm, 38 mm, and 15 mm respectively [42]. Likewise, the lateral femoral condyle for the PHF, FF are approximately 16 mm, and 60 mm respectively [42]. The anterior horn of the lateral femoral condyle radius of curvature cannot be approximated because of flatten shape [35,36]. A geometrical comparison shows that the medial articulating length is shorter and wider than the lateral condyle [43]. Geometric differences are not only restricted to the femur artic-
ulating surfaces; the mating tibial lateral plateau is concave in the lateral direction with a convex in the coronal section [35,36,43]. On the medial side of the tibia surface, the concave geometry is very slight with an upward and forward inclination of about 7° [35,36,43]. The respective menisci also contain differences that contribute to the rotational and movement in the antero-posterior direction. The medial meniscus has a deeper femoral surface and a firmer entrenchment than the lateral meniscus which limits the amount of antero-posterior distance [43]. It is these geometrical differences along with the ligaments that contribute to the rotation, sliding and rolling the knee joint. The next step is to understand the role the condyles play in the full extension, active functional, and full flexion ranges. The arc of terminal extension or ”screw home” is a common nomenclature typically used to describe the knee flexion range from 0° to 20°. The ”screw home” mechanism or locking of the knee is the rotation between the femur and tibia that occurs automatically between full extension and 10° of knee flexion [35,36,41]. The ”home screw” engagement reduces the amount of effort the quadriceps must exert when standing [41]. The rest of the range does not exhibit a typical obligatory tibial internal rotation that is common when the knee flexes [35,36]. However, the arc of terminal extension plays a vital role during heel strike in the gait process. A heel strike generates an external force and moment extension at the knee which is balance by the hamstring muscles. At heel strike, the extended knee provides the largest radius of curvature for impact absorbton and stability [37]. Spacing between the medial and lateral is minimized which allows condyles to be tighten against the tibial spine providing additional stability [37]. Further rotational stability is attained due to the collateral ligaments becoming taut at extension, and to the slope of the tibial and exposure to the larger radius of curvature [37]. The PCL provides additional restraint to hinder the anterior motion of the femur approximately 1.7 mm with respect to the tibia [37,41]. The next knee flexion range is the active functional arc which accounts for a majority of daily activities performed in
this band [35,36]. A 30° internal tibial rotation accompanies the active functional arc which typically begins from 30° to approximately 110°. This is the range where majority of the gait activity takes place. There is an external femur rotational angle approximately around 20° from extension to the end of the active functional arc range [41]. In this stage, the medial condyle moves approximately 3.6 mm posterior, which result in a net gain of 1.4 mm because of the external rotation [41]. Through a combination of sliding and rolling the lateral condyle moves in the antero-posterior direction at a greater distance than the medial condyle due to the larger radius [35,36,41,43].
There is minimal varus lift off that occurs laterally during this flexion phase. The last subdivision is the deep flexion or the squatting posture which range from 120° to 160°. This squatting position moves both of the femoral condyles in the posterior direction [35,36,43]. The posterior distance the condyles travel is approximately 10 mm at which point the lift off medially occur. This medial lift is primarily due to the limited posterior movement of the medial meniscus and the condyle riding up on the posterior horn of the meniscus [35,35,37]. The lateral condyle has almost subluxated and very little contact between the posterior horn of the condyle and the tibia [35-43]. Any functional compromise of the ligaments, menisci, or muscle groups has a tremendous impact on mobility, biomechanical stability, pain level and increased risk of osteoarthritis in of the knee joints.

1.2 Osteoarthritis

Degeneration of the joint articular cartilage to a point that the bones are in direct contact generates symptoms such as pain, swelling, stiffness and poor mobility [25]. This break down of the bone cartilage or osteoarthritis cause has yet been identified because of the etiology being a multifactorial issue [22,25,194,195,196,80]. Several high prevalent risk factors such as obesity, long term high impact sport participation, bleeding disorders, genetic inheritance, and injuries may increase the likelihood of onset degeneration or instigate pathological pathways. Researchers believe these risk factors are attributes of the mechanical stress on the joint but chemically mediated [25,194,195,196]. Degradation of the cartilage in particular induces other adverse effects such as sclerosis of the bone structure, bone cyst, osteophyte materialization and the breakdown of the subchondral bone. A look into how the cartilage structure functions is required to understand the degradation of OA. The hyaline cartilage structure can be subdivided into four layers. The cartilage top layer, which comprised
Figure 1-4: Lateral Condyle Facet Description[36]

of approximately 10% to 20% of the cartilage thickness, provides an articulating surface and shear force resistance capability [25]. This layer with the highest collagen content and highly ordered collagen fibrils provide the lowest compressive modulus compared to the other subsection [25]. Proteins in this section primary function are to protect and lubricate the surface [25]. The next layer down or the middle layer comprises of 40% to 60% of the volume. The loosely packed thicker collagen fibrils aid to produce a higher compressive modulus than the surface layer. The third layer retains the 30% remaining volume and due to the perpendicularity direction of the large collagen fibrils to the articulating surface; this section produces the highest
compressive modulus [25]. The last layer which is a calcified cartilage which directly rest upon the subchondral bone provides a foundation. The articulating cartilage surface primarily consists of extracellular matrix (ECM) is a biphasic structure that lacks blood vessels, lymphatic vessels, nerves and a sparse population of cells to produce any meaningful regeneration [25,32,33]. The solid phase of the ECM biphasic structure contains highly, ordered, compact molecular framework of collagen and proteoglycans that form a porous permeable solid matrix or interstitial intrafibrillar space for the fluid [25,32]. The fluid phase of the ECM structure fluid composition is primarily made of water and ions which accounts for 60% to 80% of the wet weight [25]. The combination of the molecular components of the two phases establishes the stiffness and viscoelastic properties of the cartilage [25,32,33]. Majority of the load transmission function of the cartilage results from the solid low permeability which forces a high pressurization of the fluid pockets or interstitial space [25]. The viscoelastic properties from the fluid phase provide a measure of stress shielding of the solid phase through reversible deformability, and load dissipation via hydraulic pressure. Within the ECM, the proteoglycans utilizes the affinity to water to generate a swelling pressure for compression resistance of the entire structure [25,33]. It is the loss of proteoglycans and the type II collagen fibrils that compromise the ECM structure to initiate a pathway to OA [22,25,32-34]. Depletion of the proteoglycans and collagen breaks down the highly ordered structure and the solid matrix becomes more permeable than the initial healthy cartilage. This results in reduction of hydraulic pressure, mechanical properties of the cartilage and key functions [25,33]. A group of proteolytic enzymes such as the matrix metalloproteinases collagenase-1 (MMP-1), and collagenase-3 (MMP-13) in particular are responsible for the degradation of both the proteoglycans and collagen fibrils [22,24,25,32-34]. Complete depletion of proteoglycans and break down of the collagen fibrils are typically seen in the late stages of OA [33]. Precipitation of OA is associated to such depletion in the articular
cartilage surface although at a much lower level [33]. It remains undetermined what precipitate the embedded cells in the cartilage called the chondrocytes to over express these degradative enzymes in OA knees. The chondrocytes have the ability to alter the metabolic response or organic matrix to address pressure, shear, and mechanical loading at any site [23,33,84,59]. These responses are documented where the matrix thickness increase in areas exposed to excess cyclical loading, mechanical loading, shear stress or hydrostatic pressure [23,25,34]. However, despite the links of mechanical loading to trigger these mechano-sensor and osmotic sensor cells, other studies are proposing a physicochemical triggering mechanism. Inflammatory cytokines such as TNF-a, IL-1b and IL-6 have been suggested to amplify MMP over expression and cartilage homeostasis disruption [22,23,25,197]. This disruption is not only contained to the encouragement of expressing degradative enzymes but also effect the chondrocyte metabolic rates to increase or inhibit MMP synthesis. In addition, the metabolic rates to increase or inhibit synthesis of collagen and proteoglycans are vulnerable to these pro-inflammatory cytokines [22,23,25,197]. An alternative perspective within the physicochemical views is the apoptosis or cell programmed death and the correlation to OA development [22,33,182]. Apoptosis of the chondrocytes has a vital role in the maintenance of homeostasis in the cartilage and the ability to repair any damage to the ECM [33,22,181]. In healthy cartilage, mechanical loading stimulate an increase of type II collagen expression while reducing the MMP-1 and MMP-3 degradation enzymes. Several studies indicate that when this ability is compromised; the resulting OA cartilage is partly due to a higher percentage of apoptotic cells [16,18,22,33,49,53,179]. Research in attempting to define what mechanism triggers these degradative enzymes to express may prove crucial. Once the degradative enzymes are reduced a homeostasis balance, improving organic matrix response, repair capabilities, and sustaining the mechanical properties of the ECM can ensure a healthy cartilage. Until a breakthrough comes to the forefront, there are medication available
to either alleviate or slow the progression of this chronic degenerative disease. Artificial implant solution or joint resurfacing are available when mobility is significantly compromised, or the threshold of pain tolerance is no longer bearable.

1.3 Knee Implants & Treatment

A few options for the 20.7 million people that have debilitating pain from OA are total knee arthroplasty (TKA) or the use of partial knee replacement products offered by several orthopaedic manufacturers [3]. Pain management through the use of nonsteroidal anti-inflammatory drugs (NSAIDs) can be used initially to alleviate painful symptoms and improve mobility by reducing inflammation [44]. Weight management programs have also shown a positive reduction of inflammation [46]. Slightly less invasive surgery such as the resurfacing of the articular surface is available but provides no long term benefits to the patient. Another option for patients with localized OA is autologous chondrocyte implantation (ACI) which inserts a patient’s own grown cartilage as surface replacement [2,6]. Additionally, the results may end in complication such as heterophic ossification [45]. An alternative treatment is viscosupplementation which is an injection that lubricates the joint providing less friction and temporary relief. Recently, TKA products have been designed to attempt to replicate the performance of the femoral, tibial and patellar functions without significantly sacrificing range of motion and impact absorption capabilities of the knee joint. The articulation surfaces of the knee joint design simulate a similarly anatomical shaped femoral component typically made of cobalt chrome alloy interacting with a ultra high molecular weight polyethylene (UHMWPE) spacer which simulates the cartilage 1-5 [2,6,121,127]. A tibial tray provides a new plateau to support the knee and the spacer. In addition, the patella posterior articulating surfaces attempt to facilitate proper flexion and extension movements of the muscles. There is also knee
replacement design that retains the ACL and PCL to provide enhanced stability, potentially improving the wear on the polyethylene. Less invasive partial replacement or unicompartmental knee arthroplasty (UKA) replaces damage surfaces isolated in a limited section of the knee only [2,6]. These treatments have made great progress in the past twenty years especially, due to the design innovations that have been described. However, research pinpointing what the daily activity, genetic disorder, or even an injury that predisposes people to OA is lagging behind medical innovation [2,6]. However, despite improvement in knee implants, there is a large population of individuals, especially outside the United States who require greater flexion range of motion to accommodate daily activities associated with religious practices and cultural.
1.4 Purpose of Research

The advances in the design of knee replacement products have lead consumers to expect increases in the potential range of motion, consistent with daily activities. The desire of patients to be able to kneel, work in the garden, and squat, have lead several orthopaedic companies to begin to address these issues. One of the activities in that this study is exploring is that involving squatting and, in particular the biomechanical requirements that squatting poses for the future knee designs.

Asian populations have a higher OA incident rate than is found in populations in Western countries. A key daily activity in Asian cultures that relates to biomechanics of the knee joint that is not prevalent in the western countries is squatting. Thus, a portion of this study will focus on a comprehensive investigation of the biomechanics of the different types of squats common to Asian culture. This focus would facilitate an improved understanding of the impact on the knee joint for the various types of squatting. There is a small body of research that has explored the effect of squatting on the knee joints. However, these published studies focused on the potential effects of the forces that are exerted when going into and out of a squat briefly [1-5]. In contrast, the present research will focus on the biomechanics of the knee during prolong squatting.

Typically, a prolong squat will necessitate shifting of the body’s weight from one foot to the other. The significance of this interest lies in the cyclic unloading and reloading of forces on the knee and the associated potential long term effects. This research will attempt to address whether or not the repetitive cyclical biomechanical loading during a squat is what makes this a risk factor for osteoarthritis. These possible links to OA may lead to strategies for mitigating the biomechanical risk factors, and also pave the way to improved designs for future knee implants.
Chapter 2

Literature Review

The squatting position accomplishes many daily tasks; however, how this action is employed have several variations. Some squats are performed from a standing position with either heels up or on in contact with the ground [8,22,160,208]. Another variation is a squat performed at the end of a gait [1]. These variations in squatting can potentially generate different biomechanic loadings and expose the knee joint to additional risks [1,22,160,188,208]. Many of the research focused on the knee joint biomechanical loading such as compressive force, anterior-posterior force, and flexion-extension moment [1,8,15,22,160,208]. Knee joint frontal or abduction-adduction moments and contact stress during deep flexion were also investigated [8,22]. The influence of heels up or down during a squat has not garnered much attention from the research community. However, a study showed that squats with heels up produced higher knee joint flexion moment than squats with heels down [22]. The squats in this study were conducted from a standing position; so the influence of gait could not be taken into account [22]. Influence of gait prior to a squatting biomechanics was another variation analyzed [1]. The flexion moment, compressive and anterior-posterior forces were squats with the heels on the ground [1]. In this published study, squats with heels up were not evaluated to confirm if larger forces may have resulted from these types of squats [1]. There was not a comprehensive research that investigated the several
variations of squats that are performed daily. Many of the research on squats were investigating for implant designs and the possible ramifications on the knee joint. The reduced contact area and exposure lead some researchers to investigate the effects of squatting on degenerative disease such as osteoarthritis [13,22,160,188].

Knee osteoarthritis a subset of arthritis is the gradual erosion of the cartilage on the joints. The limited self restoration of the cartilage produces poorer quality hyaline cartilage which cannot prevent new cracks and tears over the years [6,11-12,16-17,22-31]. This eventual erosion leads to the bones coming in contact resulting in stiffness, pain, inflammation and compromising joint function [6,11-12,16-17,22-31]. One of the ways this deterioration illicit such symptoms is the thickening of the underlying bone which causes bony growth called osteophytes or spurs development [6,11-12,16-17,22-31]. These spurs typically break off and float in the joint space. In addition, a cascading effect from cartilage deterioration inflames the synovium or joint lining due to the release of inflammation protein called cytokines [6,11-12,16-17,22-31]. In turn, these proteins exacerbate the erosion of the cartilage. Although the cause for OA has not been revealed, most data indicate that: obesity, knee injury, genetic heredity, aging, disease and disorders can play a contributive role [6,11-12,16-17,22-31]. Are one of these risk factors in the Asian culture or genome is more present in this population compared to the American cohorts? This is one of the key questions that must be addressed before an alternative solution can be proposed. Knee osteoarthritis, this chronic degenerative joint disease, has rendered approximately nine million people in the United States alone in pain or with limited mobility [9]. The devastating effect on the knee is the foremost reason for disability in the US [10]. The amounts of cartilage wear increase with each age group and is a general trend seen among these OA patients. Yet young segments of the population seem to be afflicted by this disease despite being associated mostly with the older population. Throughout the world, many population are vulnerable to the OA; however recent research indicated that a
higher prevalence in certain ethnic groups existed. This knee OA trend in the Asian populations especially when compared to the Caucasians in the United States [11-14] point to a definitive difference. The percentage of the Asian population diagnosed with knee OA in which osteophytes and joint space narrowing can be identified radiographically, ranged from two to eleven percent higher than United States Caucasian population [11-14,116]. A greater percentage of the Asian population were afflicted with bilateral OA in comparative studies [11,12]. However, a clear distinction of a higher prevalence of severe knee OA in the same age group, in the Caucasian population. Severe knee OA exhibits at least a moderate amount of osteophytes and a definite narrowing of joint space that is radiographically identified. In addition, to the cultural differences between the Western countries and Asia, there are potential risk factors that may explain the reasons a particular group is more vulnerable to unilateral, bilateral or severe knee OA.
2.1 Knee Osteoarthritis Risk Factors

Research, so far, has yet been able to unravel either the single cause or combination of factors that develop the onset of knee OA. In order to appreciate the breadth of this complex disease, a comprehensive investigation of the potential risk factors was reviewed. Knee OA is not exclusive to the older population; however, this disease is more than common among this group. Age is considered a potential risk factor for knee OA, likewise obesity must also be considered. Previous acute knee injuries increase the odds of developing OA at a later stage in life. Occupations play a contributive role due to how the knee is exposed to unusual force loadings. Naturally congenital or developmental deformities alter the knee function to predispose individuals to premature osteoarthritis. Hormones and the impact on the bone and cartilage quality also cannot be discounted in contributing to knee OA. Muscle laxity, which can alter the motion of the knee interface, is another risk factor. Genetic inheritances that predispose individuals to osteoarthritis must be included in this group of potential risk factors. Ethnicity can be potentially linked to higher knee OA for certain groups. Physical activity level and diet which impacts the health of various knee anatomy adds to the ever growing complex stew. Last but not least are certain diseases that attack the knee cartilage are understandably a powerful addition to the risk factors for knee OA. What makes knee OA, so complex is that typically, more than one of these potential factors are present. This makes it quite difficult to rule out the contribution of the individual risk factors. These risk factors are accepted as variables in the attempt to understand the higher prevalence of knee OA among the Asian population.
2.1.1 Excessive Weight or Obesity

Obesity, a commonly accepted modifiable risk factor for knee OA was theorized as an explanation why a higher prevalence of severe OA appeared in the US Caucasian population [11]. A common way fat in the body is measured, with respect to height and weight, is known as body mass index (BMI). Obesity for an adult male or female whose is a BMI that exceeds $30 \frac{kg}{m^2}$. United States obesity percentile is among the highest in the world. In comparative studies of OA between the random sampling of Caucasians in the US and Asians, the BMI was significantly wider among the two groups [11,12]. This result is not unexpected since the obese percentage or a BMI greater than $30 \frac{kg}{m^2}$ constitute approximately 3 to 4% of the populations in countries such as Japan, South Korea, and China [16]. Despite the US obesity percentage is approximately 9 times higher than the Asian countries or 34% of the population, the prevalence of knee osteoarthritis is at a lower rate [21,119]. A vast majority of the studies agrees that obesity is a key indicator of developing knee OA [47-64]. Likelihood of knee OA increases by 35% for every additional 5 unit of BMI [59,119]. Another study estimated that an obese person is 4.2 times likely to develop knee OA [45]. Despite this agreement on the strong association of obesity to knee OA, researchers have yet to identify how the etiology of knee OA is initiated by obese condition. A popular theory that many researchers subscribes to is the idea that the extra loading or mechanical stress on the knee bearing surface from the added weight contributes to OA [49,50,51]. Typically, during walking the forces on the joint is approximately three times the body weight. Excessive weight could expedite the wear and tear on the joint cartilage during walking or especially other higher impact activities such as running, and squatting [48,50,51]. Logically this line of thinking makes sense; however, when compared to long distance runners who exert 10 to 14 times their body weight on the knee joint the theory falters [56]. No such definitive
prevalence between runners and knee OA have been established. This study suggests extra mechanical loads exerted upon the knee can not solely be the cause of the deterioration of the cartilage [56,67,68]. Additionally, it also suggested that the cartilage can adapt to a higher mechanical loading without detrimental effects than expected. Another theory proposed by some researchers is the increase of inflammatory markers and abnormal amounts of insulin. The markers and insulin that accompany excessive adipose tissue can have a significant effect on the cartilage deterioration of the knee [48,49,56,180]. The adipose tissue plays another role beside energy storage, which is a highly metabolic endocrine organ [56,63]. The tissue ability to secrete agents such as adipocytokines, leptin, resistin and the adiponectin could effect the maintenance and balance of the cartilage health [180]. It is not clear whether these agents influence the joint OA progression directly or through control of the inflammation process [180,69,70,71]. It has been suggested that high levels or excess of leptin and insulin associated with high BMI individuals produce a resistance environment for both agents [56,180]. Up regulation of leptin which is associated with osteophyte formation, increase in pro-inflammatory cytokines and proportional to cartilage degeneration [56,63]. The functional role in joint deterioration, adiponectin appears to down regulate MMP-13 and reduce production of pro-inflammatory cytokines [180]. It is theorized that adipocytokines may play a vital role in controlling the inflammation process that strongly associated with osteoarthritis [180,69,70,71]. An inflammatory response is initiated from excess nutrient oxygen reactive species which begins to destroy cells [56]. A destructive cycle is initiated from an insulin resistant environment because the insulin ability to triggers certain cells to take up excess nutrient is limited. The cycle of the excess nutrient low intake increasing the insulin resistance which result in inflammation [56]. Research is continuing to investigate how inflammation impacts knee OA progression and functional capabilities [56]. Lower muscle strength in obese people has also been theorized as a potential cause
for knee OA progression. The reduction of shock absorption capability may lead to cartilage fibrillation [56,180,66]. Muscle strength, when normalized are lower in obese individuals compared to their lower BMI counterpart [56,67,72,74,75]. In particular, the quadriceps muscle were scrutinized because of the vital role of stabilizing the knee joint during loading [66,74]. These muscles play a role in reducing the impact force exerted on the knee joint during gait and descending stairs [74]. During the swing phase of the gait, the quadriceps muscle retards the descent of the leg and without this protection weaker muscles could accelerate articular cartilage deterioration [72]. It is the quadriceps muscle contraction prior to heel strike that also provides additional protection against impact forces [180,72-75]. However, linking the mechanical stress on the knee joint exerted by extra loading condition resulting from obesity to severe OA may seem logical; however, recent studies indicate it is another piece of the puzzle. A study on elderly Koreans with a much lower BMI than the Caucasian population studied in Zhang et al [11] had equivalent prevalence of severe OA [14].

2.1.2 Age Impact

Similarly, age is occasionally considered a risk factor for knee OA because of the high prevalence to knee OA [22,48,49,79-84,54,11,59,84,86,87]. Approximately almost 41% of the population in their fifties would have a radiographic knee osteoarthritis [22,49]. The percentage continues to increase to 64% and 71% for people in their sixties and seventies respectively [22,49]. Solely base on this study, the higher prevalence, demonstrated why age is considered a major risk factor for knee OA. Two schools of thoughts exist about the contribution of aging to knee OA. One is the mechanical durability of the any of the knee components that may have significant influence on the location of the weight bearing area [48,86]. There is merit behind this idea of aging increases the risk of knee OA development because over an extend time, the joints breaks down from wear and tear on [48,59]. A research study esti-
mated that even healthy knees loose one to two percentage of cartilage volume over a four and half years [17]. Additionally, the continuing degradation with age of the ligaments flexibility, muscle strength and muscle activation would result in abnormal kinematics or the shifting of normal load bearing regions [86-92]. The other school of thought is to consider not only the mechanical stress contribution to the knee joint but also genetic and cellular changes that occur with senescent [87]. One of the biological changes is the apoptosis in chondrocytes which limits the ability to synthesize and repair the cartilage matrix [87]. The extent of chondrocytes loss is debatable; however, oxidative stress, decreased levels of growth factors, advanced glycation end-products which causes brittle cartilage and increase in MMP production contributes significantly to this degradation [87]. Changes in size and structure of the molecules in the cartilage matrix bio-physical properties, also contributes to degradation [87]. Both school of thoughts concurs with the fact that the high prevalence of age and osteoarthritis remain a major independent risk factor. However, compared to the other risk factor such as obesity, age seems to play a smaller contributing role especially in the aging Asian population [14]. Asian countries are rapidly aging, and by year 2040 the rate is expected to double for people sixty five and older [54]. However, there was a noticeable difference reported in Zhang et al where the men in the study have a sharp rise in prevalence occurring for the group between the age of seventy and seventy-fours years old [11]. The contribution percentages of each of these risk factors to knee OA development is beginning to show exactly why this topic is complex.

2.1.3 Genetics

When the role of genetics and how it contributes to the potential onset of knee OA becomes enormously complicated. It is rather very difficult to quantify the contribution of genetics and to assess the impact on knee OA. Influence of genetics can be the structural protein, bone health, metabolism of the cartilage and the insulin like
growth factor genes [93]. This has not prevented researchers from attempting to find quantifiable ways to clarify possible links to knee OA onset development. A way to investigate a genetic inheritance of osteoarthritis is study families and their offsprings. It is an effective way to begin understanding the influence of the external environment and genetics impact for osteoarthritis in the joints. This allows researchers to confirm if a familial clustering of osteoarthritis in a particular joint is more prevalent than the general population. Employing this method a few studies indicated that genetic factors played a significant role in many of the knee OA cases [93]. Genetic influence on osteoarthritis can be seen when the risk of osteoarthritis increases two to three times for an individual with a sibling who underwent a total joint replacement [102]. Another way researchers used to tease out the genetic influence from the environment is to analyze monozygotic and dizygotic twins. It is through the analysis of twins that risk factor for metabolic bone disorder such as osteoporosis and associated bone mass density (BMD) have been established as significantly influenced by genetics [96]. Other studies were able to expand and investigate if BMD, bone turnover, bone geometry, bone quality inflicted on osteoporotic individuals is a precursor to knee OA [96-99,102,106-111]. A growing amount of evidence that osteoporosis as a precursor to knee OA base on the BMD association with joint space narrowing and increase of osteophytes [97-99,108-113,116]. Microfracture on the joint articulating cartilage induced by the low BMD may alter the stress capacity and predisposes individuals to the initiation of the osteoarthritic condition [97]. The stress capacity of the cartilage is important; however, the association of BMD and joint space narrowing could prove crucial. Joint space narrowing indicates the loss of cartilage volume rather than the build up of osteophytes [99,114,115,116]. This precursor for knee OA along with the strong influence of the genetics have instigated research into determining which genes and genetic variants that regulate BMD [96]. Though specific genes have yet been identified, it is believed that BMD is regulated by multiple genes, which individually
contribute to the effect to be under 10% [96]. Low or high levels of BMD does not preclude the initiation or progression of knee OA [97,99]. This indicates that other genetic factor beside the genes that regulate BMD, plays a role in the pathogenesis of knee OA. Studies of twins allow researchers to quantify the influence of either the shared environment or genetics disposition to osteoarthritis. These studies estimate that the genetic influence as low as 39% and as high 70% in terms of contribution to knee OA [102-104,106]. A quantification can be accomplish by comparing the occurrence rate of knee OA in families, siblings, monozygotic and dizygotic twins [102-104,106]. Twin studies gave insight on how genetics factored in the prevalence of the disease, the progression, osteophyte formation, joint space narrowing and cartilage volume losses [104,106]. Researchers have expanded the role of genetics and the influence on knee OA to include alignment of the bones of the lower extremity. The extreme alignment cases are understandably have a tremendous impact on knee OA; however, researchers are attempting to determine if genetics can bias a particular group to one side more than the general population.

2.1.4 Knee Alignment

Alignment of the knee is critical to how the joint respond to the impact from ground reaction forces and the distribution of the forces between the medial and lateral compartment [20,52,12,123,86]. One way to measure the alignment angle is from established bony landmarks of the hip, knee and ankle. It is these landmarks that a reference axis is established to provide angle measurements. In Figure 2-2, a vertical axis beginning from the center of the femoral head to the center of the mid-condylar which is labeled (FM) [205-206]. The other established axis extends from the center of the tibial plateau to the center of the plafond or ankle labeled (TM) [205-206]. Knee varus angle (HKA) is formed between the center lines of the femur and tibia. A neutral alignment angle forms 180° or 0° where the two axes are collinear and coinci-
dent with the load transmitted from ground reaction forces [123,205-206]. Extensive

![Figure 2-2: Knee Alignment Comparison](image)

variation from the neutral alignment such as varus alignment (bow-legged), force distribution is heavily concentrated on the medial compartment of the knee because the knee center is lateral to the load bearing axis [123,205-206]. In contrast the valgus (knocked kneed) condition the force distribution is heavily concentrated on the lateral compartment of the knee because the knee center is medial to the load bearing axis [123,180,205-206]. The external adduction moment in either condition increased during weight bearing activities which dictates the ratio between the medial and lateral compartments [20,52,86,129]. The varus alignment can double the likelihood of onset OA developing in the knees of obese individuals [20,52]. Intuitively the idea that either the valgus or varus condition could precipitate the onset of knee OA does not seem extreme; however, most research provide a stronger association to a larger contribution to the rapid progression of the disease [20,52,12,123,86]. The deterioration of one of the two compartments due not only to the accelerating cartilage loss but
also impact of misalignment on the bone structure, ligaments, menisci and osteophyte formation that propagate the progression of OA [52,68,129]. These damages are the primary reasons that varus and valgus alignment have approximately fourfold and fivefold chance of knee OA progression respectively [52,129].

A research study theorized that one of the reasons Asian were more susceptible to knee OA was because of the different varus knee alignment [18]. Tang et al theorized that the varus angle for the Asian population deviation in comparison to other ethnic groups was the primary reason for the higher prevalence of osteoarthritis. Results from the research indicated that the Asian women and men had an average of varus alignment angle of 5.4° and 4.9° respectively with a standard deviation of approximately 2.5°. The results deviation from the universally accepted varus alignment angle for a neutral knee is 3° provided potential of explaining the higher prevalence of knee OA in Asian population. However, more recent studies refute the varus alignment results from the small sample in the research by Tang et al [12]. Knee alignment comparative studies between the Asian and Caucasian group showed that the women alignment angles were comparable. However, the Asian men alignment angles were more valgus [12]. Additionally, the expected higher prevalence of medial knee OA for the Asian due to the varus condition could not be confirmed [12]. Another factor similar to the knee misalignment where the proper load bearing axis can be shifted is ligament injury.

2.1.5 Previous Injury

An injury to one of the knee ligaments in particular a vulnerable ACL or PCL can compromise the function of these passive joint stabilizer [124,126,86,130-132]. Ligaments in the knee provide a tight control on the anterior-posterior (AP) and rotational motion of the knee joint [124,126,86,130-132]. This tight control ensures dislocation does not occur while also ensuring the tibiofemoral contact location remain
the same [124,126,86,130-132]. The primary function of the knee ligaments is sensitive to injuries such as a complete tear, a partial tear, a sprain injury, or fracture of the insertion point [124]. The number of ACL injuries that occur annually in the United States are over 80,000 and is the focus of many research [124,131]. The other reason for such an intense focus is due to how commonly this ligament injury occurred amongst the population [124,131].

![Figure 2-3: The tearing of an ACL](Courtesy of //www.leonmeadmd.com/orthopedic-services-naples/sports-medicine-naples/acl-sports-medicine/)

A traumatic event illustrated in 2-3 resulting in a damage of the ACL initiate a cascade of consequences that limit the function of the knee joint. Naturally the joint becomes unstable due to the lack of control in the AP direction and also loss of rotational limiting restraint capacity [124,126,86,130-132]. The excessive sliding motion increases the shear loading and moves the load bearing contact location change to a thinner or infrequently used cartilage [124,126,86,130-132]. This new contact area is less equipped to handle the higher compressive force than the previous uninjured location [124,126,86,130-132]. The compressive force in the medial compartment is augmented by an increase in adduction moment which contribute to the accelera-
tion of knee OA [124,132]. Additionally, the homeostasis of the cartilage matrix is disrupted due to the metabolic reaction from inflammation [124]. All of these deficiencies have a profound effect on the biomechanics of the knee joint for activities such as walking. An ACL deficient knee joint loss of rotational constraints allows the contact point to move in the medio-lateral direction also [130,135]. At near full extension, the peak load is exerted on the knee joint and typically a rotational alignment occurs to allow the normal tibiofemoral contact area to absorb the load [86,135]. However, in ACL deficient knees during the “screw home” phase no external rotation occurs which implies that this loss of rotational ability exposes a different area of the articular cartilage [86,135]. In fact, a rotational offset exist where the tibia is more internally rotated in comparison to the contralateral knee [86,135]. In addition to the internal rotation reduction, a significant posterior shift happens in the medial compartment between full extension and 15° of flexion [135]. No significant changes were exhibited in the lateral compartment in ACL deficient knees [135]. In the mediolateral direction, for a range from full extension to 15° the contact point encroached near the medial tibial spine [135]. Similar observations were noted at flexion angles of 30° and 60° in the ACL deficient knees [135]. In the lateral compartment, the contact point moved away from the tibial spine and changed very little after a flexion angle of 30° [135]. This movement of the tibiofemoral contact location in both directions exposes not normally contacted cartilage areas to compressive loading, which can begin a degenerative cycle which initiate the onset of knee OA. Primary function of a damaged ACL is to provide joint stability by attempting to replicate the motion constraint and load bearing alignment.

A majority of torn ACL is surgically reconstructed as shown in Figure 2-4 from either a hamstring or patellar tendon. This stronger tendon helps to reestablish function and minimize any further deterioration of the articular cartilage [124]. Some surgeons are now incorporating multi-bundle grafts to reproduce equivalent strength and stiff-
ness of the ACL [124]. Despite the surgical intervention to repair torn ACL; the fact,

![Figure 2-4: Tendon harvest for ACL reconstruction](https://www.nytimes.com)

that the prevalence in knee OA remains a high risk factor after surgery [124,86,132-134,136]. Long term follow up studies indicated the cartilage metabolism was still effected by the injury and that the deterioration continued [124,86,132-134,136]. A study showed that 41% to 78% of the competitive soccer players with ACL injury had degenerative signs 14 years later [124,133,134]. These percentages are indicating that tibiofemoral contact location has not been restored, and the cartilage metabolism can not respond adequately to the alter load-bearing axis [124,133,134]. The high prevalence for knee OA for an injury such as the ACL demonstrate the influence of mechanical instability and potentially lead to knee OA [124].

### 2.1.6 Occupational and Rural Area

The effects of mechanical instability due to tibiofemoral contact point changing and misalignment clearly show the devastating results to the knee joint. Similarly, excessive loading on either the lateral or medial compartment is another risk factor. Exposure to excessive loading in daily activities typically occurs in an occupational
or rural setting [137-148]. Association of occupation to the prevalence of knee OA is particular strong especially if physical labor includes heavy lifting, long period of standing, climbing or squatting [137,40,49,52,53,54,11,144-148]. Odds of developing osteoarthritis in the knee due to an occupation involving one of these activities ranges from 1.5 times to 3.6 compared to a white collar job [137,40,53,144,145,147]. The chance jumps significantly when two of those occupational conditions are combined. An occupation that requires and individual to both squat and perform heavy lifting increases the odd to almost 8 times of developing knee OA [145]. The chances become even poorer of avoiding knee OA if major risk factor such as obesity is combined with squatting. This combination would result in an approximately multiplicative or in this case 14.7 times likely [144]. Similar results are observed when a comparison is made between the urban and rural areas. Typically, the rural areas are more exposed to physical labor for many years. These conditions place individuals anywhere from a 2 to 3 fold increase risk of developing knee OA compared to urban dwellers [40,139]. Physical labor plays a role in increasing the risk for knee OA; however, following that line of reasoning the expectation for manual intensive countries such as China would show men having a higher prevalence for this disease [11]. This is not the case because the Chinese women have a higher prevalence for knee OA than the men [11]. This indicates other factor play a role in developing knee OA and potentially it manifest differently in genders.

2.1.7 Gender and Hormones

Females are considered at a higher risk for the development of knee OA due to the significantly higher prevalence and having a higher severity cases than men [22,48,79,52,155,64,100,163,65,64]. After the age of 50, the odds of women developing knee OA increase to 50% higher than men [48,155,159]. It is this disparity in prevalence for knee OA, which continues for both age group that has drawn the
focus of researchers [96]. A few studies proposed theories about the differences in physical characteristics may be the cause for this higher prevalence [13,64,100,64]. Physical attributes such as pelvic dimensions, knee morphology, muscular strength, and quadriceps angle were theorized as possibilities that produced abnormal biomechanics [13,161]. The abnormal biomechanics would be the result of the joint stability compromised due to muscular strength combined with ligament laxity [96]. Thinner cartilage or less volume in women compared to men has also been added as a possible risk factor [100,64]. A thinner cartilage increases the stress on the articular surface [100,64]. A look into the knee morphology between males and females, indicates that the female femur is narrower than males and the size ratio of the lateral to medial tibial condyle is smaller [100]. These morphological differences have yet to be associated with an increased risk of knee OA. A similar study was done in an attempt to determine why Asian women and men had a higher knee OA prevalence than their Caucasian counterparts [64]. The investigation indicated that the Chinese female femoral aspect ratio were significantly smaller, and even narrower distal femur than white females [64]. It was theorized that the stress could be potentially higher than their counter parts thus leading to a higher prevalence of knee OA [64].

The increase of prevalence of OA occurs at an age when many women experience menopause which impacts the hormonal balance tremendously. One of the hormones of interest is the levels of estrogen and the possible impact on knee OA. The drop in estrogen level and the fact that estrogen receptors located in the articular cartilage have spurred research in this area [22,150,48,52,155,159,64,100,162,163]. These receptors in the articular cartilage were thought to have an influence on the cytokines and the growth factors which control the metabolism of the matrix [163]. As mention early on these cytokines mediate the synthesis and apoptosis to repair, respond and degrade the matrix to attain a homeostasis [163]. Researchers also investigated the possible role estrogen played in BMD and the possibility leading to the higher
prevalence for females [48,96]. Since high bone mass density has been documented as a mechanism for the pathogenesis of knee OA, it was theorized that because of the bone stiffness resulting in an increase of joint loading would be due to the estrogen low levels [48,96]. It was once thought that the estrogen provide a significant protection against the development of osteoarthritis; however, recent studies concluded that only borderline to moderate protection exists [22,150,48,52,155,159,64,100,162,163].

Muscular strength disparity between men and women at every age group have led researchers to investigate this as a potential trigger for the pathogenesis of the knee OA [79,51,52,155]. A low, muscle strength condition requires, the knee joint to make compensations for the mechanical loading and any shift of the load bearing axis [79,52]. It is this possible combination that may lead to structural damage and later lead to knee OA [52]. Asian men higher muscular strength has also been used to explain the rates are lower than the Asian female cohorts. Muscular strength may overcome damaging effect of squatting and kneeling in those cultures.

2.1.8 Physical Activities and Diet

Doctors, nutritionist, physical trainers all promote a balance live style that consists of exercise and proper diet. Researchers investigated links between certain several vitamins and potential of reducing the risk of knee OA [22,167,53,169,170]. In particular the antioxidant vitamin C which is an ascorbic acid has shown to be associated with OA at low levels [169,170]. In addition vitamin C, was associated with less cartilage loss and general health of the cartilage [167,53]. Levels of vitamin D has been hypothesized to dictate the ability of the bone to reduce any damage that can lead to the development of OA [22,167,53]. Similarly, vitamin E was also associated as having a protection against OA [167,53,170]. Despite this association, there is no definitive evidence that proves a protection against knee OA [167,53]. Conflicting results also existed for studies investigating a link between physical activity and knee
Some studies suggested that highly competitive sports exposes these athletes to higher risk for developing knee OA than recreational players \[52\]. However, other studies should a three fold increase for individuals with a high level of exercise compared to sedentary group \[52\]. While other studies show no adverse effect or links to the knee OA for recreational runners \[52\].

2.1.9 Diseases

Diseases such as Rheumatoid arthritis, gout, pseudogout and hemochromatosis can ravage the health and protection of the cartilage in the knee to precipitate knee OA. Congenital joint deformities are prevalence in the general population is one of the reason this topic is not explored in greater depth \[52\].

2.2 Asian Culture and Osteoarthritis

Prior to examining which risk factors that lead to the higher prevalence for knee OA, a look into the effect of culture is warranted. From a Western perspective, several factors stand out when comparing daily activities such as dining, praying, housekeeping and using the toilets. There are two things all of these activities have in common which are squatting and kneeling \[175,121,122\]. The squatting activities require a flexion angle range from 130° to 165° \[175\]. Many of the social activities in Asian homes requires either a squatting, cross legged or kneeling act \[175\]. Slowly, the western style chair are beginning to infiltrate into the Asian culture due mostly to the global economy bringing more contact with the Western world. However, the western toilets are following a similar trend although at a slower rate. The Asian style toilets require the feet to be flat, and the individual to be in a full squat through out the elimination process \[176,85\]. A primary reason this style toilet has not been replaced as quickly is the numerous health benefits \[85,176\]. Squatting during the
elimination process is faster which reduce fecal stagnation a prime cause for colon cancer [85]. This position also reduces chronic straining which lower risk for hernias, diverticulosis, and pelvic organ prolapse [85]. Lastly, it provides protection for the nerves the controls the bladder and uterus from straining [120]. Many of the household chores and interaction with toddlers requires a lot of squatting throughout the day.

Other differences include much lower BMI primarily due to the food portions, diet, and a much more active lifestyle compared to many Western countries. It is the lifestyle that predominately drive the artificial implants to be redesign to accommodate these higher flexion angle and Asian morphology for this segment of the population [121,122,127]. A majority of the TKA implant design intent was focused on the lifestyle of Western culture. As a result, the range of motion was limited between $110^\circ$ to $115^\circ$ flexion angle [121,142,143,149]. One study based upon a survey suggested that the ability to produce a full squat did not result in patient dissatisfaction [122]. This survey did not investigate if the patient would choose a choice of high flexion implant design rather than a standard design. Another by product of Western myopic design intent was the size of the implants available to Asian patients. Several studies found the morphological difference between western and Asian knees resulted in ill fitted implants [120,151-153]. Often the implants created an overhang in the medio-lateral direction which could cause soft tissue irritation and nonuniform stress distribution over the patellofemoral interface [120]. These complications may explain the higher revision rate of Asian patient compared to the Western TKA patient [120]. Recently, the design focus has shifted to providing smaller patient and those who need higher knee flexion with products to address those needs.

A few design considerations should be incorporated to ensure the TKA implant can closely replicate the natural joint motion. These design targets must not reduce the low wear ability of the polyethylene bearing surface properties [180]. The implant design allows roll back or the posterior contact motion between the femur and the
tibia during flexion [180]. This movement increases the moment arm thus reducing the muscle force [180]. Valgus and varus alignments are critical also because it directs 60% of load through the medial condyle and the rest to the lateral condyle [180]. This alignment additionally ensure that lift off is minimized which would result in edge loading and the contact stress does accelerate the deterioration of the polyethylene [180]. Lastly, tibial position and size does not inhibit the joint movement during common daily tasks. The TKA traditional or current knee implants seen in Figure 2-5 has a polyethylene tibial spine that engages with femoral cam mechanism [127]. This engagement provides stability and resistance to subluxation seen similarly in a normal knee [127].

Figure 2-5: Tibial Spine [127]

Posterior motion engages the cam approximately 75° of knee flexion and allows for roll-back [127]. As knee flexion increases, the femoral cam translates down the thicker section of the tibial spine thus reducing the stress [127,141,154]. Like wise, the anterior motion also engages the spine cam to limit anterior translation. This design also reproduces some of the key ACL functions [127]. The tibial polyethylene has a concave medial surface which provides anterior posterior stability and a natural pivoting motion. A convex lateral surface allows external rotation during flexion. Asymmetric design of the posterior shape along with condylar curvature allows for "screw home
mechanism”. Other designs incorporate the internal and external rotation into the tibial tray replicate a healthy knee motion [127]. In order to ensure efficacy of the new design, the traditional implant designs are used both as baseline performance and the initial point of designing for higher flexion implant design [127,154]. The high flexion design must produce the same capability at a minimum as the traditional design for angles less than 120°. Some of the modifications illustrated in Figure 2-6 that were made for the high flexion designs were a reduced radius of curvature to increase posterior femoral translation, and the femoral geometry to reduce polyethylene stresses [154]. A coronal view of the polyethylene insert illustrates how the medial and lateral compartments in the new implants are more conforming. This conformance with the femoral components between 0° and 30° degrees helps achieve design initiatives. This maximizes the contact area and provide great alignment to minimize either varus or valgus lift-off [154]. Additionally, the tibial post was positioned a few millimeters back which increase the moment lever arm in order to reduce the required muscle force and enhance the contact between femoral cam and post [154]. These design changes allow the new high knee flexion implants to be capable of flexion up to 155° without digging into the condyles [154].

The bending stress of the tibial spine, contact stress of the polyethylene insert and torsional stress must be thoroughly addressed to produce a design for longevity. The von Misses stresses, plastic deformation, contact stresses were determined for insert which is typically the weakest part were analyzed for both the high flexion and traditional knee implant design [127,154]. The von Misses stress within the insert increased with increasing flexion for both current and new. The deformations between the two designs were pretty close up to 110° before significant differences. This was a good indication because the new design had to be equivalent or better at the traditional flexion angles and show an improvement at higher angles. The contact stresses of the dish for the new design in the traditional angle range were similar to
the current design. Much higher stresses can be seen at the higher angles; however, the current design posterior condyles started digging into the polyethylene at angles higher than $130^\circ$. This caused the contact stress in the dish to increase dramatically. In fact, the simulation showed a complete release of the post/cam contact at $145^\circ$. This is not unexpected since the current models are designed to $120^\circ$. The amount of force required to achieve the flexion angles was also simulated. The forces for the two designs are pretty similar until about $110^\circ$. The traditional model required a lower force to achieve flexion which indicated that $110^\circ$ and $140^\circ$ the traditional model had more roll-back. A lower force may also decrease contact forces. In comparison to the traditional design, the high flexion design improved the flexion angle and achieved lower stress.

Despite these improve design changes, the contact stress is very high during deep flexion due to the decreasing contact area. This increased contact stress along with edge loading at the posterior horn accelerate wear of the polyethylene [139-141]. The polyethylene insert lift off also occurred anywhere from 0.4 mm to 1.9 mm [123]. These are the reasons why understanding the kinetics and kinematics of the knee
joint especially during activities such as squatting.

2.3 Biomechanics of Squatting

A variety of options are available to determine the kinetics of the knee joint such as model analysis, and instrumented implants. Inverse and forward solution are two types of model analysis approaches utilized to determine the forces knee joint. The widely used inverse solution applies the anthropometric measures, ground reaction forces, kinematic and link analysis to determine the knee joint force [96]. Forward solution has become recently more popular since the evolution of computer speed and capacity. In this analysis, the entire body and all movable bone segments are modeled in simulation software [96]. Inputs such as muscle moments, position, and velocities are generated from this analysis [96]. This software converges on solutions for the desired outputs that would result in equilibrium with the inputs for a time segment [96]. A major advantage for employing these analysis methods are the non-invasion application, lower cost, reduce the time for data collection and accurate anatomical assessment. However, the model accuracy is heavily dependent on the validity of the assumptions made in the indeterminate equations. Typically, the model analysis predicts higher forces when compared to instrumented implant data methods [60]. Instrumented implant data collection methods are more accurate, make fewer assumptions, and collect direct internal joint force data. This method is expensive, extremely invasive and puts the subject at more risk compared to model analysis. Since the model analysis is predominately used to evaluate the joint forces for different activities, an overview on how these results are calculated will increase understanding of this method. A force plate is used to capture the ground reaction force during a biomechanical activity. The ground reaction forces along with anthropometric data are used to calculate the equivalent muscle moment at any joint. This moment is
termed equivalent because the muscles activate to balance the torque resulting from the ground reaction forces. In the case of the knee joint, the moment arm distance is from the tibiofemoral contact to either the consolidated equivalent patella tendon or hamstring muscle. Once the moment about a joint is determined, an equivalent force can be utilized to solve for the joint contact forces. Much research has employed this analysis to investigate mechanical pathogenesis of various biological, environmental, and genetic conditions.

In particular, the research on the biomechanics of gait has increased understanding of the effects of daily activity. Recently, focus has been on understanding biomechanical effects of squatting, stair climbing, stair descending and kneeling on knee joints. The compressive or vertical load and external moment of the knee joint listed in Table 2.1 from several studies indicates the magnitudes generated during gait. Likewise, the same magnitudes can be seen in Table 2.2 for the squatting activity from several studies. A more in depth look at the equations and assumptions used for the model analysis will be covered in the latter sections of the paper.

<table>
<thead>
<tr>
<th>Author Name</th>
<th>Data Collection Type</th>
<th>Compressive Force (BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Paul$^{37,174,201}$</td>
<td>Model Analysis</td>
<td>2.7-4.3</td>
</tr>
<tr>
<td>Morrison$^{192,174}$</td>
<td>Model Analysis</td>
<td>2.1-4.0</td>
</tr>
<tr>
<td>Komistek et al.$^{174,198}$</td>
<td>Model Analysis</td>
<td>1.7-2.3</td>
</tr>
<tr>
<td>Taylor et al.$^{174,199}$</td>
<td>Telemeter</td>
<td>2.2-2.5</td>
</tr>
<tr>
<td>Fukunaga et al.$^{200}$</td>
<td>Model Analysis</td>
<td>2.8</td>
</tr>
<tr>
<td>Freeman et al.$^{36,37}$</td>
<td>Model Analysis</td>
<td>3.4</td>
</tr>
</tbody>
</table>

These publish data provide a great source for comparison of the various technique as well as to this research.
Table 2.2: Compressive Knee Joint Force During Squat

<table>
<thead>
<tr>
<th>Author Name</th>
<th>Data Collection Type</th>
<th>Compressive Force (BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dahlkvist et al.37,77</td>
<td>Model Analysis</td>
<td>5.0</td>
</tr>
<tr>
<td>Fukunaga et al.200</td>
<td>Model Analysis</td>
<td>2.8</td>
</tr>
<tr>
<td>Thambayh et al.1,15</td>
<td>Model Analysis</td>
<td>1.7-2.3</td>
</tr>
<tr>
<td>Toutoungi et al.82</td>
<td>Model Analysis</td>
<td>3.8</td>
</tr>
<tr>
<td>Smith et al.65</td>
<td>Model Analysis</td>
<td>2.8</td>
</tr>
<tr>
<td>Paul 37,174,201</td>
<td>Model Analysis</td>
<td>4.2</td>
</tr>
<tr>
<td>Komistek et al.174,198</td>
<td>Model Analysis</td>
<td>1.8-3.0</td>
</tr>
</tbody>
</table>
Chapter 3

Methods and Materials

A full kinesiology lab in the Health and Human Service facility at the University of Toledo provided tremendous capability to observe, record and analyze human motion. Motion of the subjects was tracked with attached passive reflective markers that are recorded by twelve digital camera from Motion Analysis Inc. An AMTI force platform was utilized to obtain ground reaction forces. The biomechanics data was refined and analyzed using the Cortex and Visual 3D software system. Key outputs such as the knee moment, joint reaction force, knee flexion angles, velocities, acceleration, and ground reaction forces were used to find the bone on bone contact forces. These estimated joint contact forces that were the most interesting because of wear and tear that is exacted on the cartilage. A statistical analysis was employed to determine any significance between the various types of squats. The statistical software of choice used for analysis was Minitab.

3.1 Invivo Test Protocol

An approved testing protocol (IRB#107636) was devised to simulate the possible types of squatting that occurs in the daily lives of the Asian community as close as possible in order to produce effective biomechanics data. Gait analysis research has long since establish the forces exerted on the knees during walking and running.
A testing protocol was designed to collect biomechanics data of the various squat of individuals for comparison to the gait analysis. Assuredly, the establishment of different force generated from various squats must also be confirmed or relegated as insignificant in terms of contributing force that put the knees under extreme stress. Lastly, the test protocol was devised to investigate the impact of biomechanical forces on the knees when individuals initiated weight shifting from one limb to the other during a squat. Incorporation of all these requirements necessitated a testing protocol consisting of seven activities that the subjects executed. Each of the activity was repeated three times to ensure data consistency. Each activity was intended to represent a situation that is typically executed daily in the Asian population. It is these activities that were used to generate the kinematics and kinetics data for analysis.

3.1.1 Materials and Equipment

Reflective markers were used to designate specific anatomical landmarks of the right lower limb during gait, and all of the different squats performed. The reflective marker locations were captured by twelve digital high speed cameras. The position of the markers was determined from a global lab coordinate axis established at the corner of the force plate. The data from the cameras were sampled at a rate of 100 Hz for the biomechanical trials. Ground reaction forces were captured by the AMTI force plate sampled at a rate of 60 Hz during during the trials. Additionally, an anatomical model illustrated in Figure 3-1 was generated by the Visual 3D software for coordinated visual analysis between the forces and anatomical landmark position during the trials.
3.1.2 Volunteer Group Demographic

A random group of volunteers were recruited among college students at the University of Toledo in Ohio in accordance with to IRB#107636. There was a total of six students, including three adult females and three males. The age, height and weight for the entire group are listed in the Table 3.1. None of the subjects had ever experienced lower extremity injury, joint degeneration or other conditions that may be aggravated by performing a squat. All subjects were tested in the Motion Analysis Lab of the University of Toledo department of Kinesiology. Prior to testing, and after completing the requisite Informed Consent procedure, the subjects answered a written questionnaire, which provided information about any health-related factors that may have potentially influenced their ability to walk or squat. Inquiries about any previous injuries, medications that may impair their ability to walk, stand, squat for an extended amount of time were recorded. Also, inquiry on medication for a disease that causes joint degeneration was also requested from the volunteers. A demonstration on the different types of squats, and of how to step onto the force plate while performing a squat or gait was provided. Volunteers were requested to wear clothing such as shorts or spandex to expose joint anatomical landmarks for reflective markers.
Table 3.1: Anthropometric Data of Subjects

<table>
<thead>
<tr>
<th>Classification</th>
<th>Average Age (years)</th>
<th>Average Height (ft)</th>
<th>Average Weight (lb)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Females</td>
<td>22.7</td>
<td>5’6”</td>
<td>128.3</td>
</tr>
<tr>
<td>Males</td>
<td>27.7</td>
<td>6’1”</td>
<td>189</td>
</tr>
<tr>
<td>Group</td>
<td>25.2</td>
<td>5’9”</td>
<td>158.7</td>
</tr>
</tbody>
</table>
3.1.3 Reflective Marker Instrumentation

A non invasive and effective way to track the bony landmarks was to use reflective markers, which are tracked by motion cameras [93-95]. The collected data from the reflective markers were used to construct a virtual anatomic model for biomechanical analysis. The twenty seven reflective markers measured the knee angle and velocity of the hip, ankle and knee joints for all subjects. Although the right limb was the primary target for the markers, other bony landmarks such as the left and right ASIS, posterior pelvis, as well as the iliac spine (tail bone) were tracked. The right lower limb had reflective markers placed on the: right greater trochanter, right anterior superior iliac spine, left anterior superior iliac spine, right posterior superior iliac spine, sacrum, left posterior superior iliac spine, right anterior proximal thigh, right anterior distal thigh, right posterior proximal thigh, right posterior distal thigh, right lateral femoral epicondyle, right medial femoral epicondyle, right lateral condyle of tibia, right medial condyle of tibia, right proximal patella, right distal patella, tibial tuberosity, right posterior proximal shank, right posterior distal shank, right anterior proximal shank, right anterior distal shank, right calcaneus, right first metatarsal, right fifth metatarsal, right first distal metatarsal, right lateral malleolus, and right medial malleolus. The midpoint position of the markers place medial and lateral on the femoral head and lateral and medial on the condyle of tibia represented the tibiofemoral contact point. Similarly, the midpoint between markers on the lateral and medial malleolus represented the ankle joint center. The markers place on the greater trochanter locates the center of the hip joint. These reflective marker locations provided valuable information for calculating the center of mass, knee angle, speed, velocity, position and acceleration of the leg segment during the executed protocols [1,15].

All the marker abbreviations, locations and the size of the reflective markers are...
illustrated in Figure 3-2. The markers definition and corresponding anatomical landmarks are also listed in the subsequent Table A.1.

Two coordinate systems were used to establish a reference point for the reflective markers and the ground reaction force. The global axis illustrated in Figure 3.1.3 was the lab coordinates the motion capture system used to the track reflective markers. The force plate local coordinate system origin point and the two axes of the global axis system are different. The difference in the axis centers was adjusted within the biomechanics software in order to link the two coordinate systems. The axis orientation on joints was the same as the local force plate coordinates system and the force direction adhere to that system. The force direction followed the local, coordinate convention, and the reflective markers adhere to the global axis system. Positions, accelerations, velocities, angles, and frames were tracked in the global system. Once the matter of how the leg segment was to be tracked, a virtual, model was developed for further derivations.
Figure 3-3: Two coordinate systems utilized for biomechanical simulation
3.1.4 Static Model

At the beginning of a testing session, the volunteers stood as still as possible on the force plate to capture static data. This phase of the experiment provided two key aspect for the rest of the various trials. The creation of a static model provided a baseline of the body force exerted upon the knees. Secondly, it provided a calibration test for the digital cameras that served as the basis for the automation for tracking the markers by the motion, capture system.

3.1.5 Gait Activity

One of the goals of the study was to compare the biomechanics of the subjects’ gait to various types of squat. Individuals were requested to perform a normal pace walk from one end of the lab to the other [1,15,75,76]. The starting position of the subjects varied to ensure that the pace and stride length produced natural heels strikes on the force plate. The subjects repeated the gait an additional two times for data consistency. Both video and force plate data were recorded during these trials.

3.1.6 Asian Squat at End of Gait Activity

Subjects were instructed to take a normal walking pace, step onto the force plate with arms and elbows folded horizontal, and perform a squat with heels flat on the ground [1,15,78,82]. Only the right leg was in contact with the force plate during the squat. The down portion of the squat ceased as soon as the contact between the thigh and calf muscle was felt. The subject then rose out of the squat to a standing position and proceed to walk off the force plate in the same initial direction. The gait and squat was performed two additional times for data consistency. This type of squat as mentioned in the preceding section is quite common in the Asian population when playing with children, using the toilet and performing occupational tasks [175,121-
As with the gait activity, both video and force date were recorded.

### 3.1.7 Catcher Squat at End of Gait Activity

Similar to the Asian squat activity, the subject stepped onto the force plate with arms and elbows folded horizontal; however, the squat was performed with the heels off the ground [77,78,187]. This squat is similar to a baseball, catcher position except for the fact that the legs are not flared out as much. Once contact between the between the thigh and calf muscle occurred then the subject returned to a standing position. Once again the individuals walked off the force plate at a normal pace in the same initial direction. This trial was also repeated twice for data consistency and accuracy. The force and video data were also recorded from these trials.

### 3.1.8 Shifting Catcher Squat at End of Gait Activity

The next type of squat expanded the catcher squat by adding body shifting. This shifting, catcher squat, had the individuals execute the catcher squat but remain in the squatting position even after the thigh came into contact with the calf muscle. During the five seconds of squatting that followed, the individual transferred their weight from one foot to the other. The heels were off the ground the entire time and allowed to raise slightly if loosing their balance was unavoidable. After the five second duration, the subjects rose up to a standing position and walked off the force plate at a normal pace in the initial direction. The subjects did an additional two repetitions of this trial for data consistency and accuracy. Force and video data were recorded throughout the activity.
3.1.9 Shifting Asian Squat at End of Gait Activity

Similar to the shifting catcher squat activity, in this activity the individuals remained in the squatting position even after the thigh achieved contact with the calf muscle with their heels planted on the ground for a time interval of five seconds. During that time, the subjects transferred their weight from one foot to the other. The heel remained on the ground the entire time but were allowed to raise slightly if loosing their balance was unavoidable. After the five second duration, the subjects rose up to a standing position and walked off the force plate at a normal pace in the initial direction. The subject did an additional two repetitions of the trial for data consistency and accuracy. Again, the force and video data were recorded for this activity.

3.1.10 Stand then Perform Asian Squat Activity

Requirements for this trial had the individuals beginning on the force plate for several seconds. After a brief moment, the subject folded the arms and elbows at chest height. The individuals performed a squat without the heels loosing contact with the ground. The decent phase of the squat was completed once the thigh and calf muscle came into in contact or the maximum flexion of the individuals [78]. After the brief contact, with the heels remaining firmly on the ground, the subjects rose to a standing position and walked off the force plate in the initial direction. The procedure was repeated by the individual twice to ensure data consistency and accuracy. The primary reason for this activity of the testing protocol was to investigate the influence of the lower acceleration going into a squat. Force magnitude and potentially the curve of the moment force can be significantly different when compared to the ”walk and then squat activity”. Video and force data were recorded throughout the activity.


3.1.11 Stand then Perform Catcher Squat Activity

Requirements for this trial are similar to the previous activity except that the heels were off the ground. The same sequence was initiated after a brief moment the subject folded the arms and elbows chest height. A squat was executed until the thigh and calf muscle came into contact [78]. After contact, with the heels still off the ground, the subjects rose to a standing position and walked off the force plate in the initial direction. The procedure was repeated by the individual twice to ensure data consistency and accuracy. The video and force data were recorded throughout the activity.

3.2 In vivo Biomechanics Simulation

A bio-mechanics simulation is one of the most practical way to quantify the effects of human movements especially in terms of forces, contact stress and body segment position. An approach to quantifying the various forces mentioned in bio-mechanics trials begins with a theoretical estimation of the in vivo biomechanics movement. This requires knowledge of forces being applied at the major body segments. The inverse dynamic method was utilized to estimate the forces on the knee joint because of anthropometric, kinematic and external force data availability [96,1,15]. Once these parameters were defined a simulated dynamic model can now be developed for the leg.

The dynamic model must account for the muscles, each major segments and their respective inertia and mass. The leg segments are represented by rigid bodies where the interfaces between the segments or joints can articulate in different degrees of motion. Patellar tendon and bicep femoris are the two primary muscle group that must be modeled. These two muscles have a significant influence on contact stress during many biomechanical activities. Since the muscle group wraps around a leg, a
uniaxial force vectors from an insertion point was used to estimate these muscles forces and direction in the dynamic model. Muscles play a critical role in the tibiofemoral contact location and direction of the force because of the correlation to knee angle [214,87,98].

Once leg segments, muscles or forces, and external forces are defined, applying conventional mechanics to determine equivalent knee joint internal forces and moments can be modeled. A free body diagram (FBD) was an excellent way of illustrating how the forces transpired through the various leg segments. Additionally, the FBD makes it convenient to track where the loads are applied and assumed directions. The FBD in Figure 3.2 reflects how the knee joint reacts to an external force at the foot at any given knee angle. In the FBD, the following synthesized forces are as followed: an external force applied at the foot; a force generated from the center mass of the shank being accelerated, the inertia of the shank, equilibrium moment about the knee, equilibrium anterior-posterior force and finally equilibrium superior-inferior force. In the FBD, the leg segments are represented by three rigid bodies that have relative rotation to each other. The two major rigid links represent the upper limb or femur and the lower limb which consist of the tibia and the foot. Simplification of the knee joint conversion into a free body diagram is illustrated in Figure 3.2. External ground reaction force acting against the foot was assumed to be equivalent to the exerted force from the muscles. As mentioned earlier, the muscle force was represented by a vertical force, a horizontal force along with a moment. The weight of the tibia and the forces from the mass of the leg being accelerated are additional loads that must be accounted for in the model. The center of mass is used to calculate the exerted force and inertia from the leg during movement.

Equations of translation motion from mechanics theory or Newton second law can be utilized to calculate the internal equilibrium reaction knee joint force and moment. The horizontal external ground reaction force was measured from force
plate. The next step was to determine the acceleration and mass of the shank. The mass of the shank was derived from anthropometric data that made estimation based on the percentage of total body mass location. The resultant horizontal force was the equilibrium knee joint force of the anatomical structures of the ligaments, muscles and bones. The acceleration \( a_y \) in the anterior-posterior direction was from the calculated shank center of mass. This horizontal joint reaction force can easily be mistaken for joint bone-on-bone shear force. The bone-on-bone load which combines the joint reaction force plus an active compressive force is directed at the articulating surface [96]. It is this bone-on-bone force that is central to understanding the potential effect of the magnitude and direction of this force on the knee joint. Horizontal joint reaction force was determined by summing all the forces in the anterior-posterior direction. The summation of the horizontal forces consisted of the ground reaction force and the acceleration of the shank center mass. In the calculation, the initial assumption took the anterior direction as positive which resulted in equation 3.1.
\[ F_y = GRF_y - m_{\text{shank}} \times a_y \quad (3.1) \]

Similarly, the vertical equilibrium reaction force in the knee joint began with equations of motion. The vertical ground reaction force, weight of the shank along with the acceleration of the center of mass estimated the vertical equilibrium reaction force. The weight of the shank was estimated based on anthropometric segment weight to total body weight ratio [96]. The vertical ground reaction force was also measured from the force plate and illustrated in equation 3.22. The summation of these forces is shown in equation 3.2. The superior or upward direction for the force was assumed positive in the calculation.

\[ F_z = GRF_z - m_{\text{shank}} \times g - m_{\text{shank}} \times a_z \quad (3.2) \]

The last of the three unknowns that remained to be solved for was the external muscle equilibrium moment. Application of the rotational equation of motion states that all the moments of the ground reaction forces would equal to the product of inertia and rotational acceleration of the shank passing through the shank center of mass [97,187]. Since the moment was summed about the tibiofemoral contact point and not the center mass of the shank, the parallel axis theorem was employed for this estimation [97,187]. This theorem states the summation of the moments of the external forces about the tibiofemoral contact point is equal to the components of the accelerated shank center mass plus the product of inertia and angular acceleration of the shank center of mass. It was the ability to treat the kinetic moments of the shank center of mass components, and the product inertia-rotational acceleration of the shank as free vectors that resulted in the general equation form shown in 3.3 [97,187]. These free vectors can act anywhere along the line of action during translation of the body [97,187]. One of the primary reasons the moment calculation was chosen
about the tibiofemoral point was because of the location of the two of the previous
calculated equilibrium forces were directed at that area. This simplified the moment
equation because that location reduced those two forces to zero due to the moment
arm distance being zero. This contact intersection between the femur and the tibia
was considered the tibiofemoral contact point. The moment arm measurement was
initiated and measured from this location. The moment of the external forces is
summed about the knee joint, to produce an equivalent sum of kinetic moment of the
mass of the shank. This was then multiplied by the center of mass acceleration and
the inertial components. The moment of the knee was estimated by equation 3.4.
The counter clockwise moment was assumed to be positive in the equation.

\[ \Sigma M_{knee} = \sum m_{\text{Shank}} \]  \hspace{1cm} (3.3)

\[ M_{knee} = -GRF_y \times \delta_1 - \text{Shank}_{\text{weight}} \times \delta_3 + GRF_z \times \delta_4 \]
\[ + I_G \times a_{CG} + m_{\text{shank}} \times a_z \times \delta_3 - m_{\text{shank}} \times a_z \times \delta_2 \]  \hspace{1cm} (3.4)

The distances \( \delta_1, \delta_2, \delta_3, \) and \( \delta_4 \) in the moment equation 3.4 were derived from
reflective markers placed upon anatomical landmarks. These anatomical landmarks
were recorded and tracked within the motion capture reference coordinate system.
The tibiofemoral contact point to the ground distance in the vertical direction was
calculated in equation 3.5. Tibiofemoral contact point to the shank center of mass
distance in the vertical direction was calculated in 3.6. The center of the tibial plateau
to the center mass of the shank distance in the anterior direction was calculated in
equation 3.7. Lastly, the center of the tibial plateau to the external ground force
underneath the foot distance in the anterior direction was calculated in equation 3.8.
\[
\delta_1 = \frac{RLFC_z + RLT_Pz}{2} + \frac{RMFC_z + RMTP_z}{2} - 0 \tag{3.5}
\]

\[
\delta_2 = \frac{RLFC_z + RLT_Pz}{2} + \frac{RMFC_z + RMTP_z}{2} - CG_z \tag{3.6}
\]

\[
\delta_3 = \frac{RLFC_y + RMFC_y}{2} + \frac{RLTP_y + RMTP_y}{2} - CG_y \tag{3.7}
\]

\[
\delta_4 = \frac{RLFC_y + RMFC_y}{2} + \frac{RLTP_y + RMTP_y}{2} - GRF_{coordinate} \tag{3.8}
\]

The inertia values for the shank can be estimated by determining the mass of the shank and radius of gyration. The mass of the shank was estimated from at

\[m = 0.0465 \times \text{BodyWeight}\]

while the radius of gyration becomes \[\rho_0 = 0.302 \times \left(\frac{RLFC_z + RLT_Pz + RMFC_z + RMTP_z}{2} - RLM_z\right)\]. Thus the inertia value for the shank can be derived from the equation 3.9.

\[I_G = m_{\text{shank}} \times \rho_0^2 \tag{3.9}\]

The moment and forces in the ankle seen in Figure 3-5 were also calculated in order to observe the influence on the moment about the knee joint. Same steps to calculate the moment about the knee was also followed for the ankle. The vertical, horizontal and the moment forces about the center of the ankle are the three unknown variables that required the three equation of motions. The derived equations from the FBD of the foot were rendered from Figure 3-5. The COM of the foot was estimated to be centered lengthwise between the toe and heel. The vertical COM location was centered between the malleolus and the ground. The mass of the foot was estimated using a segment to total body weight ratio from anthropometric data [96]. Similarly, the inertia for the ankle was calculated using the mass of the foot and radius of
gyration. The mass of the shank was estimated from at \( m = 0.026 \times \text{BodyWeight} \) while the radius of gyration becomes \( \rho_0 = 0.117 \times \left( \frac{RLM_z + RMM_z + RMFC_z + RMTP_z}{2} \right)^2 \). Thus, the inertia value for the shank can be derived from the equation 3.10.

\[
I_G = m_{\text{shank}} \times \rho_0^2
\]  

(3.10)

The ankle moment equation 3.11 at the center of the joint, was the initiation of the moment arm distance to the remaining forces. One key difference between the knee and ankle moment calculations was the center of pressure values utilization. Both external ground reaction forces are multiplied by their respective center of pressures generated from the force plate data. Moment arm distance, \( \xi_1 \) was defined as the horizontal distance from the center of the malleolus to the COM of the foot. Similarly, \( \xi_2 \) was the vertical distance from the center of malleolus to the COM of the foot.
\[ M_{\text{ankle}} = -GRF_y \times X_{cop} - Foot_{wt} \times \xi_1 + GRF_z \times Y_{cop} \]
\[ + \ I_G \times \alpha_{CG} + m_{foot} \times a_y \times \xi_1 - m_{foot} \times a_z \times \xi_2 \]  

\[ (3.11) \]

Equilibrium forces at the ankle joint were also calculated to verify the force from the product of the mass and acceleration of the foot could be ignored for the knee moment calculations. The contributions of external ground reaction forces were ignored since the difference was less than 5%. Equations used for this determination seen in 3.12 and 3.13 were derived from the FBD of the foot. Further usage of these forces were not required since the central interest of the research was the knee joint.

\[ F_y = m_{foot} \times a_y - GRF_y \]  

\[ (3.12) \]

\[ F_z = -m_{foot} \times g + m_{foot} \times a_z - GRF_z \]  

\[ (3.13) \]

As mentioned previously, a central part of this research was to understand the loading effects on the knee joint during movements such as squatting. In order to achieve this, the generated magnitude and direction of the bone-on-bone compressive and shear loads must be resolved. From an anatomical perspective, the muscle generates an equilibrium force to counteract the ground reaction force during a biomechanical movement. This movement can be modeled in a FBD, for better understanding of the location and direction of the forces. In the Figure 3.2, tibial plateau represents the horizontal axis as well as the line of action for the bone-on-bone shear force. A vertical or the longitudinal axis that runs the length of the tibia that intersects with the vertical axis of the femur are used for angle calculations. The angle formed between the intersection of these two axes was the knee angle. The moment arms are initiated
from the point where the bone-on-bone shear and compressive force intersect on the joint. A closer look at the FBD in Figure 3.2 revealed another complexity to account for which was the effect of the knee angle on the patella force and the moment arms. The uniaxial force representing the patella muscle force line of action forms an angle with the tibial longitudinal axis [15,214,87,98]. Intuitively, as the muscle are activated during flexion and extension activities, the sliding patella or knee pushes the patella tendon in or away from the knee joint [214,87,98]. Likewise, the changes in the line of action in the patella force effects the length of the patella moment arm because of the perpendicularity relationship. Similar, influence can be seen in the bicep femoris or hamstring muscles during flexion and extension activities. The movement of the condyles during the extension and flexion activities are influencing the line of action of the represented uniaxial force vector. Changes in the bicep femoris moment arm distance and the angles of the line of action was less drastic than the patella moment arm and line of action solely based upon the anatomic features. Studies in the past have examine this relationship between the knee angle and moment arm distance and line of action [214,87,98]. A few studies have produced moment arm distance and muscle line of actions correlations data from cadaver studies [214,87,98]. These variations with the knee joint angle can be estimated from the linear regression equations established from cadaver studies [214,87,98]. Therefore, the patella tendon force line of action direction and moment arm length were estimated by equations 3.14 and 3.15 respectively.

\[ \alpha_{\text{patella}} = -0.744 \times 10^2 - 0.575 \times 10^{-1\theta} - 0.475 \times 10^{-2\theta^2} + 0.309 \times 10^{-4\theta^3} \] (3.14)
\[ \delta_{patella} = 0.471 \times 10^1 + 0.420 \times 10^{-1\theta} - 0.896 \times 10^{-3\theta^2} + 0.477 \times 10^{-5\theta^3} \] (3.15)

Likewise, the bicep femoris force line of action direction and moment arm length was estimated by equations 3.16 and 3.17 respectively. Equations of motion for the bicep femoris force were similarly derived like the patella tendon. The Figure 3-6 was also used to derive a FBD for an activated bicep femoris.

Figure 3-6: A FBD displaying the plateau angle and longitudinal axis for bicep femoris

\[ \alpha_{bicep\, femoris} = 0.275 \times 10^1 - 0.872 \times 10^{0\theta} - 0.712 \times 10^{-3\theta^2} \] (3.16)
Linear regression for the moment arm distance to calculate the magnitudes of the represented uniaxial force of the patella tendon and the bicep femoris. Likewise, the regression equation for the muscle line of action was utilized to calculate the bone-on-bone force components. Since the external moment about the knee was resolved, the estimated force that was being exerted by the patella tendon and the bicep femoris tendon can be calculated from equation 3.19 and 3.20. These muscle force equations for the tendons were estimated by dividing the moment by their respective moment arm lengths. The moment was negative once the lower leg was in an extension movement, conversely a positive moment indicated muscle activation for flexion movement.

\[ F_{\text{patella}} = \frac{M_{\text{knee}}}{\delta_{\text{patella}}} \]  
\[ F_{\text{bicep femoris}} = \frac{M_{\text{knee}}}{\delta_{\text{bicep femoris}}} \]  

Extension-flexion moment about the knee can also be observed during activities such as squatting. In this situation, the subject initiated the squat from a stand still position. The subject performed two types of squat the first consisted of keeping the heels flat throughout the squat. The secondary squat from the stand still position had the subject maintain the heels off the ground during the entire squat to the return position or standing posture. The acceleration of the shank was so low that the calculation could have ignored the inertial values generated. There were no published data to compare either the calculation or the biomechanics laboratory data. However the biomechanics data and the calculation seen in the latter part of the results section
predicted the moment from the stand still squat very well for both types of squats.

A squat with either heels up or down, being performed at the end of a gait is another important biomechanic analysis that researchers are interested in understanding. Inertial component in the calculation can not be ignored because the acceleration of the shank COM was much higher compared to the previously mentioned squats. Moment calculation estimated the maximum torque anticipated during the entire squatting process. Interestingly, the following activities such as the walk then squat with heels up and walk then squat with heels down also depicted a better match between the data from this study than published data.

Once the external forces and moments are known the joint contact force in the vertical or compressive can additionally be determined. Likewise, the shear contact force can also be determined. The calculation used to determine the shear force and the compressive forces are illustrated in the subsequent equations. Base off the FBD the a relationship between the patella force and moment arm has been established in equations 3.19 and 3.20. A summation of the equilibrium forces in the vertical and horizontal axis becomes 3.21 and 3.22.

\[
FY_{shear} = FY_{patella} - F_Y 
\]  \hspace{1cm} (3.21)

\[
FZ_{comp} = FZ_{patella} + F_Z 
\]  \hspace{1cm} (3.22)

A force components for \( FZ_{patella} \) and \( FY_{patella} \) can be further broken down to the following equations:

\[
FY_{patella} = F_{patella} \times \sin(\beta) 
\]  \hspace{1cm} (3.23)

In equation 3.21 \( \beta \) was the angle between the patella line of action and the tibial plateau. The importance of \( \beta \) was that it represented the patella tendon direction.
with respect to the tibial plateau. This patella tendon direction or $\beta$ incorporates the two other angles. Therefore, $\beta = \gamma + \omega$ where $\gamma$ was the angle between the patella force and the longitudinal axis of the tibia. The $\omega$ is the angle between the tibial plateau and the longitudinal axis of the tibia. Typically the tibial plateau angle for men and women are $9.2^\circ$ and $7.8^\circ$ respectively. A closer look at the free body diagram in figure 3.2 illustrated these key angles and roles in establishing the magnitude and direction of the bone on bone contact force.

![Figure 3-7: A FBD displaying the plateau angle and longitudinal axis](image)

The establishment of the contact forces begins to model how the force is distributed during gait, squatting and various human movement conducted daily. Once an estimation was made on the force distribution during the gait and squatting ac-
tivity, a proper assessment of potential risk associated with those activities can be hypothesized.

\[ FZ_{\text{patella}} = F_{\text{patella}} \times \cos(\beta) \]  

(3.24)

Several assumptions were made to simplify the theoretical expression previously discussed. Frontal and transverse plane kinetics and kinematics were considered negligible, and the dynamic model was relegated to the sagittal plane [15]. The segments of the leg were considered rigid or non deformable under load which results in a constant length throughout all movements [15,96]. The tibiofemoral contact point is perpendicular to the sagittal plane, and assumed to be the instant center. Location of the tibiofemoral contact point assumed to be where the narrowest gap between the femoral condyles and the tibial plateau. As mentioned previously, the muscles and tendons are represented in the model as uniaxial approximately from the insertion sites. Any individual fibre of that may wrap around the bone structure or have differing insertion sites are not considered in the model. The ligaments are modeled as rigid links, and any elongation was considered negligible. Calculated external forces are assumed to be at equilibrium with the activated internal muscle forces. The flexion and extension moment about the knee was assumed to be counter balance by the quadriceps and hamstrings contraction respectively. The linear regression for the patella moment arm was derived for a maximum flexion angle of 120 degrees; however, the length changes for larger angles were minimal. Additionally, the linear regression for the moment arm and lines of action did not account for the potential variations of the knee anatomy. The reflective marker location and any subsequent skin movement was considered minor due to the clustering setup. Squatting activities where the thigh came into contact with the calf muscles was not reflected in the force calculations.
Table 3.2: Reflective marker definition and anatomical locations

<table>
<thead>
<tr>
<th>Marker Name</th>
<th>Anatomical Landmark</th>
</tr>
</thead>
<tbody>
<tr>
<td>RT</td>
<td>Right Greater Trochanter</td>
</tr>
<tr>
<td>RASIS</td>
<td>Right Anterior Superior Iliac Spine</td>
</tr>
<tr>
<td>LASIS</td>
<td>Left Anterior Superior Iliac Spine</td>
</tr>
<tr>
<td>RPSIS</td>
<td>Right Posterior Superior Iliac Spine</td>
</tr>
<tr>
<td>LPSIS</td>
<td>Left Posterior Superior Iliac Spine</td>
</tr>
<tr>
<td>SACRUM</td>
<td>Sacrum</td>
</tr>
<tr>
<td>RAPT</td>
<td>Right Anterior Proximal Thigh</td>
</tr>
<tr>
<td>RPPT</td>
<td>Right Posterior Proximal Thigh</td>
</tr>
<tr>
<td>RPDT</td>
<td>Right Posterior Distal Thigh</td>
</tr>
<tr>
<td>RLFC</td>
<td>Right Lateral Femoral Epicondyle</td>
</tr>
<tr>
<td>RMFC</td>
<td>Right Medial Femoral Epicondyle</td>
</tr>
<tr>
<td>RLTP</td>
<td>Right Lateral Condyle of Tibial</td>
</tr>
<tr>
<td>RMTP</td>
<td>Right Lateral Condyle of Tibial</td>
</tr>
<tr>
<td>RPPAT</td>
<td>Right Patella Anterior Proximal Tendon</td>
</tr>
<tr>
<td>RPDAT</td>
<td>Right Patella Anterior Distal Tendon</td>
</tr>
<tr>
<td>RTIBTUB</td>
<td>Right Tibial Tuberosity</td>
</tr>
<tr>
<td>RPPS</td>
<td>Right Posterior Proximal Shank</td>
</tr>
<tr>
<td>RPDS</td>
<td>Right Posterior Distal Shank</td>
</tr>
<tr>
<td>RAPS</td>
<td>Right Anterior Proximal Shank</td>
</tr>
<tr>
<td>RADS</td>
<td>Right Anterior Distal Shank</td>
</tr>
<tr>
<td>RHEEL</td>
<td>Right Calcaneus</td>
</tr>
<tr>
<td>RLF</td>
<td>Right Fifth Metatarsal</td>
</tr>
<tr>
<td>RMF</td>
<td>Right First Metatarsal</td>
</tr>
<tr>
<td>RFORE</td>
<td>Right First Distal Metatarsal</td>
</tr>
<tr>
<td>RLM</td>
<td>Right Lateral Malleolus</td>
</tr>
<tr>
<td>RMM</td>
<td>Right Medial Malleolus</td>
</tr>
</tbody>
</table>
Chapter 4

Results

There were two types of forces and moments of interest for the study of gait and various types of squats that are compiled in this section. The results in this section for the biomechanical trials were compiled into the following pattern for analysis: a flexion moment section for non-shifting squats, followed by analysis for squats with shifting, and ending with comparisons of the averages for all squat trials. The same pattern was applied for the abduction-adduction moment, compressive and AP forces. Within each of those force analysis sections, the results are reported in the following order: gait, a squat from a standing with the heels up, squat from a standing position with the heels down, walk then squat with the heels up and walk off after returning to a standing position, and lastly walk then squat with the heels down and walk off after returning to a standing position. The squat and shifting trials results which follow the force and moment sections, are compiled in the following order: stand then squat with heels up, stand then squat with heels down, walk then squat with heels up and ending with walk then squat with heels down. The flexion-extension moment section includes graphs which compared the result from the in house biomechanical customized program (BCP) to the widely used V3D software. The knee joint moments from the V3D were used for all subsequent analysis.
4.1 Knee Joint Kinetics

4.1.1 Knee Joint Flexion - Extension Moment

Figure 4-1: A comparison of knee joint moment analysis during a normal speed gait.

Moment curves based on the previously described equations for the gait of all the subjects were plotted in Figure 4-1. The BCP was compared to the biomechanics Visual 3D (V3D) simulation system and available published data. The pattern and magnitude of the average moment of the subjects curves fell comfortably well within the boundaries of published research [1,15]. Similar to published reports that gait knee joint moment has three peaks that occurred during the entire gait cycle. Heel striking the force plate created the first peak at the onset of the gait cycle as shown in Figure 4-3. A flexion moment formed the second peak due to the approaching single limb stance phase. The weight shift to the other leg which marked the approach of the terminal extension resulted in a decreasing moment. There was a slight extension
moment prior to the formation of the final peak. Knee angle during the gait cycle from flexion to terminal extension and finally back to flexion can also be observed in the Figure 4-1. The knee angle provided a crude indication of the loading of the knee during the gait phase. At heel strike, the legs were in extension, which induced a significant extension moment. Similarly, the single limb stance produced a large moment due primarily to weight shift. The average peak flexion moment of 0.2 $\frac{Nm}{kg}$ and the standard deviation for the subjects are represented graphically in Figure 4-3. Graphically, it was determined that the terminal phase of the gait cycle occurred approximately at 60% of the total cycle. The termination point of the cycle was also similar to publish data [1,15]. The gait cycle percentage began from the initiation of the striking the force plate to the brief moment prior to the toe off phase. The gait moment curves provide an excellent baseline to gage or scale moments generated from various types of squatting. The kinetics and kinematics of human gait is a
Figure 4-3: Average moment of the knee joint analysis during a normal speed gait.

A thoroughly researched topic and provides an ideal force magnitude comparison for squatting trials.

Biomechanical knee joint moment estimations from the BCP program were also compared to the V3D simulation seen in Figure 4-4. The magnitudes between the peak moment forces had slight differences; however, the two estimated simulation were within 10% of each other. Closer analysis showed that the knee joint moment curves for all the subjects rose with knee flexion.

Knee moment curve for the squatting with the heels up from a standing position produced two major flexion moment peaks close in magnitudes to one another. Initially, unlocking the knee produced a slight bump approximately at 20% of the squat cycle. This early peak corresponded with the knee unlocking at the early stages of the descent stage of the squat. The first peak concluded when the subjects reached the maximum knee flexion ability and the second corresponded to the initial movement to
Figure 4-4: The BCP and V3D moment generated for a squat from a standing position to a squat with the heels up.

Ascent from the squat followed a similar trend during the descending phase except the moment decreased with less flexion angle. The knee moment graph shown in Figure 4-6 further illustrated that peak during descent and ascent were $0.23 \frac{Nm}{kg}$ and $0.21 \frac{Nm}{kg}$ respectively. A majority of the subject had a slight dip that occurred between the two peaks on the moment curve. The magnitude of the slight drop was dependent on how fast the ascent and descent phases were executed.

Squats from a standing position with heels down had the same analysis completed for trend and comparison.

A comparison of moments for squatting down from a standing position with the heels down was also compared between the V3D and BCP simulations. Both of these moment curves displayed in Figure 4-7 had the similar relationship between knee flexion angle and moment. Once again the maximum moment coincided with the maximum achievable flexion angle. Like the previous squat with heels up, the
small flexion peak which indicated the unlocking of the knees in preparation for the descending process was also prominent. The simulation comparison of the knee flexion moment peaks were close; however, the dip was more prominent in the BCP simulation and not the V3D program. Another difference was the ascent peak in the BCP had a maximum flexion moment of 0.44 $\frac{Nm}{kg}$ compared to 0.48 $\frac{Nm}{kg}$ value for the V3D estimation. This dip estimated by BCP was not prominent in the subsequent trials with heels down from a standing position. There were no distinct peaks to mark the ascent and descent phases as seen in the previous squats. Squats with heels down from a standing position seen in Figure 4-8 had a subtle dip between the descending and ascension gap moments from most of the subjects. The average moment illustrated in Figure 4-9 had ascent and descent values of 0.23 $\frac{Nm}{kg}$ and 0.21 $\frac{Nm}{kg}$ respectively.

A dip was not prominent in this group of subject performing those squats. Knee
moment comparison between the squats performed from a standing position with heels up and down showed no significant statistical differences ($P \leq 0.09$) when analysis of variance (ANOVA) was conducted. There was approximately a 1.0% drop in average peak moment when compared to squatting with heels up from a standing position. The next series of results focused on the effect of a squat performed at the end of a gait.
Figure 4-7: The BCP and V3D moment generated for a squat from a standing position to a squat with the heels down
Figure 4-8: Knee joint moment of a squat from a standing position with heels down
Figure 4-9: Average moment of the knee joint analysis during a squat with heels down from standing position
Figure 4-10: A comparison of knee joint moment of walking then squatting with heels up off ground between BCP and V3D

Simulation estimation for the knee joint moment for a walk then squat with heels up between the BCP and V3D matched very well. The BCP program in comparison to the V3D projected a larger dip. Comparison between the two simulations seen in Figure 4-10 also expressed the relationship between the knee flexion angle and moment. Trials of the walk and then squat with heels up produced a similar extension moment seen in the gait trials during heel strike. The knee flexion angle increased after heel strike. The unlocking of the knees produced a small flexion moment. Unlocking of the knees in preparation for the descent phase of the squat occurred between knee angles of $3^\circ$ to $5^\circ$. Data from the trials in Figure 4-11 illustrated this initial flexion peak moment. This small moment dropped slightly and then rose with increase knee flexion angle similarly seen in the other squat trials. Twin peaks generated by the descent and ascent phases for each subject can be seen in Figure 4-12. These results from the subject trials produced a similar pattern to the squats with heels up from a
Figure 4-11: The knee joint position throughout the squatting with heels up at the end of gait

The magnitude of the moment increased with flexion until the maximum knee angle of the subjects was achieved. The maximum knee flexion produced the first significant peak in the moment curve. The brief time maintaining the squat showed a slight drop in the magnitude of the flexion moment. The second peak in the moment was produced during the return from the squatting to standing position. The magnitude of the flexion moment decreased with the decreasing knee flexion. These positions are graphically illustrated in Figure 4-12. Average, peak moments illustrated in Figure 4-12, were 0.25 \(\frac{N\text{m}}{kg}\) and 0.27 \(\frac{N\text{m}}{kg}\) for ascent and descent phases of the squat respectively. A similar analytical approach was conducted for the squats performed at the end of a gait with heels down.
Figure 4-12: Average moment of the knee joint analysis during a walk then squat with heels up
Trials that performed a squat with the heels down at the end of a period of walking produced moment curve with a slight dip between the ascent and descent flexion moment peaks. As seen previously in all the squat trials, the relationship between knee flexion angle and moment continued for squat with heels down executed at the end of a gait. These results summed in Figure 4-13, showed that the average ascent and descent peak forces were $0.22\frac{Nm}{kg}$ and $0.23\frac{Nm}{kg}$ respectively. Deviation from the average knee joint moment for the subjects is illustrated in Figure 4-14. There was a 34% drop in average peak moment when compared to squatting with heels up at the end of a gait. The next series of analysis investigated the effects of body weight shifts while in the squatting positions.

Figure 4-13: A knee joint moment of squat at the end of a gait with the heels off the ground.
Figure 4-14: Average moment of the knee joint analysis during a walk then squat with heels down
4.1.2 Knee Joint Moment of Shifting Squats

A shifting squat with heels up conducted from a standing position generated moments that were different than non-shifting squats in several ways. A few of the differences between the non shifting and shifting squat with heels up from a standing position are worth noting. The first peak seen in Figure 4-15 was quite similar in regards to the magnitude and pattern to the non shifting squats because the squat sequence was identical. Knee flexion angle and moment relationship remained the same up until the ascent phase. Once the maximum flexion angle and the moment was reached, the similarity ended. The moment during the ascent phase for a non shifting squat decreased with less flexion angle. However, the same phenomenon was exhibited with shifting squat except this was accomplished through body weight shifting to the left knee or the right knee moving in the adduction direction. This adduction movement created a slight extension moment in a few of the subjects.

![Figure 4-15: A knee joint moment of a shifting squat from a standing position with the heels off the ground.](image-url)
larger moment then the first peak was created as the body shifted back onto the right leg or the right knee moved in the abduction direction. The average moment peak of the second flexion seen in Figure 4-15 increased by 40% over the initial squat. The repeated abduction-adduction movement of the right knee which produced the third peak yield similar magnitude and pattern as the second moment peak. These sequences and movement are illustrated in Figure 4-15 for graphical clarity and the subjects’ deviation from the mean moment curve. The third flexion peak generated an increased knee flexion moment that was approximately 32% higher than the initial squat. Percentage of bodyweight during the movements in either the abduction or the adduction direction was summarized in Table 4.1. The third column in the table indicated ground reaction force (GRF) for the first adduction shift after the initial squat. An average of approximately 30% of the GRF was transferred to the left knee during this movement. A body shift back to the right knee or the adduction movement

Figure 4-16: A knee joint moment of squat at the end of a gait with the heels off the ground.
exerted an average of 81% of the GRF to form the second moment peak. Likewise, the repeated cycle yielded GRF percentages of 33% and 75% for the adduction and abduction movements respectively. A lower moment for the third peak corresponded with a lower GRF percentage transfer.

Table 4.1: Weight shift on the right leg for squats with heels up from a standing position

<table>
<thead>
<tr>
<th>Squat Type</th>
<th>ID</th>
<th>Adduction (% of GRF)</th>
<th>Abduction (% of GRF)</th>
<th>Adduction (% of GRF)</th>
<th>Abduction (% of GRF)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stand</td>
<td>Subj.#1</td>
<td>35.4</td>
<td>67.9</td>
<td>39.6</td>
<td>62.9</td>
</tr>
<tr>
<td>heels up</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stand</td>
<td>Subj.#2</td>
<td>27.5</td>
<td>74.2</td>
<td>37.5</td>
<td>62.5</td>
</tr>
<tr>
<td>heels up</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stand</td>
<td>Subj.#3</td>
<td>27.5</td>
<td>82.4</td>
<td>42.3.0</td>
<td>62.1</td>
</tr>
<tr>
<td>heels up</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stand</td>
<td>Subj.#4</td>
<td>37.4</td>
<td>80.8</td>
<td>39.0</td>
<td>78.0</td>
</tr>
<tr>
<td>heels up</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stand</td>
<td>Subj.#5</td>
<td>29.6</td>
<td>91.4</td>
<td>16.7</td>
<td>96.3</td>
</tr>
<tr>
<td>heels up</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stand</td>
<td>Subj.#6</td>
<td>20.4</td>
<td>88.9</td>
<td>25.0</td>
<td>87.0</td>
</tr>
<tr>
<td>heels up</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Avg.</td>
<td></td>
<td>29.60</td>
<td>80.9</td>
<td>33.3</td>
<td>74.8</td>
</tr>
</tbody>
</table>

Analysis on the effect of the heels down along with body weight shifting from a standing position was also investigated.

Similar patterns continued with a shifting squat with heels down, where the initial weight shift produced an initial moment peak of $0.21 \frac{Nm}{kg}$ preceded by larger moment peaks of $0.32 \frac{Nm}{kg}$ and $0.30 \frac{Nm}{kg}$. These moment curves presented in Figure 4-17 showed a slight decrease in peak moments when compared to the squat and body weight shift with heels up. The second and third moment peaks decreased by 14% and 7% respectively in this comparison. Both standard deviations for the group and a pictorial sequence of the squats is illustrated in Figure 4-18. Another difference between the two shifting squats was the extension moment area during the abduction
Figure 4-17: A knee joint moment of a shifting squat at the end of a gait with the heels on the ground.

...movement of the right knee. There was still an unloading of the moment on the right knee pattern seen with the shifting squat with heels up. The analysis of the amount of body shift from one leg to the other while in a squat is summarized in Table 4.2.

Percentage of the GRF for squats and shifting with heels up and down were similar for the abduction movements. However, the adduction movements were slightly lower for the heels down shifting squat. Effects of how the gait played a role with the shifting squats can be seen in the following analytical series.

A simulation comparison between the V3D and BCP program for a shifting squat with heels up performed at the end of a gait presented in Figure 4-19 showed a very close match. Shifting squat with heels up performed at the end the gait, was comparable initially to the non shifting squat with heels up at the end of a gait. Heel strike producing the extension moment peak occurred also in these trials. Initial squat average peak moment magnitude illustrated in Figure 4-22 was another similarity...
Figure 4-18: A knee joint moment of squat at the end of a gait with the heels on the ground.

between these two types of squats. Despite the magnitudes being equivalent, one difference was the dip between the descent and ascent phase was not prominent. After the initial squat, the pattern seen in previous shifting squat was the same. A few of the subjects produced a small extension moment during the shift to the left knee. Subsequent peaks were larger than the initial squat. The second and third peak moment had magnitudes of 0.41 $\frac{Nm}{kg}$ and 0.38 $\frac{Nm}{kg}$ respectively. The second peak had an increase of 44% over the initial squat. A percentage of 39% increased of the third peak moment over the primary squat. A summary of the corresponding GRF percentage for the second and third peak moments is listed in Table 4.3.

The same observation or trend was seen where the second peak had a slightly higher GRF percentage. This squat did produce the highest peak moments and GRF percentage transfer. Extension moment generated while in a squat and the pictorial sequence can be seen in Figure 4-21.
Table 4.2: Weight shift on the right leg for squats with heels down from a standing position

<table>
<thead>
<tr>
<th>Squat Type</th>
<th>ID</th>
<th>Abduction (% of GRF)</th>
<th>Adduction (% of GRF)</th>
<th>Abduction (% of GRF)</th>
<th>Adduction (% of GRF)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stand heels down</td>
<td>Subj.#1</td>
<td>37.5</td>
<td>72.5</td>
<td>42.5</td>
<td>69</td>
</tr>
<tr>
<td>Stand heels down</td>
<td>Subj.#2</td>
<td>17.5</td>
<td>80.0</td>
<td>18.3</td>
<td>82.5</td>
</tr>
<tr>
<td>Stand heels down</td>
<td>Subj.#3</td>
<td>29.1</td>
<td>84.1</td>
<td>39.6</td>
<td>86.8</td>
</tr>
<tr>
<td>Stand heels down</td>
<td>Subj.#4</td>
<td>38.5</td>
<td>73.1</td>
<td>48.9</td>
<td>62.6</td>
</tr>
<tr>
<td>Stand heels down</td>
<td>Subj.#5</td>
<td>31.5</td>
<td>79.0</td>
<td>33.3</td>
<td>77.8</td>
</tr>
<tr>
<td>Stand heels down</td>
<td>Subj.#6</td>
<td>25.0</td>
<td>76.9</td>
<td>24.1</td>
<td>76.9</td>
</tr>
<tr>
<td><strong>Avg.</strong></td>
<td></td>
<td><strong>29.80</strong></td>
<td><strong>77.60</strong></td>
<td><strong>34.50</strong></td>
<td><strong>77.10</strong></td>
</tr>
</tbody>
</table>

The last flexion-extension moment analysis was done on the shifting squat with heels down. The shifting squats with heels down executed after a period of walking, shown in Figure 4-22, provided data for comparison. This squat had the same pattern trend of two larger peak moments preceding the initial squat peak moment. The percentage increase over the initial squat for the second and third flexion moment peaks were 18%. Standard deviation and pictorial sequences are illustrated in Figure 4-23. Few of the subjects produced extension moments during the shifts to the left knee. Ground reaction force summarized in Table 4.4 showed that the percentage for the second and third moment peaks were approximately equivalent.

This section describe the results of analysis on the impact: of stand versus walking, heels up versus heels down, and shifting versus non shifting squats. All of these data results can now be compared and statistically analyzed for significance.
Figure 4-19: A knee joint moment of squat while shifting with the heels off the ground.

Figure 4-20: A knee joint moment of shifting squat at the end of a gait with the heels off the ground.
Table 4.3: Weight shift on the right leg for squats with heels up after performing a gait

<table>
<thead>
<tr>
<th>Squat Type</th>
<th>ID</th>
<th>Adduction (% of GRF)</th>
<th>Abduction (% of GRF)</th>
<th>Adduction (% of GRF)</th>
<th>Abduction (% of GRF)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walk heels up</td>
<td>Subj.#1</td>
<td>37.9</td>
<td>66.7</td>
<td>38.3</td>
<td>69.6</td>
</tr>
<tr>
<td>Walk heels up</td>
<td>Subj.#2</td>
<td>25.8</td>
<td>91.7</td>
<td>19.2</td>
<td>87.5</td>
</tr>
<tr>
<td>Walk heels up</td>
<td>Subj.#3</td>
<td>26.9</td>
<td>85.2</td>
<td>13.7</td>
<td>83.5</td>
</tr>
<tr>
<td>Walk heels up</td>
<td>Subj.#4</td>
<td>35.7</td>
<td>87.4</td>
<td>42.9</td>
<td>59.3</td>
</tr>
<tr>
<td>Walk heels up</td>
<td>Subj.#5</td>
<td>9.3</td>
<td>79.6</td>
<td>13.0</td>
<td>84.6</td>
</tr>
<tr>
<td>Walk heels up</td>
<td>Subj.#6</td>
<td>27.8</td>
<td>87.0</td>
<td>18.5</td>
<td>83.3</td>
</tr>
<tr>
<td>Avg.</td>
<td></td>
<td>27.20</td>
<td>82.90</td>
<td>24.30</td>
<td>78.0</td>
</tr>
</tbody>
</table>

Figure 4-21: A knee joint moment of a shifting squat at the end of a gait with the heels off the ground.
Figure 4-22: A knee joint moment of a shifting squat at the end of a gait with the heels on the ground.

Table 4.4: Weight shift on the right leg for squats with heels down after performing a gait

<table>
<thead>
<tr>
<th>Squat Type</th>
<th>ID</th>
<th>Abduction (% of GRF)</th>
<th>Adduction (% of GRF)</th>
<th>Abduction (% of GRF)</th>
<th>Adduction (% of GRF)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walk heels down</td>
<td>Subj.#1</td>
<td>47.5</td>
<td>73.8</td>
<td>28.3</td>
<td>71.7</td>
</tr>
<tr>
<td>Walk heels down</td>
<td>Subj.#2</td>
<td>20.0</td>
<td>95.0</td>
<td>20.8</td>
<td>95.8</td>
</tr>
<tr>
<td>Walk heels down</td>
<td>Subj.#3</td>
<td>25.8</td>
<td>79.1</td>
<td>24.2</td>
<td>76.9</td>
</tr>
<tr>
<td>Walk heels down</td>
<td>Subj.#4</td>
<td>25.8</td>
<td>79.1</td>
<td>24.2</td>
<td>78.0</td>
</tr>
<tr>
<td>Walk heels down</td>
<td>Subj.#5</td>
<td>13.6</td>
<td>72.2</td>
<td>6.8</td>
<td>88.3</td>
</tr>
<tr>
<td>Walk heels down</td>
<td>Subj.#6</td>
<td>32.4</td>
<td>76.9</td>
<td>28.7</td>
<td>78.7</td>
</tr>
<tr>
<td><strong>Avg.</strong></td>
<td></td>
<td><strong>27.50</strong></td>
<td><strong>79.30</strong></td>
<td><strong>22.20</strong></td>
<td><strong>81.50</strong></td>
</tr>
</tbody>
</table>
Figure 4-23: A knee joint moment of squat at the end of a gait with the heels on the ground.
4.1.3 Comparison of Knee Joint Flexion Moment of the Various Squats

A series of comparative analyses were conducted on the types of squats performed in the trials. Such analyses facilitate focus on the remaining variables. The results can then be isolated and effects quantified. In Figure 4-24 all squats performed with heels down were grouped together in order to understand and compare the effects of standing, walking, and shifting on knee flexion moments. This comparison graph pointed to a few observations that have an influence on the knee joint moment for squatting. Squats performed at the end of a period of walking generated higher knee joint moments. Walk then squat trials produced a 21% increase in flexion moment when compared to squats performed from a standing position. In the shifting squats, the initial squat, second and third peak moments had an increase of 29%, 30% and 20% over the shifting squats from a standing position respectively.

Figure 4-24: A comparison of the squat with heels down
Figure 4-25: A comparison of the squat with heels up

The same trend and observation can be made when looking at all types of squats performed with heels up. In Figure 4-25 the average peak moment for the squat performed at the end of the gait had increased by 20% when compared to the squat from a standing position. The shifting squats flexion moment increase was not as large between the walking and standing conditions. The squats at the end of a gait had a 0.1%, 13% and 14% flexion moment increase for the three peaks over the standing, shifting squats.

A comparison of subjects executing body weight shifts while in a squatting position is illustrated in Figure 4-26. In this comparison, the influence of the heels being either up or down can be broken down for a more in depth analysis. Shifting squats with heels up that involved gait produced that largest knee moment. This was followed by a shifting squat performed at the end of a gait with the heels up from a standing position. Comparison between the squats with heels up and down at then end of a
Figure 4-26: A comparison of the squat shifting with heels up and down gait showed a difference of 25% in the initial squat. Furthermore, the second and third peaks for the heels up were 43% and 38% higher than walk then squat with heels down trials. This trend continued when comparing the heels up versus heels down in the stand then squat trials. Results for the initial squat peak moments were equivalent; however, the second and third peak moments increased for the heels up trials. The squats with the heels up for the second and third peak moments increased approximately 10%.

The last average flexion moment illustrated in Figure 4-27, analyze the effects of the heel being up or down in non-shifting squats. A higher generated peak moment for the squats with heels up was observed in this comparison. In the walk then squat trials, heels up produced approximately 11% higher moments than the squatting condition with the heels down. However, the trials for squats from a standing position were reversed. This appeared to be primarily due to an outlier being included in the
The sagittal knee flexion moment thus far has occupied a lot of the research focus; however, the frontal plane moment cannot be completely ignored. The following analysis will look at the behavior of the frontal moment during the various squat conditions.
4.1.4 Knee Joint Abduction and Adduction Moment

Moments in the sagittal plane previously discussed are typically the main focus for most research primarily due to the higher magnitude generated. Adduction moment applies a compressive loading to the lateral condyles while conversely abduction moment applies to the loading on the medial condyles. The knee moments in the sagittal plane during squatting were higher than the moment generated from a normal gait. The question whether that phenomenon would continue in the frontal plane led to a comprehensive look at the abduction and adduction knee joint moments. Gait moment in the frontal plane displayed in Figure 4-28 remained primarily in abduction loading throughout the majority of the gait phases. The average peaked abduction moment was $0.12 \frac{Nm}{kg}$ which occurred at the phase preceding heel strike and prior to toe off. Results from these gait trials correlated well with publish data [202-203].

Figure 4-28: The abduction and adduction moment during a gait

A similar trend can be seen with the squat with heels up from a standing position.
The moments of the subjects seen in Figure 4-31 were all in the abduction direction. Peak average abduction moments were approximately $0.065 \frac{Nm}{kg}$ for this particular squat condition. The entirety of the squatting sequence was continuously exposed to an abduction moment. When the heels were up during a squat an adduction moment was generated, while conversely the squats with the heels down had an abduction moment. A similar analysis was also investigated for squats which incorporated body weight shift from one leg to the other.

Moments for the squats with heels up from a standing position were similar to the gait trials. The moments of the subjects seen in Figure 4-29 were all in abduction. Peak average abduction moments were approximately $0.125 \frac{Nm}{kg}$ for this particular squat condition. The entirety of the squatting sequence was continuously exposed to an abduction moment. The magnitude of the abduction moment increased with the knee flexion angle. During the ascent phase of the squat, the abduction moment lessened. A small dip between the descent and ascent movements occurred in many of the subjects' moment curves.

Frontal knee moment for the squatting condition with heels firmly grounded, produced a similar adduction trend. A majority of the subjects peak force was an adduction moment as illustrated in Figure 4-30. An increase in knee flexion angle resulted in an increase in abduction force with these types of squat conditions. The small dip between the descent and ascent was not prominent in this particular squat condition.

Walk then squat with heels up initially produced an adduction moment before reversing to an abduction trend. This small adduction moment coincided with the heel strike and began reversing direction once the descent phase of the squatting was initiated (see Figure 4-32). Peak abduction moments generated by the descent and ascent movements were higher than squats performed from a standing position. A dip in between the descent and ascent phase was prominent in this squatting type.
The average, peak magnitude of 0.1 \( \frac{Nm}{kg} \) was slightly less than average peak moments produced during gait.

This trend, and a similar pattern, continued for the squat performed at the end of a period of walking with heels down. Descent and ascent actions produced two major peaks during the entire squatting activity. A dip can be seen in Figure 4-32 for many of the subjects. The average, peaks moments were less than the walk then squat with heels up trials. The average, peak moments were 0.06 \( \frac{Nm}{kg} \) for both. The next step was to analyze the influence of adding body shifts while in a squat.
Figure 4-30: The abduction and adduction moment during a squat with heels down from standing position
Figure 4-31: The abduction and adduction moment during a squat with heels up at the end of a gait
Figure 4-32: The abduction and adduction moment during a squat with heels down at the end of a gait
4.1.5 Abduction-Adduction Moment for Shifting Squats

Body weight shift from one leg to the next during a squat as expected had a tremendous influence on the moment curve.

The average moment peak of a shifting squat with heels down from a standing position in Figure 4-33 had few key differences when compared to a non shifting squat. Although the peak magnitude was slightly lower than then non shifting squat, a cyclical pattern emerged from the data. Shifting the body weight from one side to the other resulted in a frontal moment moving into the abduction and adduction phase in a sinusoidal pattern. The knee was now subjected to an adduction frontal moment additionally. An equivalent adduction moment in terms of average, peak magnitude was being applied to the knee during body weight shifts. The abduction and the adduction moment peak at 0.05 $\frac{Nm}{kg}$ and 0.03 $\frac{Nm}{kg}$ respectively.

Figure 4-33: The abduction and adduction moment during a shifting squat with heels up from standing position
The shifting squat with heels down from a standing position seen in Figure 4-34 produced similar patterns exhibited in the shifting squat with the heels up. The key difference was that the peak moments for the abduction and adduction were not equivalent.

Shifting the body weight from one side to the other resulted in a frontal moment moving into the abduction and adduction phase in a sinusoidal pattern. The abduction and the adduction moment peaks were $0.05 \frac{Nm}{kg}$ and $0.01 \frac{Nm}{kg}$ respectively.

Figure 4-34: The abduction and adduction moment during a shifting squat with heels down from standing position

Walk then squat and shift trials shown in Figure 4-35 crossed back and forth generating abduction and adduction moments. The heel strike produced a slight adduction moment followed by an abduction moment created by the squat with the heels up. Abduction movement or body weight shift to the left knee generated an abduction moment. Likewise, the adduction movement of the knee generated an almost equivalent adduction moment. Frontal average moment for shifting squat
Figure 4-35: The abduction and adduction moment during a shifting squat with heels up at the end of a gait

with heels up at the end of a gait shown in Figure 4-35 had an average abduction and adduction peak of 0.05 and 0.02 $\frac{Nm}{kg}$ respectively. These average magnitudes were rough estimations due to trying to align moment peaks with different time sequences to execute both the shift and the squat.

Trends similar to those described for the previous squat were seen in these trials with heels down after heel strike. Frontal moment for shifting squat with heels down at the end of a gait shown in Figure 4-36 had an average abduction and adduction peak of 0.05 $\frac{Nm}{kg}$ and a negligible amount respectively. The average curves were difficult to estimate due to the various timing difference of executing the squats. A lot of these misalignments were pronounced in shifting squats. A majority of the knees in this trial was exposed to both abduction and adduction moments during the body shift while squatting.
Figure 4-36: The abduction and adduction moment during a shifting squat with heels down at the end of a gait
4.1.6 Comparison of Abduction-Adduction Knee Moment for Various Squats

A comparative analysis was conducted for isolation on key differences of the various squats. The comparison of the squats with heels being down seen in Figure 4-37

isolated the effects of body shifting and prior actions before a squat for further analysis. The squats without the body weight shifting, only produced adduction moments and had no unloading of the moment when compared to the shifting counterparts. Peak moment magnitudes of squats from a standing position produced a 22% increase in force when compared to the squats performed after walking. The magnitudes for the shifting squats from both standing and walking activities were equivalent.

A similar comparison seen in Figure 4-38 compared the squats with heels up. This comparative graph allowed for a few key observations. Like the previous comparative
Figure 4-38: The abduction and adduction moment of squats with heels up analysis the squats without the body weight shifting, only produced abduction moment. A larger cycling of the moment between the abduction and adduction direction was also prominent in these trials. A notable difference was the peak magnitude for the squats executed after a gait. Squats performed after a gait in both the shifting and non-shifting conditions produced higher peak moments than squat from a standing position. The peak moments for non shifting performed after a gait increased by 31% over the squats performed from a standing position. An increased of 50% in peak moment for the walk then squat with shifting over the stand with shifting squats.

The previous analysis focused on the effects of body shifting while squatting and the influence of standing versus walking prior to squatting. A comparison of the influence of heel position was the other variable isolated in Figure 4-39 for further analysis. A slight higher moment was observed for the heels up shifting squat trials.
However, the non-shifting squats with heels down produced an increase of 31% peak moment over the heels up non-shifting squats. Another notable observation was the larger encroachment of the heels up shifting squat in the adduction direction. Conversely, the non-shifting squats with heels up or down generated only an abduction moment at higher magnitudes than the shifting squat counterparts.

Another analysis was done for the squats performed after a gait, to collaborate the preceding results. The walking then squat trials illustrated in Figure 4-40 isolated the effects of heel position. The shifting squats for both heels up and down were equivalent. However, the heels up for the non-shifting squat produced a 22% higher moment over the heels down trials. Walk then squat trials had more influence when the heels were up. Conversely the stand then squat trials were more influenced when the heels were down.
Figure 4-40: The abduction and adduction moment of walk-squat-walk
4.1.7 Knee Joint Bone on Bone Compressive Force

Once the flexion moment values were determined, the derived compressive force equation was used for each trial condition for comparison and analysis. Calculated results for the gait trial illustrated in Figure 4-41 showed the subjects had compressive loads ranging from 1.6 BW to 2.1 BW for subjects. The average and range of the present data fell within published gait values [1,15,172,189]. Compressive loads started to reduce as the knee flexion reached approximately 30 deg or the initiation phase of the toe-off. This compressive load range seen for the gait trials did not deviate much from loads calculated for many of the squatting sequences. The knee joint compressive loads are summarized in Table 4.5 along with the average for the

![Knee Joint Compressive Force for Gait](image)

Figure 4-41: A compressive force of the knee joint analysis during a normal speed gait.
Squats performed from a standing position with the heels up produced compressive force curves that resemble the flexion moment pattern. The compressive curve shown in the Figure 4-42 increased until it reached the first peak which coincided with the end of the squat. A slight drop in the compressive force could be seen prior to the initiation of the ascent phase. The subjects return to the standing position caused a reduction in compressive force due to a reducing knee flexion angle. This decrease in force continued back to the original force level. Magnitudes between the two types of squatting from a standing position were not statistically significantly different \( p \leq 0.05 \). The compressive loading data summarized in Table 4.6 were slightly lower than some of the publish data \([1,139,160,172,200]\).

Compressive loads for performing a squat from a standing position with heels down produced similar magnitude range when compared to the gait sequence. In particular, the research focused on the compressive loads at the descent and ascent phases of all the squat trials. As the knee flexion or descent phase was initiated, the compressive loads increased and then flatten off from approximately 60 deg to maximum flexion of the subjects. Conversely, the load reduced once the ascent angle reached 60 deg again.
Figure 4-42: The compressive force generated from a standing still position to a squat with the heels up and became unloaded post toe-off phase. This compressive loading pattern exhibited in Figure 4-43 shows the slight rise in the descent and ascent phase of the squat.

The compressive load for squats with heels up produced a similar pattern and magnitude as the squats with heels down at the end of gait. These similarities shown in Figure 4-44 also include the drop to zero after the descent phase. The two squat are summarized and compared in Table 4.7.

Squats with heels down performed at the end of gait sequence also produce similar compressive load magnitudes as the previously discussed squats from a standing position. Majority of the compressive loads for subject illustrated in Figure 4-45 produced a similar pattern seen in the squats from a standing position. However,
Table 4.6: Compressive knee joint force for squats with heels up and down from standing position

<table>
<thead>
<tr>
<th>Squat Type</th>
<th>Identification</th>
<th>Descent (BW)</th>
<th>Ascent (BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Squat with heels down</td>
<td>Subject 1</td>
<td>0.90</td>
<td>0.88</td>
</tr>
<tr>
<td>Squat with heels down</td>
<td>Subject 2</td>
<td>0.95</td>
<td>0.95</td>
</tr>
<tr>
<td>Squat with heels down</td>
<td>Subject 3</td>
<td>0.85</td>
<td>1.01</td>
</tr>
<tr>
<td>Squat with heels down</td>
<td>Subject 4</td>
<td>0.79</td>
<td>0.87</td>
</tr>
<tr>
<td>Squat with heels down</td>
<td>Subject 5</td>
<td>0.85</td>
<td>0.96</td>
</tr>
<tr>
<td>Squat with heels down</td>
<td>Subject 6</td>
<td>0.67</td>
<td>0.94</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td></td>
<td><strong>0.84</strong></td>
<td><strong>0.94</strong></td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 1</td>
<td>0.85</td>
<td>0.90</td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 2</td>
<td>0.81</td>
<td>0.95</td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 3</td>
<td>0.72</td>
<td>1.04</td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 4</td>
<td>0.91</td>
<td>0.98</td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 5</td>
<td>0.83</td>
<td>1.2</td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 6</td>
<td>0.79</td>
<td>0.90</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td></td>
<td><strong>0.82</strong></td>
<td><strong>1.0</strong></td>
</tr>
</tbody>
</table>

occasionally the compressive loads would drop to zero after the descent phase due to a lateral body weight shift toward the left leg.
Figure 4-43: The compressive force generated from a standing still position to a squat with the heels down
Figure 4-44: The compressive force generated from a squat with heels down at the end of gait
Figure 4-45: The compressive force generated from a squat with heels down at the end of gait
Table 4.7: Compressive knee joint force for squats with heels up and down at the end of a gait

<table>
<thead>
<tr>
<th>Squat Type</th>
<th>Identification</th>
<th>Descent (BW)</th>
<th>Ascent (BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walk then squat with heels down</td>
<td>Subject 1</td>
<td>0.82</td>
<td>0.77</td>
</tr>
<tr>
<td>Walk then squat with heels down</td>
<td>Subject 2</td>
<td>1.5</td>
<td>1.56</td>
</tr>
<tr>
<td>Walk then squat heels down</td>
<td>Subject 3</td>
<td>0.81</td>
<td>1.08</td>
</tr>
<tr>
<td>Walk then squat heels down</td>
<td>Subject 4</td>
<td>0.61</td>
<td>1.03</td>
</tr>
<tr>
<td>Walk then squat heels down</td>
<td>Subject 5</td>
<td>1.39</td>
<td>0.62</td>
</tr>
<tr>
<td>Walk then squat heels down</td>
<td>Subject 6</td>
<td>0.85</td>
<td>0.75</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td></td>
<td><strong>1.0</strong></td>
<td><strong>0.97</strong></td>
</tr>
<tr>
<td>Walk then squat with heels up</td>
<td>Subject 1</td>
<td>0.83</td>
<td>0.83</td>
</tr>
<tr>
<td>Walk then squat with heels up</td>
<td>Subject 2</td>
<td>1.50</td>
<td>1.48</td>
</tr>
<tr>
<td>Walk then squat with heels up</td>
<td>Subject 3</td>
<td>1.14</td>
<td>1.13</td>
</tr>
<tr>
<td>Walk then squat with heels up</td>
<td>Subject 4</td>
<td>0.84</td>
<td>1.03</td>
</tr>
<tr>
<td>Walk then squat with heels up</td>
<td>Subject 5</td>
<td>1.20</td>
<td>0.73</td>
</tr>
<tr>
<td>Walk then squat with heels up</td>
<td>Subject 6</td>
<td>0.88</td>
<td>0.88</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td></td>
<td><strong>1.07</strong></td>
<td><strong>1.01</strong></td>
</tr>
</tbody>
</table>
4.1.8 Compressive Forces for Shifting Squat

The patterns exhibited in Figures 4-46 showed that compressive loading cyclically approaching zero. The unloading of the right knee occurred during the shift in the adduction direction. This was the result of weight shifting from the measured right leg to left and back again to the right side. Each successive adduction movement generated higher peak compressive knee joint force than the initial squat. Another markedly difference when compared to the non shifting squats from a standing position, there was an increase in compressive magnitude. This trend and values are listed in Table 4.8 for each subject.

![Knee Joint Compressive Force (Stand then Squat & Shift with Heels Up)](figure)

Figure 4-46: The compressive force generated from a standing still position to a shifting squat with the heels up
Table 4.8: Compressive knee joint force for shifting squats with heels up from standing position

<table>
<thead>
<tr>
<th>Squat Type</th>
<th>ID</th>
<th>Descent (BW)</th>
<th>Adduction (BW)</th>
<th>Abduction (BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shift squat w/ heels up</td>
<td>Subj.#1</td>
<td>1.07</td>
<td>1.15</td>
<td>1.13</td>
</tr>
<tr>
<td>Shift squat w/ heels up</td>
<td>Subj.#2</td>
<td>1.15</td>
<td>1.75</td>
<td>1.73</td>
</tr>
<tr>
<td>Shift squat w/ heels up</td>
<td>Subj.#3</td>
<td>0.95</td>
<td>1.12</td>
<td>1.21</td>
</tr>
<tr>
<td>Shift squat w/ heels up</td>
<td>Subj.#4</td>
<td>0.92</td>
<td>1.40</td>
<td>1.30</td>
</tr>
<tr>
<td>Shift squat w/ heels up</td>
<td>Subj.#5</td>
<td>1.09</td>
<td>1.15</td>
<td>0.96</td>
</tr>
<tr>
<td>Shift squat w/ heels up</td>
<td>Subj.#6</td>
<td>0.94</td>
<td>.98</td>
<td>1.09</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td></td>
<td><strong>1.02</strong></td>
<td><strong>1.26</strong></td>
<td><strong>1.24</strong></td>
</tr>
</tbody>
</table>

Compression force data of the shifting squat with heels down from a standing position presented in Figure 4-47 also showed the higher peak compressive force for subsequent abduction movements. Shifting of the body weight to the left leg or in adduction direction unloaded the reaction compressive force on the right knee joint. Peak compressive force resulted from shifting back to the right leg is listed in Table 4.9.

Table 4.9: Compressive knee joint force for shifting squats with heels down from standing position

<table>
<thead>
<tr>
<th>Squat Type</th>
<th>ID</th>
<th>Descent (BW)</th>
<th>Adduction (BW)</th>
<th>Abduction (BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shift squat w/ heels down</td>
<td>Subj.#1</td>
<td>1.07</td>
<td>1.17</td>
<td>0.80</td>
</tr>
<tr>
<td>Shift squat w/ heels down</td>
<td>Subj.#2</td>
<td>1.08</td>
<td>1.06</td>
<td>1.48</td>
</tr>
<tr>
<td>Shift squat w/ heels down</td>
<td>Subj.#3</td>
<td>0.73</td>
<td>1.00</td>
<td>0.82</td>
</tr>
<tr>
<td>Shift squat w/ heels down</td>
<td>Subj.#4</td>
<td>0.81</td>
<td>1.40</td>
<td>1.40</td>
</tr>
<tr>
<td>Shift squat w/ heels down</td>
<td>Subj.#5</td>
<td>1.24</td>
<td>1.56</td>
<td>1.88</td>
</tr>
<tr>
<td>Shift squat w/ heels down</td>
<td>Subj.#6</td>
<td>0.87</td>
<td>1.3</td>
<td>1.25</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td></td>
<td><strong>0.97</strong></td>
<td><strong>1.25</strong></td>
<td><strong>1.27</strong></td>
</tr>
</tbody>
</table>

The same analysis was also conducted on shifting squats at the end of a gait for each subject. The first knee joint compressive peak force shown in Figure 4.1.8 was
Figure 4-47: The compressive force generated from a standing still position to a shifting squat with the heels up solely due to the heel strike. This initial peak had a similar pattern seen in the gait trials previously discussed. A drop in force magnitude that followed this initial peak occurred when the subject stopped momentarily prior to initiating the squat. The second peak was created by a squat with the heels up. Shifting the body weight generated the third and fourth compressive force peaks. The values for each subject are listed in Table 4.10. These compressive force peaks generated from the shifting of the body weight were higher than the non-shifting squat. The last peak was generated after the subject returned to a standing position and walked off the force platform.

Results for the squat and then shift performed at the end of a gait were similar to the previous trials. Four peaks were generated by a squat which consists of heel strike,
Figure 4-48: The compressive force generated for walk then squat and shift with heels up

squat, shifting motion and walk off activities. The compressive force peak shown in Figure 4.1.8 had higher compressive force peaks than the initial squats. The average compressive force peaks are listed in Table 4.11 confirmed both magnitude and trend seen in the previous shifting squats with heels up. The next section compared the compressive force under various squatting conditions, to analyze the differences.
Table 4.10: Compressive knee joint force for shifting squats with heels up executed after gait

<table>
<thead>
<tr>
<th>Squat Type</th>
<th>ID</th>
<th>Descent (BW)</th>
<th>Adduction (BW)</th>
<th>Adduction (BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shift squat w/ heels up</td>
<td>Subj.#1</td>
<td>0.97</td>
<td>1.08</td>
<td>1.29</td>
</tr>
<tr>
<td>Shift squat w/ heels up</td>
<td>Subj.#2</td>
<td>1.77</td>
<td>2.04</td>
<td>1.86</td>
</tr>
<tr>
<td>Shift squat w/ heels up</td>
<td>Subj.#3</td>
<td>1.09</td>
<td>1.02</td>
<td>1.04</td>
</tr>
<tr>
<td>Shift squat w/ heels up</td>
<td>Subj.#4</td>
<td>0.93</td>
<td>1.42</td>
<td>1.36</td>
</tr>
<tr>
<td>Shift squat w/ heels up</td>
<td>Subj.#5</td>
<td>1.28</td>
<td>1.30</td>
<td>1.42</td>
</tr>
<tr>
<td>Shift squat w/ heels up</td>
<td>Subj.#6</td>
<td>0.73</td>
<td>1.13</td>
<td>1.14</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td></td>
<td><strong>1.12</strong></td>
<td><strong>1.33</strong></td>
<td><strong>1.35</strong></td>
</tr>
</tbody>
</table>

Table 4.11: Compressive knee joint force for shifting squats with heels down executed after gait

<table>
<thead>
<tr>
<th>Squat Type</th>
<th>ID</th>
<th>Descent (BW)</th>
<th>Adduction (BW)</th>
<th>Abduction (BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shift squat w/ heels down</td>
<td>Subj.#1</td>
<td>0.91</td>
<td>1.07</td>
<td>1.3</td>
</tr>
<tr>
<td>Shift squat w/ heels down</td>
<td>Subj.#2</td>
<td>1.9</td>
<td>1.73</td>
<td>2.0</td>
</tr>
<tr>
<td>Shift squat w/ heels down</td>
<td>Subj.#3</td>
<td>0.93</td>
<td>1.08</td>
<td>1.46</td>
</tr>
<tr>
<td>Shift squat w/ heels down</td>
<td>Subj.#4</td>
<td>0.72</td>
<td>1.28</td>
<td>1.23</td>
</tr>
<tr>
<td>Shift squat w/ heels down</td>
<td>Subj.#5</td>
<td>1.40</td>
<td>1.01</td>
<td>-</td>
</tr>
<tr>
<td>Shift squat w/ heels down</td>
<td>Subj.#6</td>
<td>0.75</td>
<td>1.05</td>
<td>1.10</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td></td>
<td><strong>1.10</strong></td>
<td><strong>1.20</strong></td>
<td><strong>1.42</strong></td>
</tr>
</tbody>
</table>
Figure 4-49: The compressive force generated for walk then squat and shift with heels down
4.1.9 Comparison of Compressive Forces for Squats

All of the average compressive knee joint forces for both the gait and various types of non shifting squats were compared to discover any differences in magnitude and patterns. The Figure 4-50 compared average compressive force force for gait and all types of squats performed with the heels up. The compressive force from the gait was significantly larger than the non shifting squats performed at the end of the gait or from a standing position. In particular, the descent and ascent compressive forces of the squatting performed at the end of a gait were 1.5 times lower than the gait average single limb stance compressive force. The squat from a standing position average compressive was 1.39 times lower than the gait trails. There was a 34% drop in when the squat with shifting from a standing position when compared to the gait compressive force. However, the average compressive force for squat with shifting at the end of a gait walking was 7% lower than the single limb stance compressive force. Squats from a standing position were approximately 25% higher than the compressive forces from the squats executed at the end of a gait.

Similar results can be seen in Figure 4-51 when squats with the heels down are compared. Compressive forces from the squats that performed after a gait produced similar percentage drop when compared to the gait trial results. The squats executed after a gait were within 3% of the squats performed from a standing position. Shifting squats executed at the end of a gait and from a standing position produced 10% and 23% lower compressive forces respectively than the gait trials.

All the shifting squats seen in the Figure 4-52 were plotted together to isolate and analyze the influence of heels up versus heels down. There was no significant difference between the heels up and down for shifting squats perform at the end of a gait. The same observation was seen for non shifting squats for both heels on and off the ground. Similar results were confirmed when comparing the influence of heels
Figure 4-50: The compressive force generated for squats with the heels up position for non shifting squats seen in Figure 4-53.
Figure 4-51: The compressive force generated for squats with the heels down
Figure 4-52: The compressive force generated for squats with the heels down
Figure 4-53: The compressive force generated for squats with the heels up
4.1.10 Shear Forces for Squats

This research also focused on the anterior-posterior shear force on the knee joint after exploring the moments and compressive forces. The knee joint shear force illustrated in Figure 4-54 had a pattern resembling the gait moment and compressive forces. Shear force values had the same pattern and magnitude as published data [1,15]. Shear forces in the gait trial displayed graphically in Figure 4-54 were directed posteriorly. These posterior shear forces that are summarized in Table 4.12 are comparable to publish data in terms force direction also [1,15].

Figure 4-54: The shear force generated from a standing still position to a squat with the heels down

The squats with the heels up from a standing position, shown in Figure 4-55 had the shear forces directed posteriorly. The descent phase into the squatting process
Figure 4-55: The shear force generated from a standing still position to a squat with the heels up
drove the shear forces toward the anterior direction. Only one subject shear force changed from the posterior direction to an anterior load. The magnitudes for each subject are listed in Table 4.12 for side by side comparison for the squats with heels down. One key difference between published data and these results was the change of force direction from posterior to anterior during the squatting process.

Shear force analysis for squats with heels down performed from a standing position graphically illustrated in Figure 4-56 slightly differed than publish data [15]. A majority of the subjects’ shear forces stayed in the posterior direction throughout the squatting sequence. However, the shear forces started to shift slightly toward the anterior direction as the knee reached maximum flexion for half of the subjects. The
Table 4.12: Shear knee joint force for squats with heels up and down from standing position

<table>
<thead>
<tr>
<th>Squat Type</th>
<th>ID</th>
<th>Descent (BW)</th>
<th>Ascent (BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Squat with heels down</td>
<td>Subject 1</td>
<td>0.20</td>
<td>0.22</td>
</tr>
<tr>
<td>Squat with heels down</td>
<td>Subject 2</td>
<td>0.19</td>
<td>0.10</td>
</tr>
<tr>
<td>Squat with heels down</td>
<td>Subject 3</td>
<td>0.20</td>
<td>0.23</td>
</tr>
<tr>
<td>Squat with heels down</td>
<td>Subject 4</td>
<td>0.26</td>
<td>0.14</td>
</tr>
<tr>
<td>Squat with heels down</td>
<td>Subject 5</td>
<td>0.26</td>
<td>0.27</td>
</tr>
<tr>
<td>Squat with heels down</td>
<td>Subject 6</td>
<td>0.27</td>
<td>0.25</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td></td>
<td><strong>0.23</strong></td>
<td><strong>0.19</strong></td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 1</td>
<td>0.22</td>
<td>0.13</td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 2</td>
<td>0.20</td>
<td>0.26</td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 3</td>
<td>0.28</td>
<td>0.25</td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 4</td>
<td>0.25</td>
<td>0.12</td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 5</td>
<td>0.25</td>
<td>0.13</td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 6</td>
<td>0.14</td>
<td>0.21</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td></td>
<td><strong>0.22</strong></td>
<td><strong>0.18</strong></td>
</tr>
</tbody>
</table>

Shear force went from a posterior to an anterior direction which produced a very small anterior force.

A key difference which is listed in Table 4.12, was the higher magnitude for squats with the heels down from the standing position. The trend corresponded with higher compressive forces generated from the same squats. The peak forces occurred during either the descent or ascent phase of the squats. This indicated the subtle weight shift during either of those phases.

Squat with heels up after a gait produced results that merged aspects of both the gait and squats from a standing position. The heel strike illustrated in Figure 4-57, generated a posterior shear force. A second peak shear force accompanied the descent into a squat. In general, the descent phase drove the shear force magnitudes toward zero. Initiation of the ascent phase generated the third shear force peak in the squatting process. Last peak was generated by the toe off phase after returning to a
The shear force generated from a standing still position to a squat with the heels down. There were a few subjects that had a definitive force directional change from posterior to anterior. These magnitudes are listed in Table 4.13 for comparison to the heel position down.

Walking then squatting with heels down graphically illustrated in Figure 4-58 was very comparable to publish data [15]. The same protocol was done as the publish data being referenced [15]. Results of this of squatting showed a definitive change of direction from posterior to anterior of the shear force. Magnitude of the shear forces were statistically the same as the squats from a standing position and equivalent to published data [15].
Figure 4-57: The shear force generated from a squat with heels down at the end of gait
Figure 4-58: The compressive force generated from a squat with heels down at the end of gait
Table 4.13: Shear knee joint force for squats with heels up and down executed at end of gait

<table>
<thead>
<tr>
<th>Squat Type</th>
<th>Identification</th>
<th>Descent (BW)</th>
<th>Ascent (BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walk-Squat with heels down</td>
<td>Subject 1</td>
<td>0.14</td>
<td>0.27</td>
</tr>
<tr>
<td>Walk-Squat with heels down</td>
<td>Subject 2</td>
<td>0.17</td>
<td>0.20</td>
</tr>
<tr>
<td>Walk-Squat with heels down</td>
<td>Subject 3</td>
<td>-0.33</td>
<td>0.31</td>
</tr>
<tr>
<td>Walk-Squat with heels down</td>
<td>Subject 4</td>
<td>-0.07</td>
<td>0.30</td>
</tr>
<tr>
<td>Walk-Squat with heels down</td>
<td>Subject 5</td>
<td>0.32</td>
<td>0.27</td>
</tr>
<tr>
<td>Walk-Squat with heels down</td>
<td>Subject 6</td>
<td>0.26</td>
<td>0.21</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td></td>
<td><strong>0.22</strong></td>
<td><strong>0.27</strong></td>
</tr>
<tr>
<td>Squat with heels up</td>
<td>Subject 1</td>
<td>0.26</td>
<td>0.24</td>
</tr>
<tr>
<td>Walk-Squat with heels up</td>
<td>Subject 2</td>
<td>0.28</td>
<td>0.31</td>
</tr>
<tr>
<td>Walk-Squat with heels up</td>
<td>Subject 3</td>
<td>0.28</td>
<td>0.26</td>
</tr>
<tr>
<td>Walk-Squat with heels up</td>
<td>Subject 4</td>
<td>0.20</td>
<td>0.22</td>
</tr>
<tr>
<td>Walk-Squat with heels up</td>
<td>Subject 5</td>
<td>0.36</td>
<td>0.25</td>
</tr>
<tr>
<td>Walk-Squat with heels up</td>
<td>Subject 6</td>
<td>0.30</td>
<td>0.20</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td></td>
<td><strong>0.28</strong></td>
<td><strong>0.25</strong></td>
</tr>
</tbody>
</table>
4.1.11 Shear Forces for Shifting Squats

The shear force generated from a standing still position to a shifting squat with the heels up (Figure 4-59), was similar to the previous squatting sequence. Similar pattern in terms of magnitude and direction from posterior to anterior were the same as the squatting with the heels down at the end of the gait. These shifting squats had the same influence on the shear force magnitude and direction as the non shifting squats.

Figure 4-59: The shear force generated from a standing still position to a shifting squat with the heels up

The shifting squat with heels up from a standing position shown in Figure 4-59, was similar to the previous squatting sequence. Similar pattern in terms of magnitude and direction from posterior to anterior were the same as the squatting with the heels down at the end of the gait. These shifting squats had the same influence on the shear force magnitude and direction as the non shifting squats.

Shear forces for squats with heels down while executing a body weight shift from the right heel to the left heel is represented in Figure 4-60. For a majority of the subjects, the shear remained in a posterior direction throughout the shifts. One notable occurrence was a change of direction from the posterior toward the anterior
Figure 4-60: The shear force generated from a standing still position to a shifting squat with the heels down during a shift to the left heel. Shear force magnitude ranges were similar to previous squats from a standing position.

Squatting with heels up and producing a body weight shift to the left heel at the end of a gait is represented in Figure 4-61. Some of the shear forces changed from posterior to anterior direction during the shifting phase of the body weight. Shear force magnitudes for this protocol in exhibited a slightly higher force than the heels down squatting sequence.

Shear force for shifting squat with the heels down executed at the end of a gait is illustrated in Figure 4-62. For a majority of the subjects, the shear remained in a posterior direction throughout the shifts. Same previous occurrence seen in shifting
Figure 4-61: The shear force generated from a shifting squat with heels down at the end of gait squats was change of direction from the posterior toward the anterior during a shift to the left heel. Shear force magnitudes range were similar to previous squats from a standing position.
Figure 4-62: The shear force generated from a shifting squat with heels down at the end of gait
Chapter 5

Discussion

There are three cases in particular that can be applied broadly and also have relatively a respectable correlation with knee osteoarthritis. Excessive weight, age, and job functions had a higher correlation for knee OA. As mentioned earlier, the Western countries BMI are much higher than the Asian counterparts which cannot explain the difference in knee OA prevalence. This overloading of the knee joints exists not only in the obese group but rural area jobs that where lifting excessive loads is performed constantly. Interestingly the osteoarthritis also shows a strong connection with jobs that require an extraordinary amount of squatting or kneeling which required higher knee flexion angle. Bio-mechanical loading of the knee joint over decades in the average person causes wear of the knee joint remains unclear; however, the infliction of knee OA to the older population has a higher prevalence. Age is not a factor that can be pointed to as a solution to the reason why the Asian culture has a higher knee osteoarthritis prevalence. The other contributing factors to knee OA such as the genetics, knee alignment, previous alignment, diseases, physical activities and gender are difficult to distinguish their weighted factor for either of the two cultures. The basis of this research was to understand which particular biomechanical loading activity and wear of the knee joint can lead to osteoarthritis.

However, the daily squatting exhibited in the Asian culture is one of the most
glaring biomechanical activity difference seen between the Western and Eastern culture. What makes this activity so much more severe compared to other activities such as a running, gait or standing is the drastic reduction in cartilage contact area. High contact stress resulted at higher flexion due to the reduction of the cartilage [188]. High loading and fatigue cycling are two phenomena that possibly can precipitate the disintegration of the cartilage especially when the contact area has been reduced. The major forces in the sagittal plane, such as the moment, compression and shear forces, had a high magnitude or cycling loading for the various squats. Additionally, the moment in the coronal plane was also observed to indicate if either the cyclical or high loading occurred.

The various of squats generated forces that are comparable to loads from gait activities. The flexion moment for all of squats which occurred at maximum knee angles generated higher moments than gait trials. Gait trials provide an excellent benchmark comparison for squatting primarily due to the volume of available research. This common activity exposes the knee cartilage to potential wear and tear. Flexion moments for the gait produced lower values especially when compared to the shifting squat trials. Published study theorized that the magnitude of the moment and cyclical shear force difference during squatting at the end of a gait as a key reason for knee osteoarthritis [15]. A similar result was also seen in Figure 4-58 and Figure 4-13 of the present study. The results of the current study closely matching publish data validated the moment, compression and shear forces equations for the squatting trials. There was ANOVA statistical significance (P<.005) of compressive loading during the shifting squats. Although statistically the shifting squat influence on the moment was not significantly different; however, the trend indicates that larger sample size could reverse the results. A similar result existed for the squats performed at the end of a gait. The statistical significance difference between the squats performed with heels up or down were negligible; however, squats with heels up had higher moment
magnitudes. Higher moment magnitudes were seen for squats with heels up were seen throughout the entire research. The vertical ground reaction forces of the squats with heels up and down were equivalent. This implied the moment arm distance played a key role in the magnitude differences. Relationship between the vertical ground reach forces and the moment arm length was analyzed for each subject. This analysis focused on the anterior-posterior shear force, vertical ground reaction force, and the corresponding moments for the non shifting squats with heels up and down. The non shifting squats with heels up shown in the Figure 5-1 were slightly higher in the squats from a standing position. The magnitude difference was apparent in the squats performed at the end of the gait. Plots of the vertical ground reaction force and the moment arm length showed the relationship between the various squats with heels up and down. Moment arm length was measured from the center of pressure (COP) to the lateral condyle of the right leg both vertically and anterior-posteriorly. Moment arm length for the heels up for a squat from a standing position seen in Figure 5-2 achieved a peak of 0.35 m. This corresponded with a knee vertical reaction force of almost one times the body weight. Squatting with heels down had a shorter moment arm length for equivalent knee vertical reaction force. The difference in moment arm length was based on the physical degrees of freedom the heels placed upon the subjects. When the heels are up, the subject can rock more anteriorly thus resulting in a much larger moment arm length. Conversely, the heels down limited the anterior motion of the knee and moved slightly posteriorly during the descent phase of the squat. Additionally, the squats with heels up place the thigh more horizontal which maintained the increased moment arm length throughout the entire process. These same physical degrees of freedom limitation can be seen in Figure 5-3 for the anterior-posterior shear force graph. Heels up allowed the subjects knees to move anteriorly which resulted in the knee lowering closer to the ground. Squat with the heels down produced a shallower drop when compared to heels up condition. This moment arm
length played the role in differences seen in the abduction moment with heels up versus heels down. Initially the knee shear force was anteriorly directed and at the apex of the squat changed to a posteriorly direction for both squat conditions. Squats performed at the end of a gait had similar results seen previously. Moment arm length illustrated in Figure 5-4 for the squats with heels up executed at the end of a gait showed the disparity in distances between the two types of conditions. Additionally, the shallower dip associated with the heels down seen in Figure 5-5 exhibited the same trend as the squats from a standing position. However, the knee shear forces started posteriorly and reversed direction to anterior at the apex of the squatting process.

Figure 5-1: Four Types of non shifting squats for subject 1
Figure 5-2: Knee joint vertical force and moment arm length for squats from standing position
Figure 5-3: Knee joint anterior posterior force for non shifting squats from standing position
Figure 5-4: Knee joint vertical reaction force for non shifting squats performed at end of gait
Figure 5-5: Knee joint vertical reaction force for non shifting squats performed at end of gait
The knee joint moments for the squat with heels up and down of the second subject shown in Figure 5-6 had slight magnitude differences. Squats from the standing position had a larger moment magnitude. Once more a more detailed look at the how the moment arm length influenced this magnitude was conducted. Moments arm length shown in Figure 5-7 for the squats with heels up was larger due to the ability of the knee to move anteriorly. These results were very similar to the previously reviewed standing squat trials. Shear force for the standing squats illustrated in Figure 5-8 initially was posterior then changed anteriorly. The same shallow dip seen in the previous shear force graphs was present here.

The knee joint moment graph showed that there was a slight difference between the squats performed at the end of a gait with heels up and down. The two moment arm

Figure 5-6: Four Types of non shifting squats for subject 1
lengths in the Figure 5-9 were also equivalent. This relationship strengthen the theory on what drove the moment magnitude difference between the two heel condition was the moment arm length. Walk then squat trials shear forces were initiated posteriorly and changed toward the anterior direction at the apex of the squat before returning back to the posterior orientation. As shown in Figure 5-10, the shallow dip of the squat with heels down also continued.
Figure 5-8: Knee joint anterior posterior force for non shifting squats from standing position
Figure 5-9: Knee joint vertical reaction force for non shifting squats performed at end of gait
Figure 5-10: Knee joint vertical reaction force for non shifting squats performed at end of gait
The following graphs (Figure 5-11 to Figure 5-30) of the cohorts reiterated the point of how reliant and how influential the moment arm length was on the difference between the magnitudes from the squats with heels up and down. Most of the subject generated a shear force that initiated in the posterior direction and reversed during squatting phase anteriorly prior to returning back to the posterior orientation.

Figure 5-11: Four Types of non shifting squats for subject 1
Figure 5-12: Knee joint vertical force and moment arm length for squats from standing position
Figure 5-13: Knee joint anterior posterior force for non shifting squats from standing position
Figure 5-14: Knee joint vertical reaction force for non shifting squats performed at end of gait
Figure 5-15: Knee joint vertical reaction force for non shifting squats performed at end of gait.
Figure 5-16: Four Types of non shifting squats for subject 1
Figure 5-17: Knee joint vertical force and moment arm length for squats from standing position
Figure 5-18: Knee joint anterior posterior force for non shifting squats from standing position
Figure 5-19: Knee joint vertical reaction force for non shifting squats performed at end of gait
Figure 5-20: Knee joint vertical reaction force for non shifting squats performed at end of gait
Figure 5-21: Four Types of non shifting squats for subject 1
Figure 5-22: Knee joint vertical force and moment arm length for squats from standing position
Figure 5-23: Knee joint anterior posterior force for non shifting squats from standing position
Figure 5-24: Knee joint vertical reaction force for non shifting squats performed at end of gait
Figure 5-25: Knee joint vertical reaction force for non shifting squats performed at end of gait
Figure 5-26: Four Types of non shifting squats for subject 1

The statistical analysis result above accounted for moments generated during periods of ascending and descending. Shifting squats in particular generated fluctuated flexion moments that on occasions begin to generate extension moment. A statistical analysis looked at the significance of standing, walking, heels up, heels down and body weight shifting squats. Shifting squats were statistically significantly \((P<0.005)\) different from the other types of squat in terms of the magnitude of cyclical flexion moments.

A more in depth look at data generated from one of the subjects in Table 5.1 provides a broad perspective of how moments, compressive loading and shear forces behaved for the various biomechanical activities. Moments were typically higher for
the squatting trials than walking. While the squats performed from a standing position produced slightly higher moments than the gait trials, a vast majority of the other types of squat moments were significantly higher. Only the shifting squats showed a cyclical moment loading and almost complete unloading. This cyclical moment exposed the small contact area in the knee during the natural shifting while in a squat. The sagittal plane was exposed to a higher moment magnitude and also exhibited a cyclical loading at a time when the contact stress is exceptionally high. Moment magnitudes in the coronal plane were not as high as the flexion moment; however, a similar cyclical loading pattern was observed in the shifting squats. Additionally, the moments were significantly higher than the other types of squat and
gait trials. A cyclical moment in the coronal plane was present in the shifting squat trials. Another potential mechanical load that can produce a cyclical loading is the compression or vertical loading on the knee joint.

Compression load of the knee during the squatting generated were roughly equivalent to the values generated from the gait trials. The higher moments from the squatting trials would have indicated an expected comparative increase in the compression loading. The lower compression loading was primarily due to the lower body segment acceleration which resulted in the unexpectedly force equivalent to the gait trials with lower moments. Shifting squats did produce a more pronounced cyclical loading and unloading when compared to the gait trials. Compression loading still

Figure 5-28: Knee joint anterior posterior force for non shifting squats from standing position
confirmed a cyclical loading, and unlike the gait cycle this occurred at a much smaller contact area. Shear force was the last load analyzed to understand the magnitude, any cyclical behavior and potential link to knee osteoarthritis.

Shear loading mostly directed posteriorly of the knee joint for gait activity was very similar to publish data [1,15]. Magnitudes of the shear load of the gait cycle were not significantly different than the squatting trials [1,15]. Direction of the shear load did change toward the anterior side for shifting squats and squats that maintained heel contact on the ground for a few of the subjects. The magnitude for the shifting squats at the end of the gait with heels down generated higher forces and cyclical loading.
A material that is constantly exposed to a fatigue loading becomes vulnerable to potential failure. Knee osteoarthritis is not a disease that occurs instantly from overloading or impact but deterioration marked by a slow wearing and continual biomechanical compromising. Shifting while squatting data in this research showed how a cyclical loading occurred in the knee joint in the flexion moment, compressive force, shear force and the coronal moment. The knee joint was exposed to these forces at a high angle flexion where the contact area was quite small. This begins to shed light of why research shows the fatigue cycles or repetitive loading have very little correlation between runners and knee OA [207]. The cartilage contact area is maximized during running and is one of the reasons for the low correlation to knee OA.
The cartilage contact area absorbs much of the cyclical loading from running due to a larger volume and small knee flexion angle. In the squats, the cartilage contact area is significantly reduced while exposed to higher moments and lower compressive force. Similar phenomenon can be seen in the strong correlation of obesity and knee OA which is typically characterized as a load overwhelming a too small of an area. A shifting squat not only produced a higher moments but also equivalent impact forces seen in a gait activity with a smaller contact area. Non shifting squats although produced slightly higher moments when compared to gait cycles; however, did not expose the knee joint to the devastating cyclical loading.

Table 5.1: Compressive knee joint force for shifting squats with heels up executed after gait

<table>
<thead>
<tr>
<th>ID</th>
<th>Compressive Force (BW)</th>
<th>Flexion Moment (Nm/kg)</th>
<th>Adduction Moment (Nm/kg)</th>
<th>A/P Shear Force (BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 1</td>
<td>0.90</td>
<td>0.27</td>
<td>0.15</td>
<td>0.26</td>
</tr>
<tr>
<td>0.6</td>
<td>3</td>
<td>1.09</td>
<td>1.02</td>
<td>1.04</td>
</tr>
<tr>
<td>0.2</td>
<td>2</td>
<td>0.93</td>
<td>1.42</td>
<td>1.36</td>
</tr>
<tr>
<td>0.5</td>
<td>2</td>
<td>1.28</td>
<td>1.30</td>
<td>1.42</td>
</tr>
<tr>
<td>0.7</td>
<td>2</td>
<td>0.73</td>
<td>1.13</td>
<td>1.14</td>
</tr>
<tr>
<td>1</td>
<td>23</td>
<td>1.12</td>
<td>1.33</td>
<td>1.35</td>
</tr>
</tbody>
</table>
Conclusion

Within the research community interest in the potential link between biomechanical factors associated with squatting and osteoarthritis has been increasing. This activity requires a high knee flexion which results in a smaller cartilage area being exposed to the forces and moment generated from the weight of the body. Asian populations perform many types of squatting in daily life and also have a higher prevalence of knee OA. This higher prevalence of knee osteoarthritis and designing artificial implants to meet patients needs for an increase in knee flexion have spurred research in this area. The research on a knee joint that undergoes a squat and the potential ramifications are limited. While this study provides useful information; however, this simulation represents a person performing a squat to retrieve an object on the ground. In the Asian culture, people performing squats is a common daily occurrence. This type of squatting typically, requires a body shift from one from leg to the other for comfort and relief from stiffness. A major focus of this research was to conduct a comprehensive investigation on the biomechanics of the different types of squats. This comprehensive investigation may help to discern if a squat can produce detrimental forces or loads that could initiate knee osteoarthritis. A central premise of this research was people stay in a squat position much longer than test simulations. Additionally, these prolonged squats had frequent shifting of the body for comfort or
completing a task. This shifting in a prolong squat would continuously generate a cyclical loading on the knee joint. A clearer understanding of biomechanical loading can potentially show the connection with the initiation of knee osteoarthritis. These possible links to knee OA may lead to mitigation of the biomechanical risk factors and also pave a way to improving the design of future knee implants.

One of the main issues identified early on in the research process was the ability of the subjects to perform squatting that generated high knee flexion exhibited in the Asian countries. These squats were simulated in this study for estimation of the loading and behavior of the knee response. This discrepancy was more evident when comparing knee flexion and the ability to squat while maintaining the heels on the ground throughout the entire process. There were two subjects that were able to execute the squats without any issues and achieved flexion moments that were significantly higher than the rest of the group. The effect of squats that had contact between the calf and back of the thigh in terms of contribution remains unknown. Skin marker artifact was another obstacle that was difficult to control [93-95]. To help with this a cluster of markers arranged around a bony landmark of interest was used and significantly reduced movement of the markers [93-95].

A majority of the squatting that was done generated flexion moments that tremendously exceeded the moments observed from the gait trials. The higher moments did not result in higher compressive loads primarily due to the lower shank acceleration during the squatting phase. The shear force between the tibia and femur occasionally generated a cyclical loading for a few of the squats. Shifting squat generated the most prominent cyclical shear loading. This shear cyclical loading can also be seen reflected in the squatting flexion moment. The shifting moment in particular produced a high cyclical magnitude moment in the frontal plane. This moment in the abduction and adduction direction reflects the shift of body weight. Another observation that was seen in the shifting squats was the compressive force loading
and unloading of the knee joint. This cyclical compressive force produced similar to the impacts force absorbed during gait. It is the assertion of this research study that the cyclical loading on the cartilage material can potentially lead the onset of osteoarthritis. Going into a squat requires a high flexion angle, and results in a posterior small contact area primarily due to the shape of the condyles. When this small cartilage contact area is exposed to cyclical compressive, shear, flexion moment and coronal moment at the same time, wear is inevitable. The knee joint cartilage is exposed to impact loading from the compressive force while a shear force fluctuates between anterior and posterior direction. When a high contact stress area is exposed to cyclical anterior-posterior loading, adduction-abduction moment, and compressive loading, the cartilage matrix structure is overwhelmed and become damage.
References


[2] Knee,
http://www.pediatric-orthopedics.com/Topics/Bones/Knee/knee.html

[3] Sports Injury Clinic,
http://www.sportsinjuryclinic.net/cybertherapist/kneeanatomy.php


[9] American Academy of Orthopaedic Surgeons,
http://orthoinfo.aaos.org/topic.cfm?topic=a00399

[10] Centers for Disease Control Prevention,
http://www.cdc.gov/chronicdisease/resources/publications/AAG/arthritis.htm


[22] Sarzi-Puttini, Piercarlo., Cimmino, Marco.A., Scarpa, Raffaele., Capo-


[32] **Akizuki, S., Mow, VC., Muller, F., Pita, JC., Howell, DS.**, Biomechanical


Vrezas I, Elsner G, Bolm-Audorff U, Abolmaali N, Seidler A, Case-control study of knee osteoarthritis and lifestyle factors considering their in-


OConnor, Mary I., Advanced glycation end products increases matrix
metalloproteinase-1,-3,and-13 and TNF-a in human osteoarthritic chondrocytes.


[111] Jones, G., Ding, C., Scott, F., Glisson, M., Cicuttini, FM., Early radiographic osteoarthritis is associated with substantial changes in cartilage volume and tibial bone surface area in both males and females Osteoarthritis Cartilage, 2004, 12(2), 169–74.


[119] Segal, Neil A., Zimmerman, Bridget M., Brubaker, M., Obesity and Knee Osteoarthritis Are Not Associated With Impaired Quadriceps Specific Strength in Adults Arthritis & Rheumatism, 1999, 42(1), 17–24. 178 said 64


[122] Narayan, K., Thomas, G., Kumar, R., Is extreme flexion of the knee after

[123] Liu, F., Ohdera, T., Miyamoto, H., Wasielewski, R. C., et al., In vivo kinematic determination of total knee arthroplasty from squatting to standing *The Knee*, 2009, 16(2), 116–120.


[132] Chaudhari, Ajit M. W., Briant, Paul L., Bevill, Scott L., Koo, Seungbum., Andriacchi, Thomas P., Knee Kinematics, Cartilage Morphology, and Os-


187


[170] Li, S., Micheletti, R., Role of Diet in Rheumatic Disease Rheum Dis Clin N Am., 2011, 37, 119–133.


[207] Chakravarty, Eliza F., Hubert, Helen B., Fries, James F., Long Distance Running and Knee Osteoarthritis A Prospective Study American Journal of Preventive Medicine, 2008, 35(2), 133–138

Appendix A

Biomechanics Derivation

The center mass location of the shank is calculated with the following equation A.1 [186]. Similar equations were also utilized to calculate the center of mass location seen in Figure 3.2.

\[ CG_y = \frac{RMM_y + RLM_y}{2} + 0.567 \times (RLTP_y - \frac{RMM_y + RLM_y}{2}) \] (A.1)

\[ CG_z = \frac{RMM_z + RLM_z}{2} + 0.567 \times (\frac{RLFC_z + RLTP_z}{2} + \frac{RMFC_z + RMTP_z}{2} - \frac{RMM_z + RLM_z}{2}) \] (A.2)

\[ CG_x = \frac{RMM_x + RLM_x}{2} + 0.567 \times (RLTP_x - \frac{RMM_x + RLM_x}{2}) \] (A.3)

Once the position of the COM is known then the velocity and acceleration can be determined. The following equation calculates the velocity and acceleration during the testing protocol.
In equation A.6 the $\Delta t^2$ is the digital video frame rate which is 100 Hz. Similar equation is utilized to calculate the accelerations along the other axis.

\[
a_{xi} = \frac{CGx_{i+1} - 2 \times CGx_i + CGx_{i-1}}{\Delta t^2} \quad (A.4)
\]

\[
a_{zi} = \frac{CGz_{i+1} - 2 \times CGz_i + CGz_{i-1}}{\Delta t^2} \quad (A.5)
\]

\[
a_{yi} = \frac{CGy_{i+1} - 2 \times CGy_i + CGy_{i-1}}{\Delta t^2} \quad (A.6)
\]

Passive marker attached to anatomical landmarks allow the knee flexion to be calculated during the testing protocol. The subsequent equations were used to determine the knee flexion at each phase.

\[
Thigh_{angle} = \arctan\left(\frac{GT_z - RLFC_z}{GT_x - RLMFC_x + RLFCC_x}\right) \times \frac{180}{\pi} \quad (A.7)
\]

\[
Tibia_{angle} = \arctan\left(\frac{RLTP_z - RLM_z}{RMTLP_x + RLTP_x - RMMPP_x + RLM_x}\right) \times \frac{180}{\pi} \quad (A.8)
\]

\[
Knee_{flexion} = Thigh_{angle} - Tibia_{angle} \quad (A.9)
\]

The angular position of the center of mass during the various activities is equal to the tibia angle calculated in equation A.8. The same acceleration formula previous used to determine the shank center of mass acceleration is utilized again. The angular acceleration is determined in equation A.10.

\[
\alpha_{CG} = \frac{CG\theta_{i+1} - 2 \times CG\theta_i + CG\theta_{i-1}}{\Delta t^2} \quad (A.10)
\]
Table A.1: Reflective marker definition and anatomical locations

<table>
<thead>
<tr>
<th>Marker Name</th>
<th>Anatomical Landmark</th>
</tr>
</thead>
<tbody>
<tr>
<td>RT</td>
<td>Right Greater Trochanter</td>
</tr>
<tr>
<td>RASIS</td>
<td>Right Anterior Superior Iliac Spine</td>
</tr>
<tr>
<td>LASIS</td>
<td>Left Anterior Superior Iliac Spine</td>
</tr>
<tr>
<td>RPSIS</td>
<td>Right Posterior Superior Iliac Spine</td>
</tr>
<tr>
<td>LPSIS</td>
<td>Left Posterior Superior Iliac Spine</td>
</tr>
<tr>
<td>SACRUM</td>
<td>Sacrum</td>
</tr>
<tr>
<td>RAPT</td>
<td>Right Anterior Proximal Thigh</td>
</tr>
<tr>
<td>RPPT</td>
<td>Right Posterior Proximal Thigh</td>
</tr>
<tr>
<td>RPDT</td>
<td>Right Posterior Distal Thigh</td>
</tr>
<tr>
<td>RLFC</td>
<td>Right Lateral Femoral Epicondyle</td>
</tr>
<tr>
<td>RMFC</td>
<td>Right Medial Femoral Epicondyle</td>
</tr>
<tr>
<td>RLTP</td>
<td>Right Lateral Condyle of Tibial</td>
</tr>
<tr>
<td>RMTP</td>
<td>Right Lateral Condyle of Tibial</td>
</tr>
<tr>
<td>RPPAT</td>
<td>Right Patella Anterior Proximal Tendon</td>
</tr>
<tr>
<td>RPDAT</td>
<td>Right Patella Anterior Distal Tendon</td>
</tr>
<tr>
<td>RTIBTUB</td>
<td>Right Tibial Tuberosity</td>
</tr>
<tr>
<td>RPPS</td>
<td>Right Posterior Proximal Shank</td>
</tr>
<tr>
<td>RPDS</td>
<td>Right Posterior Distal Shank</td>
</tr>
<tr>
<td>RAPS</td>
<td>Right Anterior Proximal Shank</td>
</tr>
<tr>
<td>RADS</td>
<td>Right Anterior Distal Shank</td>
</tr>
<tr>
<td>RHEEL</td>
<td>Right Calcaneus</td>
</tr>
<tr>
<td>RLF</td>
<td>Right Fifth Metatarsal</td>
</tr>
<tr>
<td>RMF</td>
<td>Right First Metatarsal</td>
</tr>
<tr>
<td>RFORE</td>
<td>Right First Distal Metatarsal</td>
</tr>
<tr>
<td>RLM</td>
<td>Right Lateral Malleolus</td>
</tr>
<tr>
<td>RMM</td>
<td>Right Medial Malleolus</td>
</tr>
</tbody>
</table>
Appendix B

BCP Program

The BCP is a program utilizing the capability of the Mathcad software to conduct theoretical estimations of the moments, anterior-posterior and vertical forces. This program also calculated the center of gravity of the leg segments and moment are distances which are base on knee flexion angles. The program begins by importing the anatomical land markers position data captured by high speed digital cameras for gaits and squats movements. A variety of equations are then employed to solve the moments and forces for each of the activity movements.
Figure B-1: BCP raw data input from force plate and marker sensors
Figure B-2: BCP input data array
Figure B-3: Continued BCP input data array
Figure B-4: BCP knee flexion angle equations
Equations for the center of gravity location of the leg and inertia

\[ CO_X = \text{RM}_X + 0.56 \left( \frac{\text{LM}_X + \text{RM}_X}{2} - \text{RM}_X \right) \]

\[ CO_Y = \frac{\text{RM}_Y + \text{LM}_Y}{2} + 0.56 \left( \frac{\text{RM}_Y + \text{LM}_Y}{2} - \frac{\text{RM}_Y + \text{LM}_Y}{2} \right) \]

Location of the center mass of the leg

\[ CO_k = \mathbf{I}_{ib_k} \]

\[ \mathbf{c}_o = \left( \begin{array}{c} c_{1ib} \ c_{2ib} \end{array} \right) \]

Figure B-5: Equations for the center of gravity location of the leg and inertia
External Forces Data

\[ F_{X_{\text{external}}} = GRF_x - \frac{k_{\text{smooth}}(\text{Time},CGX_{\text{accel}},2)}{m_{\text{shk}}} m_{\text{shk}} \]  \hspace{1cm} \text{anterior posterior direction}

\[ F_{Y_{\text{external}}} = GRF_y - \frac{k_{\text{smooth}}(\text{Time},CGY_{\text{accel}},2)}{1000} m_{\text{shk}} \]  \hspace{1cm} \text{medial lateral direction}

\[ F_{Z_{\text{external}}} = GRF_z - \frac{CGZ_{\text{accel}}}{1000} m_{\text{shk}} - m_{\text{shank}} \]  \hspace{1cm} \text{vertical direction}

Figure B-6: Knee reaction force calculations
Figure B-7: Flexion moment calculations
Figure B-8: Flexion moment curve comparisons of raw data, smoothing of raw data, V3D, and knee angle
Figure B-9: Patella tendon line of action and moment arm equations
Figure B-10: Patella tendon line of action and moment arm for rest of subjects
Figure B-11: Patella force equations
Figure B-12: Anterior-Posterior knee force equations
Figure B-13: Compressive knee force equations
Figure B-14: Shear and compressive force graphs
Figure B-15: Shear and compressive force graphs
Figure B-16: Shear and compressive force graphs
Figure B-17: Shear and compressive force graph of all subjects