MUSCLE CONTRIBUTIONS TO KNEE LOADS IN HEALTHY YOUNG WOMEN

MUSCLE FUNCTION CONTRIBUTIONS TO KNEE LOADING DURING STATIC AND DYNAMIC ACTIVITIES IN HEALTHY YOUNG WOMEN

By

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ABSTRACT

Lower extremity muscles are primary contributors to knee loading during static and dynamic activities. The underlying philosophy of this thesis contends that a more robust understanding of the relationships between muscle function and the loading environment in the healthy knee is needed for clinicians to prescribe effective lower extremity muscle training programs. The primary objectives were to (i) identify changes in knee mechanics and muscle function during gait, static squats and lunges after lower extremity neuromuscular fatigue, (ii) explore relationships between knee loading, and peak knee extensor and flexor strength and power, and (iii) identify isometric standing muscle training postures which elicit minimal medial knee loads.

In the studies addressing objective (i), isotonic lower extremity fatigue reduced the peak knee extension moment during gait, decreased vastus lateralis activation, and decreased mean knee moments during static lunging. These findings indicated that fatigue altered muscle recruitment strategies and may have concurrently altered the knee loading environment during these submaximal tasks. However, these biomechanical alterations are not consistent with those in knee osteoarthritis or ligament injury. In the studies addressing objective (ii), peak knee extension torque and power were important in describing healthy knee loading during the early and late stance phases of gait, respectively. Finally in the studies addressing objective (iii), isometric muscle training postures were identified which targeted quadriceps and hamstrings muscle activation without overloading the medial compartment of the knee. Muscle activation levels required to increase lower limb torque production capacity can be achieved by

performing squats and lunges and these postures elicit lower knee moments than those experienced during gait.

This work expands the understanding of healthy knee function and the relationships between muscle function and mechanical knee loading. These insights will contribute toward the future creation of muscle training programs and guidelines for knee OA or ACL injury.

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LIST OF ABBREVIATIONS AND SYMBOLS

ACL	Anterior cruciate ligament
ADL	Activities of daily living
ANOVA	Analysis of variance
BF	Biceps Femoris
BL	Baseline
BMI	Body mass index
CI	Co-activation index
CMR	Common mode rejection
EMG	Electromyography
EMG_{AG}	Magnitude of electromyography signal from the agonist muscle
EMG_{ANT}	Magnitude of electromyography signal from the antagonist muscle
Ext Lat Ang	Extended Lateral Angle
Ext	Knee extension
Flex	Knee flexion
KAM	Knee adduction moment
KAM1	First peak of the knee adduction moment during gait
KAM2	Second peak of the knee adduction moment during gait
KFM	Knee flexion moment
KOOS	Knee Injury and Osteoarthritis Outcome Score
MPF	Median power frequency
MVC	Maximum voluntary contraction
MVIC	Maximum voluntary isometric contraction
PF1	Post fatigue 1
PF2	Post fatigue 2
RF	Rectus Femoris
RMS	Root mean square
SD	Standard deviation
SENIAM	Surface Electromyography for the Non-Invasive Assessment of a Muscle
VL	Vastus Lateralis

DECLARATION OF ACADEMIC ACHIEVEMENT

The following summarizes all author contributions to each manuscript in this thesis.

For all four manuscripts: Heather Longpré developed the research questions, applied for and obtained ethics approval, designed and executed the protocols, managed and supervised all interactions with study participants, managed all data collection, storage and transfer, conducted and lead analysis of the data, and prepared all manuscripts for publication. Dr. Monica Maly advised on all research questions, protocols, data and statistical analyses, and interpretation of the study findings, and assisted with editing each manuscript.

For Chapter 2, titled "Biomechanical Changes at the Knee after Lower Limb Fatigue in Healthy Young Women": Dr. Jim Potvin assisted with refining the research question, interpreting the study findings, and editing the manuscript.

For Chapter 3, titled "Muscle Activation during Squatting and Lunging after Lower Extremity Fatigue in Healthy Young Women": Dr. Stacey Acker assisted with refining the research question and editing the manuscript. Dr. Acker also assisted during data collection when more than one study operator was required.

For Chapter 4, titled "Identifying Yoga-Based Strengthening Exercises for Knee Osteoarthritis with Minimal Medial Knee Loads": Ayesha Johnson, as the MacMobilize laboratory Research Assistant, assisted with data collection when more than one study operator was required and with compiling raw data for analysis which was then conducted by Heather Longpré.

Chapter One

Introduction

1.1 Motivation

The motivation for this thesis comes from a desire to understand the mechanisms for knee loading during various static and dynamic activities, with a particular focus on the contributions of muscle function. In doing so, this work explores muscle function using measures beyond lower extremity strength to develop a more robust understanding of the role of knee musculature in joint loading.

This work contributes to the body of knowledge surrounding healthy knee function. A clear understanding of the mechanisms supporting healthy knee function during load bearing activities must be ascertained before identifying potential risk factors for, or abnormalities associated with, knee injuries or joint degeneration. It is through the study of individuals with healthy, pain free knees that we may gain the fundamental knowledge necessary for planning treatment and rehabilitation protocols for those with injured or diseased joints.

1.2 Key Topics

1.2.1 The Knee Adduction Moment

Loading of the tibiofemoral joint during human movement has received considerable attention to promote an understanding of the mechanical factors contributing to the onset and progression of pathologies of the knee joint. The magnitude and distribution of load between the medial and lateral compartments during ambulation relate to degenerative knee disease (Foroughi, Smith, & Vanwanseele, 2009), joint injury (Hewett et al., 2005), and healthy aging (Kerrigan, Todd, Della Croce, Lipsitz, & Collins, 1998). External knee moments are often calculated through inverse dynamics using measures of ground reaction forces and moments, estimates of anthropometric variables, and limb motion collected during three-dimensional motion analysis (Winter, 2009). For example, medial knee load can be estimated by the external knee adduction moment (KAM). The KAM refers to the moment tending to rotate the tibia toward varus alignment on the femur and is primarily derived from the three-dimensional ground reaction forces acting on the foot and the inertial characteristics of the lower extremity. A KAM is produced when the vertical ground reaction force vector passes medially to the knee centre and will increase with a greater ground reaction force and/or moment arm, that is, the perpendicular distance between the force vector and the knee joint centre.

The KAM during gait generally consists of a two-peak waveform, the first peak occurring in early stance shortly after heel-strike, and the second peak occurring toward terminal stance shortly before toe-off. The shape of this curve is driven by the shape of the vertical ground reaction force and may change with age, knee ligament injury, or knee OA pathology (Chen et al., 2003; Hewett et al., 2005; Mundermann, Dyrby, & Andriacchi, 2005). A large magnitude first peak of the KAM curve during gait is related to increased knee osteoarthritis (OA) severity (Sharma et al., 1998) and often occurs after anterior cruciate ligament (ACL) injury (Butler, Minick, Ferber, & Underwood, 2009;

Hewett et al., 2005). Meanwhile, a reduced or indistinguishable second peak KAM during gait has been reported in studies of both healthy aging and knee OA (Chen et al., 2003; Hurwitz, Ryals, Case, Block, & Andriacchi, 2002). While it is clear that knee loading patterns are important in aging, knee OA, and ACL injury, there is a gap in understanding the key factors which contribute to and alter these loads in healthy knees.

1.2.2 Muscle Activation and Coactivation

External loads at the knee must be counterbalanced by internal knee loads, which are equal in magnitude but opposite in direction of action. These internal loads are produced by the muscles, soft tissues, and other joint structures, though the individual contributions of these structures are unclear and vary between individuals (Bennell, Wrigley, Hunt, Lim, & Hinman, 2013). Nevertheless, biomechanical modeling has demonstrated that active muscle contributions from several muscle groups in conjunction with passive supports are necessary to maintain dynamic stability at the knee (Schipplein & Andriacchi, 1991). Force contributions of the muscles that may provide support to the knee while the joint undergoes loads are of particular interest because they can be improved by targeted muscle training (Fransen & McConnell, 2008; Roddy, Zhang, & Doherty, 2005). Improving the force tolerance of the knee muscles may protect the joint against soft-tissue damage when undergoing strenuous activities (Bennell et al., 2013).

Lower extremity muscle activation patterns during human activities are of interest as they provide insights into the muscular contributions during different tasks. Muscle activation during static and dynamic tasks can be measured through electromyography (EMG), which captures the electrical potential generated by neurological activity.

Activation changes in the muscles surrounding the knee following a lower extremity exercise program are unclear and have been investigated primarily in the quadriceps (Narici, Roi, Landoni, Minetti, & Cerretelli, 1989; Thorstensson, Karlsson, Viitasalo, Luhtanen, & Komi, 1976).

Muscle coactivation refers to the simultaneous activation of two or more muscles, typically an agonist and an antagonist, around a joint. Muscle coactivation is required to perform many activities. For example, during gait, muscle coactivation between the quadriceps and hamstrings is required to maintain equilibrium and counteract the KAM (Buchanan & Lloyd, 1997; Lloyd & Buchanan, 1996; Schipplein & Andriacchi, 1991). Coactivation of opposing muscle groups around the knee reduces the total net moment acting at the joint and increases the resistance to angular joint motion (Busse, Wiles, & van Deursen, 2006; Heiden, Lloyd, & Ackland, 2009). While increasing coactivation between the quadriceps and hamstrings may produce higher net knee compressive forces (Knarr, Zeni, & Higginson, 2012; Messier et al., 2011), selective coactivation may also even the relative distribution of load between the medial and lateral compartments of the knee by activating the muscles with moments arms that counteract these loads (Palmieri-Smith, McLean, Ashton-Miller, & Wojtys, 2009). Likewise, coactivation of the quadriceps and hamstrings may be beneficial in mediating the loads sustained by the ligaments supporting the knee, minimizing the risk for ligament injuries such as anterior cruciate ligament (ACL) tears (O'Connor, 1993). Therefore, in this thesis, higher coactivation between the quadriceps and hamstrings is considered to be favourable in healthy, young individuals.

1.2.3 Muscle Strength

Lower extremity muscle strength refers to the maximum force or torque generating capability of muscles. Deficits in strength refer to maximum voluntary force or torque productions that are lower than the expected capacity of a muscle for a given age, sex, and health or disease state. Deficits in quadriceps strength are associated with increased pain and disability in knee OA (Brandt et al., 2000; O'Reilly, Jones, Muir, & Doherty, 1998; Shakoor, Furmanov, Nelson, Li, & Block, 2008), while lower extremity muscle strength preserves bone density, quality of life, and reduces the risk for chronic diseases such as heart disease and diabetes by improving glycemic control (Seguin & Nelson, 2003). Since muscle strength varies with body size, affecting both the maximum force and the moment arm of the force, it is recommended that the absolute strength value be normalized to body mass (kg) (Bennell et al., 2013). Lower extremity muscle strength can be improved with targeted muscle training (Lange & Vanwanseele, 2008; Pelland et al., 2004).

Strength of the knee extensor muscles is typically almost twice that of the knee flexors in healthy women (Calmels, Nellen, van der Borne, Jourdin, & Minaire, 1997; Pincivero, Gandaio, & Ito, 2003). Maximum torque output of both the knee extensors and flexors significantly decreases with healthy aging (Lindle et al., 1997; Murray, Duthie, Gambert, Sepic, & Mollinger, 1985). Quadriceps weakness compared to healthy controls is prevalent in ACL injury and knee OA and may be caused by increased pain, body mass, joint effusion, muscle atrophy, or a reduced ability to activate muscle fibres (Lewek, Rudolph, & Snyder-Mackler, 2004; O'Reilly et al., 1998; Palmieri-Smith, Thomas, & Wojtys, 2008; Rice, McNair, & Lewis, 2011; Segal & Glass, 2011; Slemenda et al., 1997). Large magnitude torque output of the quadriceps is associated, in general, with higher muscle activation amplitude measured using EMG (Alkner, Tesch, & Berg, 2000a; Bilodeau, Schindler-Ivens, Williams, Chandran, & Sharma, 2003; Karlsson & Gerdle, 2001), however this relationship is not well established for the hamstrings muscles (Mohamed, Perry, & Hislop, 2002).

An understanding of the association between lower extremity muscle weakness and knee loading is important given the role of the muscles in supporting joint stability through a variety of physical activities. While knee loading rates during gait were higher for individuals with lower quadriceps strength in 37 healthy women (Mikesky, Meyer, & Thompson, 2000), little association existed between quadriceps strength and the peak knee adduction moment during gait in individuals with knee OA (Calder et al., 2013; Lim et al., 2009). Quadriceps weakness is related to a reduced external knee flexion moment and a reduced knee flexion angle during gait after ACL reconstruction (Lewek, Rudolph, Axe, & Snyder-Mackler, 2002). Multiple randomized controlled studies have shown that strengthening knee muscles does not alter the peak KAM in women with knee OA (Bennell et al., 2014; Foroughi et al., 2011; Lim, Hinman, Wrigley, Sharma, & Bennell, 2008), which suggests that individuals who develop muscle weakness through age, pathology, or injury could maintain their knee biomechanics, possibly through altered muscle recruitment patterns.

1.2.4 Alternative Measures of Muscle Function beyond Strength

While much attention has been given to the role of quadriceps strength and weakness in the development and progression of knee pathologies and injuries, knowledge gaps exist around alternative measures of muscle function. The thesis adds to the body of knowledge surrounding muscle contributions to knee loading by investigating other measures of muscle function beyond strength. These include neuromuscular fatigue, hamstrings activation and coactivation, and knee extensor and flexor power.

Neuromuscular fatigue of the quadriceps, defined as a reduction in the torque producing capabilities of the muscle group, may occur after long durations and high repetitions of an activity, and some studies have suggested that altered knee loads may result (Murdock & Hubley-Kozey, 2012; Thomas, McLean, & Palmieri-Smith, 2010). It is important to explore muscle dysfunction in young healthy individuals to develop a clearer understanding of the loading changes experienced by the knee after prolonged engagement in physical activity, muscle training or lower extremity rehabilitation.

Much of the knee literature regarding lower extremity muscles has focused on the quadriceps, particularly the relationships between quadriceps weakness and knee injury and disease. However, the hamstrings, which may be overlooked by other studies, likely produce meaningful contributions to knee loading patterns during static and dynamic activities, and may be important to consider in aging, pathology, or injury as a contributor to quadriceps-hamstrings coactivation. Forces produced by the hamstrings during knee motion substantially reduce the load sustained by the ACL and reduce tibial motion against the femur (Herzog & Read, 1993). This finding indicates the importance of studying the role of hamstrings in static and dynamic exercises to reduce the risk for ligament injuries. This thesis aims to evaluate the hamstrings contributions to different activities by considering the activation and the coactivation between the quadriceps and hamstrings.

Power is defined as the rate of rotational torque development through a range of motion. Knee extensor power is an important indicator of functional ability assessed during activities such as walking and stair climbing in healthy older adults (Bean et al., 2002). Bean et. al (2002), evaluated 45 mobility-limited adults and found that knee power explained the variance in these physical performance measures better than knee strength. Furthermore, knee extensor power is a key contributor to knee loads during gait for individuals with knee pathology, explaining a significant portion of the variance in the first KAM peak in 53 adults with clinical knee OA (Calder et al., 2013). These studies suggest that knee extensor power could be an important measure of muscle function, however no studies have evaluated the contributions of knee flexor power to knee loading and physical performance.

1.3 Objectives

The purpose of this work was to explore how muscle function influences the knee loading environment, and whether these altered knee loading patterns mimic, and potentially identify the risk for, knee pathology. Contributions of lower extremity neuromuscular fatigue, muscle activation and coactivation, strength and power to measures of knee loads were investigated in young healthy women. Specifically, this thesis aimed to:

1) Identify changes in knee kinematics and kinetics, and quadriceps and hamstrings muscle activations during gait before and after neuromuscular fatigue of the knee flexors and extensors.
2) Evaluate knee mechanics and quadriceps and hamstrings muscle activation during squatting and lunging before and after lower extremity neuromuscular fatigue.
3) Examine the relationships between measures of medial knee loading during gait with knee flexor and extensor strength and power.
4) Identify standing yoga postures which elicit higher muscle activations with minimal medial compartment knee loads.

1.4 Benefits and Impact

The most prominent implication of this work is an increased understanding of the role of the lower extremity musculature in mediating load distribution across the knee during static and dynamic tasks. These findings are important for individuals who have or are at risk of developing knee pathologies, those with knee injuries, and those who participate in lower extremity muscle training.

1.5 Thesis Organization and Structure

This thesis has been prepared in the manuscript format as detailed in the *Guide for the Preparation of Theses* developed by the School of Graduate Studies for McMaster

University. This thesis is comprised of six chapters, containing four original manuscripts that have either been published or submitted for publication in peer-reviewed journals. These manuscripts are preceded by an introductory chapter and are followed by a discussion and conclusions chapter. For chapters 2, 3, and 4, data were collected from the same convenience sample of 25 individuals from the university population, while for chapter 5; data were collected from a different convenience sample of 30 individuals from the same population.

Chapter two presents a published manuscript exploring the effects of lower extremity fatigue on knee biomechanics during gait in healthy young women. Knee kinematics and kinetics, and leg muscle activations data during gait, before and after neuromuscular fatigue of the quadriceps and hamstrings, were collected and analyzed. The results of this study capture the biomechanical changes at the knee after fatigue during a submaximal dynamic task.

Chapter three contributes new evidence on the effects of lower extremity neuromuscular fatigue on muscle activation and the KAM during activities with higher demands on lower extremity muscles compared to gait. This study analyzes biomechanical changes in the lower limbs before and after fatigue during typical lower body muscle training exercises. These findings also offer insights about which muscle training exercise better challenged the muscles supporting the knee and the loads experienced by the joint during squatting and lunging.

Chapter four explores how muscle strength and power are related to medial compartment knee loading, and whether the relationship is different between knee flexors and knee extensors. This manuscript examines the relationships between the peaks of the KAM during gait and knee flexor and extensor strength and power. The findings quantify the contributions of muscle strength and power to the variance in the peaks of the KAM during gait in healthy young women.

Chapter five focuses on the development of a yoga-based exercise program for lower extremity muscle training that minimizes overloading at the knee. Knee kinetics and quadriceps and hamstrings activations are examined during gait and various standing yoga postures. Yoga postures eliciting higher muscle activations with minimal KAMs in healthy young women are identified. The knee loads experienced during the yoga postures are also compared with the peak experienced during gait.

The concluding chapter summarizes the major findings and impact of each manuscript as they apply to the overarching thesis goal. This summary is followed by discussions of the key innovations and implications, major limitations and constraints, directions for future research, and final personal reflections.

1.6 References

Alkner, B. A., Tesch, P., & Berg, H. E. (2000). Quadriceps EMG/force relationship in knee extension and leg press. Medicine and Science in Sports and Exercise, 32(2), 459-463.

Bean, J. F., Kiely, D. K., Herman, S., Leveille, S. G., Mizer, K., Frontera, W. R., & Fielding, R. A. (2002). The relationship between leg power and physical performance in Mobility-Limited older people. Journal of the American Geriatrics Society, 50(3), 461-467.

Bennell, K. L., Kyriakides, M., Metcalf, B., Egerton, T., Wrigley, T. V., Hodges, P. W., Hunt, M. A., Roos, E. M., Forbes, A., Ageberg, E., & Hinman, R. S. (2014). Neuromuscular versus quadriceps strengthening exercise in patients with medial knee osteoarthritis and varus malalignment: A randomized controlled trial. Arthritis & Rheumatology, 66(4), 950-959.

Bennell, K. L., Wrigley, T. V., Hunt, M. A., Lim, B., & Hinman, R. S. (2013). Update on the role of muscle in the genesis and management of knee osteoarthritis. Rheumatic Diseases Clinics of North America, 39(1), 145-176.

Bilodeau, M., Schindler-Ivens, S., Williams, D. M., Chandran, R., & Sharma, S. S. (2003). EMG frequency content changes with increasing force and during fatigue in the quadriceps femoris muscle of men and women. Journal of Electromyography and Kinesiology, 13(1), 83-92. doi:10.1016/S1050-6411(02)00050-0

Brandt, K. D., Heilman, D. K., Slemenda, C., Katz, B. P., Mazzuca, S., Braunstein, E. M., & Byrd, D. (2000). A comparison of lower extremity muscle strength, obesity, and depression scores in elderly subjects with knee pain with and without radiographic evidence of knee osteoarthritis. The Journal of Rheumatology, 27(0315-162; 0315-162; 8), 1937-1946.

Buchanan, T. S., & Lloyd, D. G. (1997). Muscle activation at the human knee during isometric flexion-extension and varus-valgus loads. Journal of Orthopaedic Research, 15(1), 11-17.

Busse, M. E., Wiles, C. M., & van Deursen, R. W. (2006). Co-activation: Its association with weakness and specific neurological pathology. Journal of Neuroengineering and Rehabilitation, 3, 26. doi:10.1186/1743-0003-3-26

Butler, R. J., Minick, K. I., Ferber, R., & Underwood, F. (2009). Gait mechanics after ACL reconstruction: Implications for the early onset of knee osteoarthritis. British Journal of Sports Medicine, 43(5), 366-370.

Calder, K. M., Acker, S. M., Arora, N., Beattie, K. A., Callaghan, J. P., Adachi, J. D., & Maly, M. R. (2013). Knee power is an important parameter in understanding medial knee joint load in knee osteoarthritis. Arthritis Care & Research, 66(5), 687-694.

Calmels, P. M., Nellen, M., van der Borne, I., Jourdin, P., & Minaire, P. (1997). Concentric and eccentric isokinetic assessment of flexorextensor torque ratios at the hip, knee, and ankle in a sample population of healthy subjects. Archives of Physical Medicine and Rehabilitation, 78(11), 1224-1230.

Chen, C. P., Chen, M. J., Pei, Y., Lew, H. L., Wong, P., & Tang, S. F. (2003). Sagittal plane loading response during gait in different age groups and in people with knee osteoarthritis. American Journal of Physical Medicine & Rehabilitation, 82(4), 307-312.

Foroughi, N., Smith, R. M., Lange, A. K., Baker, M. K., Fiatarone Singh, M. A., & Vanwanseele, B. (2011). Lower limb muscle strengthening does not change frontal plane moments in women with knee osteoarthritis: A randomized controlled trial. Clinical Biomechanics, 26(2), 167-174.

Foroughi, N., Smith, R., & Vanwanseele, B. (2009). The association of external knee adduction moment with biomechanical variables in osteoarthritis: A systematic review. The Knee, 16(5), 303-309. doi:10.1016/j.knee.2008.12.007

Fransen, M., & McConnell, S. (2008). Exercise for osteoarthritis of the knee. Cochrane Database of Systematic Reviews (Online), (4)(4), CD004376. doi:10.1002/14651858.CD004376.pub2

Heiden, T. L., Lloyd, D. G., & Ackland, T. R. (2009). Knee extension and flexion weakness in people with knee osteoarthritis: Is antagonist cocontraction a factor? The Journal of Orthopaedic and Sports Physical Therapy, 39(11), 807-815. doi:10.2519/jospt.2009.3079

Herzog, W., & Read, L. J. (1993). Lines of action and moment arms of the major force-carrying structures crossing the human knee joint. Journal of Anatomy, 182(Pt 2), 213-230.

Hewett, T. E., Myer, G. D., Ford, K. R., Heidt, R. S., Colosimo, A. J., McLean, S. G., van den Bogert, A. J., Paterno, M. V., & Succop, P. (2005). Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: A prospective study. The American Journal of Sports Medicine, 33(4), 492-501.

Hurwitz, D. E., Ryals, A. B., Case, J. P., Block, J. A., & Andriacchi, T. P. (2002). The knee adduction moment during gait in subjects with knee osteoarthritis is more closely correlated with static alignment than radiographic disease severity, toe out angle and pain. Journal of Orthopaedic Research: Official Publication of the Orthopaedic Research Society, 20(0736-0266; 1), 101-107.

Karlsson, S., & Gerdle, B. (2001). Mean frequency and signal amplitude of the surface EMG of the quadriceps muscles increase with increasing torque—a study using the continuous wavelet transform. Journal of Electromyography and Kinesiology, 11(2), 131-140.

Kerrigan, D. C., Todd, M. K., Della Croce, U., Lipsitz, L. A., & Collins, J. J. (1998). Biomechanical gait alterations independent of speed in the healthy elderly: Evidence for specific limiting impairments. Archives of Physical Medicine and Rehabilitation, 79(3), 317-322.

Knarr, B. A., Zeni, J. A., Jr, & Higginson, J. S. (2012). Comparison of electromyography and joint moment as indicators of co-contraction. Journal of Electromyography and Kinesiology: Official Journal of the International Society of Electrophysiological Kinesiology, doi:10.1016/j.jelekin.2012.02.001

Lange, A. K., & Vanwanseele, B. (2008). Strength training for treatment of osteoarthritis of the knee: A systematic review. Arthritis Care & Research, 59(10), 1488-1494.

Lewek, M., Rudolph, K., Axe, M., & Snyder-Mackler, L. (2002). The effect of insufficient quadriceps strength on gait after anterior cruciate ligament reconstruction. Clinical Biomechanics, 17(1), 56-63.

Lewek, M. D., Rudolph, K. S., & Snyder-Mackler, L. (2004). Quadriceps femoris muscle weakness and activation failure in patients with symptomatic knee osteoarthritis. Journal of Orthopaedic Research: Official Publication of the Orthopaedic Research Society, 22(1), 110-115.

Lim, B., Hinman, R. S., Wrigley, T. V., Sharma, L., & Bennell, K. L. (2008). Does knee malalignment mediate the effects of quadriceps strengthening on knee adduction moment, pain, and function in medial knee osteoarthritis? A randomized controlled trial. Arthritis Care & Research, 59(7), 943-951.

Lim, B., Kemp, G., Metcalf, B., Wrigley, T. V., Bennell, K. L., Crossley, K. M., & Hinman, R. S. (2009). The association of quadriceps strength with the knee adduction moment in medial knee osteoarthritis. Arthritis Care & Research, 61(4), 451-458.

Lindle, R., Metter, E., Lynch, N., Fleg, J., Fozard, J., Tobin, J., Roy, T. A., Hurley, B. (1997). Age and gender comparisons of muscle strength in 654 women and men aged 20–93 yr. Journal of Applied Physiology, 83(5), 1581-1587.

Lloyd, D., & Buchanan, T. (1996). A model of load sharing between muscles and soft tissues at the human knee during static tasks. Journal of Biomechanical Engineering, 118(3), 367-376.

Messier, S. P., Legault, C., Loeser, R. F., Van Arsdale, S. J., Davis, C., Ettinger, W. H., & DeVita, P. (2011). Does high weight loss in older adults with knee osteoarthritis affect bone-on-bone joint loads and muscle forces during walking? Osteoarthritis and Cartilage, 19(3), 272-280. doi:10.1016/j.joca.2010.11.010

Mikesky, A. E., Meyer, A., & Thompson, K. L. (2000). Relationship between quadriceps strength and rate of loading during gait in women. Journal of Orthopaedic Research, 18(2), 171-175.

Mohamed, O., Perry, J., & Hislop, H. (2002). Relationship between wire EMG activity, muscle length, and torque of the hamstrings. Clinical Biomechanics, 17(8), 569-579.

Mundermann, A., Dyrby, C. O., & Andriacchi, T. P. (2005). Secondary gait changes in patients with medial compartment knee osteoarthritis: Increased load at the ankle, knee, and hip during walking. Arthritis and Rheumatism, 52(0004-3591; 9), 2835-2844.

Murdock, G., & Hubley-Kozey, C. (2012). Effect of a high intensity quadriceps fatigue protocol on knee joint mechanics and muscle activation during gait in young adults. European Journal of Applied Physiology, 112(2), 439-449. doi:10.1007/s00421-011-1990-4

Murray, M. P., Duthie, E. H., Gambert, S. R., Sepic, S. B., & Mollinger, L. A. (1985). Age-related differences in knee muscle strength in normal women. Journal of Gerontology, 40(3), 275-280.

Narici, M., Roi, G., Landoni, L., Minetti, A., & Cerretelli, P. (1989). Changes in force, cross-sectional area and neural activation during strength training and detraining of the human quadriceps. European Journal of Applied Physiology and Occupational Physiology, 59(4), 310-319.

O'Connor, J. J. (1993). Can muscle co-contraction protect knee ligaments after injury or repair? The Journal of Bone and Joint Surgery.British Volume, 75(1), 41-48.

O'Reilly, S. C., Jones, A., Muir, K. R., & Doherty, M. (1998). Quadriceps weakness in knee osteoarthritis: The effect on pain and disability. Annals of the Rheumatic Diseases, 57(10), 588-594.

Palmieri-Smith, R. M., McLean, S. G., Ashton-Miller, J. A., & Wojtys, E. M. (2009). Association of quadriceps and hamstrings cocontraction patterns with knee joint loading. Journal of Athletic Training, 44(3), 256-263. doi:10.4085/1062-6050-44.3.256; 10.4085/1062-6050-44.3.256

Palmieri-Smith, R. M., Thomas, A. C., & Wojtys, E. M. (2008). Maximizing quadriceps strength after ACL reconstruction. Clinics in Sports Medicine, 27(3), 405-424. doi:10.1016/j.csm.2008.02.001

Pelland, L., Brosseau, L., Wells, G., MacLeay, L., Lambert, J., Lamothe, C., Robinson, V., Tugwell, P. (2004). Efficacy of strengthening exercises for osteoarthritis (part I): A meta-analysis. Physical Therapy Reviews, 9(2), 77-108.

Pincivero, D. M., Gandaio, C. B., & Ito, Y. (2003). Gender-specific knee extensor torque, flexor torque, and muscle fatigue responses during maximal effort contractions. European Journal of Applied Physiology, 89(2), 134-141.

Rice, D. A., McNair, P. J., & Lewis, G. N. (2011). Mechanisms of quadriceps muscle weakness in knee joint osteoarthritis: The effects of prolonged vibration on torque and muscle activation in osteoarthritic and healthy control subjects. Arthritis Research & Therapy, 13(5), R151. doi:10.1186/ar3467

Roddy, E., Zhang, W., & Doherty, M. (2005). Aerobic walking or strengthening exercise for osteoarthritis of the knee? A systematic review. Annals of the Rheumatic Diseases, 64(4), 544-548.

Schipplein, O., & Andriacchi, T. (1991). Interaction between active and passive knee stabilizers during level walking. Journal of Orthopaedic Research, 9(1), 113-119.

Segal, N. A., & Glass, N. A. (2011). Is quadriceps muscle weakness a risk factor for incident or progressive knee osteoarthritis? The Physician and Sportsmedicine, 39(4), 44-50. doi:10.3810/psm.2011.11.1938

Seguin, R., & Nelson, M. E. (2003). The benefits of strength training for older adults. American Journal of Preventive Medicine, 25(3), 141-149.

Shakoor, N., Furmanov, S., Nelson, D. E., Li, Y., & Block, J. A. (2008). Pain and its relationship with muscle strength and proprioception in knee OA: Results of an 8-week home exercise pilot study. Journal of Musculoskeletal & Neuronal Interactions, 8(1), 35-42.

Sharma, L., Hurwitz, D. E., Thonar, E. J., Sum, J. A., Lenz, M. E., Dunlop, D. D., Schnitzer, T. J., Kirwan-Mellis, G., & Andriacchi, T. P. (1998). Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis. Arthritis and Rheumatism, 41(0004-3591; 7), 1233-1240.

Slemenda, C., Brandt, K. D., Heilman, D. K., Mazzuca, S., Braunstein, E. M., Katz, B. P., & Wolinsky, F. D. (1997). Quadriceps weakness and osteoarthritis of the knee. Annals of Internal Medicine, 127(0003-4819; 0003-4819; 2), 97-104.

Thomas, A. C., McLean, S. G., & Palmieri-Smith, R. M. (2010). Quadriceps and hamstrings fatigue alters hip and knee mechanics. Journal of Applied Biomechanics, 26(2), 159-170.

Thorstensson, A., Karlsson, J., Viitasalo, J., Luhtanen, P., & Komi, P. (1976). Effect of strength training on EMG of human skeletal muscle. Acta Physiologica Scandinavica, 98(2), 232-236.

Winter, D. A. (2009). Biomechanics and motor control of human movement (4th ed.). Hoboken, N.J.: Wiley.

Chapter Two

Biomechanical Changes at the Knee after Lower Limb Fatigue in Healthy Young Women

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This manuscript presents an evaluation of the biomechanical changes experienced at the knee during gait after lower limb fatigue in healthy young women. The biomechanical measures evaluated include knee moments and angles, and quadriceps and hamstrings activations and co-activation. This study contributes to the overall understanding of knee loading changes during a submaximal dynamic activity after functional impairment of quadriceps and hamstrings peak torque. More challenging, static activities will be explored in later manuscripts, as will other measures of muscle function beyond peak torque.

2.1 Abstract

Background: The purpose of this study was to identify changes in knee kinematics, kinetics and stiffness that occur during gait due to lower limb neuromuscular fatigue.

Methods: Kinematic, kinetic and electromyographic measures of gait were collected on healthy, young women (n=20) before and after two bouts of fatigue. After baseline gait analysis, two bouts of fatiguing contractions were completed. Fatigue was induced using sets of 50 isotonic knee extensions and flexions at 50% of the peak torque during a maximum voluntary isometric contraction. Fatigue was defined as a drop in knee extension or flexion maximum voluntary isometric torques of at least 25% from baseline. Gait analyses were completed after each bout of fatigue. Dynamic knee stiffness was calculated as the change in knee flexion moment divided by the change in knee flexion angle from 3-15% of the gait cycle. Co-activations of the biceps femoris and rectus femoris muscles were calculated from 3-15% and 40-52% of gait. Repeated measures analyses of variance assessed differences in discrete gait measures, knee torques, and electromyography amplitudes between baseline and after each bout of fatigue.

Findings: Fatigue decreased peak isometric torque. Fatigue did not alter knee adduction moments, knee flexion angles, dynamic knee stiffness, or muscle co-activation. Fatigue reduced the peak knee extension moment.

Interpretation: While neuromuscular fatigue of the knee musculature alters the sagittal plane knee moment in healthy, young women during walking, high intensity fatigue is not consistent with known mechanical environments implicated in knee pathologies or injuries.

Keywords: Neuromuscular Fatigue, Gait, Knee Biomechanics, Knee Osteoarthritis, Dynamic Knee Stiffness, Electromyography

2.2 Introduction

A decrease in the strength of the quadriceps, defined as a reduction in maximum torque generating capacity (Enoka, 2002), is thought to alter knee biomechanics and impair the ability to mediate loads across the knee (Lewek, Rudolph, & Snyder-Mackler, 2004; Rice, McNair, & Lewis, 2011; Slemenda et al., 1997; Slemenda et al., 1998; J. A. Zeni, Rudolph, & Higginson, 2009). For example, an elevated peak knee adduction moment and increased knee stiffness have been noted concurrently with muscle dysfunctions, including reduced knee extensor strength, activation failure, and impaired coordination in severe knee osteoarthritis (OA) (J. A. Zeni et al., 2009). Decreased strength of the knee extensors and flexors is also common with anterior cruciate ligament (ACL) injury and may be associated with an increased external knee flexion moment and internal knee rotation during the loading response portion of gait (Andriacchi & Dyrby, 2005; Berchuck, Andriacchi, Bach, & Reider, 1990; Palmieri-Smith, Thomas, & Wojtys, 2008).

Fatigue refers to a reduction in the efficiency and force generating capacity of muscles after prolonged exposure to activity (Gandevia, Allen, & McKenzie, 1995). Thus, fatigue could lead to changes in knee loads. Fatigue of the quadriceps reduced the knee flexion angle and the peak knee extensor moment during gait in healthy, young participants (Parijat & Lockhart, 2008), and decreased the external knee flexion moment and increased the knee adduction moment during gait in sedentary, young participants (Murdock & Hubley-Kozey, 2012; Walter, D'Lima, Colwell, & Fregly, 2010).

Men exhibit faster fatigue rates compared to women. Mechanisms for fatigue resistance in women may include a lower absolute muscle mass providing the same amount of work for a given task, a lower proportion of the more fatigable type II muscle fibres, or differences in blood flow and muscle metabolism (Clark, Collier, Manini, & Ploutz-Snyder, 2005; Hicks, Kent-Braun, & Ditor, 2001; Pincivero, Green, Mark, & Campy, 2000; Pincivero, Gear, Sterner, & Karunakara, 2000; Pincivero, Gandaio, & Ito, 2003). Since men tend to have greater muscle mass and absolute strength than women (Clark et al., 2005; Hicks et al., 2001; Hunter, Critchlow, & Enoka, 2004; Kent-Braun, Ng, Doyle, & Towse, 2002; Pincivero et al., 2000; Pincivero et al., 2000; Pincivero et al., 2003), they use a lower proportion of their strength reserve in performing the same task. Therefore, women may be more susceptible to fatigue-induced decreases in strength and the resultant changes at the knee. It is uncertain whether altered gait mechanics, due to lower limb fatigue, would occur in a sample of only young, healthy women (Cortes, Quammen, Lucci, Greska, & Onate, 2012; Lucci, Cortes, Van Lunen, Ringleb, & Onate, 2011; Murdock & Hubley-Kozey, 2012; Parijat & Lockhart, 2008; A. C. Thomas, McLean, & Palmieri-Smith, 2010).

Fatigue-induced co-activation of opposing muscle groups may affect knee joint loads. Since knee joint reaction moments are net quantities, co-activation of opposing muscle groups will reduce the total net joint moment. This will translate into an apparent decrease in torque generating capacity of the muscles surrounding the joint (Bennell, Hunt, Wrigley, Lim, & Hinman, 2008; Busse, Wiles, & van Deursen, 2006; Ebenbichler et al., 1998; Heiden, Lloyd, & Ackland, 2009). Furthermore, co-activation increases the resistance to angular joint motion (rotational stiffness). This stiffness represents the rotational resistance contributions from the muscles and other soft tissues controlling knee excursion (Dixon, Hinman, Creaby, Kemp, & Crossley, 2010; J. A. Zeni et al., 2009; J. A. Zeni Jr & Higginson, 2009). Higher stiffness, due to greater antagonistic hamstrings activity during the loading portion of gait, occurs in severe knee OA compared to healthy individuals (J. A. Zeni Jr & Higginson, 2009).

This study aimed to identify the biomechanical changes at the knee that occur during gait in response to neuromuscular fatigue of the knee extensor and flexor muscles in healthy, young women. Our goal was to determine whether the effects of fatigue on knee kinetics and kinematics would further our understanding of the contributions of muscle dysfunction to knee loading patterns. We hypothesized that neuromuscular fatigue around the knee would increase the knee adduction moment, increase co-activation of the quadriceps and hamstrings, and increase dynamic knee stiffness during gait.

2.3 Methods

2.3.1 Participants

A convenience sample of 20 healthy women (age 18 to 30 years) completed this study. Inclusion criteria included regular physical activity, and no contraindications to exercise on the Physical Activity Readiness Questionnaire (S. Thomas, Reading, & Shephard, 1992). Exclusion criteria included pregnancy and a history of knee pain, injury, or surgery. Participants were recruited from the local student population. Written, informed consent was obtained from participants and this study was approved by the McMaster University Faculty of Health Sciences Human Research Ethics Board.

2.3.2 Design

Participants came to the laboratory for two visits, one week apart (Figure 2.1). The first visit was an orientation to familiarize participants with the equipment and protocol. This was incorporated in the study design to facilitate maximal performance during the second visit.

The second visit consisted of gait analyses before and after two separate bouts of fatigue. Baseline (BL) gait analysis was performed to establish points of reference prior to fatigue. This was followed by a measure of baseline isometric peak knee extensor and flexor torque, followed by the first round of fatiguing contractions. Upon reaching fatigue, participants performed the first post fatigue gait analysis (PF1). This was followed by a second round of fatiguing contractions to ensure that participants remained fatigued or were re-fatigued to account for the possibility of fatigue recovery. The final post fatigue gait analysis (PF2) was performed quickly after achieving fatigue.

2.3.3 Measures

2.3.3.1 Gait Analysis

Marker motion during gait was measured using a Vicon MX 8 camera motion capture system sampling at 100 Hz (Vicon Motion Systems, Oxford, UK) and synchronized with three force platforms measuring ground reaction forces and moments sampled at 1000 Hz (Ives & Wigglesworth, 2003) (Advanced Mechanical Technology Inc., Watertown, MA). Eighteen reflective markers were affixed to the right and left anterior and posterior superior iliac spines, mid thighs, lateral epicondyles, mid shanks, lateral malleoli, calcanei, and 2nd metatarsal heads. Six additional reflective markers were affixed to the left and right iliac crest, greater trochanters, medial epicondyles, and medial malleoli during static standing calibration trials as digital landmarks. These 6 additional markers were removed before beginning the gait trials, but the other 18 markers remained affixed throughout the protocol including all gait analyses and fatiguing contractions. The marker placement and gait protocols used in this study were based on Vicon's plug-in-gait lower limb model, with the addition of the markers used for digital landmarking.

Gait analysis was used to capture external knee joint moments and angles. Participants walked barefoot at self-selected speeds. Gait trials were considered successful when the right foot alone fell in full contact with one of the force platforms. Gait trials were taken from heel strike of the right foot to the following heel strike of the same foot, therefore 0% of the gait cycle refers to the initial heel strike of the right foot, and 100% of the gait cycle represents the subsequent heel strike of that same foot. Five gait trials were collected for each participant at BL, PF1, and PF2. Gait trials were completed within a 10-minute window of achieving fatigue to avoid recovery (Cheng & Rice, 2005; Parijat & Lockhart, 2008). Kinematic and kinetic gait variables were calculated (C-Motion, Inc., Germantown, MD, USA) using inverse dynamics (Winter, 1984). Marker and force platform data were filtered with a dual-pass fourth order Butterworth low-pass filter at a cut-off frequency of 6 Hz. Knee joint moments were calculated using the Joint Coordinate System floating axis model (Grood & Suntay, 1983). Joint moments were normalized to body mass, and moments and angles were time normalized to the gait cycle (C-Motion, Inc., Germantown, MD, USA). Discrete measures including peaks, maximums, and minimums were extracted from joint moment and angle waveforms from 5 gait trials.

2.3.3.2 Electromyography

Electromyographic (EMG) signals were collected during gait and during peak torque measurements to determine lower limb muscle contributions to these activities. The EMG was pre-amplified through dual differential amplifiers with input impedance >100,000 MΩ, Common Mode Rejection Ratio >100 dB at 65 Hz and equivalent input noise of <1.2 μV Root Mean Square nominal. Each participant-mounted amplifier had an input impedance of 31KΩ, signal-to-noise ratio of >50 dB, and a gain of 2000 (MA 300, Motion Lab Systems, Inc., Baton Rouge, LA, USA). EMG data were bandpass filtered between 20-450 Hz, synchronized with the motion capture system, and sampled at 1000

Hz. Activity of the rectus femoris, vastus lateralis, and biceps femoris was monitored using stainless steel surface electrodes with a 17 mm inter-electrode distance affixed to the skin along the orientation of the muscle fibers. The rectus femoris and biceps femoris are antagonists, biarticulating the knee and hip. Altered vastus lateralis function resulted in significant changes in knee kinematics during stair climbing; therefore this muscle was selected as a possible explanatory variable for altered knee mechanics during gait (Hinman, Bennell, Metcalf, & Crossley, 2002; Pincivero, Gandhi, Timmons, & Coelho, 2006). A reference electrode was placed over the tibial tubercle. Electrode locations were determined through palpation according to the SENIAM (Surface Electromyography for the Non-Invasive Assessment of a Muscle) guidelines (www.seniam.org, Enschede, Netherlands). The skin was prepared by shaving the electrode locations, vigorously wiping them with rubbing alcohol, and applying electrode gel between the skin and the electrodes. Electrode placement was verified by having the participant perform contractions to elicit activity in each muscle. A quiet trial to establish a baseline activity level for each muscle was obtained with participants lying supine and relaxed.

2.3.3.3 Peak Torque and Fatigue

Strength and fatigue were assessed using a dynamometer (Biodex Medical Systems, Inc., Shirley, NY, USA) with a leg extension attachment. Participants were positioned for testing knee extension and flexion according to the Biodex Multi-Joint System Setup/Operation Manual. Participants performed an isokinetic warm up of 50 contractions at 60°/sec. Strength was measured as the peak knee extension and flexion torque obtained during a set of 5 maximum voluntary isometric contractions (MVICs), at 60° of knee flexion, held for 5 seconds each with rests of 5 seconds between each contraction. Peak torque for knee extensors and flexors during maximal efforts typically occurs after the first repetition, but within the first 5 repetitions for women (Pincivero et al., 2003). Peak torque was measured at baseline, and after each bout of fatiguing contractions to test for quadriceps or hamstrings fatigue.

2.3.3.4 Fatigue Protocol

Neuromuscular fatigue was induced in the right leg using repetitive dynamic isotonic knee extensions and flexions throughout the knee range of motion on the dynamometer. An isotonic fatigue protocol was chosen because it provides greater insight into functional capacity than isokinetic exercises, and is an effective and efficient method of inducing neuromuscular fatigue (Cheng & Rice, 2005; McNeil & Rice, 2007; Stauber, Barill, Stauber, & Miller, 2000). Participants performed sets of 50 extensions and flexions at 50% of their peak torque during MVIC. Peak torque decreases occur primarily within the first 40-50 contractions and are minimal past 50 knee extension repetitions as individuals enter the stable endurance phase (Larsson, Karlsson, Eriksson, & Gerdle, 2003; Lindström, Karlsson, & Gerdle, 1995). Fatigue was assessed by comparing peak isometric knee extension and flexion torques, after the first and second bouts of fatigue, to the baseline values. Fatigue was defined as a drop in either peak isometric knee extension or flexion torque of at least 25% from the baseline value. Measures of peak quadriceps torque, before and after fatiguing contractions, are a reliable indicator of knee extensors fatigue (Larsson et al., 2003), and fatigue recovery of the knee extensor muscles does not occur within the first 10 minutes after fatigue (Cheng & Rice, 2005). Participants performed up to 4 sets of fatiguing contractions in a row until fatigue was achieved. If fatigue was not achieved within 4 sets of fatiguing contractions, participants were withdrawn from the study to avoid lower limb injury.

2.3.3.5 Dynamic Stiffness

Dynamic knee stiffness was calculated as the change in knee flexion moment divided by the change in knee flexion angle during the weight acceptance portion of the gait cycle (3-15%) (J. A. Zeni Jr & Higginson, 2009). Dynamic knee stiffness was taken as the slope of a linear regression line fitted to the data points.

2.3.4 Data Analysis

The EMG signals were full-wave rectified and low pass filtered at 6 Hz (4th order Butterworth filter). Baseline activity obtained from the bias trial was removed from each signal, and the signals obtained from gait trials were amplitude normalized to % MVIC, meaning that 100% muscle activation is equivalent to the peak muscle activation achieved during MVIC. EMG signals obtained from gait trials were also time normalized to the gait cycle (C-Motion, Inc., Germantown, MD, USA). EMG amplitude data from all participants were ensemble averaged to obtain one curve per muscle for all participants at BL, PF1, and PF2. Mean and peak EMG amplitudes during knee loading (3-15% of gait) and unloading (40-52% of gait) were extracted at BL, PF1, and PF2. Signals for each muscle during MVIC were passed through a Hanning window and power spectrums were calculated for the middle 2 seconds (2048 samples) of the 5 second MVIC with a fast Fourier transform (Matlab, the MathWorks, Natick, MA, USA; MyoResearch XP Master Edition, Noraxon USA Inc., Scottsdale, AZ, USA). For each muscle, the median power frequency (MPF) was calculated as the frequency that divided the total power in half.

Co-activation of the biceps femoris and rectus femoris was calculated across the loading response portion of the gait cycle, from 3-15% of gait, and across the unloading portion of the gait cycle, from 40-52% of gait. These two muscles were compared since they both articulate the hip and knee, and EMG fatigue patterns are different in mono- and bi-articular muscles (Ebenbichler et al., 1998). The co-activation index was calculated using the following equation:

$$CI = \frac{\int EMG_{ANT}}{\int EMG_{ANT} + EMG_{AG}} \times 100\%$$

where EMG_{ANT} is the magnitude of EMG from the antagonist (lower activity muscle), and EMG_{AG} is the is the magnitude of EMG from the agonist (higher activity muscle) (Kellis, Arabatzi, & Papadopoulos, 2003).

The mean of 5 gait trials provided one representative curve for each participant for knee moment, knee angle, and muscle activation data. These data from all participants were ensemble averaged to obtain one curve for each variable representing all participants at BL, PF1, and PF2.

Repeated measures analyses of variance and pair-wise comparisons among estimated marginal means were used to assess differences in discrete gait measures, knee extension and flexion torques, and EMG amplitudes between BL, PF1, and PF2 ($\alpha = 0.05$). A Bonferroni adjustment accounted for multiple comparisons.

2.4 Results

Four participants did not achieve a 25% decrease in MVIC after 4 sets of 50 fatiguing contractions and their data were removed from the study. These participants were not different from the rest of the sample in age, body mass, BMI, or baseline peak extension and flexion torques ($p>0.05$). Twenty participants were considered to be fatigued as they demonstrated at least the criterion 25% decrease in flexor or extensor MVIC torque. These 20 participants were (in mean (standard deviation)) 23.2 (3.1) years old, had a body mass of 63.2 (9.7) kg, and had a BMI of 23.2 (3.0) kg/m^2. The fatigue protocol significantly decreased peak isometric torque from BL (mean [95% Confidence Interval], extension: 173.5 [159.1, 187.9] Nm, flexion: 82.4 [75.4, 89.4] Nm) to PF1 (extension: 129.9 [116.4, 143.3] Nm, flexion: 58.1 [53.0, 63.2] Nm) and from BL to PF2 (extension: 128.6 [115.4, 141.9] Nm, flexion: 60.7 [54.4, 67.1] Nm) ($p < 0.001$) (Table 2.1).

Gait speed and stride length were not altered by fatigue (Table 2.1). Knee joint moments and angles during gait are shown in Figure 2.2. Fatigue did not alter first or second peak knee adduction moments, or peak knee flexion angles (Table 2.1). Peak knee flexion moments during gait were unchanged with fatigue; however fatigue did decrease the peak knee extension moment from BL (mean [95% Confidence Interval], -0.30 [-0.37, -0.22] Nm/kg) to PF1 (-0.24 [-0.31, -0.16] Nm/kg), and from BL to PF2 (-0.24 [-0.32, -0.17] Nm/kg) ($p<0.001$). This decrease occurred between 40-50% of the gait cycle, when the heel leaves the ground, preparing for push off.

Vastus lateralis data from one participant were excluded. Due to the high level of physical exertion during the data collection and the length of data collection time, the vastus lateralis electrode shifted during data collection on this participant. This occurred after the first set of fatiguing contractions. The EMG amplitudes as % MVIC are displayed over the gait cycle in Figure 2.3. Mean and peak EMG amplitudes (% MVIC) were not different between BL, PF1, and PF2 across either 3-15% or 40-52% of gait (Table 2.2). The MPF of rectus femoris decreased 6.3 Hz (8.7%) ($p=0.05$) from BL to PF1 (Table 2.2). No differences in MPF were found between time points for other muscles. Dynamic knee stiffness was not significantly increased with fatigue (Figure 2.2, Table 2.1). Muscle co-activation between the biceps femoris and rectus femoris was not altered with fatigue (Table 2.2).

2.5 Discussion

This study sought to identify biomechanical changes at the knee during gait before and after an isotonic fatigue protocol in healthy, young women. Two sets of quadriceps and hamstrings fatiguing contractions did not increase the peak knee adduction moment, dynamic knee stiffness, or anterior-posterior muscle co-activation around the knee; however these contractions decreased the external peak knee extension moment in late stance in these strong, young women. The key implication of this study is that young women with high lower limb muscle strength retain enough strength reserve after a fatigue protocol, which resulted in reductions to maximum torque generating capacity, to perform a submaximal activity such as gait without eliciting potentially damaging knee biomechanics. Peak knee extensor and flexor moments do not exceed approximately 0.6 Nm/kg in young, healthy individuals during typical overground walking (Riley, Paolini, Della Croce, Paylo, & Kerrigan, 2007). Using the mean body mass as an example from the current participant sample (body mass = 63.2 kg) this equates to a peak torque of 37.9 Nm, which is well below the amount of strength retained after the fatigue protocol. This is likely related to the low muscle activation demands required for gait. For example, rectus femoris activity during gait in young, healthy individuals does not typically exceed approximately 20% MVIC during the knee loading portion of the gait cycle (Arsenault, Winter, & Marteniuk, 1986).

The gait changes observed after the fatigue protocol, which resulted in a reduction to maximum torque generation in the current study, do not imitate any obvious knee joint injury or pathology. Individuals with knee OA and joint effusion demonstrate a reduced late stance knee extensor moment similar to that associated with the fatigue protocol in the current study. However, individuals with knee OA, but without knee effusion, do not demonstrate changes in this moment (Rutherford, Hubley-Kozey, & Stanish, 2012). In other pathologies, a reduced knee extensor moment in late stance is typically coupled with alterations in the knee flexion angle, frontal knee moments, and/or quadriceps activity (Henriksen, Graven-Nielsen, Aaboe, Andriacchi, & Bliddal, 2010; Rutherford et al., 2012). Murdock and Hubley-Kozey (2012) reported an increased post fatigue external knee adduction moment between early and mid stance, and decreased post fatigue external knee flexion moment during early stance, however no differences were observed during late stance. While the values and waveforms obtained in the current study are similar to Murdock and Hubley-Kozey (2012), the current study found fewer post fatigue gait differences.

The participants in the current study demonstrated substantially greater quadriceps torque at baseline with the mean (standard deviation) being 173.5 (30.7) Nm compared to 143.6 (29.7) Nm in Murdock and Hubley-Kozey (2012), highlighting the difference between a physically active sample and a sedentary sample. Furthermore, Murdock and Hubley-Kozey used a maximal effort isokinetic fatiguing protocol and obtained a 40% decrease in quadriceps torque, while the current study used an isotonic fatiguing protocol at 50% of maximal effort and fatigue criteria of a 25% decrease in maximum torque of either the knee extensors or flexors to avoid participant injury. This suggests that a fatigue protocol, which results in a 25% reduction in maximum isometric torque of the knee flexors or extensors in a healthy, strong sample of women, does not result in altered knee joint mechanics and muscle activation patterns during gait within a 10-minute window after fatigue. A significant shift in the median power spectrum toward lower frequencies was found for rectus femoris between BL and PF1 in the current study. This has been used as a measure of neuromuscular fatigue in the quadriceps during isometric contractions (Bilodeau, Schindler-Ivens, Williams, Chandran, & Sharma, 2003; Masuda, Masuda, Sadoyama, Inaki, & Katsuta, 1999; Murdock & Hubley-Kozey, 2012; Pincivero et al., 2006); however there is a lack of consistency in the specific quadriceps muscles that undergo these spectral shifts. Furthermore, the rate of change of EMG frequency over a sustained isometric contraction may be a better indicator of quadriceps fatigue (Mannion & Dolan, 1996) and no relationship has been reported between fatigue and median frequency shifts for biceps femoris.

Fatigue of the knee flexors or extensors in the current study may have minimized the biomechanical changes found by previous studies, which fatigued the extensors exclusively. Parijat and Lockhart (2008) observed mechanical changes due to fatigue shortly after the knee begins to accept load reporting a reduced knee flexion angle at heel strike, and a reduced external knee extension moment in early-mid stance and just before toe-off. Conversely, the current study showed mechanical changes at the knee occurring in preparation for push-off. This discrepancy could be explained by differences in the sex, type of fatigue, muscle groups, and level of physical activity between the participants in these studies. Men experience a greater peak external knee flexion moment immediately following heel strike during gait than women (Kerrigan, Todd, & Della Croce, 1998). The gender difference in this study compared to previous studies may have attenuated the results of fatigue at this point in the stance cycle because the lower absolute knee flexion moment at baseline in women leaves a lower margin for a decrease in this measure after fatigue. Murdock and Hubley-Kozey (2012) studied 20 sedentary young adults, half of whom were women, while Parijat and Lockhart (2008) evaluated 10 men and 6 women who were healthy, but no further details on physical fitness were given. In the current study, all participants were physically active, young women. Furthermore, participants in the current study walked barefoot to eliminate alterations in gait patterns caused by footwear including inflated lower extremity joint loads (Shakoor & Block, 2006), while other studies may have analyzed gait while wearing shoes (Murdock & Hubley-Kozey, 2012; Parijat & Lockhart, 2008).

The decreased peak extension moment during late stance was not explained by a difference in co-activation or knee angle before and after fatigue. A possible explanation is an increase in the acceleration of the lower limb during this portion of stance, as a mechanism to decrease the duration of load applied to the knee. Participants retained at least 75% of their maximum strength, leaving plenty of reserve for gait (Table 2.1). However, post fatigue gait did not require a significant increase in muscle activation (average or peak) (Table 2.2, Figure 2.3) to maintain pre fatigue mechanics. This indicates that the strenuous fatigue protocol did not alter the knee biomechanics in a way that is consistent with gait patterns associated with ACL injury or knee OA.

Dynamic knee stiffness during knee loading was not altered by the fatigue protocol. Mechanical knee stiffness is poorly correlated with self-reported knee stiffness (Dixon et al., 2010); however it indicates the amount of torque being used to rotate the knee in the sagittal plane. The minimal biomechanical changes in this study were observed only during knee unloading, pre-swing. This measure of knee stiffness focuses on the knee loading portion of gait; therefore it did not capture the portion of the gait cycle that was altered by the fatigue protocol.

Quadriceps fatigue has previously been induced by repetitive isokinetic contractions at maximal effort (Murdock & Hubley-Kozey, 2012; Parijat & Lockhart, 2008; Pincivero, Gear, & Sterner, 2001). This study aimed to better simulate fatigue caused by repetitive performance of activities of daily living (ADLs). ADLs have been characterized as bursts of activity requiring varying velocities of joint motion, under relatively constant loads applied to both knee extensors and flexors (Cheng & Rice, 2005), however activities such as walking are not strictly isokinetic or isotonic tasks as contraction velocities, and both internal and external loads, change throughout the cyclical motion. As a step toward a more functionally relevant representation of these ADLs, an isotonic fatigue protocol was used where the load was held constant while undergoing variations in angular velocity, applied to knee extensors and flexors. The current study attempted to induce fatigue in a similar manner to that induced by ADLs and ensure that fatigue was maintained throughout the gait trials by analyzing gait within 10-minutes after achieving fatigue; however achieving fatigue quickly in a laboratory setting may elicit faster rates of recovery compared to fatigue caused by repetitive ADLs. The high intensity, short duration, and repetitive contractions confined to the knee in this study are not particularly representative of the fatigue that affects individuals with ACL injury or knee OA. Typically, fatigue occurs from low intensity, long duration, and repetitive loads on the knee, through ADLs such as squatting, lunging, lifting, or long distance walking. This is difficult to reproduce in a laboratory setting, however future studies should attempt to quantify biomechanical changes induced by low intensity, repetitive fatigue.

Analysis techniques in the currently study differ from previous studies of knee mechanics during gait. The current study used discrete analyses to identify differences, before and after the fatigue protocol, between waveform peaks and means during specific portions of the gait cycle. This analysis method disregards differences between waveform shape, magnitude, and timing. Our sample size limited the statistical sensitivity of our analysis; therefore variances in discrete parameters were analyzed rather than performing full-wave analyses, such as principal component analysis, which is an effective tool for biomechanical analysis with large sample sizes (Deluzio, Wyss U.P., Zee B., Costigan P.A., & Sorbie C., 1997; Deluzio & Astephen, 2007). This discrete analysis did not detect substantial differences in waveform peaks and means, likely due to variability between participants.

The mechanism for fatigue alters the rate of change in muscle activation in both young and older adults, and varies between muscle groups. In young adults, the rectus femoris, vastus lateralis, and vastus medialis display different activation properties under isometric and dynamic conditions (Callahan, Foulis, & Kent-Braun, 2009; Pincivero et al., 2006). Relative to the absolute strength of the participants in the current study, tasks such as walking are low demand in terms of strength requirements. The strong, healthy participants clearly maintained enough strength reserve after a fatigue protocol that resulted in a reduction to maximum torque generating capacity to perform this activity without substantial changes to their knee biomechanics, which could be considered harmful. Alterations in knee mechanics due to fatigue may be more prominent in higher demand tasks such as running or jumping.

In conclusion, this study indicates that a high intensity lower limb fatigue-inducing activity did not alter the mechanical environment of the knee joint during walking in young, active women in a way that reflects an increased risk for ACL injury.

2.6 References

Andriacchi, T. P., & Dyrby, C. O. (2005). Interactions between kinematics and loading during walking for the normal and ACL deficient knee. Journal of Biomechanics, 38(2), 293-298. doi:10.1016/j.jbiomech.2004.02.010

Arsenault, A. B., Winter, D. A., & Marteniuk, R. G. (1986). Is there a 'normal' profile of EMG activity in gait? Medical & Biological Engineering & Computing, 24(4), 337-343.

Bennell, K. L., Hunt, M. A., Wrigley, T. V., Lim, B. W., & Hinman, R. S. (2008). Role of muscle in the genesis and management of knee osteoarthritis. Rheumatic Diseases Clinics of North America, 34(3), 731-754.

Berchuck, M., Andriacchi, T. P., Bach, B. R., & Reider, B. (1990). Gait adaptations by patients who have a deficient anterior cruciate ligament. The Journal of Bone and Joint Surgery.American Volume, 72(6), 871-877.

Bilodeau, M., Schindler-Ivens, S., Williams, D. M., Chandran, R., & Sharma, S. S. (2003). EMG frequency content changes with increasing force and during fatigue in the quadriceps femoris muscle of men and women. Journal of Electromyography and Kinesiology, 13(1), 83-92. doi:10.1016/S1050-6411(02)00050-0

Busse, M. E., Wiles, C. M., & van Deursen, R. W. (2006). Co-activation: Its association with weakness and specific neurological pathology. Journal of Neuroengineering and Rehabilitation, 3, 26. doi:10.1186/1743-0003-3-26

Callahan, D. M., Foulis, S. A., & Kent-Braun, J. A. (2009). Age-related fatigue resistance in the knee extensor muscles is specific to contraction mode. Muscle & Nerve, 39(5), 692-702. doi:10.1002/mus.21278

Cheng, A. J., & Rice, C. L. (2005). Fatigue and recovery of power and isometric torque following isotonic knee extensions. Journal of Applied Physiology (Bethesda, Md.: 1985), 99(4), 1446-1452. doi:10.1152/japplphysiol.00452.2005

Clark, B. C., Collier, S. R., Manini, T. M., & Ploutz-Snyder, L. L. (2005). Sex differences in muscle fatigability and activation patterns of the human quadriceps femoris. European Journal of Applied Physiology, 94(1), 196-206.

Cortes, N., Quammen, D., Lucci, S., Greska, E., & Onate, J. (2012). A functional agility short-term fatigue protocol changes lower extremity mechanics. Journal of Sports Sciences, 30(8), 797-805. doi:10.1080/02640414.2012.671528

Deluzio, K. J., & Astephen, J. L. (2007). Biomechanical features of gait waveform data associated with knee osteoarthritis: An application of principal component analysis. Gait & Posture, 25(1), 86-93.

Deluzio, K. J., Wyss U.P., Zee B., Costigan P.A., & Sorbie C. (1997). Principal component models of knee kinematics and kinetics: Normal vs pathological gait patterns. Human Movement Science, 16, 201-2.

Dixon, S. J., Hinman, R. S., Creaby, M. W., Kemp, G., & Crossley, K. M. (2010). Knee joint stiffness during walking in knee osteoarthritis. Arthritis Care & Research, 62(1), 38-44. doi:10.1002/acr.20012

Ebenbichler, G., Kollmitzer, J., Quittan, M., Uhl, F., Kirtley, C., & Fialka, V. (1998). EMG fatigue patterns accompanying isometric fatiguing knee-extensions are different in mono- and bi-articular muscles. Electroencephalography and Clinical Neurophysiology/ Electromyography and Motor Control, 109(3), 256-262.

Enoka, R. M. (2002). In Robertson L. D., Thomas R. T., Ewing A. S. and Sexton J. (Eds.), Neuromechanics of human movement (Third ed.). United States of America: Human Kinetics. doi:0-7360-0251-0

Gandevia, S., Allen, G., & McKenzie, D. (1995). Central fatigue. Advances in Experimental Medicine and Biology, 384, 281.

Grood, E. S., & Suntay, W. J. (1983). A joint coordinate system for the clinical description of three-dimensional motions: Application to the knee. Journal of Biomechanical Engineering, 105, 136.

Heiden, T. L., Lloyd, D. G., & Ackland, T. R. (2009). Knee extension and flexion weakness in people with knee osteoarthritis: Is antagonist cocontraction a factor? The Journal of Orthopaedic and Sports Physical Therapy, 39(11), 807-815. doi:10.2519/jospt.2009.3079

Henriksen, M., Graven-Nielsen, T., Aaboe, J., Andriacchi, T. P., & Bliddal, H. (2010). Gait changes in patients with knee osteoarthritis are replicated by experimental knee pain. Arthritis Care & Research, 62(4), 501-509. doi:10.1002/acr.20033

Hicks, A. L., Kent-Braun, J., & Ditor, D. S. (2001). Sex differences in human skeletal muscle fatigue. Exercise and Sport Sciences Reviews, 29(3), 109-112.

Hinman, R. S., Bennell, K. L., Metcalf, B. R., & Crossley, K. M. (2002). Delayed onset of quadriceps activity and altered knee joint kinematics during stair stepping in individuals with knee osteoarthritis. Archives of Physical Medicine and Rehabilitation, 83(8), 1080-1086.

Hunter, S. K., Critchlow, A., & Enoka, R. M. (2004). Influence of aging on sex differences in muscle fatigability. Journal of Applied Physiology (Bethesda, Md.: 1985), 97(5), 1723-1732. doi:10.1152/japplphysiol.00460.2004

Ives, J. C., & Wigglesworth, J. K. (2003). Sampling rate effects on surface EMG timing and amplitude measures. Clinical Biomechanics, 18(6), 543-552. doi:10.1016/S0268-0033(03)00089-5

Kellis, E., Arabatzi, F., & Papadopoulos, C. (2003). Muscle co-activation around the knee in drop jumping using the co-contraction index. Journal of Electromyography and Kinesiology : Official Journal of the International Society of Electrophysiological Kinesiology, 13(3), 229-238.

Kent-Braun, J. A., Ng, A. V., Doyle, J. W., & Towse, T. F. (2002). Human skeletal muscle responses vary with age and gender during fatigue due to incremental isometric exercise. Journal of Applied Physiology (Bethesda, Md.: 1985), 93(5), 1813-1823. doi:10.1152/japplphysiol.00091.2002

Kerrigan, D. C., Todd, M. K., & Della Croce, U. (1998). Gender differences in joint biomechanics during walking: Normative study in young adults. American Journal of Physical Medicine & Rehabilitation / Association of Academic Physiatrists, 77(1), 2-7.

Larsson, B., Karlsson, S., Eriksson, M., & Gerdle, B. (2003). Test–retest reliability of EMG and peak torque during repetitive maximum concentric knee extensions. Journal of Electromyography and Kinesiology, 13(3), 281-287. doi:10.1016/S1050-6411(03)00022-1

Lewek, M. D., Rudolph, K. S., & Snyder-Mackler, L. (2004). Quadriceps femoris muscle weakness and activation failure in patients with symptomatic knee osteoarthritis. Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society, 22(1), 110-115.

Lindström, B., Karlsson, S., & Gerdle, B. (1995). Knee extensor performance of dominant and non-dominant limb throughout repeated isokinetic contractions, with special reference to peak torque and mean frequency of the EMG. Clinical Physiology, 15(3), 275-286.

Lucci, S., Cortes, N., Van Lunen, B., Ringleb, S., & Onate, J. (2011). Knee and hip sagittal and transverse plane changes after two fatigue protocols. Journal of Science and Medicine in Sport / Sports Medicine Australia, 14(5), 453-459. doi:10.1016/j.jsams.2011.05.001

Mannion, A. F., & Dolan, P. (1996). Relationship between myoelectric and mechanical manifestations of fatigue in the quadriceps femoris muscle group. European Journal of Applied Physiology and Occupational Physiology, 74(5), 411-419.

Masuda, K., Masuda, T., Sadoyama, T., Inaki, M., & Katsuta, S. (1999). Changes in surface EMG parameters during static and dynamic fatiguing contractions. Journal of Electromyography and Kinesiology : Official Journal of the International Society of Electrophysiological Kinesiology, 9(1), 39-46.

McNeil, C. J., & Rice, C. L. (2007). Fatigability is increased with age during velocity-dependent contractions of the dorsiflexors. The Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 62(6), 624-629.

Murdock, G., & Hubley-Kozey, C. (2012). Effect of a high intensity quadriceps fatigue protocol on knee joint mechanics and muscle activation during gait in young adults. European Journal of Applied Physiology, 112(2), 439-449. doi:10.1007/s00421-011-1990-4

Palmieri-Smith, R. M., Thomas, A. C., & Wojtys, E. M. (2008). Maximizing quadriceps strength after ACL reconstruction. Clinics in Sports Medicine, 27(3), 405-424. doi:10.1016/j.csm.2008.02.001

Parijat, P., & Lockhart, T. E. (2008). Effects of quadriceps fatigue on the biomechanics of gait and slip propensity. Gait & Posture, 28(4), 568-573.

Pincivero, D. M., Gandaio, C. B., & Ito, Y. (2003). Gender-specific knee extensor torque, flexor torque, and muscle fatigue responses during maximal effort contractions. European Journal of Applied Physiology, 89(2), 134-141.

Pincivero, D. M., Gear, W. S., Sterner, R. L., & Karunakara, R. A. J. G. (2000). Gender differences in the relationship between quadriceps work and fatigue during high-intensity exercise. The Journal of Strength & Conditioning Research, 14(2), 202.

Pincivero, D. M., Green, R. C., Mark, J. D., & Campy, R. M. (2000). Gender and muscle differences in EMG amplitude and median frequency, and variability during maximal voluntary contractions of the quadriceps femoris. Journal of Electromyography and Kinesiology, 10(3), 189-196.

Pincivero, D. M., Gandhi, V., Timmons, M. K., & Coelho, A. J. (2006). Quadriceps femoris electromyogram during concentric, isometric and eccentric phases of fatiguing dynamic knee extensions. Journal of Biomechanics, 39(2), 246-254. doi:10.1016/j.jbiomech.2004.11.023

Pincivero, D. M., Gear, W. S., & Sterner, R. L. (2001). Assessment of the reliability of high-intensity quadriceps femoris muscle fatigue. Medicine and Science in Sports and Exercise, 33(2), 334-338.

Rice, D. A., McNair, P. J., & Lewis, G. N. (2011). Mechanisms of quadriceps muscle weakness in knee joint osteoarthritis: The effects of prolonged vibration on torque and muscle activation in osteoarthritic and healthy control subjects. Arthritis Research & Therapy, 13(5), R151. doi:10.1186/ar3467

Riley, P. O., Paolini, G., Della Croce, U., Paylo, K. W., & Kerrigan, D. C. (2007). A kinematic and kinetic comparison of overground and treadmill walking in healthy subjects. Gait & Posture, 26(1), 17-24. doi:10.1016/j.gaitpost.2006.07.003

Rutherford, D. J., Hubley-Kozey, C. L., & Stanish, W. D. (2012). Knee effusion affects knee mechanics and muscle activity during gait in individuals with knee osteoarthritis. Osteoarthritis and Cartilage / OARS, Osteoarthritis Research Society, 20(9), 974-981. doi:10.1016/j.joca.2012.05.014

Shakoor, N., & Block, J. A. (2006). Walking barefoot decreases loading on the lower extremity joints in knee osteoarthritis. Arthritis and Rheumatism, 54(9), 2923-2927. doi:10.1002/art.22123

Slemenda, C., Brandt, K. D., Heilman, D. K., Mazzuca, S., Braunstein, E. M., Katz, B. P., & Wolinsky, F. D. (1997). Quadriceps weakness and osteoarthritis of the knee. Annals of Internal Medicine, 127(0003-4819; 0003-4819; 2), 97-104.

Slemenda, C., Heilman, D. K., Brandt, K. D., Katz, B. P., Mazzuca, S. A., Braunstein, E. M., & Byrd, D. (1998). Reduced quadriceps strength relative to body weight: A risk factor for knee osteoarthritis in women? Arthritis and Rheumatism, 41(0004-3591; 0004-3591; 11), 1951-1959.

Stauber, W., Barill, E., Stauber, R., & Miller, G. (2000). Isotonic dynamometry for the assessment of power and fatigue in the knee extensor muscles of females. Clinical Physiology, 20(3), 225-233.

Thomas, S., Reading, J., & Shephard, R. J. (1992). Revision of the physical activity readiness questionnaire (PAR-Q). Canadian Journal of Sport Sciences, 17(4), 338-345.

Thomas, A. C., McLean, S. G., & Palmieri-Smith, R. M. (2010). Quadriceps and hamstrings fatigue alters hip and knee mechanics. Journal of Applied Biomechanics, 26(2), 159-170.

Walter, J. P., D'Lima, D. D., Colwell, C. W.,Jr, & Fregly, B. J. (2010). Decreased knee adduction moment does not guarantee decreased medial contact force during gait. Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society, 28(10), 1348-1354. doi:10.1002/jor.21142

Winter, D. A. (1984). Biomechanics of human movement with applications to the study of human locomotion. Critical Reviews in Biomedical Engineering, 9(4), 287-314.

Zeni, J. A.,Jr, & Higginson, J. S. (2009). Dynamic knee joint stiffness in subjects with a progressive increase in severity of knee osteoarthritis. Clinical Biomechanics (Bristol, Avon), 24(4), 366-371. doi:10.1016/j.clinbiomech.2009.01.005

Zeni, J. A., Rudolph, K., & Higginson, J. S. (2009). Alterations in quadriceps and hamstrings coordination in persons with medial compartment knee osteoarthritis. Journal of Electromyography and Kinesiology, 20(1), 148-154. doi:10.1016/j.jelekin.2008.12.003

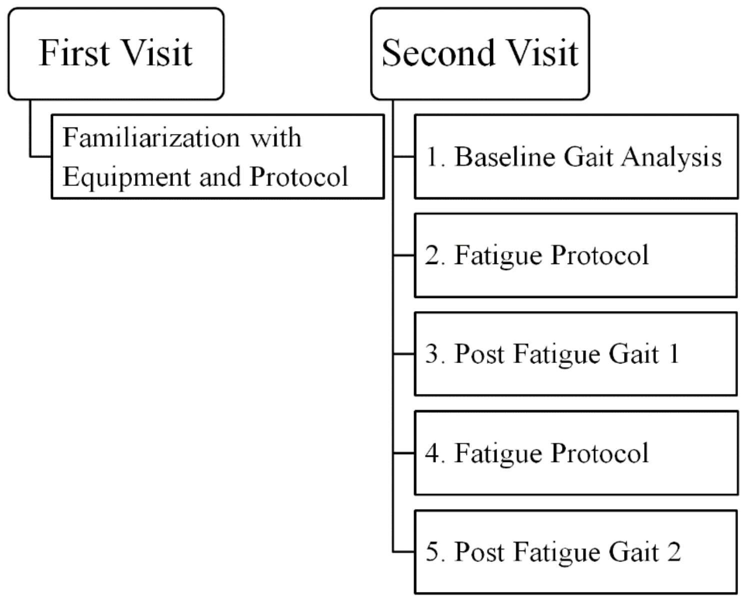

Figure 2. 1. Study design flow chart.

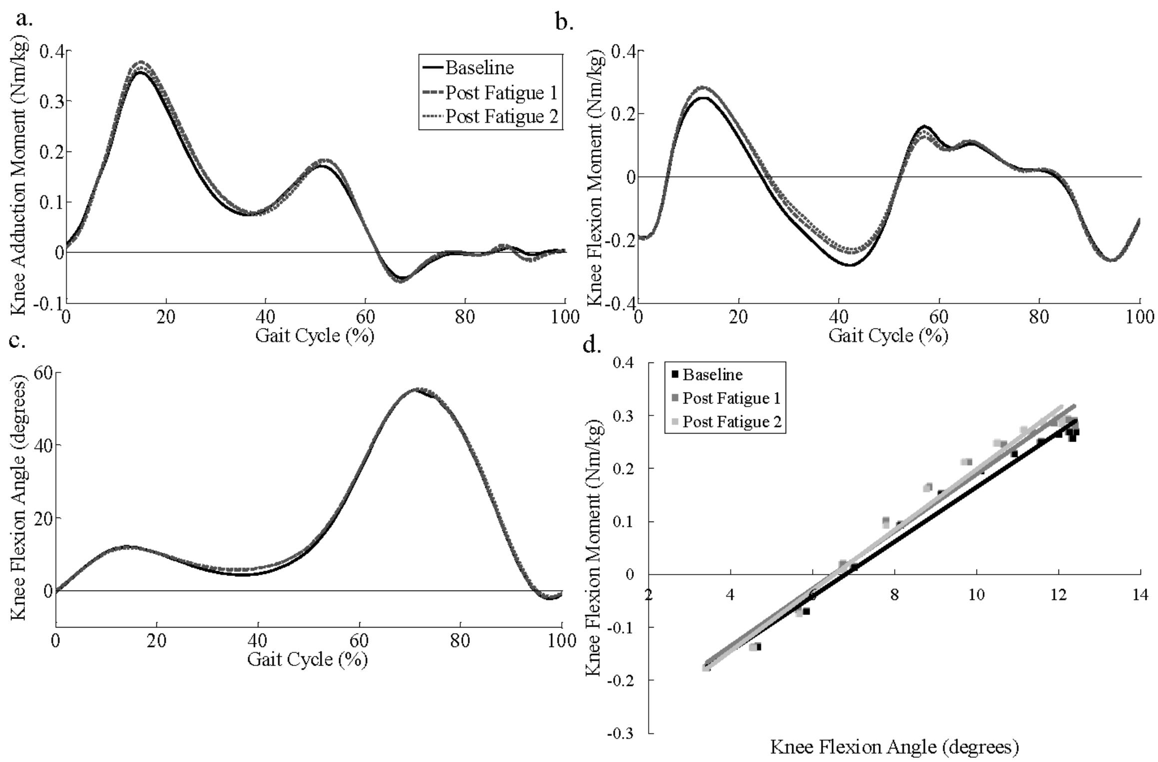

Figure 2. 2. Ensemble averaged a. knee adduction moment (Nm/kg, positive moment represents an external knee adduction moment), b. knee flexion moment (Nm/kg, positive moment represents an external knee flexion moment), and c. knee flexion angle (degrees) over the gait cycle at BL, PF1, and PF2, as well as d. dynamic knee stiffness (calculated as the slope of the regression lines) during the loading response portion of gait.

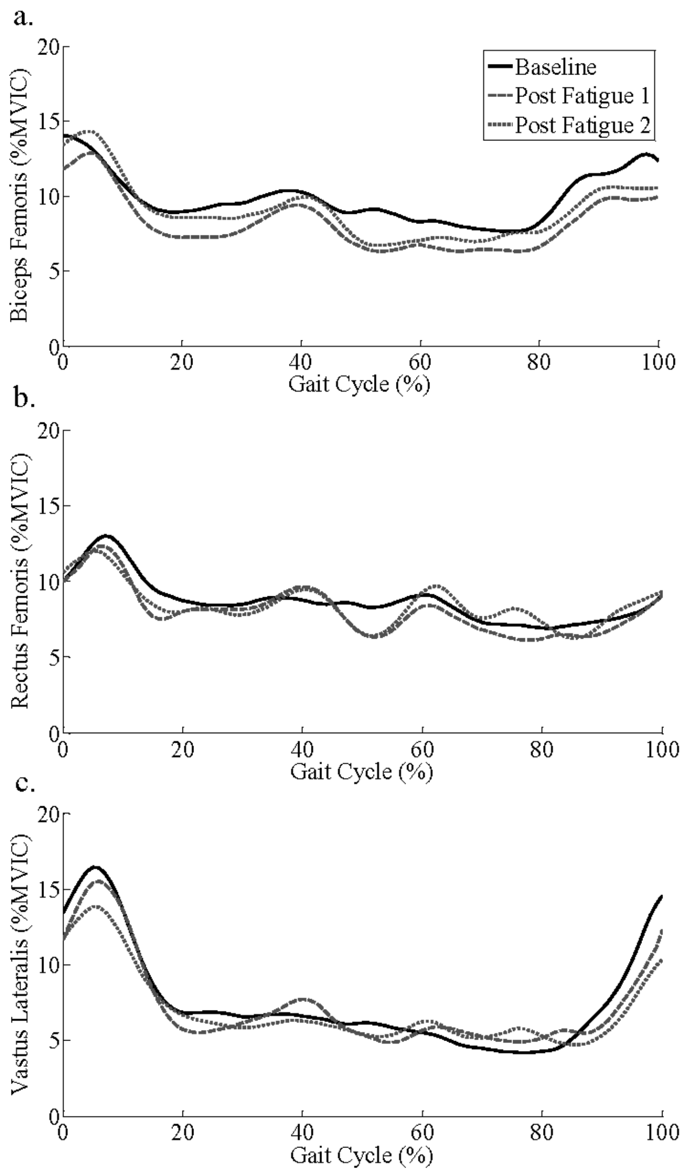

Figure 2. 3. Muscle activation (% MVIC) of the a. Biceps Femoris, b. Rectus Femoris, and c. Vastus Lateralis over the gait cycle at BL, PF1, and PF2, where n=20 for Rectus Femoris and Biceps Femoris, n=19 for Vastus Lateralis.

Table 2. 1. Mean (standard deviation) of peak torque and gait measures before and after neuromuscular fatigue (n=20).

	Baseline	Post-Fatigue 1	Post-Fatigue 2	p-value
Peak isometric torque (Nm) – Extension	173.5 (30.7)	129.9 (28.7)	128.6 (28.3)	$p < 0.001^{a,b}$
Peak isometric torque (Nm) – Flexion	82.4 (14.9)	58.1 (10.9)	60.7 (13.6)	$p < 0.001^{a,b}$
Gait speed (m/s)	1.24 (0.15)	1.25 (0.13)	1.24 (0.13)	NS
Stride length (m)	1.37 (0.13)	1.37 (0.16)	1.37 (0.13)	NS
Knee Adduction Moment (Nm/kg)				
First Peak	0.37 (0.12)	0.39 (0.11)	0.38 (0.12)	NS
Second Peak	0.20 (0.12)	0.22 (0.10)	0.23 (0.12)	NS
Dynamic Knee Stiffness (Nm/degree)	0.05 (0.03)	0.06 (0.02)	0.06 (0.02)	NS
Knee Flexion Angle (degrees)				
First Peak	12.51 (4.49)	12.63 (4.41)	12.30 (4.02)	NS
Second Peak	56.44 (3.79)	56.16 (3.91)	56.07 (3.81)	NS
Sagittal Knee Moment (Nm/kg)				
First Peak Flexion	0.30 (0.23)	0.30 (0.17)	0.30 (0.16)	NS
First Peak Extension	-0.30 (0.16)	-0.24 (0.16)	-0.24 (0.15)	$p < 0.001^{a,b}$
Second Peak Flexion	0.18 (0.14)	0.16 (0.06)	0.16 (0.06)	NS

NS = non-significant, Significant difference between [a]Baseline and Post-Fatigue 1, [b]Baseline and Post-Fatigue 2, [c]Post-Fatigue 1 and Post-Fatigue 2

Table 2. 2. EMG Activation (% MVIC) and Co-activation indices (%) during knee loading (3-15% gait) and unloading (40-52% gait), and Median Power Frequency (Hz) during MVIC. Measures are reported at baseline, after the first bout of fatigue, and after the second bout of fatigue. Values are reported as mean (standard deviation), n=20 for Rectus Femoris and Biceps Femoris, n=19 for Vastus Lateralis.

	Baseline	Post Fatigue 1	Post Fatigue 2
EMG Activation (% MVIC)			
Rectus Femoris			
3-15% gait Average	11.7 (15.8)	10.6 (8.2)	10.6 (9.5)
3-15% gait Peak	13.5 (16.2)	12.8 (9.5)	12.7 (10.8)
40-52% gait Average	8.5 (15.8)	8.0 (8.6)	8.0 (9.3)
40-52% gait Peak	10.0 (16.7)	10.3 (10.8)	10.0 (11.4)
Vastus Lateralis			
3-15% gait Average	13.6 (7.2)	13.1 (6.9)	11.8 (5.2)
3-15% gait Peak	17.0 (8.1)	16.4 (9.0)	14.1 (6.2)
40-52% gait Average	6.3 (7.2)	6.3 (7.1)	5.8 (4.5)
40-52% gait Peak	8.0 (9.4)	8.4 (10.0)	6.7 (5.1)
Biceps Femoris			
3-15% gait Average	11.4 (9.5)	10.7 (7.0)	12.0 (8.2)
3-15% gait Peak	13.9 (9.7)	13.3 (8.3)	14.9 (9.3)
40-52% gait Average	9.4 (9.3)	7.8 (5.8)	8.5 (6.7)
40-52% gait Peak	11.5 (11.1)	9.9 (6.9)	10.4 (7.5)
Co-activation Index (%)			
3-15% gait	33.3 (12.8)	31.0 (11.7)	29.1 (14.2)
40-52% gait	27.7 (14.5)	28.3 (12.7)	28.6 (14.9)
Median Power Frequency (Hz)			
Rectus Femoris	72.7 (15.6)	66.4 (8.4)*	67.4 (19.1)
Vastus Lateralis	73.6 (14.9)	69.3 (12.2)	70.5 (9.7)
Biceps Femoris	64.4 (11.8)	69.2 (12.4)	67.7 (10.7)

* Significant difference ($p<0.05$) between Baseline and Post Fatigue 1 values.

** Significant difference ($p<0.05$) between Baseline and Post Fatigue 2 values.

Chapter Three

Muscle Activation and Knee Biomechanics during Squatting and Lunging after Lower Extremity Fatigue in Healthy Young Women

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The findings in this manuscript build on those presented in Chapter 2 by evaluating muscle activation and joint loading changes in the knee after neuromuscular fatigue of the lower limbs during more challenging, static exercises. These results contribute to a comprehensive understanding of the biomechanical changes at the knee after peak torque impairment of the muscles supporting the knee during functional isometric exercises.

3.1 Abstract

Muscle activations and knee joint loads were compared during squatting and lunging before and after lower extremity neuromuscular fatigue. Electromyographic activations of the rectus femoris, vastus lateralis and biceps femoris, and the external knee adduction and flexion moments were collected on 25 healthy women (mean age 23.5 years, BMI of 23.7 kg/m^2) during squatting and lunging. Participants were fatigued through sets of 50 isotonic knee extensions and flexions, with resistance set at 50% of the peak torque achieved during a maximum voluntary isometric contraction. Fatigue was defined as a decrease in peak isometric knee extension or flexion torque ≥25% from baseline. Co-activation indices were calculated between rectus femoris and biceps femoris; and between vastus lateralis and biceps femoris. Fatigue decreased peak isometric extension and flexion torques ($p<0.05$), mean vastus lateralis activation during squatting and lunging ($p<0.05$), and knee adduction and flexion moments during lunging ($p<0.05$). Quadriceps activations were greater during lunging than squatting ($p<0.05$). Thus, fatigue altered the recruitment strategy of the quadriceps during squatting and lunging. Lunging challenges quadriceps activation more than squatting in healthy, young women.

Keywords: Strength Training, Occupational Health, Rehabilitation, Muscle Fatigue, Repetitive Motion Disorders

3.2 Introduction

Squatting and lunging are important components of lower limb muscle training and rehabilitation programs. These exercises are often performed with the goal of increasing the force generating capacity of the lower limb muscles, particularly the quadriceps, reducing the risk for joint injury, and training balance in functional postures (Ebben et al., 2009). However, repetitive occupational exposure to squats and lunges may increase the risk for joint pathology, particularly at the knee (Amin et al., 2008). While moderate knee joint loading contributes to articular cartilage health (Griffin & Guilak, 2005), high intensity or long duration loading such as during occupational tasks requiring repetitive knee flexion may be excessive, causing joint breakdown (Richmond et al., 2013). The discrepancy between the use of squats and lunges as rehabilitation exercises, and the potential risks of performing these exercises in an occupational setting may be explained by the duration of exposure. It remains unclear whether prolonged or repetitive demands on the knee flexors and extensors change the muscle activation patterns and knee loads required to remain successful in completing the tasks.

Fatigue, characterized by a reduction in the efficiency and force generating capacity of muscles after prolonged exposure to activity (Gandevia, Allen, & McKenzie, 1995), could alter the motor control and joint loading requirements of certain tasks. Knee loads can be estimated by evaluating the external moments at the knee. The knee adduction moment (KAM) and knee flexion moment (KFM) differ for individuals with knee pathologies and injuries compared to healthy controls and may be important in the initiation and progression of joint degeneration (Andriacchi & Mundermann, 2006; Baliunas et al., 2002; Deluzio & Astephen, 2007; Foroughi, Smith, & Vanwanseele, 2009; Rudolph, Axe, Buchanan, Scholz, & Snyder-Mackler, 2001; Sharma et al., 1998). These load measures are determined largely by the orientation and magnitude of the vertical ground reaction force relative to the knee center and by the actions of the muscles supporting the joint (Shelburne, Torry, & Pandy, 2006). The quadriceps, hamstrings, and gastrocnemius muscles provide stability to the knee by resisting joint excursions (Winby, Lloyd, Besier, & Kirk, 2009). Thus, an alteration in the activation of muscles of the knee may alter knee loads. Neuromuscular fatigue of the muscles around the knee is associated with a loss of postural control during quiet standing (Gribble & Hertel, 2004; Salavati, Moghadam, Ebrahimi, & Arab, 2007), altered muscle activation strategies, and altered knee mechanics during dynamic tasks such as jumping (Ortiz et al., 2010; A. C. Thomas, McLean, & Palmieri-Smith, 2010). It is unclear how knee mechanics are affected by neuromuscular fatigue during isometric exercises such as squatting and lunging, which are recommended for muscle training rehabilitation for those with knee pain (American Geriatrics Society Panel on Exercise and Osteoarthritis, 2001; Bennell, Wrigley, Hunt, Lim, & Hinman, 2013).

The purpose of this study was to compare muscle and knee loading measures during static squatting and lunging exercises before and after neuromuscular fatigue of the knee extensor and flexor muscles in healthy, young women. Muscle measures included quadriceps and hamstrings activation and co-activation patterns, and the KAM and KFM. We hypothesized that neuromuscular fatigue around the knee would increase lower-limb muscle activations, co-activations of the quadriceps and hamstrings, and knee moments, and that these variables would be higher for lunging than squatting.

3.3 Methods

3.3.1 Participants

A sample of 25 healthy women (age 18 to 30 years) was recruited from the university population. This study focused on women alone as they are more susceptible to knee injuries than men, particularly anterior cruciate ligament (ACL) injury (Prodromos, Han, Rogowski, Joyce, & Shi, 2007). Participants were physically active and had no contraindications to exercise on the Physical Activity Readiness Questionnaire (S. Thomas, Reading, & Shephard, 1992). Exclusion criteria included pregnancy and a history of knee pain, injury, or surgery. This study was approved by the McMaster University Faculty of Health Sciences Human Research Ethics Board.

3.3.2 Design

Participants visited the laboratory twice, one week apart. Written, informed consent was obtained at each visit. The first visit served to familiarize participants with the equipment and protocol to facilitate maximal performance during the second visit (Figure 3.1).

During the second visit, baseline squats and lunges were performed prior to fatigue. These were followed by measures of baseline isometric peak knee extensor and flexor torques, followed by the first round of fatiguing contractions. After fatigue, participants performed the first set of post fatigue squats and lunges (PF1), followed by a second round of fatiguing contractions to ensure that participants had not recovered. Another set of post fatigue squats and lunges (PF2) were performed after fatigue was verified.

3.3.3 Squats and Lunges

Participants performed three squats and three lunges at baseline, PF1, and PF2. During the squats, feet were approximately hip width apart, toes pointed forward, and the right foot alone was in full contact with a force platform. Participants were encouraged to squat without flexing their knees past their toes until their thighs were approximately parallel to the floor while keeping their body mass distributed evenly through the left and right feet. Lunges were performed by stepping forward with the right leg (lead leg) so that the right foot was in full contact with the force platform. Participants were encouraged to lower their torso vertically and bend both knees until the front thigh was approximately parallel to the floor, with the left heel off the ground. Participants were asked to keep their toes pointed forward, their right knee directly above their right ankle. Each squat and lunge was held static for 2 seconds.

A Vicon MX 8 camera motion capture system sampling at 100 Hz measured marker position during squats and lunges (Vicon Motion Systems, Oxford, UK). Eighteen reflective markers were positioned on the legs as required for Vicon's plug-in-gait lower limb marker set (Vicon Motion Systems, Oxford, UK) for the duration of the protocol. Six additional markers were affixed to the left and right iliac crest, greater trochanters, medial epicondyles, and medial malleoli during static standing calibration trials as digital landmarks but were removed before beginning the squats and lunges. These additional markers were used to define the shank and thigh coordinate systems of the leg in accordance with the models described by Grood and Suntay (1983). Knee flexion/extension and ankle transmalleolar axes were generated between the medial and lateral epicondyles and malleoli, rather than using knee and ankle width measurements and hip joint center estimations as described by the plug-in-gait model. Further, the greater trochanters were palpated and identified during static calibration rather than their locations being estimated based on the location of the anterior superior iliac spinae and the leg length of the participant. The 2 second static portion of each trial was identified as the interval in which the vertical velocity of one of the posterior pelvis marker was less than 0.1 m/s.

3.3.4 Knee Flexion Angle

To confirm that the participants achieved the same knee flexion before and after fatigue, the knee flexion angle was calculated from the kinematic motion of the thigh and shank segments (C-Motion, Inc., Germantown, MD, USA).

3.3.5 Peak Torque

Peak torque was measured at baseline, and after each bout of fatiguing contractions, to test for quadriceps or hamstrings fatigue. These measurements were assessed using a dynamometer positioned for testing knee extension and flexion (Biodex Multi-Joint System Setup/Operation Manual, Biodex Medical Systems, Inc., Shirley, NY, USA). Participants performed an isokinetic warm up of 50 contractions at 60°/s. Strength was measured as the peak extension and flexion torque during 5 maximum voluntary isometric contractions (MVICs) at 60° of knee flexion (Knapik, Wright, Mawdsley, & Braun, 1983; Pincivero, Gandaio, & Ito, 2003), held for 5 seconds each. Rests between each contraction were 5 seconds (Symons, Vandervoort, Rice, Overend, & Marsh, 2005). Baseline peak torque was measured after performing baseline squats and lunges, whereas post fatigue peak torques were measured immediately prior to performing PF1 and PF2 squats and lunges. This difference in data collection order increased the convenience of the experimental setup by minimizing the number of times that each participant was positioned on the dynamometer. It is not believed that this difference in data collection order would affect the study outcomes.

3.3.6 Fatigue

Neuromuscular fatigue was induced in the right leg using sets of 50 dynamic isotonic knee extensions and flexions, performed alternately, throughout the knee range of motion at 50% MVIC peak torque on a dynamometer (Biodex Medical Systems, Inc., Shirley, NY, USA). An isotonic fatigue protocol provides insight into functional capacity and is an effective method of inducing neuromuscular fatigue (Stauber, Barill, Stauber, & Miller, 2000). Peak torque decreases occur primarily within the first 40-50 contractions and are minimal past 50 knee extension repetitions as individuals enter the stable endurance phase (Larsson, Karlsson, Eriksson, & Gerdle, 2003). Fatigue was defined as a 25% drop in either peak isometric knee extension or flexion torque from the baseline value. Measures of peak quadriceps torque, before and after fatiguing contractions, are a reliable indicator of knee extensor fatigue (Larsson et al., 2003).

3.3.7 Dependent Measures

3.3.7.1 Electromyography

EMG signals were collected from the right rectus femoris, vastus lateralis, and biceps femoris during squats, lunges, and peak torque measurements. The right leg was both the fatigued leg and the lead leg during lunges. The rectus femoris and biceps femoris are antagonists, biarticulating the knee and hip, and along with the vastus lateralis are key contributors to motion during squatting and lunging (Dionisio, Almeida, Duarte, & Hirata, 2008; Ebben et al., 2009). Muscle activation was monitored using stainless steel surface electrodes (MA-411 EMG Preamplifiers, MA 300, Motion Lab Systems, Inc., Baton Rouge, LA, USA) affixed to the skin along the orientation of the muscle fibers (www.seniam.org, Enschede, Netherlands). A reference electrode was placed over the tibial tubercle. The EMG data were bandpass filtered between 20-450 Hz and sampled at 1000 Hz. The motion capture and EMG data were time synchronized through Vicon Nexus (Vicon Motion Systems, Oxford, UK). Skin preparation included shaving, vigorous wiping with rubbing alcohol, and applying electrode gel between the skin and the electrodes. A quiet trial with participants lying supine and relaxed established a baseline activity level for each muscle.

3.3.7.2 Knee Moments

Ground reaction forces and moments were measured during squatting and lunging using two force platforms sampling at 1000 Hz (Advanced Mechanical Technology Inc., Watertown, MA) that were time synchronized with the motion capture and EMG systems. External KAMs and KFMs were calculated using inverse dynamics and the Joint Coordinate System floating axis model (Grood & Suntay, 1983; Winter, 1984) (C-Motion, Inc., Germantown, MD, USA). Knee moments were normalized to body mass.

3.3.8 Data Analysis

The EMG, motion, and force platform data were extracted for the 2 second static portion of the squats and lunges. Mean knee moments and knee flexion angles were calculated during these 2 seconds. The EMG signals were full-wave rectified and low pass filtered at 6 Hz (4th order Butterworth filter). Baseline activity obtained from the quiet trial was removed from each signal, and the signals obtained during squatting and lunging were amplitude normalized to %MVIC. Mean muscle activation data from all participants was ensemble averaged to obtain one value per muscle for all participants at baseline, PF1, and PF2. Co-activation indices were calculated during squatting and lunging between biceps femoris and rectus femoris and between biceps femoris and vastus lateralis (Kellis, Arabatzi, & Papadopoulos, 2003).

Two-factor repeated measures analyses of variance (ANOVAs) consisting of time (baseline, PF1, and PF2) and task (squats, lunges) were used to determine whether differences existed within each of the dependent variables. Separate ANOVAs were performed for EMG amplitudes (biceps femoris, rectus femoris, vastus lateralis) and co-activations (biceps femoris and rectus femoris, femoris and vastus lateralis) (SPSS Version 15, SPSS Inc., Chicago, Illinois, USA) ($\alpha = 0.05$). Mauchly's test was used to validate the assumption of sphericity. Significant interactions were explored using separate one-way ANOVAs and pair-wise comparisons with a Bonferroni adjustment for multiple comparisons. One-way repeated measures ANOVAs and pair-wise comparisons among estimated marginal means with a Bonferroni adjustment for multiple comparisons were used to assess differences in knee extension and flexion torques, knee moments, and knee flexion angle between baseline, PF1, and PF2 (SPSS Version 15, SPSS Inc., Chicago, Illinois, USA) ($\alpha = 0.05$).

3.4 Results

All participants met the fatigue criterion of a minimum 25% decrease in flexor or extensor MVIC torque. The participants were (in mean ± standard deviation) 23.5 ± 3.6 years old, had a body mass of 63.3 ± 9.5 kg, and had a BMI of 23.7 ± 3.2 kg/m^2. No differences were found between PF1 and PF2 among any of the muscle activation or torque measures, therefore only PF1 values are reported and compared with baseline (Table 3.1). The fatigue protocol decreased peak isometric torque from baseline to PF1 by an average of 43.3 Nm for the knee extensor muscles ($p<0.05$) and 23.8 Nm for the knee flexor muscles ($p<0.05$).

Figure 3.3 shows a time-series of muscle activation during squatting and lunging. A significant time/muscle interaction ($F_{2.157, 51.756} = 6.938$, $p<0.05$) emerged for EMG amplitude, however Mauchly's test indicated that the assumption of sphericity had been violated ($\chi^2(9) = 75.249$, $p<0.05$), therefore degrees of freedom were corrected using Greenhouse-Geisser estimates of sphericity ($\varepsilon = 0.539$). Fatigue decreased vastus lateralis activation during squatting from baseline (31.4 %MVIC) to PF1 (22.3 %MVIC) ($p<0.05$) (Figure 3.4). Likewise, fatigue decreased vastus lateralis activation during lunging (lead leg) from baseline (40.2 %MVIC) to PF1 (28.2 %MVIC) ($p<0.05$). Mean EMG amplitudes were not different between baseline and PF1 for the other muscles during squatting and lunging.

A significant main effect of task indicated that overall EMG amplitudes were higher during lunging than squatting ($F_{1, 24} = 9.426$, $p<0.05$). A significant task/muscle interaction ($F_{2, 48} = 8.653$, $p<0.05$) indicated that at all time points, vastus lateralis activation was higher for lunges than squats, and rectus femoris activation was higher during lunges than squats after fatigue. Muscle co-activation indices were not altered with fatigue or task ($p>0.05$).

Knee moments and the knee flexion angle were calculated for a subgroup of 12 participants. As the hip flexion angle increased during squatting and lunging, the anterior superior iliac spinae markers became obstructed in the other 13 participants. This prohibited the calculation of a hip joint center. No differences were found between PF1 and PF2 among any of the joint moment or angle measures, therefore only PF1 values are reported compared to baseline (Table 3.2). In the subgroup, KAMs and KFMs were higher for lunging than squatting ($p<0.05$) and decreased after fatigue for lunging ($p<0.05$). Figure 3. 2 presents time-series ground reaction force and moment data during a single squat and lunge for a representative participant, highlighting the 2-second static segment that was extracted for analysis of knee moments. Knee flexion angles were higher for lunging than squatting at baseline ($p<0.05$) but were unchanged after fatigue. Post hoc power analyses of all significant test results revealed powers greater than 80% for all tests, except for the analysis of the difference in KFMs between squatting and lunging, which was underpowered at 61%.

3.5 Discussion

Fatigue was associated with a decrease in vastus lateralis activation during squatting and lunging and decreased knee moments in the lead limb during lunging. Activations of the biceps femoris and rectus femoris, co-activations of the quadriceps and hamstrings were not altered by fatigue. Quadriceps activations and knee moments were higher during lunging than squatting. The key implications from these findings are that lower limb fatigue resulted in a decrease in vastus lateralis activation in young, healthy women and that knee joint loads may decrease concurrently with this impairment in quadriceps activation.

Lower extremity fatigue created an altered loading environment in the knee during squatting and lunging. Contrary to our hypotheses, these alterations included decreased external frontal and sagittal moments at the knee and decreased vastus lateralis activation. These decreases were not due to a change in squat and lunge depth between conditions, as indicated by the knee flexion angle. An evaluation of lunges performed to volitional fatigue in 10 young, healthy participants found that during ascent and descent, EMG activation of the vastus lateralis and biceps femoris significantly increased from baseline (Pincivero, Aldworth, Dickerson, Petry, & Shultz, 2000). The discrepancy with the current study may indicate that post fatigue requirements of the thigh muscles differ between static and dynamic tasks. Vastus lateralis activation may have decreased after fatigue in the current study to move toward increasing co-activation with hamstrings. This strategy may be an effort to control knee excursion by increasing resistance to angular joint motion. An increase in co-activation of these opposing muscle groups would reduce the total net joint moment experienced by the knee, creating an apparent decrease in torque generating capacity of the muscles surrounding the knee (Bennell et al., 2013; Heiden, Lloyd, & Ackland, 2009). The trend toward a post fatigue increase in co-activation between the vastus lateralis and biceps femoris may therefore have contributed to an apparent decrease in knee moments; however, the increase in co-activation in the current study was not significant.

A change in quadriceps activation after fatigue has the potential to alter knee joint loads since the quadriceps are a primary contributor to knee flexion-extension moments and a primary resistor of knee abduction-adduction moments during isometric and dynamic motion (Buchanan & Lloyd, 1997; Shelburne et al., 2006). This muscle-load relationship may explain, at least in part, the decrease in the KAM and KFM, although the possibility exists that altered knee loads were experienced due to a change in knee moment demand. Lower extremity neuromuscular fatigue alters knee moments during low demand tasks such as walking in healthy young adults (Longpré, Potvin, & Maly, 2013; Murdock & Hubley-Kozey, 2012; Parijat & Lockhart, 2008); however some inconsistency in joint loading changes after fatigue exists between these studies. All of these studies fatigued the quadriceps and reported a decrease in the knee flexion/extension moment at different points during the gait cycle; however one study reported a higher post-fatigue peak KAM during gait (Murdock & Hubley-Kozey, 2012) while another reported this measure to be unchanged by fatigue (Longpré et al., 2013). Similarly between these studies, post-fatigue activations of the quadriceps trended toward lower values than those at baseline, which is consistent with the current study. Given the higher intensity and demand of the squats and lunges performed in the current study, a greater decrease in vastus lateralis activation may be associated with more substantial decreases in the KAM and KFM than those experienced during post-fatigue gait.

Few studies have evaluated muscle activity differences between squatting and lunging. Quadriceps and hamstrings activations were higher for lunges than squats during dynamic lunging and squatting, weighted with barbells, particularly at higher knee flexion angles (Ebben et al., 2009; Stuart, Meglan, Lutz, Growney, & An, 1996). However, in the current study participants performed these tasks statically and without weights. Activation values in the current study are supported by previous studies, which have evaluated unloaded, or lightly-loaded, static squatting (Escamilla, 2001; Isear, Erickson, & Worrell, 1997) and lunging (Boudreau et al., 2009). Greater vastus lateralis activation during lunging than squatting may indicate that more muscular effort is required of the quadriceps in lunging. This finding may be important for exercise programs designed to increase lower extremity force production capability by provoking increased muscle activation.

To perform the same squatting and lunging tasks after fatigue with a significant reduction in vastus lateralis muscle activity, increased demand may have been placed on other quadriceps muscles such as the vastus medialis and vastus intermedialis. Vastus medialis is an important contributor during static squatting (Dionisio et al., 2008), and may have experienced a change in activation after fatigue; however EMG was not collected on this muscle. To achieve accurate MVICs from participants, knee flexions and extensions were performed on a dynamometer using a thigh strap to limit hip motion. When properly positioned, the thigh strap was tightened over the vastus medialis, which precluded EMG monitoring of this muscle. Previous studies report no difference between vastus medialis and vastus lateralis activations during lower body resistance training

(Escamilla, 2001; Signorile et al., 1994), except that vastus medialis activation decreases at high squat depths (Caterisano et al., 2002).

Having fatigued only the right leg, participants may have shifted the burden to the left leg during squats and lunges after fatigue, potentially increasing the loads experienced by the left knee and muscle activations in order to maintain the static posture. While the current study did not evaluate muscle activations and knee loads in the contralateral limb, an evaluation of 17 healthy, young participants found no difference in vertical ground reaction force asymmetry before and after fatigue during a squat exercise, instead reporting a trend toward increasing symmetry between limbs after fatigue (Hodges, Patrick, & Reiser, 2011). While KFMs were higher during lunging than squatting at baseline, caution should be taken in the interpretation of these results, as this study was underpowered to reveal these differences. This difference was no longer present after fatigue, as the KFM during lunging decrease significantly from baseline.

Low intensity, long duration fatigue achieved during occupational or recreational squatting, lunging, lifting, or long distance running is difficult to reproduce in a laboratory setting. Contrary to studies that induced lower extremity fatigue using repetitive isokinetic or isometric contractions, the current study used high intensity, short duration repetitive isotonic contractions where the load is held constant and angular velocity changes throughout the range of motion. This method may better represent fatigue caused by the repetitive performance of activities of daily living which require bursts of activity to moderate varying velocities of joint motion (Cheng & Rice, 2005).

In conclusion, lower extremity fatigue alters the recruitment strategy of the quadriceps and the adduction and flexion knee moments during squatting and lunging. Future studies should explore the contributions of other lower extremity muscles to knee loading during squatting and lunging and the bilateral effects of lower extremity neuromuscular fatigue. Lunging challenges quadriceps activation better than squatting in young, healthy women and may be more useful for training and rehabilitation programs focused on lower extremity muscle training.

3.6 References

American Geriatrics Society Panel on Exercise and Osteoarthritis. (2001). Exercise prescription for older adults with osteoarthritis pain: Consensus practice recommendations: A supplement to the AGS clinical practice guidelines on the management of chronic pain in older adults. Journal of the American Geriatrics Society, 49(10), 1400.

Amin, S., Goggins, J., Niu, J., Guermazi, A., Grigoryan, M., Hunter, D. J., . . . Felson, D. T. (2008). Occupation-related squatting, kneeling, and heavy lifting and the knee joint: A magnetic resonance imaging-based study in men. The Journal of Rheumatology, 35(8), 1645-1649.

Andriacchi, T. P., & Mundermann, A. (2006). The role of ambulatory mechanics in the initiation and progression of knee osteoarthritis. Current Opinion in Rheumatology, 18(1040-8711; 5), 514-518.

Baliunas, A. J., Hurwitz, D. E., Ryals, A. B., Karrar, A., Case, J. P., Block, J. A., & Andriacchi, T. P. (2002). Increased knee joint loads during walking are present in subjects with knee osteoarthritis. Osteoarthritis and Cartilage / OARS, Osteoarthritis Research Society, 10(1063-4584; 7), 573-579.

Bennell, K. L., Wrigley, T. V., Hunt, M. A., Lim, B., & Hinman, R. S. (2013). Update on the role of muscle in the genesis and management of knee osteoarthritis. Rheumatic Diseases Clinics of North America, 39(1), 145-176.

Boudreau, S. N., Dwyer, M. K., Mattacola, C. G., Lattermann, C., Uhl, T. L., & McKeon, J. M. (2009). Hip-muscle activation during the lunge, single-leg squat, and step-up-and-over exercises. Journal of Sport Rehabilitation, 18(1), 91.

Buchanan, T. S., & Lloyd, D. G. (1997). Muscle activation at the human knee during isometric flexion-extension and varus-valgus loads. Journal of Orthopaedic Research, 15(1), 11-17.

Caterisano, A., Moss, R., Pellinger, T., Woodruff, K., Lewis, V., Booth, W., & Khadra, T. (2002). The effect of back squat depth on the EMG activity of 4 superficial hip and thigh muscles. The Journal of Strength & Conditioning Research, 16(3), 428-432.

Cheng, A. J., & Rice, C. L. (2005). Fatigue and recovery of power and isometric torque following isotonic knee extensions. Journal of Applied Physiology (Bethesda, Md.: 1985), 99(4), 1446-1452. doi:10.1152/japplphysiol.00452.2005

Deluzio, K. J., & Astephen, J. L. (2007). Biomechanical features of gait waveform data associated with knee osteoarthritis: An application of principal component analysis. Gait & Posture, 25(1), 86-93.

Dionisio, V. C., Almeida, G. L., Duarte, M., & Hirata, R. P. (2008). Kinematic, kinetic and EMG patterns during downward squatting. Journal of Electromyography and Kinesiology : Official Journal of the International Society of Electrophysiological Kinesiology, 18(1), 134-143. doi:10.1016/j.jelekin.2006.07.010

Ebben, W. P., Feldmann, C. R., Dayne, A., Mitsche, D., Alexander, P., & Knetzger, K. J. (2009). Muscle activation during lower body resistance training. International Journal of Sports Medicine, 30(1), 1-8. doi:10.1055/s-2008-1038785; 10.1055/s-2008-1038785

Escamilla, R. F. (2001). Knee biomechanics of the dynamic squat exercise. Medicine and Science in Sports and Exercise, 33(1), 127-141.

Foroughi, N., Smith, R., & Vanwanseele, B. (2009). The association of external knee adduction moment with biomechanical variables in osteoarthritis: A systematic review. The Knee, 16(5), 303-309. doi:10.1016/j.knee.2008.12.007

Gandevia, S., Allen, G., & McKenzie, D. (1995). Central fatigue. Advances in Experimental Medicine and Biology, 384, 281-294.

Gribble, P. A., & Hertel, J. (2004). Effect of lower-extremity muscle fatigue on postural control. Archives of Physical Medicine and Rehabilitation, 85(4), 589-592.

Griffin, T. M., & Guilak, F. (2005). The role of mechanical loading in the onset and progression of osteoarthritis. Exercise and Sport Sciences Reviews, 33(0091-6331; 0091-6331; 4), 195-200.

Grood, E. S., & Suntay, W. J. (1983). A joint coordinate system for the clinical description of three-dimensional motions: Application to the knee. Journal of Biomechanical Engineering, 105(2), 136-44.

Heiden, T. L., Lloyd, D. G., & Ackland, T. R. (2009). Knee extension and flexion weakness in people with knee osteoarthritis: Is antagonist cocontraction a factor? The Journal of Orthopaedic and Sports Physical Therapy, 39(11), 807-815. doi:10.2519/jospt.2009.3079

Hodges, S. J., Patrick, R. J., & Reiser, R. F. (2011). Effects of fatigue on bilateral ground reaction force asymmetries during the squat exercise. The Journal of Strength & Conditioning Research, 25(11), 3107-3117.

Isear, J., Erickson, J., & Worrell, T. (1997). EMG analysis of lower extremity muscle recruitment patterns during an unloaded squat. Medicine and Science in Sports and Exercise, 29(4), 532-539.

Kellis, E., Arabatzi, F., & Papadopoulos, C. (2003). Muscle co-activation around the knee in drop jumping using the co-contraction index. Journal of Electromyography and Kinesiology : Official Journal of the International Society of Electrophysiological Kinesiology, 13(3), 229-238.

Knapik, J. J., Wright, J. E., Mawdsley, R. H., & Braun, J. (1983). Isometric, isotonic, and isokinetic torque variations in four muscle groups through a range of joint motion. Physical Therapy, 63(6), 938-947.

Larsson, B., Karlsson, S., Eriksson, M., & Gerdle, B. (2003). Test–retest reliability of EMG and peak torque during repetitive maximum concentric knee extensions. Journal of Electromyography and Kinesiology, 13(3), 281-287. doi:10.1016/S1050-6411(03)00022-1

Longpré, H. S., Potvin, J. R., & Maly, M. R. (2013). Biomechanical changes at the knee after lower limb fatigue in healthy young women. Clinical Biomechanics, 28(4), 441-447.

Murdock, G., & Hubley-Kozey, C. (2012). Effect of a high intensity quadriceps fatigue protocol on knee joint mechanics and muscle activation during gait in young adults. European Journal of Applied Physiology, 112(2), 439-449. doi:10.1007/s00421-011-1990-4

Ortiz, A., Olson, S. L., Etnyre, B., Trudelle-Jackson, E. E., Bartlett, W., & Venegas-Rios, H. L. (2010). Fatigue effects on knee joint stability during two jump tasks in women. Journal of Strength and Conditioning Research / National Strength & Conditioning Association, 24(4), 1019-1027. doi:10.1519/JSC.0b013e3181c7c5d4; 10.1519/JSC.0b013e3181c7c5d4

Parijat, P., & Lockhart, T. E. (2008). Effects of quadriceps fatigue on the biomechanics of gait and slip propensity. Gait & Posture, 28(4), 568-573.

Pincivero, D. M., Aldworth, C., Dickerson, T., Petry, C., & Shultz, T. (2000). Quadriceps-hamstring EMG activity during functional, closed kinetic chain exercise to fatigue. European Journal of Applied Physiology, 81(6), 504-509. doi:10.1007/s004210050075

Pincivero, D. M., Gandaio, C. B., & Ito, Y. (2003). Gender-specific knee extensor torque, flexor torque, and muscle fatigue responses during maximal effort contractions. European Journal of Applied Physiology, 89(2), 134-141.

Prodromos, C. C., Han, Y., Rogowski, J., Joyce, B., & Shi, K. (2007). A meta-analysis of the incidence of anterior cruciate ligament tears as a function of gender, sport, and a knee injury–reduction regimen. Arthroscopy: The Journal of Arthroscopic & Related Surgery, 23(12), 1320-1325. e6.

Richmond, S. A., Fukuchi, R. K., Ezzat, A., Schneider, K., Schneider, G., & Emery, C. A. (2013). Are joint injury, sport activity, physical activity, obesity, or occupational activities predictors for osteoarthritis? A systematic review. The Journal of Orthopaedic and Sports Physical Therapy, 43(8), 515-524. doi:10.2519/jospt.2013.4796; 10.2519/jospt.2013.4796

Rudolph, K. S., Axe, M. J., Buchanan, T. S., Scholz, J. P., & Snyder-Mackler, L. (2001). Dynamic stability in the anterior cruciate ligament deficient knee. Knee Surgery, Sports Traumatology, Arthroscopy, 9(2), 62-71.

Salavati, M., Moghadam, M., Ebrahimi, I., & Arab, A. M. (2007). Changes in postural stability with fatigue of lower extremity frontal and sagittal plane movers. Gait & Posture, 26(2), 214-218. doi:10.1016/j.gaitpost.2006.09.001

Sharma, L., Hurwitz, D. E., Thonar, E. J., Sum, J. A., Lenz, M. E., Dunlop, D. D., Schnitzer, T. J., Kirwan-Mellis, G., & Andriacchi, T. P. (1998). Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis. Arthritis and Rheumatism, 41(0004-3591; 7), 1233-1240.

Shelburne, K. B., Torry, M. R., & Pandy, M. G. (2006). Contributions of muscles, ligaments, and the ground-reaction force to tibiofemoral joint loading during normal gait. Journal of Orthopaedic Research, 24(10), 1983-1990.

Signorile, J. F., Weber, B., Roll, B., Caruso, J. F., Lowensteyn, I., & Perry, A. C. (1994). An electromyographical comparison of the squat and knee extension exercises. The Journal of Strength & Conditioning Research, 8(3), 178-183.

Stauber, W., Barill, E., Stauber, R., & Miller, G. (2000). Isotonic dynamometry for the assessment of power and fatigue in the knee extensor muscles of females. Clinical Physiology, 20(3), 225-233.

Stuart, M. J., Meglan, D. A., Lutz, G. E., Growney, E. S., & An, K. (1996). Comparison of intersegmental tibiofemoral joint forces and muscle activity during various closed kinetic chain exercises. The American Journal of Sports Medicine, 24(6), 792-799.

Symons, T. B., Vandervoort, A. A., Rice, C. L., Overend, T. J., & Marsh, G. D. (2005). Effects of maximal isometric and isokinetic resistance training on strength and functional mobility in older adults. The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 60(6), 777-781.

Thomas, S., Reading, J., & Shephard, R. J. (1992). Revision of the physical activity readiness questionnaire (PAR-Q). Canadian Journal of Sport Sciences, 17(4), 338-345.

Thomas, A. C., McLean, S. G., & Palmieri-Smith, R. M. (2010). Quadriceps and hamstrings fatigue alters hip and knee mechanics. Journal of Applied Biomechanics, 26(2), 159-170.

Winby, C. R., Lloyd, D. G., Besier, T. F., & Kirk, T. B. (2009). Muscle and external load contribution to knee joint contact loads during normal gait. Journal of Biomechanics, 42(14), 2294-2300.

Winter, D. A. (1984). Biomechanics of human movement with applications to the study of human locomotion. Critical Reviews in Biomedical Engineering, 9(4), 287-314.

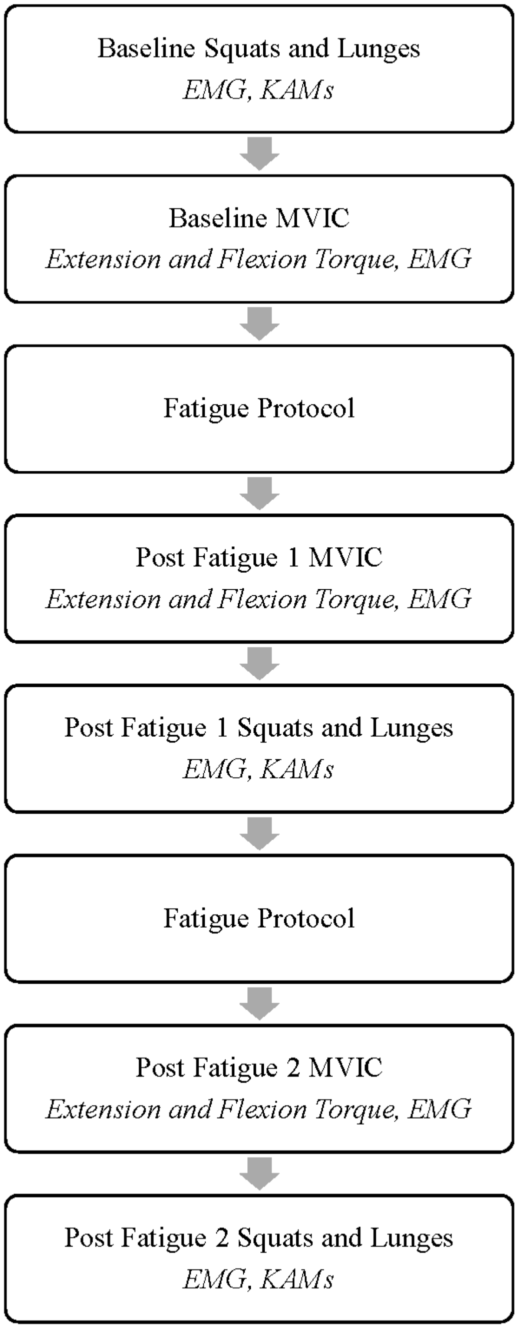

Figure 3. 1. Study design flow chart for the second laboratory visit. Measures collected at each step are shown in italics. Note: External knee adduction moments (KAMs) were collected on a subgroup (n=12).

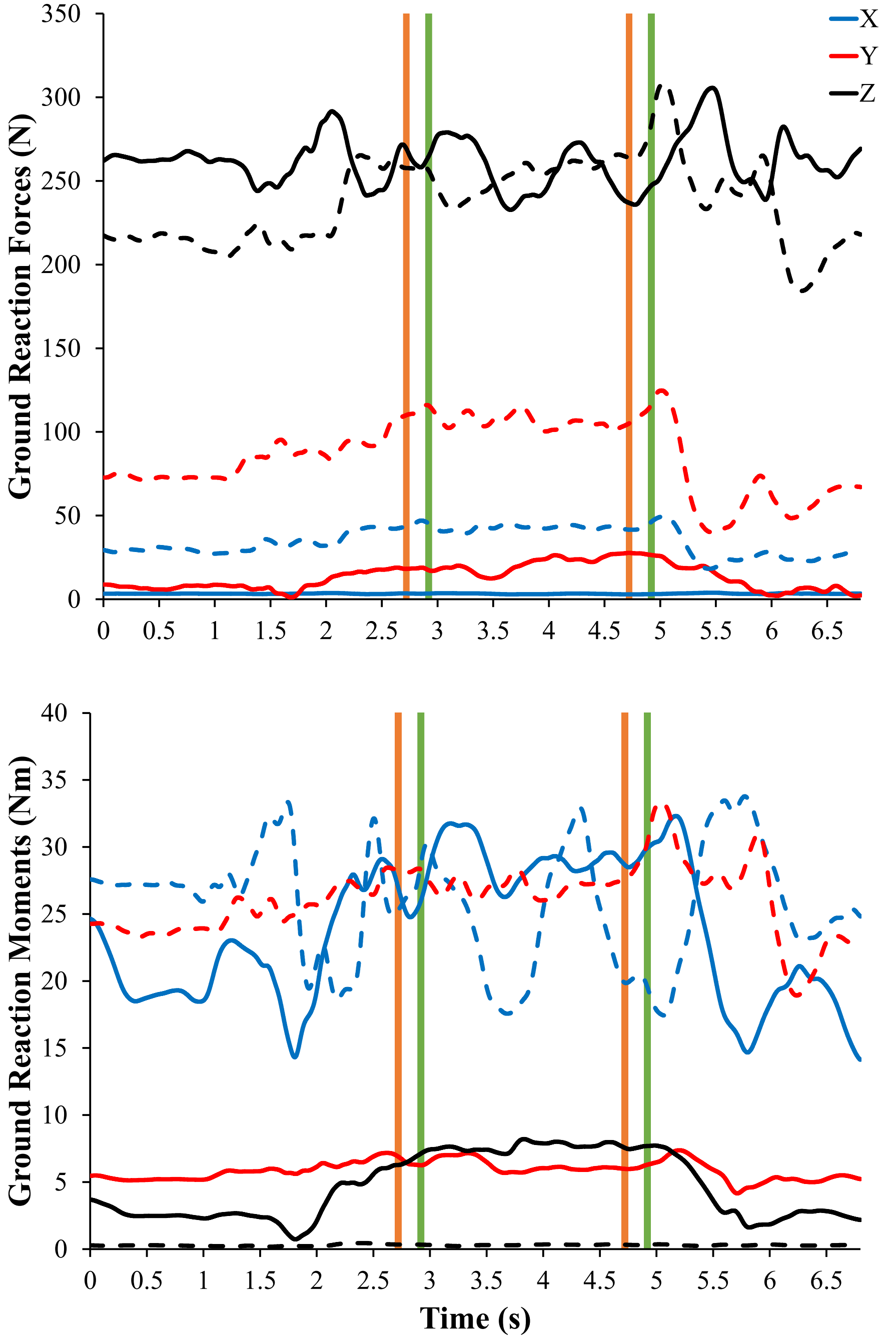

Figure 3. 2. Time-series ground reaction forces (N, top panel) and moments (Nm, bottom panel) during a single squat (solid lines) and lunge (dashed lines) for a representative participant. The 2-second static segment occurred from 2.7-4.7 seconds for squatting (orange vertical lines), and from 2.9-4.9 seconds for lunging (green vertical lines). Approximate representative directions of force based on the orientation of the foot and shank during the squat and lunge:
X – medial-lateral, Y – anterior-posterior, Z – proximal-distal

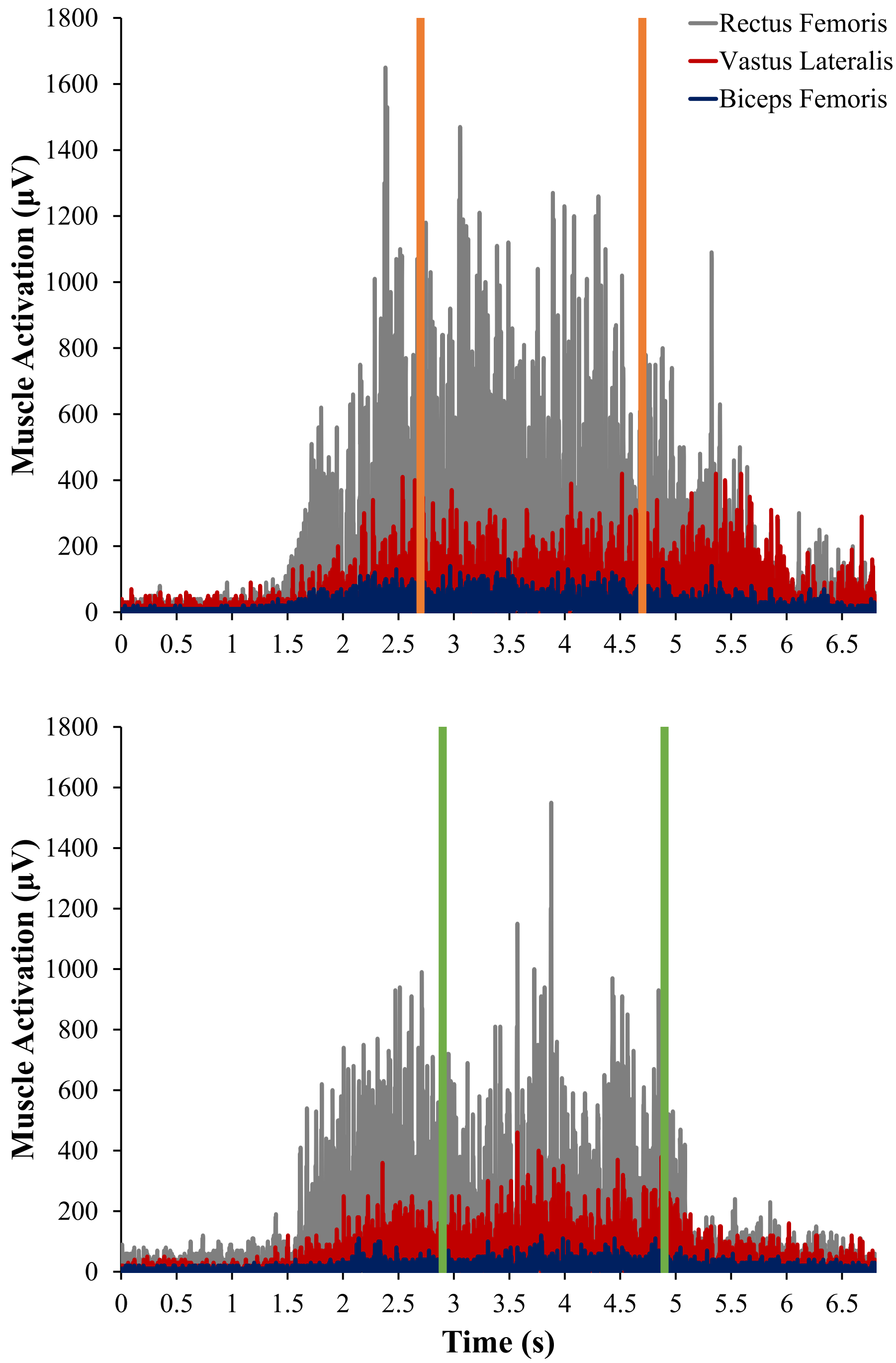

Figure 3. 3. Time-series, unfiltered, rectified muscle activation (µV) data during a single squat (top panel) and lunge (bottom panel) for a representative participant. The 2-second static segment occurred from 2.7-4.7 seconds for squatting (orange vertical lines), and from 2.9-4.9 seconds for lunging (green vertical lines).

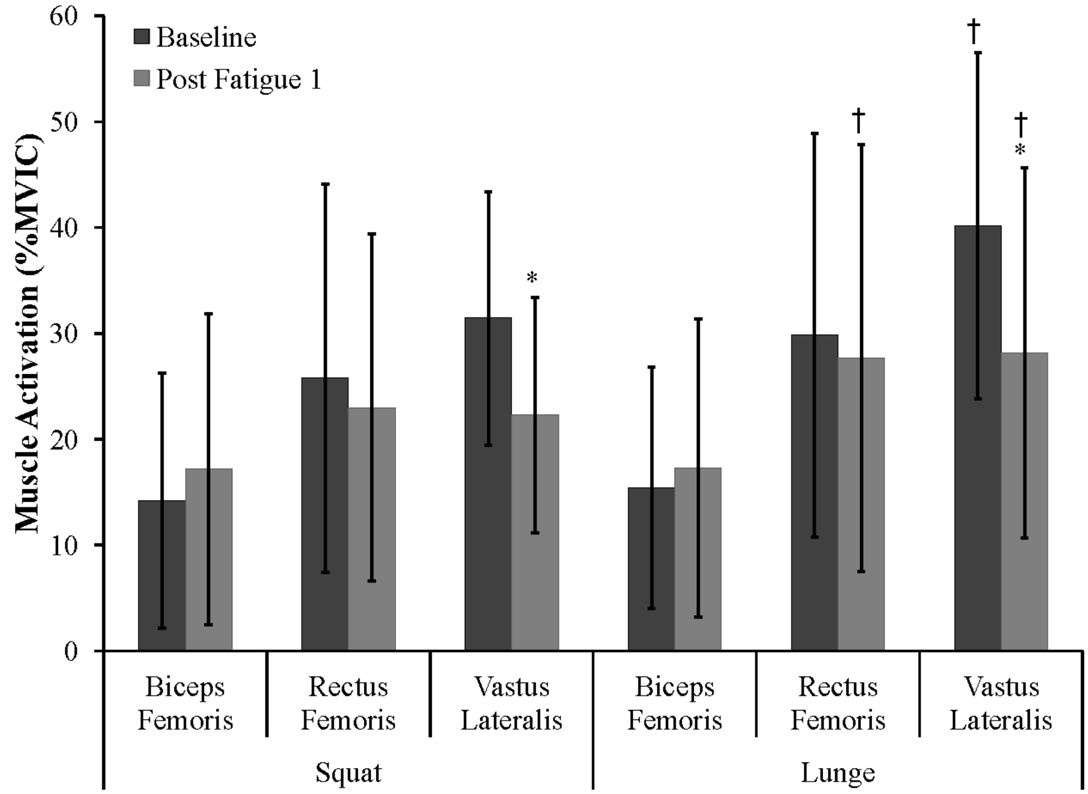

Figure 3. 4. Muscle activation (%MVIC) during static squatting and lunging reported at baseline and after the first bout of fatigue. Values are means (n=25) with standard deviations represented as error bars.
***Significant difference (p<0.05) from baseline value.**
† Significant difference (p<0.05) between squatting and lunging at the indicated time point.

Table 3. 1. Mean EMG activation (% MVIC) and co-activation indices (%) during the static portion (2 seconds) of squatting (right leg) and lunging (lead leg). Also presented are the peak isometric torques (Nm) for knee extensions (ext) and flexions (flex) (n=25). Measures are reported at baseline and after the first bout of fatigue as mean (standard deviation) [95% Confidence Interval].

	Baseline		Post Fatigue	
	Squat	*Lunge*	*Squat*	*Lunge*
Mean EMG Activation (%MVIC)				
BF	14.2 (12.0) [9.2, 19.2]	15.4 (11.4) [10.7, 20.1]	17.2 (14.7) [11.1, 23.3]	17.3 (14.1) [11.5, 23.1]
RF	25.8 (18.3) [18.2, 33.4]	29.8 (19.1) [22.0, 37.7]	23.0 (16.4) [16.2, 29.8]	27.7 (20.2) [19.4, 36.0]†
VL	31.4 (12.0) [26.5, 36.4]	40.2 (16.4) [33.4, 46.9]†	22.3 (11.1) [17.7, 26.9]*	28.2 (17.5) [21.0, 35.4]*†
Co-activation Index (%)				
BF/RF	27.8 (11.1) [23.2, 32.4]	28.4 (10.8) [24.0, 32.9]	25.7 (12.4) [20.6, 30.9]	25.1 (12.2) [20.1, 30.1]
BF/VL	27.5 (10.3) [23.2, 31.7]	25.2 (10.3) [20.9, 29.4]	30.2 (13.2) [24.7, 35.6]	29.0 (13.2) [23.6, 34.4]
	Baseline		Post Fatigue	
Peak Isometric Torque (Nm)				
Ext.	170.3 (29.0) [158.3, 182.3]		127.0 (29.2) [114.9, 139.0]*	
Flex.	81.3 (14.5) [75.3, 87.2]		57.5 (10.7) [53.0, 61.9]*	

BF = biceps femoris, RF = rectus femoris, VL = vastus lateralis
*Significantly different ($p < 0.05$) from baseline value.
†Significantly different ($p < 0.05$) from squatting.

Table 3. 2. Mean external knee adduction moment (Nm/kg), external knee flexion moment (Nm/k), and knee flexion angle (degrees) during the static portion (2 seconds) of squatting (right leg) and lunging (lead leg) for the subgroup (n=12) of participants. Measures are reported at baseline and after the first bout of fatigue as mean (standard deviation) [95% Confidence Interval].

	Baseline		Post Fatigue	
	Squat	*Lunge*	*Squat*	*Lunge*
Knee Adduction Moment (Nm/kg)	0.03 (0.11) [-0.03, 0.09]	0.50 (0.13) [0.43, 0.57]†	0.04 (0.10) [-0.02, 0.1]	0.42 (0.12) [0.35, 0.49]*†
Knee Flexion Moment (Nm/kg)	0.39 (0.15) [0.30, 0.49]	0.55 (0.20) [0.42, 0.68]†	0.34 (0.15) [0.25, 0.44]	0.44 (0.17) [0.34, 0.55]*
Knee Flexion Angle (degrees)	72.1 (11.7) [64.6, 79.5]	84.7 (7.5) [80.0, 89.5]†	72.7 (10.9) [65.8, 79.6]	79.9 (9.1) [74.0, 85.7]

*Significantly different ($p < 0.05$) from baseline value.
†Significantly different ($p < 0.05$) from squatting.

Chapter Four

Knee Extensor Torque and Power are related to Peak Medial Knee Loads in Young Healthy Women

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Publication Status: Submitted as a Short Communication to *Gait and Posture*

This manuscript extends beyond the traditional measure of peak torque as an evaluation of muscle function, examining the contributions of peak power production to knee loading during gait. Furthermore, peak torque and peak power contributions of both the quadriceps and hamstrings muscle groups are considered. These results support a more robust understanding of muscle function contributions to knee biomechanics during a key activity of daily living.

4.1 Abstract

The purpose of this study was to examine the contributions of knee extensor and flexor strength and power to the variance in the peaks of the knee adduction moments in young women. Knee adduction moments during gait were measured from motion and force data on 25 healthy women (mean of 23.5 years). The first and second knee adduction moment peaks were averaged from five walking trials per participant and were normalized to body mass (Nm/kg). Peak isometric knee extensor and flexor torque (Nm/kg), and peak isotonic knee extensor and flexor power were measured on a dynamometer and were normalized to body mass (W/kg). Correlations and linear regressions were used to examine the relationship of peak knee extension and flexion torques and powers with the two peaks of the knee adduction moment, controlling for gait speed. The knee adduction moment peaks were significantly correlated with peak knee extension torque and power ($p<0.05$). Peak knee extension torque explained 25% of the variance in the first peak ($p<0.01$). Peak knee extension power explained 12% of the variance in the second peak ($p=0.05$). The magnitude and rate of lower extremity torque production are important in understanding medial compartment loads in healthy knees. Knee extensor strength is an important contributor to the peak knee adduction moment during gait just after heel-strike, while knee extensor power relates with the second peak as the foot prepares for push-off. Knee flexor strength and power did not contribute to the variance in the peak knee adduction moments.

Keywords: Muscle Strength; Osteoarthritis, Knee; Gait; Isotonic Contraction

4.2 Introduction

The knee adduction moment (KAM) during gait acts as a surrogate for medial tibiofemoral loading (Foroughi, Smith, & Vanwanseele, 2009). The KAM waveform generally consists of two peaks (KAM1 and KAM2) which increase with osteoarthritis (OA) severity (Bennell et al., 2011; Miyazaki et al., 2002). The peaks occur as the body's center of gravity accelerates downward. KAM1 occurs in early stance during load acceptance, when the foot is braking vertically. KAM2 occurs during late stance as the foot prepares for vertical propulsion. This suggests that KAM1 and KAM2 may be controlled by different mechanisms.

Reduced strength in the knee extensors and flexors refers to a decline in torque output during maximum voluntary contractions. Strength deficits are common in knee OA (Bennell, Wrigley, Hunt, Lim, & Hinman, 2013) and have been postulated to hinder load distribution across the knee (Rice, McNair, & Lewis, 2011). However, little association was found between quadriceps strength and the peak KAM in knee OA individuals (Lim et al., 2009). The KAM peaks reflect the downward acceleration of the foot and tibia and may be better represented by a measure accounting for the rate of moment development.

Knee power, which refers to the rate of torque production through a range of motion, might better indicate the quadriceps and hamstrings ability to mediate loads across the knee during dynamic tasks (Sayers, 2007; Sayers & Gibson, 2010). Knee extensor power is more useful than strength in explaining physical performance in healthy aging (Bean et al., 2002)(Sayers & Gibson, 2010) and clinical knee OA (Calder et al., 2013). Knee flexor contributions to peak KAMs and relationships between strength and power with KAM2 remain unclear. The purpose of this study was to examine the extent to which knee extensor and flexor strength and power explain the variance in KAM1 and KAM2 in young women.

4.3 Methods

4.3.1 Participants and Design

Healthy, physically active young women, with no history of knee pain, injury, or surgery participated. KAMs were measured during gait analysis, followed by measures of peak isometric knee torque, and peak isotonic knee power. Participants gave written, informed consent and the McMaster University Faculty of Health Sciences Human Research Ethics Board approved the study.

4.3.2 Dependent Measures

4.3.2.1 Knee Adduction Moments

External KAMs were calculated during barefoot, self-selected speed gait. An 8 camera motion capture system sampling at 100 Hz measured marker motion (Vicon Motion Systems, Oxford, UK) was synchronized with three force platforms measuring ground reaction forces and moments sampling at 1000 Hz (Advanced Mechanical Technology Inc., Watertown, MA). Based on Vicon's plug-in-gait model, eighteen reflective markers were affixed to the lower limbs. Six additional markers were affixed to the left and right iliac crest, greater trochanters, medial epicondyles, and medial malleoli during standing calibration trials as digital landmarks.

Marker and force platform data were filtered with a dual-pass fourth order Butterworth low-pass filter at a cut-off frequency of 6 Hz. KAMs were calculated with the Joint Coordinate System floating axis model (Grood & Suntay, 1983) (C-Motion, Inc., Germantown, MD, USA). KAMs were amplitude normalized to body mass (Nm/kg) and time normalized to percent gait cycle. KAM1 and KAM2 were averaged from 5 gait trials per participant.

4.3.3 Independent Measures

4.3.3.1 Peak Knee Extension and Flexion Torques

Extension and flexion torques were measured in the right leg using a dynamometer (Biodex Multi-Joint System Setup/Operation Manual, Biodex Medical Systems, Inc., Shirley, NY, USA). Peak torques were obtained during a set of 5 (5-second) maximum voluntary isometric contractions (MVICs), at 60° of knee flexion. These torques were normalized to body mass for each participant (Nm/kg) (Bennell et al., 2013).

4.3.3.2 Peak Knee Extension and Flexion Powers

Participants performed repetitive isotonic extensions and flexions of the right leg throughout the knee range of motion on the dynamometer at 50% of their peak torque during MVIC. Torque and angular velocity were measured and multiplied to calculate power at each frame of data (Sayers, 2007). The 5 peak extension and flexion powers were extracted, normalized to body mass for each participant (W/kg), and averaged to represent one peak knee extension and flexion power value.

4.3.4 Covariate

Gait speed (m/s) was calculated between the first and second heel strikes of the right foot and was examined as a potential covariate since a reduced gait speed may lower the peak KAM (Mundermann, Dyrby, Hurwitz, Sharma, & Andriacchi, 2004).

4.3.5 Data Analyses

Pearson's correlation coefficients described the bivariate relationships between the covariate, dependent and independent measures. Separate stepwise linear regressions for each of KAM1 and KAM2 assessed the amount of variance in the dependent measures explained by peak torques and powers while controlling for gait speed. Entry and removal p-values were 0.05 and 0.1, respectively (SPSS Version 15, SPSS Inc., Chicago, Illinois, USA).

4.4 Results

Twenty-five women participated and were (in mean (standard deviation)) 23.5 (3.6) years old, had a body mass of 63.3 (9.5) kg, and had a BMI of 23.7 (3.2) kg/m^2. KAM1 and KAM2 were significantly correlated with peak extension torque and power (r range: 0.37-0.53) (Table 4.1, Figure 4.1). Neither KAM1 nor KAM2 were related to peak flexion torque or power, or gait speed.

Stepwise linear regression analyses for KAM1 and KAM2 (Table 4.2) revealed variance inflation factors of 6.23 and 5.94 for peak flexion power and peak extension power, respectively, indicating multicollinearity. Peak flexion power was removed from the analysis as it was not correlated with either dependent variable. Peak extension torque explained 25% of the variance in KAM1 ($p<0.01$). Gait speed, peak flexion torque, and peak extension power did not add significantly to the model. Peak extension power explained 12% of the variance in KAM2 ($p=0.05$). Gait speed, and peak extension and flexion torque did not add significantly to the model.

4.5 Discussion

Both magnitude and rate of lower limb force production are important in understanding peak KAMs in healthy knees. Strength and power contributions to knee loading during gait differ between young healthy women and women with knee OA (Calder et al., 2013). It is unclear whether these contributions change with age, training, or as a consequence of joint disease.

KAM1 was higher in women with greater knee extensor strength. Strength may be important in braking during heel-strike. Increased muscle strength around the knee may result in higher coactivation across the joint, potentially increasing tibiofemoral forces during gait (Bennell et al., 2013). While reports of extensor strength effects on radiographic OA progression are mixed (Segal et al., 2009; Slemenda et al., 1998), higher quadriceps strength protects against the development of symptomatic knee OA (Segal et al., 2009). Further work should evaluate the relationship between quadriceps strength and muscle contributions to knee loading through direct measures of joint contact forces.

Knee extensor power was positively correlated with KAM2 and explained 12% of its variance. This suggests that the magnitude of KAM2, produced in preparation for push-off from the ground, has more association with power produced by the quadriceps than strength. Nevertheless, KAM2 is likely also correlated with several other factors such as varus-valgus knee angle, body mass, and toe-out angle during gait. While power related to KAM1 in women with knee OA (Calder et al., 2013), decelerating joint moments quickly during gait may provide a protective mechanism. Increased extensor power may increase the knees ability to resist external excursions, decreasing the risk of injury (Bean et al., 2002; Sayers, 2007). However, it is not possible to know from the results of the current study what effects knee power have on KAM1 or KAM2.

Knee flexor strength and power did not contribute significantly to the variance in peak KAM. This supports previous research observing that the quadriceps and gastrocnemius contributed primarily to tibiofemoral loading during gait while hamstrings contributed minimally (Shelburne, Torry, & Pandy, 2006). For a submaximal task like walking, peak flexor strength may not be reflected in the contribution of the hamstrings to the peak KAMs. Longitudinal data should determine whether increases in knee strength and power alter peak KAMs, however it is clear that both extensor strength and power are important contributors to peak KAMs in this sample of young healthy women.

4.6 References

Bean, J. F., Kiely, D. K., Herman, S., Leveille, S. G., Mizer, K., Frontera, W. R., & Fielding, R. A. (2002). The relationship between leg power and physical performance in Mobility-Limited older people. Journal of the American Geriatrics Society, 50(3), 461-467.

Bennell, K. L., Bowles, K., Wang, Y., Cicuttini, F., Davies-Tuck, M., & Hinman, R. S. (2011). Higher dynamic medial knee load predicts greater cartilage loss over 12 months in medial knee osteoarthritis. Annals of the Rheumatic Diseases, 70(10), 1770-1774.

Bennell, K. L., Wrigley, T. V., Hunt, M. A., Lim, B., & Hinman, R. S. (2013). Update on the role of muscle in the genesis and management of knee osteoarthritis. Rheumatic Diseases Clinics of North America, 39(1), 145-176.

Calder, K. M., Acker, S. M., Arora, N., Beattie, K. A., Callaghan, J. P., Adachi, J. D., & Maly, M. R. (2014). Knee power is an important parameter in understanding medial knee joint load in knee osteoarthritis. Arthritis Care & Research, 66(5), 687-94.

Foroughi, N., Smith, R., & Vanwanseele, B. (2009). The association of external knee adduction moment with biomechanical variables in osteoarthritis: A systematic review. The Knee, 16(5), 303-309. doi:10.1016/j.knee.2008.12.007

Grood, E. S., & Suntay, W. J. (1983). A joint coordinate system for the clinical description of three-dimensional motions: Application to the knee. Journal of Biomechanical Engineering, 105, 136.

Lim, B., Kemp, G., Metcalf, B., Wrigley, T. V., Bennell, K. L., Crossley, K. M., & Hinman, R. S. (2009). The association of quadriceps strength with the knee adduction moment in medial knee osteoarthritis. Arthritis Care & Research, 61(4), 451-458.

Miyazaki, T., Wada, M., Kawahara, H., Sato, M., Baba, H., & Shimada, S. (2002). Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis. Annals of the Rheumatic Diseases, 61(7), 617-622.

Mundermann, A., Dyrby, C. O., Hurwitz, D. E., Sharma, L., & Andriacchi, T. P. (2004). Potential strategies to reduce medial compartment loading in patients with knee osteoarthritis of varying severity: Reduced walking speed. Arthritis and Rheumatism, 50(0004-3591; 4), 1172-1178.

Rice, D. A., McNair, P. J., & Lewis, G. N. (2011). Mechanisms of quadriceps muscle weakness in knee joint osteoarthritis: The effects of prolonged vibration on torque and muscle activation in osteoarthritic and healthy control subjects. Arthritis Research & Therapy, 13(5), R151. doi:10.1186/ar3467

Sayers, S. P. (2007). High-speed power training: Anovel approach to resistance training in older men and women. abrief review and pilot study. The Journal of Strength & Conditioning Research, 21(2), 518-526.

Sayers, S. P., & Gibson, K. (2010). A comparison of high-speed power training and traditional slow-speed resistance training in older men and women. The Journal of Strength & Conditioning Research, 24(12), 3369-3380.

Segal, N. A., Torner, J. C., Felson, D., Niu, J., Sharma, L., Lewis, C. E., & Nevitt, M. (2009). Effect of thigh strength on incident radiographic and symptomatic knee osteoarthritis in a longitudinal cohort. Arthritis and Rheumatism, 61(0004-3591; 0004-3591; 9), 1210-1217.

Shelburne, K. B., Torry, M. R., & Pandy, M. G. (2006). Contributions of muscles, ligaments, and the ground-reaction force to tibiofemoral joint loading during normal gait. Journal of Orthopaedic Research, 24(10), 1983-1990.

Slemenda, C., Heilman, D. K., Brandt, K. D., Katz, B. P., Mazzuca, S. A., Braunstein, E. M., & Byrd, D. (1998). Reduced quadriceps strength relative to body weight: A risk factor for knee osteoarthritis in women? Arthritis and Rheumatism, 41(0004-3591; 0004-3591; 11), 1951-1959.

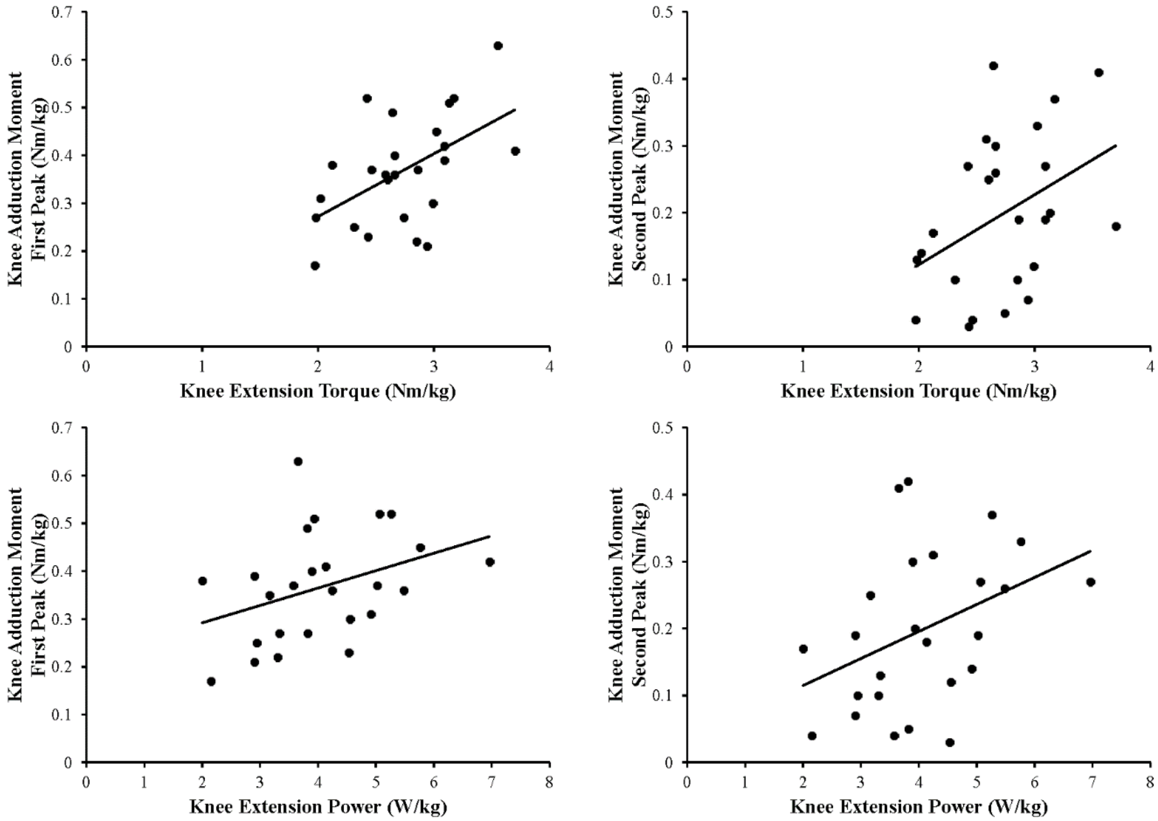

Figure 4. 1. Relationships between dependent variables (KAM1 and KAM2) and knee extension torque and power.

Table 4. 1. Descriptive statistics (mean, standard deviation (SD), and range) and Pearson's correlation coefficients (2-tailed p-values, significant relationships (α = 0.05) in italics) between the covariate, independent, and dependent variables.

Variable	KAM1	KAM2	Gait Speed	Ext Torque	Flex Torque	Ext Power	Flex Power
Mean (SD)	0.37 (0.11)	0.20 (0.12)	1.25 (0.16)	2.72 (0.45)	1.30 (0.22)	4.05 (1.16)	2.47 (0.86)
[Range]	[0.17-0.63]	[0.03-0.42]	[0.95-1.58]	[1.97-3.70]	[0.91-1.69]	[2.00-6.96]	[1.15-4.44]
KAM1 (Nm/kg)	1	0.82 (*<0.01*)	-0.10 (0.64)	0.53 (*<0.01*)	<-0.01 (0.99)	0.37 (0.07)	0.18 (0.40)
KAM2 (Nm/kg)		1	-0.12 (0.58)	0.40 (*0.05*)	0.09 (0.65)	0.40 (*0.05*)	0.29 (0.16)
Gait Speed (m/s)			1	-0.08 (0.71)	-0.08 (0.69)	-0.22 (0.30)	-0.36 (0.07)
Ext Torque (Nm/kg)				1	0.21 (0.32)	0.32 (0.12)	0.18 (0.40)
Flex Torque (Nm/kg)					1	0.20 (0.33)	0.39 (*0.05*)
Ext Power (W/kg)						1	0.88 (*<0.01*)
Flex Power (W/kg)							1

KAM1 = Knee adduction moment first peak, KAM2 = Knee adduction moment second peak, Ext = Knee extension, Flex = Knee Flexion.

Table 4. 2. Summary of forward sequential linear regression models of the first and second peak knee adduction moments.

Dependent Variable: Knee Adduction Moment First Peak (Nm/kg)							
Model	Analysis	R^2	Adjusted R^2	F change	Standardized β Coefficient	Unstandardized β Coefficient (95% Confidence Interval)	p-value
1	Ext Torque	0.28	0.25	8.91	0.53	0.13 (0.04, 0.22)	<0.01
Excluded Variables:							
Gait Speed							0.75
Flex Torque							0.53
Ext Power							0.24
Dependent Variable: Knee Adduction Moment Second Peak (Nm/kg)							
Model	Analysis	R^2	Adjusted R^2	F change	Standardized β Coefficient	Unstandardized β Coefficient (95% Confidence Interval)	p-value
1	Ext Power	0.16	0.12	4.37	0.40	0.04 (0.00, 0.08)	0.05
Excluded Variables:							
Gait Speed							0.87
Ext Torque							0.95
Flex Torque							0.14

Ext = Knee extension, Flex = Knee Flexion.

Chapter Five

Identifying Yoga-Based Strengthening Exercises for Knee Osteoarthritis with Minimal Medial Knee Loads

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Publication Status: Submitted for publication to *Clinical Biomechanics*

This manuscript evaluates measures of knee loading and muscle activation during several yoga-based isometric lower extremity muscle training postures. These results identify postures which elicit muscle activations within muscle training guidelines while minimizing potentially damaging knee loads. Recommended exercises for muscle training are identified, as are potential exercises to avoid when minimizing the risk of knee OA and ACL injury development and progression are priorities.

5.1 Abstract

Objective. A low impact, yoga-based, muscle strengthening program may be effective for the management of symptomatic knee osteoarthritis. This study aimed to identify static standing yoga postures with minimal medial knee loads and maximal quadriceps and hamstrings muscle activations to determine the appropriateness of these exercise for knee osteoarthritis.

Methods. Knee adduction moments and electromyography of the vastus lateralis, rectus femoris, vastus medialis, biceps femoris, and semitendinosus of both legs were collected on healthy, young women (n=30, age 24.4 (5.4) years, BMI 23.1 (3.7) kg/m^2) during six static, standing yoga postures. These included two squatting postures, two lunging postures, a hamstrings stretch, and a single-leg balance posture. Average knee adduction moments and muscle activations were calculated during the five second static portion of each posture. A two-factor repeated measures analysis of variance was used to identify differences in muscle amplitudes and knee adduction moment between postures and legs.

Results. The wide legged squat (Goddess) and lunge with trunk upright (Warrior) produced the lowest knee adduction moments ($p \leq 0.006$), while the single-leg balance posture elicited a higher knee adduction moment than all other postures ($p<0.05$). Quadriceps activations were highest during squat and lunge postures ($p \leq 0.001$). Hamstrings activations were highest during the hamstrings stretch ($p \leq 0.003$).

Conclusion. Single-leg balance postures may elicit large knee adduction moments and do not evoke muscle activation levels within the recommended guidelines for muscle training. Squatting and lunging postures could improve leg strength without overloading the knee. Future work should evaluate these exercises in people with knee osteoarthritis.

Keywords: Muscle Strength; Yoga; Osteoarthritis, Knee; Biomechanics; Electromyography

5.2 Introduction

Lower extremity muscle training exercise is recommended to improve strength and coordination for individuals with knee pathologies such as osteoarthritis (OA) (K. L. Bennell, Wrigley, Hunt, Lim, & Hinman, 2013). Increased strength, that is the capacity for muscles around the knee to generate torque, is associated with reduced pain and improved function in OA (K. L. Bennell et al., 2013), and may help to dissipate loads applied across the knee (Mikesky et al., 2006). Yoga exercise programs, consisting of a series of physical postures, are modifiable for a variety of physical capabilities. Evidence supports the implementation of a low impact, low velocity, lower extremity muscle training program such as yoga to reduce knee pain and improve self-reported physical function in individuals with knee OA (Fransen & McConnell, 2008). Yoga improves strength, flexibility and knee OA symptoms (Bukowski, Conway, Glentz, Kurland, & Galantino, 2006; Cheung, Wyman, & Resnick, 2012; Ebnezar, Nagarathna, Yogitha, & Nagendra, 2012; Kolasinski et al., 2005), reduces the risk of cardiovascular disease, and improves quality of life (Ross & Thomas, 2010). However, there remains a need to identify safe, comfortable yoga postures which concurrently target the muscles surrounding the knee while minimizing potentially damaging knee loads. While some work has assessed the physical demands of standing yoga postures for seniors, the biomechanical factors implicated in OA pathology have not been explored in a clinical population (Wang et al., 2013).

The external knee adduction moment (KAM) can be used as a representation of medial compartment tibiofemoral loading. The KAM magnitude increases with knee OA severity (N. Foroughi, Smith, & Vanwanseele, 2009) and may increase the rate of OA progression (K. L. Bennell et al., 2011; Miyazaki et al., 2002). A large magnitude of the KAM at baseline was associated with greater cartilage loss after 12-months in 144 participants with radiographic knee OA (K. L. Bennell et al., 2011). Individuals with knee OA may utilize movement strategies to minimize KAM exposure during gait to avoid pain or discomfort at the knee. Examples include reducing gait speed (Mundermann, Dyrby, Hurwitz, Sharma, & Andriacchi, 2004), walking barefoot (Shakoor & Block, 2006), aligning the center of mass of the body over the weight-bearing foot during single-leg stance, and externally rotating the foot (D. J. Rutherford, Hubley-Kozey, Deluzio, Stanish, & Dunbar, 2008). Many of these strategies can be emphasized in yoga. An exercise program for knee pathologies could utilize these strategies to minimize KAM exposure in an effort to limit pain and disease progression.

Loss of quadriceps strength is implicated in the initiation and progression of symptomatic knee OA (N. A. Segal, Findlay, Wang, Torner, & Nevitt, 2012; Slemenda et al., 1998). Weakness of the quadriceps is strongly associated with knee pain and disability, and is thought to impair the ability to dampen loads across the knee (Lewek, Rudolph, & Snyder-Mackler, 2004; O'Reilly, Jones, Muir, & Doherty, 1998; Rice, McNair, & Lewis, 2011; Slemenda et al., 1997; Slemenda et al., 1998; Zeni, Rudolph, & Higginson, 2009). Regular participation in a quadriceps strengthening program is effective in improving muscle strength, which in turn reduces pain and improves mobility (Fransen & McConnell, 2008; Jenkinson et al., 2009; Roddy, Zhang, & Doherty, 2005). In women, longitudinal data suggests that higher quadriceps and hamstrings strength protects against the onset of knee OA symptoms (N. A. Segal et al., 2009). Furthermore, participation in a resistance and aerobic exercise program improves mobility in knee OA as measured by the distance walked in 6 minutes (Messier et al., 2004).

Quadriceps strengthening could be further augmented with hamstrings strengthening for sustained improvements in OA symptoms. The hamstrings contribute to knee motion and strengthening these muscles with the quadriceps can further improve pain and mobility (D. T. Felson, 2011; Fransen & McConnell, 2008; Topp, Woolley, Hornyak, Khuder, & Kahaleh, 2002). Quadriceps and hamstrings strength can be increased through a targeted strengthening program (Pelland et al., 2004). Isometric strengthening program guidelines for people with OA pain suggest daily participation in low to moderate intensity submaximal contractions, achieving 40%-60% of maximum voluntary contraction (MVC) of quadriceps and hamstrings (American Geriatrics Society Panel on Exercise and Osteoarthritis, 2001; K. L. Bennell et al., 2013; Ratamess et al., 2009).

This study aimed to compare knee mechanics and thigh muscle activations between static yoga postures in healthy, young women. Our goal was to identify which standing yoga postures minimized medial compartment knee loading while maximizing quadriceps and hamstrings activations. We hypothesized that the lunging postures would produce minimal KAMs and the highest quadriceps and hamstrings activations compared to other postures. The findings from this work will guide the development of a yoga-based muscle training exercise program for individuals with knee OA which limits medial load exposure at the knee.

5.3 Methods

5.3.1 Participants

Thirty healthy women (age 18 to 40 years) were recruited from the local university population. This study focused on women alone because knee moments associated with knee OA pathology differ between men and women (McKean et al., 2007) and knee OA affects women more frequently than men (D. T. Felson et al., 2000). Women with knee OA were not selected for this exploratory study to avoid exposing these individuals to large knee loads. Participants were physically active and had no contraindications to exercise on the Physical Activity Readiness Questionnaire (Thomas, Reading, & Shephard, 1992). Exclusion criteria included pregnancy and a history of knee pain, injury, or surgery. This study was approved by the McMaster University Faculty of Health Sciences Human Research Ethics Board. Written, informed consent was obtained from all participants.

5.3.2 Protocol

Participants were instrumented for motion capture and lower extremity muscle activation monitoring before performing maximum muscle activations, walking trials, and six standing yoga postures (Figure 5.1). The yoga included two squatting postures, two lunging postures, a standing hamstrings stretch posture, and a single-leg balance posture. Standing postures were chosen in order to calculate the moments acting at the knee. Participants were barefoot and given ample opportunity to observe a demonstration, learn, and practice each posture. Each posture was held static for 10 seconds and repeated three times. Posture order was randomized. During each posture, each foot was in full contact with a separate force platform, except for the single-leg balance posture. The primary leg was the forward leg during the lunging postures and the standing hamstrings stretch posture (Extended Lateral Angle, Warrior, Triangle), the straight leg during the single-leg balance posture (Tree), and the left leg during the squatting postures (Chair and Goddess). The secondary leg was the back leg during the lunging postures, the bent leg during the single-leg balance posture, and the right leg during the squatting postures.

5.3.2.1 Squatting Postures

"Chair" posture was performed with the feet in neutral rotation (toes pointed forward), slightly less than hip width apart. Participants kept their torso upright and flexed their shoulders to 180°. "Goddess" posture was performed with the feet externally rotated approximately 45°, slightly greater than shoulder width apart. Participants kept their torso upright, abducted their shoulders to 90°, externally rotated their shoulders to

90°, and flexed their elbows to 90°. For both squatting postures, participants were encouraged to squat down, without allowing their knees to travel anterior to their toes, until their thighs were approximately parallel to the floor, keeping their body mass distributed evenly through both feet.

5.3.2.2 Lunging Postures

The two lunging postures used the same leg position but different trunk and arm positions. "Extended Lateral Angle" posture was performed by stepping forward with the left foot. The left foot was in neutral rotation. The left knee was flexed until the thigh was parallel to the floor such that the thigh and shank were perpendicular to one another. The right foot was externally rotated approximately 45°, and the heel of the left foot was in line with the inner arch of the right foot. The right knee was in full extension. Participants placed their left hand beside the medial aspect of the left ankle, without grasping the ankle or touching the force platform. This position required the torso to be flexed and rotated to the right such that the torso was oriented as close to parallel to the sagittal plane as possible. Finally, participants abducted the right shoulder 90° such that the right arm was vertical. During "Warrior" posture participants kept the pelvis and torso in the coronal plane and flexed both shoulders to 180° with the arms positioned overhead.

5.3.2.3 Standing Hamstrings Stretch Posture

"Triangle" posture was performed by stepping forward with the left foot less than one meter, with the left foot oriented in neutral rotation, and the right foot externally rotated approximately 45°. The heel of the left foot was aligned with the inner arch of the right foot. Both knees were in extension. Participants placed the left hand on the medial aspect of the left ankle, without grasping the ankle or touching the force platform. The torso was flexed and rotated to the right, parallel to the sagittal plane. Finally, the right shoulder was externally rotated 180° with the right arm in full extension.

5.3.2.4 Single-Leg Balance Posture

“Tree” posture was performed by standing with the feet in neutral rotation, hip width apart, with the hands in front of the chest and the palms pressed together. Participants slowly transferred their body mass onto the right leg, raised their left leg, and externally rotated the hip, placing the plantar surface of the left foot on the medial right shank.

5.3.3 Measures

5.3.3.1 Muscle Activation

The mean electromyography (EMG) activation amplitude of the left and right leg rectus femoris, vastus lateralis, vastus medialis, biceps femoris, and semitendinosus relative to their activations during a maximum voluntary isometric contraction (MVIC) were collected during each static yoga posture. Muscle activation was monitored using 10 silver/silver chloride surface electrodes with a 20 mm inter-electrode distance affixed to the skin along the orientation of the muscle fibers. Electrode locations were determined through palpation according to the SENIAM guidelines (www.seniam.org, Enschede, Netherlands). The EMG was pre-amplified through participant-mounted dual differential amplifiers with input impedance >100 MΩ, CMR >100 dB, signal-to-noise ratio <1 μV RMS of baseline noise, base gain of 200, and a final gain of 500 (Desktop DTS, Noraxon USA Inc., Scottsdale, AZ, USA). The EMG data were bandpass filtered between 10-500 Hz, time synchronized with motion capture data, and sampled at 1500 Hz. Skin preparation included shaving and vigorous wiping with rubbing alcohol. A quiet trial with participants lying supine and relaxed established a baseline activity level for each muscle.

Participants completed MVICs of the knee flexors and extensors on each leg to elicit maximal muscle activation. These MVICs were completed individually for each leg using a dynamometer (Biodex Medical Systems, Inc., Shirley, NY, USA) with the knee attachment. Participants were positioned for testing knee extension and flexion with the dynamometer tilt at 0°, the seat back tilt at 85°, and the dynamometer axis of rotation through the lateral femoral condyle on a sagittal plane (Biodex Multi-Joint System Setup/Operation Manual, Biodex Medical Systems, Inc., Shirley, NY, USA). Participants performed a warm up of 30 submaximal contractions on the dynamometer. Maximum activation amplitude was obtained from each muscle during sets of 3 MVICs, at 15°, 45°, and 55° of knee flexion, held for 3 seconds each with rests of 5 seconds between each contraction (D. J. Rutherford, Hubley-Kozey, & Stanish, 2011). The single peak activation from each muscle was identified from these contractions to represent the maximum excitation used for normalization.

5.3.3.2 Knee Adduction Moments

The mean KAM during gait and yoga postures was calculated using three motion capture camera banks (Optotrak Certus, Northern Digital Inc., Waterloo, ON, Canada) at a sample rate of 100 Hz to track the position of the participants' body segments. Seven rigid body clusters each containing 3 infrared light emitting diodes recorded kinematic data and were attached to the sacrum, lateral thighs, lateral shanks, and lateral sides of the feet with Velcro straps and medical-grade adhesive. A four marker digitizing probe established the locations of body landmarks relative to the rigid body marker clusters, including the left and right anterior and posterior superior iliac spines, iliac crests, greater trochanters, the medial and lateral femoral and tibial epicondyles of the knee, the medial and lateral malleoli, the calcanei, and the first and fifth metatarsal heads. The motion capture system was synchronized with 3 in-ground force platforms recording ground reaction forces and moments sampled at 1000 Hz (Advanced Medical Technology Inc., Watertown, MA, USA).

Gait analysis of the right limb was completed on 5 barefoot walking trials at self-selected speeds. Gait trials were performed immediately prior to the yoga postures and were considered successful when the right foot alone contacted one of the force platforms.

5.3.4 Data Analysis

The EMG, force platform, and motion capture data were extracted. The middle 5 seconds of the static portion of the yoga postures was selected for analysis to minimize postural adjustments at the start or near the end of the data collection (Wahl & Behm, 2008). In this window, the EMG signals were full-wave rectified and low pass filtered at 6 Hz with a 4th order Butterworth filter. Baseline activity obtained from the quiet trial was removed from each signal, and the signals obtained during the yoga postures were amplitude normalized to %MVIC.

Marker and force platform data were filtered with a dual pass 4th order Butterworth low pass filter at a cut-off frequency of 6 Hz. Inverse dynamics was used to calculate the external KAMs for the right leg during the gait trials and for both the left and right legs during the yoga postures using the Joint Coordinate System floating axis model (C-Motion Inc., Germantown, MD, USA) (Grood & Suntay, 1983; Winter, 1984). KAMs were normalized to body mass and were time normalized to one stride for the walking trials. The maximum KAM from each of the 5 gait trials was averaged to provide one representative peak KAM value for each participant.

For each participant, an average of the means of three yoga trials represented the muscle activations and KAM during each yoga posture. To establish group means for gait and each yoga posture, the mean muscle activation and KAM data from all participants were ensemble averaged.

A two-factor repeated measures analysis of variance (ANOVA) with the factors posture (Chair, Extended Lateral Angle, Goddess, Tree, Triangle, and Warrior) and leg (primary and secondary) was used to identify differences within EMG amplitudes and KAMs ($\alpha=0.05$) (SPSS Version 15, SPSS Inc., Chicago, Illinois, USA). Mauchly's test was used to validate the assumption of sphericity. Significant interactions were explored using separate post-hoc one-way ANOVAs and pair-wise comparisons among estimated marginal means with a Bonferroni adjustment for multiple comparisons. Analysis of the KAM for Tree posture was only explored for the primary leg.

5.4 Results

Thirty women participated and were (mean (standard deviation)) 24.4 (5.4) years old, had a body mass of 61.5 (10.0) kg, had a BMI of 23.1 (3.7) kg/m^2, and had a peak KAM of 0.42 (0.16) Nm/kg during gait. Nine women regularly participated in yoga an average of 2 days per week (range: 1-6 days per week) for an average of 60 minutes per session (range: 40-75 minutes per session). Of the 21 women who did not regularly participate in yoga, 19 had tried yoga previously, an average of 24 times (range: 2-75 times).

A significant main effect of posture was found for all dependent measures, however Mauchly's test indicated that the assumption of sphericity had been violated ($\chi2(14) \geq 40.90$, $p<0.001$), therefore degrees of freedom were corrected using Greenhouse-Geisser estimates of sphericity ($\varepsilon \leq 0.695$).

Activations of the quadriceps were higher during the squats and lunges (Chair, Goddess, Extended Lateral Angle and Warrior) than during the hamstrings stretch (Triangle) and balance (Tree) postures ($p \leq 0.001$). Hamstrings activations were higher during Triangle than during Chair and Goddess ($p \leq 0.03$). Goddess and Warrior produced lower KAMs when both legs were grouped compared to Extended Lateral Angle and Triangle ($p \leq 0.006$). Since the legs were grouped for this section of the analysis, these results did not include Tree. All pair-wise comparisons showing significant differences between yoga postures for KAMs and EMG activations are presented in Tables 5.1, 5.2, and 5.3.

A significant main effect of leg indicated that vastus medialis activation ($F1, 29 = 4.363$, $p=0.046$) and the KAM ($F1, 29 = 60.233$, $p<0.001$) were higher in the primary leg than the secondary leg when collapsed across all yoga postures.

A significant posture/leg interaction was identified for the KAMs and all muscle activations except for biceps femoris. However Mauchly's test indicated that the assumption of sphericity had been violated ($\chi2(14) \geq 45.22$, $p<0.001$), therefore degrees of freedom were corrected using Greenhouse-Geisser estimates of sphericity ($\varepsilon \leq 0.604$). The KAMs were higher for the primary leg than the secondary leg for both squats and lunges (Chair ($p=0.01$), Goddess ($p<0.001$), Extended Lateral Angle ($p<0.001$), Warrior ($p<0.001$)), and were higher for the secondary leg than the primary leg during Triangle

($p<0001$) (Table 5.4). Tree produced a higher KAM than all other postures ($p<0.001$) (Figure 5.2). The lowest KAMs were experienced during Chair and Goddess.

Quadriceps activations were highest during Chair, Goddess, and Warrior for the primary leg and during Chair and Goddess for the secondary leg (Figure 5.2). The primary leg experienced higher vastus medialis and vastus lateralis activations than the secondary leg during Extended Lateral Angle (vastus medialis $p<0.001$, vastus lateralis $p=0.005$) and Warrior (vastus medialis $p=0.001$, vastus lateralis $p<0.001$) (Table 5.5). Hamstrings activations were highest during Triangle and Extended Lateral Angle for the primary leg and during Triangle for the secondary leg (Figure 5.2, Table 5.6).

5.5 Discussion

This study is the first in a series to identify exercises that are appropriate for medial knee OA from a biomechanical perspective. The current study identified that the squatting postures, Chair and Goddess, produced the lowest KAMs, indicating that these exercises might be least likely to facilitate knee OA progression because these impart small magnitudes of medial compartment loading. It may be desirable to limit KAM exposure among people with established knee OA given the relationship between the KAM and knee OA progression (K. L. Bennell et al., 2011; Creaby et al., 2010; Miyazaki et al., 2002). In particular, older adults, people with medial compartment joint space narrowing, obese individuals or those who experience higher knee loads due to occupational tasks may benefit from exercises that minimize the KAM.

These data show that in healthy women, the knee experiences lower KAMs during standing yoga postures than the peak experienced during gait. During gait at a self-selected speed over level ground, peak KAMs in young healthy women are typically between 0.3-0.5 Nm/kg (Moisio, Sumner, Shott, & Hurwitz, 2003). In the current study, the average peak KAM obtained during gait was 0.42 ± 0.16 Nm/kg; while during all yoga postures mean KAMs were less than 0.3 Nm/kg. The magnitude of moments found in the current study were lower than reported in previous work. Wang et al. (2003) reported average KAM of more than 0.4 Nm/kg during Tree posture, and average KAMs of more than 0.3 Nm/kg for the secondary leg during Warrior posture in 20 older adults. This discrepancy with the results from the current study may point to a difference in age. The current study included young women; while Wang et al. investigated seniors. In the current study, Tree posture produced the highest KAM of all postures; however it is likely an underestimate because the force applied by the secondary leg on the tibia of the primary leg has been neglected in calculating the KAM. Many exercise programs include single-leg activities (Ageberg, Nilsdotter, Kosek, & Roos, 2013; K. Bennell et al., 2013). These exercises require further evaluation because they may increase medial loading. Alternative exercises for challenging balance including simple modifications such as positioning the feet in tandem (Williams, Brand, Hill, Hunt, & Moran, 2010).

This study identified which yoga postures might be most effective in targeting lower extremity muscle activation in an effort to increase strength in the muscles around the knee. Quadriceps activations were highest during Chair, Goddess, and Warrior, ranging between 21%-40% MVIC. While this was shown in a healthy young sample, it reinforces the suggestion that squats and lunges may be effective for muscle training in an OA population (Lange & Vanwanseele, 2008; Pelland et al., 2004). Guidelines for isometric knee extensor and flexor strength training exercises for knee OA recommend that novice intensity be approximately 30% of the MVC and that contractions be held for no longer than 6 seconds (American Geriatrics Society Panel on Exercise and Osteoarthritis, 2001; K. L. Bennell et al., 2013; Ratamess et al., 2009). Quadriceps activations for both legs during squatting and lunging postures fit within these guidelines. This study provides insights into how strengthening programs produce positive effects on pain and mobility in knee OA (American Geriatrics Society Panel on Exercise and Osteoarthritis, 2001; Fransen & McConnell, 2008; Roddy et al., 2005). Exercises eliciting high muscle activations likely strengthen the musculature without eliciting high medial loads in the knee.

Hamstrings activations during squats and lunges were lower than the recommended guidelines (6%-21% MVIC) (American Geriatrics Society Panel on Exercise and Osteoarthritis, 2001; K. L. Bennell et al., 2013; Ratamess et al., 2009). Older knee OA adults, however, could reach the recommended 30% MVC during these tasks. Higher quadriceps-hamstrings activations and co-activations have been reported for knee OA adults compared to young and older healthy adults during the same task (Astephen, Deluzio, Caldwell, Dunbar, & Hubley-Kozey, 2008; Hortobágyi et al., 2005). During level walking and stair ascent and descent, vastus lateralis activations were twice as high and vastus lateralis/biceps femoris co-activation ratios were 2.5 times higher for 26 individuals with knee OA than for 20 young, healthy individuals (Hortobágyi et al., 2005). This work suggests that squat and lunge postures have the potential to elicit muscle activations within the guidelines for individuals with knee OA.

While quadriceps strengthening and neuromuscular training programs may not alter external KAMs (K. L. Bennell et al., 2010; N. Foroughi et al., 2011), these programs improve peak torque generation and clinical performance measures (Ageberg et al., 2013; K. L. Bennell et al., 2013; Lange & Vanwanseele, 2008; Pelland et al., 2004; Roddy et al., 2005). Furthermore, a strengthening program may slow the progression of joint space narrowing and protect tibiofemoral cartilage in knee OA through improved resistance to mechanical compression (Dahlberg, Roos, Svensson, Leander, & Tiderius, 2003; Mikesky et al., 2006). A 12-week neuromuscular lower extremity training program consisting of closed chain postural stability, functional and lower extremity muscle strengthening exercises improved all subscales of the Knee Injury and Osteoarthritis Outcome Score (KOOS) in 49 individuals with unilateral knee OA, and increased knee extensor strength in both legs (Ageberg et al., 2013). These programs may effectively treat knee symptoms because they elicit a physiological response within the muscle, which over time increases maximum torque generation capability, without overloading the medial compartment of the knee.

Extended Lateral Angle and Warrior postures elicited a KAM in the primary leg and an external knee abduction moment in the secondary leg. Conversely, during Triangle posture the primary leg experienced a knee abduction moment and the secondary leg experienced a KAM, suggesting that it will be important to consider knee load exposure in both legs while targeting muscle training in one leg. Postures with minimal KAMs and higher muscle activations (Chair, Goddess, and Warrior postures) present opportunities for muscle training in both legs.

The KAMs for the squatting postures, which are symmetrical in the frontal plane, were statistically different between the left and right legs. These moments were small (less than -0.1 Nm/kg) and negative, indicating slight external abduction moments at the knee. While these moments were statistically different between legs, they do not represent a clinical difference (50). Goddess experienced the greater difference in KAM between right and left legs (0.06 Nm/kg or 0.37 %BW*ht), however this falls below the minimal detectable change (0.59 %BW*ht) for the peak KAM during gait in individuals with knee OA at a moderate (75%) confidence level (Birmingham, Hunt, Jones, Jenkyn, & Giffin, 2007). Therefore, the differences in KAMs between legs during Chair and Goddess postures are not considered to be clinically relevant.

Typically, yoga consists of a series of standing, seated, supine, and prone positions. This study captures knee moments for standing postures only, resulting in uncertainty in the total medial joint load exposure during a yoga session. It is also unclear whether knee loading changes with increased repetitions or duration of these postures. Finally, the ability for older adults with knee OA to perform these exercises with proper positioning may limit the practicality of this program. Future studies should explore variations in repetitions and duration of squat and lunge postures on KAMs and muscle activations, and investigate the viability of such an exercise program in individuals with knee OA.

In conclusion, squatting and lunging postures have good potential as strengthening exercises for knee OA, requiring high muscle activation amplitudes, with minimal KAM. These exercises could improve torque generating capability without overloading the medial compartment of the knee. Single-leg balance exercises such as Tree posture should be avoided as they may elicit higher KAMs and do not evoke quadriceps and hamstrings activations recommended for muscle training. This point is particularly relevant for individuals who experience excessive medial knee loading due to alignment, pathology, or occupation, but who wish to improve knee symptoms, physical functioning, and lower extremity strength through a muscle training exercise program.

5.6 References

Ageberg, E., Nilsdotter, A., Kosek, E., & Roos, E. M. (2013). Effects of neuromuscular training (NEMEX-TJR) on patient-reported outcomes and physical function in severe primary hip or knee osteoarthritis: A controlled before-and-after study. BMC Musculoskeletal Disorders, 14(1), 232.

American Geriatrics Society Panel on Exercise and Osteoarthritis. (2001). Exercise prescription for older adults with osteoarthritis pain: Consensus practice recommendations: A supplement to the AGS clinical practice guidelines on the management of chronic pain in older adults. J Am Geriatr Soc, 49(10), 1400.

Astephen, J. L., Deluzio, K. J., Caldwell, G. E., Dunbar, M. J., & Hubley-Kozey, C. L. (2008). Gait and neuromuscular pattern changes are associated with differences in knee osteoarthritis severity levels. Journal of Biomechanics, 41(4), 868-876.

Bennell, K. L., Bowles, K., Wang, Y., Cicuttini, F., Davies-Tuck, M., & Hinman, R. S. (2011). Higher dynamic medial knee load predicts greater cartilage loss over 12 months in medial knee osteoarthritis. Annals of the Rheumatic Diseases, 70(10), 1770-1774.

Bennell, K. L., Wrigley, T. V., Hunt, M. A., Lim, B., & Hinman, R. S. (2013). Update on the role of muscle in the genesis and management of knee osteoarthritis. Rheumatic Diseases Clinics of North America, 39(1), 145-176.

Bennell, K., Wrigley, T., Kyriakides, M., Metcalf, B., Roos, E., Egerton, T., Hunt, M., Hodges, P., Forbes, A., Ageberg, E., & Hinman, R. (2013). Comparison of neuromuscular and quadriceps strengthening exercise in people with medial knee osteoarthritis and varus malalignment: Randomised controlled trial. Osteoarthritis and Cartilage, 21, Supplement(0), S273-S274. doi:http://dx.doi.org.libaccess.lib.mcmaster.ca/10.1016/j.joca.2013.02.573

Bennell, K. L., Hunt, M. A., Wrigley, T. V., Hunter, D. J., McManus, F. J., Hodges, P. W., & Hinman, R. S. (2010). Hip strengthening reduces symptoms but not knee load in people with medial knee osteoarthritis and varus malalignment: A randomised controlled trial. Osteoarthritis and Cartilage / OARS, Osteoarthritis Research Society, 18(5), 621-628. doi:10.1016/j.joca.2010.01.010

Birmingham, T. B., Hunt, M. A., Jones, I. C., Jenkyn, T. R., & Giffin, J. R. (2007). Test–retest reliability of the peak knee adduction moment during walking in patients with medial compartment knee osteoarthritis. Arthritis Care & Research, 57(6), 1012-1017.

Bukowski, E. L., Conway, A., Glentz, L. A., Kurland, K., & Galantino, M. L. (2006). The effect of iyengar yoga and strengthening exercises for people living with osteoarthritis of the knee: A case series. International Quarterly of Community Health Education, 26(3), 287-305. doi:10.2190/IQ.26.3.f

Cheung, C., Wyman, J., & Resnick, B. (2012). Is yoga effective for knee osteoarthritis in older women? Osteoarthritis and Cartilage, 20, S280.

Creaby, M., Wang, Y., Bennell, K., Hinman, R., Metcalf, B., Bowles, K., & Cicuttini, F. (2010). Dynamic knee loading is related to cartilage defects and tibial plateau bone area in medial knee osteoarthritis. Osteoarthritis and Cartilage, 18(11), 1380-1385.

Dahlberg, L., Roos, E., Svensson, J., Leander, P., & Tiderius, C. (2003). Exercise: Medicine for knee cartilage? Arthritis Research & Therapy, 5(Suppl 3), 168.

Ebnezar, J., Nagarathna, R., Yogitha, B., & Nagendra, H. R. (2012). Effects of an integrated approach of hatha yoga therapy on functional disability, pain, and flexibility in osteoarthritis of the knee joint: A randomized controlled study. The Journal of Alternative and Complementary Medicine, 18(5), 463-472.

Felson, D. T. (2011). Osteoarthritis in 2010: New takes on treatment and prevention. Nature Reviews. Rheumatology, 7(2), 75-76. doi:10.1038/nrrheum.2010.228

Felson, D. T., Lawrence, R. C., Dieppe, P. A., Hirsch, R., Helmick, C. G., Jordan, J. M., Kington, R. S., Lane, N. E., Nevitt, M. C., Zhang, Y., Sowers, M., McAlindon, T., Spector, T. D., Poole, A. R., Yanovski, S. Z., Ateshian, G., Sharma, L., Buckwalter, J. A., Brandt, K. D., & Fries, J. F. (2000). Osteoarthritis: New insights. part 1: The disease and its risk factors. Annals of Internal Medicine, 133(0003-4819; 8), 635-646.

Foroughi, N., Smith, R. M., Lange, A. K., Baker, M. K., Fiatarone Singh, M. A., & Vanwanseele, B. (2011). Lower limb muscle strengthening does not change frontal plane moments in women with knee osteoarthritis: A randomized controlled trial. Clinical Biomechanics, 26(2), 167-174.

Foroughi, N., Smith, R., & Vanwanseele, B. (2009). The association of external knee adduction moment with biomechanical variables in osteoarthritis: A systematic review. The Knee, 16(5), 303-309. doi:10.1016/j.knee.2008.12.007

Fransen, M., & McConnell, S. (2008). Exercise for osteoarthritis of the knee. Cochrane Database of Systematic Reviews (Online), (4)(4), CD004376. doi:10.1002/14651858.CD004376.pub2

Grood, E. S., & Suntay, W. J. (1983). A joint coordinate system for the clinical description of three-dimensional motions: Application to the knee. Journal of Biomechanical Engineering, 105, 136.

Hortobágyi, T., Westerkamp, L., Beam, S., Moody, J., Garry, J., Holbert, D., & DeVita, P. (2005). Altered hamstring-quadriceps muscle balance in patients with knee osteoarthritis. Clinical Biomechanics, 20(1), 97-104.

Jenkinson, C. M., Doherty, M., Avery, A. J., Read, A., Taylor, M. A., Sach, T. H., Silcocks, P., & Muir, K. R. (2009). Effects of dietary intervention and quadriceps strengthening exercises on pain and function in overweight people with knee pain: Randomised controlled trial. British Medical Journal (Clinical Research Ed.), 339, b3170. doi:10.1136/bmj.b3170

Kolasinski, S. L., Garfinkel, M., Tsai, A. G., Matz, W., Van Dyke, A., & Schumacher, H. R. (2005). Iyengar yoga for treating symptoms of osteoarthritis of the knees: A pilot study. Journal of Alternative and Complementary Medicine (New York, N.Y.), 11(4), 689-693. doi:10.1089/acm.2005.11.689

Lange, A. K., & Vanwanseele, B. (2008). Strength training for treatment of osteoarthritis of the knee: A systematic review. Arthritis Care & Research, 59(10), 1488-1494.

Lewek, M. D., Rudolph, K. S., & Snyder-Mackler, L. (2004). Quadriceps femoris muscle weakness and activation failure in patients with symptomatic knee osteoarthritis. Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society, 22(1), 110-115.

McKean, K. A., Landry, S. C., Hubley-Kozey, C. L., Dunbar, M. J., Stanish, W. D., & Deluzio, K. J. (2007). Gender differences exist in osteoarthritic gait. Clinical Biomechanics, 22(4), 400-409.

Messier, S. P., Loeser, R. F., Miller, G. D., Morgan, T. M., Rejeski, W. J., Sevick, M. A., Ettinger, W.H. Jr., Pahor, M., & Williamson, J. D. (2004). Exercise and dietary weight loss in overweight and obese older adults with knee osteoarthritis: The arthritis, diet, and activity promotion trial. Arthritis and Rheumatism, 50(5), 1501-1510. doi:10.1002/art.20256

Mikesky, A. E., Mazzuca, S. A., Brandt, K. D., Perkins, S. M., Damush, T., & Lane, K. A. (2006). Effects of strength training on the incidence and progression of knee osteoarthritis. Arthritis and Rheumatism, 55(0004-3591; 0004-3591; 5), 690-699.

Miyazaki, T., Wada, M., Kawahara, H., Sato, M., Baba, H., & Shimada, S. (2002). Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis. Annals of the Rheumatic Diseases, 61(7), 617-622.

Moisio, K. C., Sumner, D. R., Shott, S., & Hurwitz, D. E. (2003). Normalization of joint moments during gait: A comparison of two techniques. Journal of Biomechanics, 36(4), 599-603.

Mundermann, A., Dyrby, C. O., Hurwitz, D. E., Sharma, L., & Andriacchi, T. P. (2004). Potential strategies to reduce medial compartment loading in patients with knee osteoarthritis of varying severity: Reduced walking speed. Arthritis and Rheumatism, 50(0004-3591; 4), 1172-1178.

O'Reilly, S. C., Jones, A., Muir, K. R., & Doherty, M. (1998). Quadriceps weakness in knee osteoarthritis: The effect on pain and disability. Annals of the Rheumatic Diseases, 57(10), 588-594.

Pelland, L., Brosseau, L., Wells, G., MacLeay, L., Lambert, J., Lamothe, C., Robinson, V., & Tugwell, P. (2004). Efficacy of strengthening exercises for osteoarthritis (part I): A meta-analysis. Physical Therapy Reviews, 9(2), 77-108.

Ratamess, N. A., Alvar, B. A., Evetoch, T. E., Housh, T. J., Ben Kibler, W., Kraemer, W. J., & Triplett, N. T. (2009). Progression models in resistance training for healthy adults. Medicine and Science in Sports and Exercise, 41(3), 687-708.

Rice, D. A., McNair, P. J., & Lewis, G. N. (2011). Mechanisms of quadriceps muscle weakness in knee joint osteoarthritis: The effects of prolonged vibration on torque and muscle activation in osteoarthritic and healthy control subjects. Arthritis Research & Therapy, 13(5), R151. doi:10.1186/ar3467

Roddy, E., Zhang, W., & Doherty, M. (2005). Aerobic walking or strengthening exercise for osteoarthritis of the knee? A systematic review. Annals of the Rheumatic Diseases, 64(4), 544-548.

Ross, A., & Thomas, S. (2010). The health benefits of yoga and exercise: A review of comparison studies. The Journal of Alternative and Complementary Medicine, 16(1), 3-12.

Rutherford, D. J., Hubley-Kozey, C. L., & Stanish, W. D. (2011). Maximal voluntary isometric contraction exercises: A methodological investigation in moderate knee osteoarthritis. Journal of Electromyography and Kinesiology, 21(1), 154-160.

Rutherford, D. J., Hubley-Kozey, C. L., Deluzio, K. J., Stanish, W. D., & Dunbar, M. (2008). Foot progression angle and the knee adduction moment: A cross-sectional investigation in knee osteoarthritis. Osteoarthritis and Cartilage / OARS, Osteoarthritis Research Society, 16(8), 883-889. doi:10.1016/j.joca.2007.11.012

Segal, N. A., Findlay, C., Wang, K., Torner, J. C., & Nevitt, M. C. (2012). The longitudinal relationship between thigh muscle mass and the development of knee osteoarthritis. Osteoarthritis and Cartilage, 20(12):1534-40.

Segal, N. A., Torner, J. C., Felson, D., Niu, J., Sharma, L., Lewis, C. E., & Nevitt, M. (2009). Effect of thigh strength on incident radiographic and symptomatic knee osteoarthritis in a longitudinal cohort. Arthritis and Rheumatism, 61(0004-3591; 0004-3591; 9), 1210-1217.

Shakoor, N., & Block, J. A. (2006). Walking barefoot decreases loading on the lower extremity joints in knee osteoarthritis. Arthritis and Rheumatism, 54(9), 2923-2927. doi:10.1002/art.22123

Slemenda, C., Brandt, K. D., Heilman, D. K., Mazzuca, S., Braunstein, E. M., Katz, B. P., & Wolinsky, F. D. (1997). Quadriceps weakness and osteoarthritis of the knee. Annals of Internal Medicine, 127(0003-4819; 0003-4819; 2), 97-104.

Slemenda, C., Heilman, D. K., Brandt, K. D., Katz, B. P., Mazzuca, S. A., Braunstein, E. M., & Byrd, D. (1998). Reduced quadriceps strength relative to body weight: A risk factor for knee osteoarthritis in women? Arthritis and Rheumatism, 41(0004-3591; 0004-3591; 11), 1951-1959.

Thomas, S., Reading, J., & Shephard, R. J. (1992). Revision of the physical activity readiness questionnaire (PAR-Q). Canadian Journal of Sport Sciences, 17(4), 338-345.

Topp, R., Woolley, S., Hornyak, J.,3rd, Khuder, S., & Kahaleh, B. (2002). The effect of dynamic versus isometric resistance training on pain and functioning among adults with osteoarthritis of the knee. Archives of Physical Medicine and Rehabilitation, 83(9), 1187-1195. Retrieved from http://eutils.ncbi.nlm.nih.gov

Wahl, M. J., & Behm, D. G. (2008). Not all instability training devices enhance muscle activation in highly resistance-trained individuals. The Journal of Strength & Conditioning Research, 22(4), 1360-1370.

Wang, M., Sean, S., Hashish, R., Samarawickrame, S. D., Kazadi, L., Greendale, G. A., & Salem, G. (2013). The biomechanical demands of standing yoga poses in seniors: The yoga empowers seniors study (YESS). BMC Complementary and Alternative Medicine, 13(1), 8.

Williams, S. B., Brand, C. A., Hill, K. D., Hunt, S. B., & Moran, H. (2010). Feasibility and outcomes of a home-based exercise program on improving balance and gait stability in women with lower-limb osteoarthritis or rheumatoid arthritis: A pilot study. Archives of Physical Medicine and Rehabilitation, 91(1), 106-114.

Winter, D. A. (1984). Biomechanics of human movement with applications to the study of human locomotion. Critical Reviews in Biomedical Engineering, 9(4), 287-314.

Zeni, J. A., Rudolph, K., & Higginson, J. S. (2009). Alterations in quadriceps and hamstrings coordination in persons with medial compartment knee osteoarthritis. Journal of Electromyography and Kinesiology, 20(1), 148-154. doi:10.1016/j.jelekin.2008.12.003

Table 5. 1. P-values for pair-wise comparisons of the KAM among yoga postures. Mean KAM values in Nm/kg are provided in parentheses beside each posture for reference. Significant differences (α=0.05) between postures are highlighted in bold text. Note: Ext Lat Ang = Extended Lateral Angle.

		Goddess	**Extended Lateral Angle**	**Warrior**	**Triangle**	**Tree**
Primary Leg KAM (Nm/kg)	Chair (-0.01)	0.728	**<0.001**	**<0.001**	0.226	**<0.001**
	Goddess (-0.03)	1	**<0.001**	**<0.001**	>0.999	**<0.001**
	Ext Lat Ang (0.11)	-	1	>0.999	**<0.001**	**<0.001**
	Warrior (0.09)	-	-	1	**0.003**	**<0.001**
	Triangle (-0.10)	-	-	-	1	**<0.001**
	Tree (0.29)	-	-	-	-	1
Secondary Leg KAM (Nm/kg)	Chair (-0.04)	0.093	>0.999	**<0.001**	**<0.001**	NA
	Goddess (-0.09)	1	>0.999	**<0.001**	**<0.001**	NA
	Ext Lat Ang (-0.06)	-	1	**<0.001**	**<0.001**	NA
	Warrior (-0.29)	-	-	1	**<0.001**	NA
	Triangle (0.18)	-	-	-	1	NA

Table 5. 2. P-values for pair-wise comparisons of quadriceps EMG activations among yoga postures. Mean EMG values in %MVIC are provided in parentheses beside each posture for reference. Significant differences (α=0.05) between postures are highlighted in bold text. Note: Ext Lat Ang = Extended Lateral Angle.

		Goddess	Extended Lateral Angle	Warrior	Triangle	Tree
Primary Leg Rectus Femoris Activation (%MVIC)	Chair (27.7)	>0.999	**<0.001**	>0.999	**<0.001**	0.391
	Goddess (26.9)	1	**<0.001**	>0.999	**<0.001**	0.471
	Ext Lat Ang (17.3)	-	1	**0.001**	**<0.001**	0.664
	Warrior (25.7)	-	-	1	**<0.001**	>0.999
	Triangle (10.9)	-	-	-	1	**0.001**
	Tree (21.7)	-	-	-	-	1
Secondary Leg Rectus Femoris Activation (%MVIC)	Chair (31.5)	0.229	**<0.001**	**<0.001**	**<0.001**	**<0.001**
	Goddess (27.5)	1	0.086	**0.011**	**<0.001**	**<0.001**
	Ext Lat Ang (22.6)	-	1	>0.999	>0.999	**<0.001**
	Warrior (22.3)	-	-	1	**0.026**	**<0.001**
	Triangle (20.1)	-	-	-	1	**0.001**
	Tree (11.0)	-	-	-	-	1
Primary Leg Vastus Medialis Activation (%MVIC)	Chair (30.4)	>0.999	>0.999	>0.999	**<0.001**	**<0.001**
	Goddess (34.1)	1	**0.011**	>0.999	**<0.001**	**<0.001**
	Ext Lat Ang (27.0)	-	1	0.186	**<0.001**	**<0.001**
	Warrior (33.4)	-	-	1	**<0.001**	**<0.001**
	Triangle (13.7)	-	-	-	1	>0.999
	Tree (16.3)	-	-	-	-	1
Secondary Leg Vastus Medialis Activation (%MVIC)	Chair (33.5)	>0.999	**<0.001**	**<0.001**	**<0.001**	**<0.001**
	Goddess (34.5)	1	**<0.001**	**<0.001**	**<0.001**	**<0.001**
	Ext Lat Ang (17.7)	-	1	0.232	0.085	0.377
	Warrior (21.1)	-	-	1	**0.001**	**0.015**
	Triangle (15.0)	-	-	-	1	>0.999
	Tree (14.0)	-	-	-	-	1
Primary Leg Vastus Lateralis Activation (%MVIC)	Chair (35.7)	0.757	**0.008**	>0.999	**<0.001**	**<0.001**
	Goddess (38.9)	1	**<0.001**	>0.999	**<0.001**	**<0.001**
	Ext Lat Ang (28.9)	-	1	0.071	**<0.001**	**<0.001**
	Warrior (35.1)	-	-	1	**<0.001**	**<0.001**
	Triangle (14.7)	-	-	-	1	>0.999
	Tree (16.2)	-	-	-	-	1
Secondary Leg Vastus Lateralis Activation (%MVIC)	Chair (37.9)	>0.999	**<0.001**	**<0.001**	**<0.001**	**<0.001**
	Goddess (39.1)	1	**<0.001**	**<0.001**	**<0.001**	**<0.001**
	Ext Lat Ang (21.2)	-	1	>0.999	0.069	**0.048**
	Warrior (23.8)	-	-	1	**<0.001**	**<0.001**
	Triangle (15.8)	-	-	-	1	>0.999
	Tree (14.9)	-	-	-	-	1

Table 5. 3. P-values for pair-wise comparisons of hamstrings EMG activations among yoga postures. Mean EMG values in %MVIC are provided in parentheses beside each posture for reference. Significant differences (α=0.05) between postures are highlighted in bold text. Note: Ext Lat Ang = Extended Lateral Angle.

		Goddess	Extended Lateral Angle	Warrior	Triangle	Tree
Primary Leg Biceps Femoris Activation (%MVIC)	Chair (16.0)	>0.999	**0.002**	>0.999	**<0.001**	>0.999
	Goddess (17.1)	1	>0.999	>0.999	0.081	>0.999
	Ext Lat Ang (18.6)	-	1	**0.033**	>0.999	0.111
	Warrior (16.5)	-	-	1	**0.002**	>0.999
	Triangle (19.5)	-	-	-	1	**0.026**
	Tree (14.2)	-	-	-	-	1
Secondary Leg Biceps Femoris Activation (%MVIC)	Chair (16.0)	0.141	0.126	0.332	**<0.001**	>0.999
	Goddess (17.0)	1	0.762	>0.999	**0.001**	>0.999
	Ext Lat Ang (19.6)	-	1	>0.999	>0.999	>0.999
	Warrior (20.8)	-	-	1	>0.999	>0.999
	Triangle (20.8)	-	-	-	1	>0.999
	Tree (18.6)	-	-	-	-	1
Primary Leg Semitendinosus Activation (%MVIC)	Chair (8.3)	>0.999	0.387	>0.999	**0.039**	>0.999
	Goddess (7.8)	1	0.101	>0.999	**0.036**	>0.999
	Ext Lat Ang (11.6)	-	1	0.142	0.541	>0.999
	Warrior (8.0)	-	-	1	**0.035**	>0.999
	Triangle (15.0)	-	-	-	1	0.236
	Tree (9.7)	-	-	-	-	1
Secondary Leg Semitendinosus (%MVIC)	Chair (6.9)	>0.999	**0.046**	**0.001**	**0.039**	0.074
	Goddess (6.7)	1	**0.007**	**<0.001**	**0.012**	**0.048**
	Ext Lat Ang (9.2)	-	1	0.813	>0.999	0.642
	Warrior (11.7)	-	-	1	>0.999	>0.999
	Triangle (10.6)	-	-	-	1	>0.999
	Tree (13.4)	-	-	-	-	1

Table 5. 4. Mean knee adduction/abduction moments (Nm/kg) during the static (5 seconds) yoga postures (n=30). Knee adduction moments are positive values. Measures are reported for both legs as mean (standard deviation) [95% Confidence Interval]. Significant differences (α=0.05) between the primary and secondary legs are highlighted in bold text.

	Primary Leg	Secondary Leg	P-value
Knee Adduction/Abduction Moment (Nm/kg)			
Chair	-0.01 (0.07) [-0.03, 0.02]	-0.04 (0.08) [-0.07, -0.02]	**0.01**
Goddess	-0.03 (0.08) [-0.06, 0.00]	-0.09 (0.09) [-0.13, -0.06]	**<0.001**
Extended Lateral Angle	0.11 (0.13) [0.07, 0.16]	-0.06 (0.11) [-0.10, -0.02]	**<0.001**
Warrior	0.09 (0.12) [0.05, 0.14]	-0.29 (0.13) [-0.34, -0.24]	**<0.001**
Triangle	-0.10 (0.18) [-0.16, -0.03]	0.18 (0.14) [0.13, 0.23]	**<0.001**
Tree†	0.29 (0.10) [0.26, 0.33]		

†During *Tree* posture knee adduction moments are measured for the primary leg only.

Table 5. 5. Mean quadriceps EMG activation amplitudes, expressed as percentage of activation achieved during a maximal voluntary isometric contraction (% MVIC) during the static (5 seconds) yoga postures (n=30). Measures are reported as mean (standard deviation) [95% Confidence Interval]. Significant differences (α=0.05) between the primary and secondary legs are highlighted in bold text.

	Primary Leg	Secondary Leg	P-value
Rectus Femoris Activation (%MVIC)			
Chair	27.7 (11.4) [23.1, 31.9]	31.5 (12.2) [26.9, 36.0]	0.061
Goddess	26.9 (11.5) [22.6, 31.2]	27.5 (12.7) [22.8, 32.3]	0.785
Extended Lateral Angle	17.3 (6.9) [14.7, 19.9]	22.6 (12.1) [18.1, 27.1]	**0.018**
Warrior	25.7 (11.8) [21.3, 30.1]	22.3 (11.2) [18.1, 26.5]	0.192
Triangle	10.9 (5.6) [8.8, 13.0]	20.1 (11.3) [15.9, 24.4]	**<0.001**
Tree	21.7 (11.5) [17.4, 25.9]	11.0 (5.3) [9.0, 13.0]	**<0.001**
Vastus Medialis Activation (%MVIC)			
Chair	30.4 (12.4) [25.8, 35.0]	33.5 (13.4) [28.5, 38.5]	0.241
Goddess	34.1 (12.0) [29.6, 38.6]	34.5 (12.9) [29.7, 39.3]	0.890
Extended Lateral Angle	27.0 (10.4) [23.1, 30.9]	17.7 (8.7) [14.5, 21.0]	**<0.001**
Warrior	33.4 (14.3) [28.0, 38.7]	21.1 (10.7) [17.1, 25.1]	**0.001**
Triangle	13.7 (6.7) [11.1, 16.1]	15.0 (7.7) [12.1, 17.8]	0.322
Tree	16.3 (7.8) [13.4, 19.2]	14.0 (6.3) [11.7, 16.4]	0.114
Vastus Lateralis Activation (%MVIC)			
Chair	35.7 (15.6) [29.9, 41.5]	37.9 (12.3) [33.3, 42.5]	0.508
Goddess	38.9 (17.2) [32.5, 45.3]	39.1 (15.8) [33.2, 45.0]	0.951
Extended Lateral Angle	28.9 (11.6) [24.6, 33.2]	21.2 (13.9) [16.0, 26.4]	**0.005**
Warrior	35.1 (13.8) [29.9, 40.2]	23.8 (9.6) [20.2, 27.4]	**<0.001**
Triangle	14.7 (8.7) [11.5, 18.0]	15.8 (8.3) [12.7, 19.0]	0.316
Tree	16.2 (7.7) [13.4, 19.1]	14.9 (8.0) [11.9, 17.9]	0.269

Table 5. 6. Mean hamstrings EMG activation amplitudes, expressed as percentage of activation achieved during a maximal voluntary isometric contraction (% MVIC) during the static (5 seconds) yoga postures (n=30). Measures are reported as mean (standard deviation) [95% Confidence Interval]. Significant differences (α=0.05) between the primary and secondary legs are highlighted in bold text.

	Primary Leg	Secondary Leg	P-value
Biceps Femoris Activation (%MVIC)			
Chair	16.0 (9.2) [12.5, 19.4]	16.0 (12.9) [11.2, 20.8]	0.982
Goddess	17.1 (10.0) [13.3, 20.8]	17.0 (13.8) [11.8, 22.1]	0.970
Extended Lateral Angle	18.6 (9.1) [15.1, 22.0]	19.6 (13.4) [14.6, 24.6]	0.648
Warrior	16.5 (9.1) [13.0, 19.9]	20.8 (12.4) [16.1, 25.4]	0.067
Triangle	19.5 (9.0) [16.2, 22.9]	20.8 (12.7) [16.0, 25.5]	0.515
Tree	14.2 (9.6) [10.6, 17.8]	18.6 (8.8) [15.3, 21.9]	**0.010**
Semitendinosus Activation (%MVIC)			
Chair	8.4 (7.4) [5.6, 11.1]	6.9 (4.4) [5.3, 8.5]	0.287
Goddess	7.8 (6.3) [5.4, 10.2]	6.7 (4.3) [5.1, 8.4]	0.365
Extended Lateral Angle	11.6 (8.2) [8.5, 14.6]	9.2 (6.4) [6.8, 11.6]	0.059
Warrior	8.0 (5.6) [5.9, 10.1]	11.7 (7.1) [9.0, 14.3]	**0.025**
Triangle	15.0 (12.7) [10.3, 19.8]	10.6 (8.0) [7.6, 13.5]	**0.012**
Tree	9.7 (9.6) [6.1, 13.2]	13.4 (11.3) [9.2, 17.7]	0.075

Figure 5. 1. Yoga Postures. Top row, from left to right: Chair posture, Goddess posture, Triangle posture. Bottom row, from left to right: Extended Lateral Angle posture, Warrior posture, Tree posture.

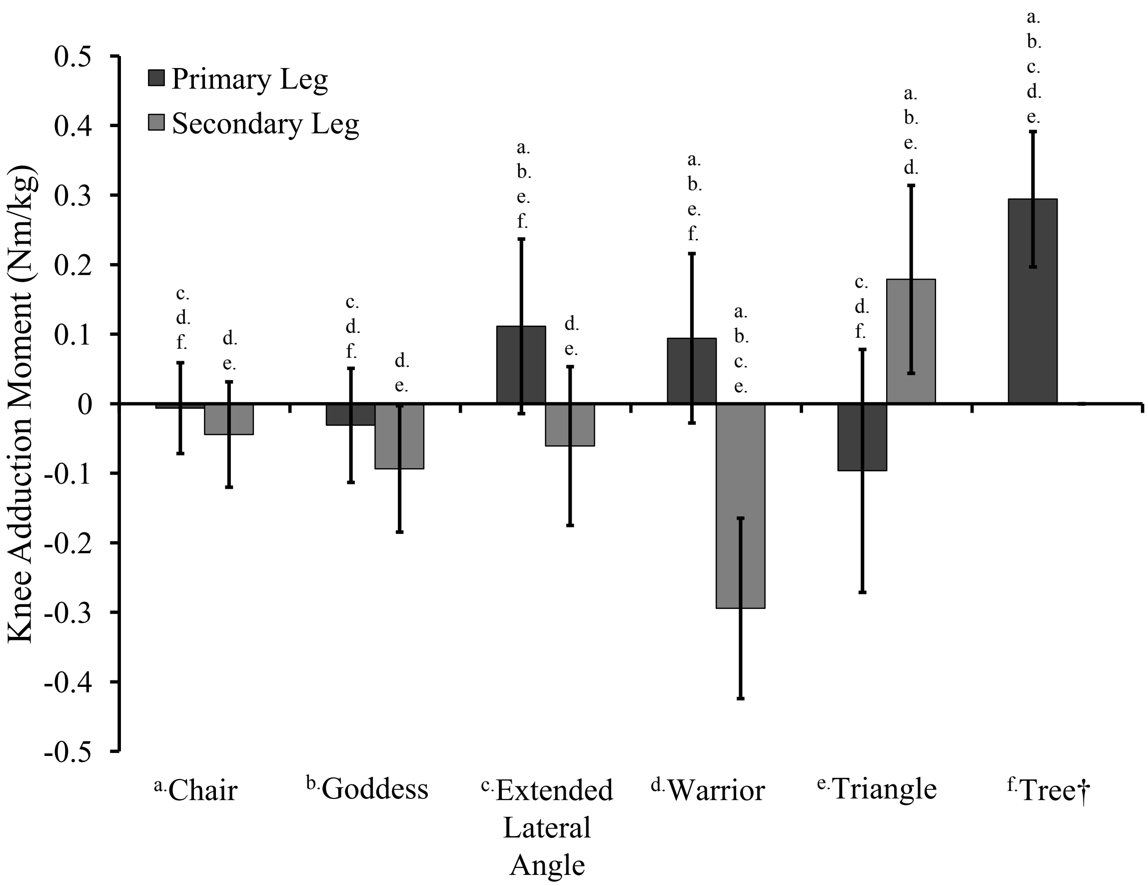

Figure 5. 2. Mean knee adduction/abduction moments (Nm/kg) for both legs during the static (5 seconds) yoga postures (n=30). Knee adduction moments are positive values. Significant differences (p < 0.05) in each leg indicated from a. Chair, b. Goddess, c. Extended Lateral Angle, d. Warrior, e. Triangle, and f. Tree. †During Tree posture knee adduction moments are measured for the primary leg only.

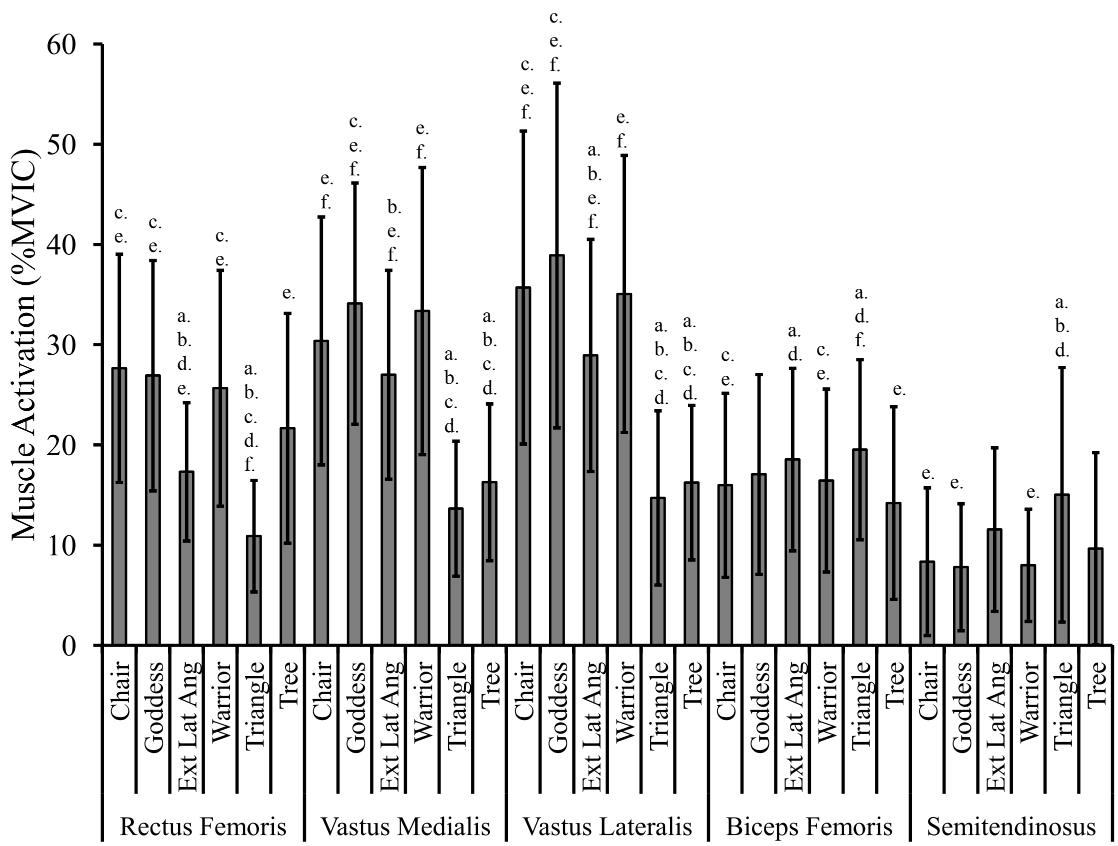

Figure 5. 3. Mean EMG activation amplitudes, expressed as a percentage of activation achieved during a maximal voluntary isometric contraction (%MVIC) for the primary leg during the static (5 seconds) yoga postures (n=30). Significant differences ($p < 0.05$) in each leg indicated from a. Chair, b. Goddess, c. Extended Lateral Angle, d. Warrior, e. Triangle, and f. Tree.

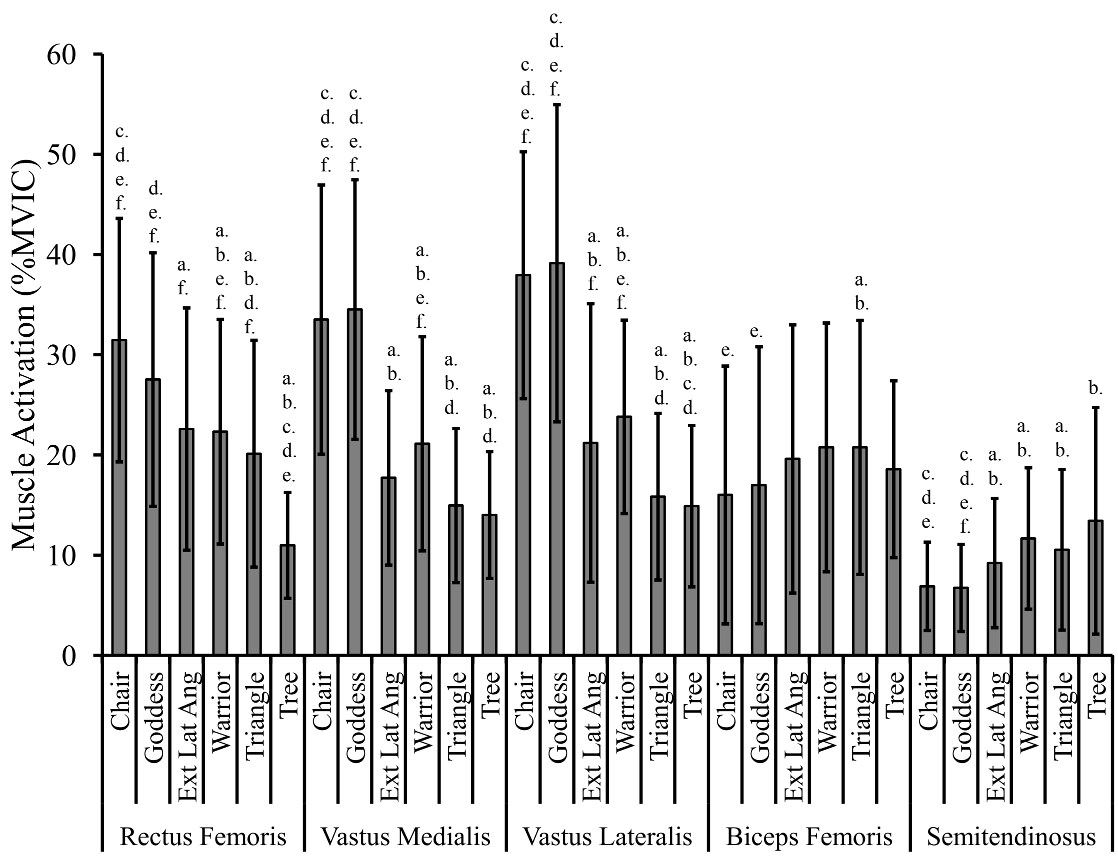

Figure 5. 4. Mean EMG activation amplitudes, expressed as a percentage of activation achieved during a maximal voluntary isometric contraction (%MVIC) for the secondary leg during the static (5 seconds) yoga postures (n=30). Significant differences ($p < 0.05$) in each leg indicated from a. Chair, b. Goddess, c. Extended Lateral Angle, d. Warrior, e. Triangle, and f. Tree.

Chapter Six

Discussion

The overarching goal of this work was to (i) examine how lower extremity muscle function is related to the knee loading environment, and (ii) examine knee loading patterns which may be implicated in anterior cruciate ligament (ACL) injury and knee osteoarthritis (OA) during static and dynamic activities. This thesis extended beyond the traditional use of lower extremity peak torque as the sole measure of muscle function to include lower extremity power, quadriceps and hamstrings activations and coactivations, and the impairment of peak torque through neuromuscular fatigue. Establishing the role of muscle function in the knee loading environment of healthy knees is important to identify differences or changes which may occur in injured or diseased knees. Further, this thesis explored the suitability of static and dynamic lower extremity muscle training activities for exercise programs which look to minimize medial knee load exposure.

The results of the studies highlight the relationships between higher lower extremity peak torque and power and higher peak medial knee loads during gait and suggest that young healthy women retain enough peak torque reserve after neuromuscular fatigue of the muscles surrounding the knee to perform submaximal tasks such as gait and isometric squatting and lunging. The altered muscle patterns and knee moments observed after fatigue, namely decreases in lateral quadriceps activation and knee moments, suggest that the young women in these studies have substantial neuromuscular redundancy to alter the recruitment strategies of the muscles supporting the knee without increasing joint loading. Squatting and lunging appear to challenge lower extremity muscle activation better than other static standing postures, while eliciting lower medial knee loads than the peak during gait. These findings suggest that these isometric postures may be ideal activities for targeting improvements in muscle strength (i.e., peak torque production) without the risk for provoking knee mechanics implicated in ACL injury or knee OA.

6.1 Summary of Major Findings and Key Implications of Individual Studies

This thesis comprised four interrelated studies that examined the influence of muscle function on the knee joint, with particular attention paid to knee loading during static and dynamic activities in healthy young women. The first study identified changes in knee biomechanics during gait after lower extremity neuromuscular fatigue. While the isotonic fatigue protocol significantly decreased the peak isometric knee flexion and extension torques, it did not alter the peaks of the external knee adduction moment (KAM), the dynamic knee stiffness, or quadriceps and hamstrings muscle activations during gait. However, lower extremity fatigue did reduce the peak external knee extension moment during terminal stance of gait. The major implication from this study is that healthy young women retained enough strength after lower extremity fatigue to perform a submaximal task such as gait without substantial alterations to their knee biomechanics.

The second study explored more demanding activities by comparing quadriceps and hamstrings muscle activations and external knee moments during static squats and lunges before and after lower extremity neuromuscular fatigue. Peak knee flexion and extension torques were decreased with fatigue. The activation of the vastus lateralis was decreased after fatigue for both activities, as were the mean KAM and the mean knee flexion moment during lunging. These results indicated that fatigue altered the recruitment strategies of the quadriceps in performing both tasks and that this change in muscle activation strategy may have concurrently altered the loading environment at the knee during lunging.

The third study broadened the scope of muscular influences on knee moments during gait by examining the contributions of both knee flexor and extensor peak torque and power to the peak KAM values. Interestingly, peak knee extension torque was a significant contributor to the first peak KAM, while peak knee extension power was a significant contributor to the second peak KAM during gait. These results suggested that both the peak magnitude of lower extremity torque generation and the rate at which it is produced are important in describing healthy knee loading just after heel strike and just as the foot is preparing for push off during gait, respectively. Importantly, this study also indicated that knee flexor torque and power were not important contributors to the submaximal activity of walking in young healthy women.

The final study aimed to identify standing, static postures which targeted quadriceps and hamstrings muscle activation without overloading the medial compartment of the knee, evaluated using the KAM. The wide-legged squat and lunge with trunk upright postures elicited the greatest amplitude of quadriceps and hamstrings activation while maintaining the lowest KAMs. These activities were recommended for muscle training exercise programs aiming to increase torque production capacity without overloading the medial compartment of the knee. Further, it was recommended that the single leg balance posture be avoided for individuals with knee injuries or at risk of knee pathologies as it elicited a higher KAM in young healthy women.

6.2 Thesis Philosophy

The work in this thesis subscribes to the underlying philosophy that informed exercise prescription requires scientific basis. Prescribed lower extremity muscle training programs should elicit improvements in muscle function without contributing to a damaging knee load environment. Specifically, lower extremity muscle training exercise prescription should be derived from a fundamental understanding of the biomechanical relationships between knee loading and muscle function. This understanding must be supported by evidence identifying (i) factors that could be damaging to the knee loading environment, (ii) exercises that minimize these potentially damaging factors, (iii) the criteria required for muscle training improvements, and (iv) exercises that elicit the muscle mechanics required for muscle training improvements. Each of the four studies presented in this thesis contributes evidence in support of this philosophy.

This thesis contributes new research adding to a fundamental understanding of the potential risks associated with a lower limb exercise program and the identification of effective lower limb exercises. These factors are shown in blue in Figure 6. 1. This collection of studies forms the basis of a future research program investigating the longitudinal effects of lower limb exercise interventions on muscle and joint mechanics in healthy and pathological populations (Figure 6. 1, shown in red).

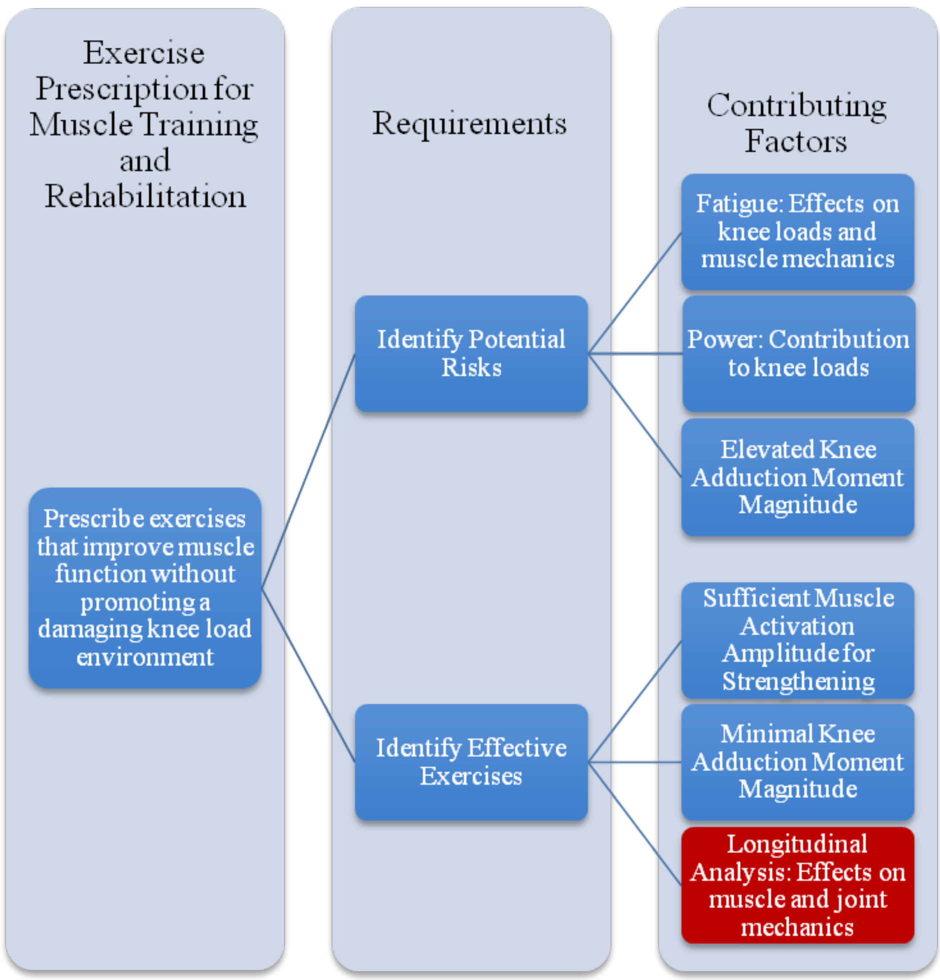

Figure 6. 1. Thesis philosophy flowchart. Blue boxes identify factors addressed in this thesis. Red boxes identify priorities for future research, which are featured in section 6.5 of this thesis.

Potential risks to knee health including lower extremity fatigue, power, and an elevated KAM during static and dynamic activities were evaluated in this thesis. Knee mechanics are critical to joint health given their association with the onset and progression of knee pathologies ranging from ligamentous injury to degeneration (Andriacchi & Dyrby, 2005; Bennell et al., 2011).

The first two studies in this thesis examined knee mechanics and quadriceps and hamstrings activation before and after lower extremity fatigue. Fatigue did not elicit knee moments or muscle activity that are typically implicated in the onset or progression of knee OA or ACL injury, suggesting that gait, and isometric squats and lunges may be appropriate exercises for promoting a favourable knee loading environment. Chapter 4 explored the contributions of strength and power to the KAM during gait, an identified risk factor in the onset and progression of knee OA. Power may pose a possible risk to increased knee loads due to a potential for increased rate of loading, or torque generation. In the healthy sample, both strength and power were found to contribute to the variance in the KAM peaks; strength contributed in early stance, and power contributed in late stance. Chapters 2, 3, and 5 monitored the KAM in healthy women during static and dynamic activities. Single-limb static postures were identified as a potential risk to joint health as they elicited a substantially higher KAM than other postures and gait. The combination of these results suggest that individuals with higher peak torque and peak power capacity may be at greater risk of experiencing higher peak KAMs, and certain static postures utilized by some exercise programs should be avoided to limit exposure to these moments.

Chapters 3 and 5 support the overall philosophy of this thesis by identifying effective exercises based on measureable physiological characteristics. These studies identified exercises that challenge the quadriceps and hamstrings indicated by activation levels at or approaching the established guidelines for muscle training. Isometric squatting and lunging postures elicited higher quadriceps and hamstrings activation amplitudes than other isometric standing postures and gait, with significantly lower mean KAMs. These studies highlighted easy-to-perform, modifiable exercises which can be implemented at no financial cost to the individual.

The evaluation of risks associated with lower extremity exercise and the identification of effective exercises is particularly meaningful for rehabilitation and muscle training programs for individuals with pathologies such as knee OA. For those with knee pain and degeneration or with previous knee injuries, this thesis highlights exercises that avoid potentially damaging knee loading conditions while challenging the quadriceps and hamstrings. Future research can now examine the longitudinal effects of a lower extremity exercise program built on these fundamental findings.

6.3 Contribution and Implications of the Thesis Overall

6.3.1 Muscle Training Exercises to Improve Lower Extremity Torque Production

Strategies to improve lower extremity muscle function are important for health professionals aiming to reduce the likelihood of developing or progressing ACL injury and knee OA. This thesis evaluated the influence of muscle function on measures of medial knee loading to determine whether muscle training posed any risk for increasing loads in the medial knee. Both physiological and neurological changes occur in response to targeted muscle training. Some measureable physiological characteristics can be changed with resistance training such as muscle strength, muscle power, and muscular hypertrophy, or an increase in the size of skeletal muscle (Charette et al., 1991; Ratamess et al., 2009). The relationships between changes in these factors with knee loading during activities such as walking are important in preventing joint injury and degeneration.

Some neural adaptations occur predominately during the early stages of a resistance training program. These include increases in neural drive and changes in the interaction between agonist and antagonist muscle groups (Gabriel, Kamen, & Frost, 2006; Sale, 1988). These changes are most often evaluated using surface electromyography (EMG) to measure muscle activations and coactivation between muscle groups. Therefore, particularly during the early stages of a resistance training program, increases in neural drive such as muscle activation may demonstrate potential for eventual increases in lower extremity torque production.

Current guidelines for muscle training or resistance training exercises vary for age group and health status. For example, for healthy adults between the ages of 18-65 years, it is recommended that participation in strengthening exercises be undertaken at least 2 days of the week (Haskell et al., 2007; Tremblay et al., 2011). A muscle strengthening program for healthy adults should include 8-10 exercises targeting major muscle groups performed with substantial resistance such that 8-12 repetitions of each exercises elicits volitional fatigue (Haskell et al., 2007). Strength training guidelines for healthy older adults include 1-3 sets of 8-12 repetitions of isotonic resistance exercises at 60-80% of an individual's one-repetition-maximum completed 2-3 days per week (Ratamess et al., 2009). Healthy older adults are also encouraged to combine this training regimen with exercises geared toward balance training to improve or maintain mobility and prevent falls (Tremblay et al., 2011).

Muscle strength training guidelines differ for individuals with knee OA compared to those for healthy adults. For knee pathology, guidelines reduce the volume and intensity of isotonic resistance exercises, and place greater emphasis on isometric or static postures (Bennell, Wrigley, Hunt, Lim, & Hinman, 2013). For example, daily participation in isometric strengthening exercises is recommended for individuals with knee OA at low to moderate intensity (40-60% of the maximum voluntary contraction) (American Geriatrics Society Panel on Exercise and Osteoarthritis, 2001). These guidelines suggest 1-10 submaximal isometric contractions per muscle group with a contraction time from 1-6 seconds. Strengthening guidelines for those with knee pathologies are more conservative than those for healthy adults because of the increased risk for knee pain during strenuous exercise and increased prevalence of muscle weakness (Lewek, Rudolph, & Snyder-Mackler, 2004; Segal & Glass, 2011; Slemenda et al., 1997). This reduced capacity for lower extremity torque production in individuals with knee OA compared to healthy controls limits the intensity and duration of strengthening exercises.

These guidelines provide individuals and health providers a starting point for planning a muscle training exercise program. However, further research is required to (i) define the types of activities, (ii) assess the appropriateness of the exercises for various muscle groups, and (iii) improve our understanding of the dose-response relationship to different exercises. The work in this thesis contributes to the advancement of these guidelines by examining biomechanical quantities during the performance of a range of dynamic and static activities such as gait, squatting, lunging, and balance postures. Chapters 3 and 5 demonstrated that lower extremity strengthening exercises intended to minimize knee biomechanics patterns implicated in ACL injury and knee degeneration should include double-limb squatting and lunging activities. Held isometrically for 5 seconds, these activities elicit muscle activations at or near the recommended range for muscle training in young healthy women, while maintaining a low magnitude KAM. Chapters 2 and 3 also highlighted that lower extremity neuromuscular fatigue has little impact on knee biomechanics during submaximal tasks. Therefore, short-term fatigue achieved through a high impact, strenuous muscle training exercise session may not produce abnormal or undesirable knee biomechanics in young healthy women. This finding is helpful in clarifying the appropriate intensity of exercise for this population.

6.3.2 Role of Muscles in Knee Loading during Static and Dynamic Activities

This thesis offers key insights into the role of the quadriceps and hamstrings in mediating loading in the knee. Chapters 2 and 3 found that impairing the torque generating capacity of these muscles through fatigue decreased the peak knee extension moment during gait, decreased vastus lateralis activation during squatting and lunging, and decreased the mean KAM and mean knee flexion moment during lunging. Therefore, after neuromuscular fatigue of the quadriceps and hamstrings, the knee experienced slightly reduced joint loading during the performance of submaximal tasks. This could be due to the reduction in activation from the lateral quadriceps or to compensation from muscles other than those evaluated in these studies.

The fourth chapter of this thesis emphasizes the role of quadriceps in mediating medial knee loads estimated by the KAM. Knee extensor torque and power explained significant portions of the variance in the first and second peaks of the KAM during gait, while knee flexor torque and power were unrelated to measures of joint loading. Though this study found that KAM peaks are higher for individuals with greater quadriceps torque and power, the results from Chapter 2 demonstrate that even after significant impairments to knee extensor torque, the peak KAM during gait remained unchanged. This finding suggests that short term changes in knee extensor torque may not produce significant increases in peak KAMs during submaximal activities; however these data suggest that, in the future, we should ask the question whether increasing quadriceps torque and power over a longer time period elicits higher joint loads. On the other hand, quadriceps weakness has been implicated in knee OA (Bennell et al., 2013; Lewek et al., 2004; Rice, McNair, & Lewis, 2011; Slemenda et al., 1997), ACL injury (Palmieri-Smith, Thomas, & Wojtys, 2008; Shelbourne & Gray, 1997), and is known to occur as women age (Hurley, Rees, & Newham, 1998; Young, Stokes, & Crowe, 1984). Therefore, quadriceps strengthening in healthy young women may promote prevention of future muscle weakness. This hypothesis should be explored longitudinally in a future study with particular attention on whether any changes in the KAM are clinically relevant.

Little evidence was found for a connection between hamstrings function and medial knee loads. Chapter 4 found no associations between knee flexor torque and power with the peak KAMs and chapters 2 and 3 found no alterations in hamstrings activations or coactivations with the quadriceps during both static and dynamic activities after fatigue. These findings are consistent with others who have found that hamstrings have a lesser role in knee joint loading than other muscle groups, particularly the quadriceps, during activities such as gait (Shelburne, Torry, & Pandy, 2006). Previous studies have identified the importance for young women to develop and maintain torque generating capacity in the quadriceps and hamstrings as they age so that they are able to adjust to increasing demands while performing various tasks without an increased risk for injury (Devan, Pescatello, Faghri, & Anderson, 2004; Grace, Sweetser, Nelson, Ydens, & Skipper, 1984). These studies support the idea of increasing redundancy in the muscular system supporting the knee joint through an increased capacity for torque production. For example, as one muscle fatigues, other muscles are able to compensate by changing the torque produced in order to maintain the knee loading environment. The work in this thesis suggests that efforts to increase or maintain peak hamstrings torque will not have adverse effects on medial knee loading during submaximal activities of daily living.

Finally, Chapters 3 and 5 highlight potential isometric quadriceps and hamstrings strengthening activities that elicit minimal KAM values, and thus are likely to minimize the potential for joint injury and degeneration. Activations of these muscle groups during squatting and lunging were equal to, or approaching, the recommended guidelines (Bennell et al., 2013; Kraemer et al., 2002). Meanwhile, mean KAM values during these activities were well below the peaks experienced during gait. It is recommended that lower extremity muscle training for individuals with or at risk for ACL injury or knee OA include variations on these exercises as major components of a balanced program.

6.3.3 Impact on Clinical Practice and Research

Clinical Practice

Uncertainty exists around appropriate dosage (type and frequency) of exercise prescription for knee OA and ACL injury (Pisters et al., 2007). Treatment programs developed by clinical practitioners often utilize a combination of aerobic and strengthening exercises for the lower extremities (Fransen & McConnell, 2008), however the biomechanical evidence of the risks and effectiveness for many of these exercises is minimal. Using quantifiable biomechanical measures, this thesis identifies exercises that minimize the risk of exposing the knee to unnecessary medial loading, while eliciting muscle activations within the guidelines for muscle training. These exercises are particularly beneficial for prescription by clinicians because they are easy to learn, have no associated costs, and are modifiable for various ages and abilities. Clinicians can build modifiable and progressive exercise programs around a combination of walking, squats, and lunges. Variations on squats and lunges include altering the width between the feet, holding weights, increasing or decreasing the knee flexion angle, and holding onto a balance support.

Perhaps more important for clinicians to note, Chapter 5 identifies isometric activities that may create a harmful environment for the knee or may not be effective. Clinicians should be wary of prescribing exercises or exercise programs that incorporate single-legged tasks as they may elicit higher medial knee loads in the support limb. While lower extremity stretching may increase flexibility, it may not be useful as a means for muscle building, as demonstrated by the low quadriceps and hamstrings activation levels during Triangle posture.

The findings outlined in Chapters 2 and 3 indicate that lower extremity neuromuscular fatigue may not be a substantial risk factor for altered knee loads, particularly in low demand activities such as gait and isometric postures. Clinicians can draw upon these findings in building an exercise program. For example, aerobic lower limb exercises with the potential to induce fatigue may be performed prior to isometric strengthening exercises without the risk of eliciting higher KAMs.

Determining the effects of changes in strength and power on knee loads requires additional longitudinal study. However, the results from Chapter 4 indicate that clinicians may be able to use baseline measures of extensor strength and power to inform them of a patient's potential risk of experiencing higher peak medial knee loads when walking. Importantly, these measures must be interpreted with caution as peak extensor torque and power only explained a small, but significant, portion of the variance in the first and second KAM peaks. The magnitude of the KAM during gait is also heavily influenced by factors such as body mass, frontal plane static alignment, and toe-out angle (Andrews, Noyes, Hewett, & Andriacchi, 1996; Foroughi, Smith, & Vanwanseele, 2009; Hurwitz, Ryals, Case, Block, & Andriacchi, 2002; Segal, Yack, & Khole, 2009), which should be evaluated by clinicians in combination with measures of strength and power.

Research

Research priorities for the future include the assessment of an exercise program emphasizing squatting and lunging postures to determine the effects on knee loading over an extended period of time. This should be explored in healthy participant groups as well as those with knee pathologies and knee injuries to evaluate whether knee loading response to these exercises differ between these groups of individuals. While the work in this thesis has identified potential exercises that elicit minimal medial knee loads, long term effects from this type of repetitive light loading cannot be determined without longitudinal analysis. In individuals with knee OA or ACL injury, an effective longitudinal exercise program could be partially defined by a reduction in the peak loading experienced by the knee during activities of daily living. Conversely, in healthy individuals, experiencing no increase in measures of knee loading may, in part, determine the effectiveness of the exercise program. Longitudinal outcomes are paramount to determining the effectiveness of such a program on knee health. This thesis established the potential for this type of exercise program based on biomechanical measures of knee loading and muscle activity.

Chapter 4 established that strength and power are both important in healthy young women for the two KAM peaks. These findings may contribute to the development of a model that covers the majority of the variance for the KAM during gait based on simple and cost-effective measures. Researchers may further speculate how the measures contributing to the variability in the KAM differ with age and knee health status (e.g. pathology, injury, etc.). While the work in this thesis examined the variability of the peaks of the KAM, questions remain around the importance of strength and power to the KAM waveform during gait as a whole, as well as the KAM during other activities and exercises.

6.4 Limitations, Assumptions, and Constraints

6.4.1 Knee Adduction Moment Measurements

Throughout this thesis, the KAM was used as a surrogate for medial compartment knee loading. Direct measurement of the in vivo joint contact force in the medial compartment of the knee non-invasively is not possible; therefore these forces must be estimated through other measures. The external KAM, estimated from the ground reaction force on the foot and the inertial properties of the lower limb, represents a net moment on all joint structures tending to rotate the tibia toward varus alignment on the femur.

Data from the relatively few knees that are instrumented with force-measuring implants show agreement with the external KAM measured using motion capture and force platforms for various patterns of gait (Kim et al., 2009; Zhao et al., 2007). However, slight gait modifications, such as a medial thrust of the knee, reduced the external KAM but did not always reduce the medial compartment knee joint force (Walter, D'Lima, Colwell, & Fregly, 2010). During some activities a combination of the KAM and the external knee flexion moment may better represent the load experienced by the medial compartment of the knee (Walter et al., 2010). This point is particularly relevant for movements involving extensive interior/exterior rotation of the shank. This previous work does highlight the possibility that the static exercises requiring greater knee flexion such as squats and lunges recommended in this thesis work could be eliciting higher medial compartment knee loads than those captured by the KAM due to a larger magnitude knee flexion angle.

6.4.2 Electromyography and Measures of Muscle Activation

As with estimating the magnitude of load experienced by the medial compartment of the knee, direct in vivo measurement of the force contributions of the quadriceps and hamstrings to joint loading requires invasive methods such as attaching a strain gauge to a tendon. The relationship between muscle force and muscle activation is not fully understood, however there is evidence that muscle activation increases non-linearly with increasing muscle force during isometric an isokinetic activities (Alkner, Tesch, & Berg, 2000; Bilodeau, Schindler-Ivens, Williams, Chandran, & Sharma, 2003; Karlsson & Gerdle, 2001; Perry & Bekey, 1981). Patterns of muscle activation during static and dynamic tasks are estimated using EMG to record and analyze neuromuscular activation through myoelectric signals. The EMG used throughout this thesis was non-invasive surface EMG, and as such, is limited to the measurement of activations from superficial muscles. Therefore, activations of deeper muscles that may contribute to joint motion and stability were overlooked.

It is important to note that the same posture, or joint angles, can be attained using different muscles or varied proportions of muscle activations, particularly for submaximal tasks (Cram & Kasman, 1998). Two of the studies contained within this thesis examined differences in EMG activation levels between tasks, and two others analyzed EMG differences during the same task before and after neuromuscular fatigue. A key limitation with surface EMG is the ability to only measure a proportion of the muscles that contribute to the movement pattern. For example, in the studies of this thesis the primary movements involved knee flexion and extension. The muscles from the quadriceps and hamstrings groups were monitored since they contribute heavily to these motions, while secondary movers such as the gastrocnemius, semimembranosus, and vastus intermedius were disregarded. This selection of muscles makes the comparison of muscle contributions between movements and fatigued states somewhat limited, as the recruitment patterns may have changed in muscles that were not being monitored. It is likely that more than one particular combination of muscles is able to produce the same knee loading environment and postural joint configuration for the postures studied in this thesis. Future studies should work toward modeling the knee loading environment during squatting and lunging under various states of muscle activation and co-activation around the joint.

6.4.3 Fatigue Effects on Muscle Activation, Implications for Interpretation

Two of the studies in this thesis examined muscle activations of the quadriceps and hamstrings, using surface EMG, before and after lower extremity fatigue, which was defined by a drop in peak torque output during isometric contractions. In Chapter 3, a potential link was hypothesized between the decrease in vastus lateralis activation during squatting and lunging, and the concurrent decrease in net external knee moments after fatigue. As noted in the previous section, the relationship between surface EMG amplitude and absolute force exerted by a muscle is complex and this relationship is further complicated by the introduction of neuromuscular fatigue (Bigland-Ritchie & Woods, 1984; Cifrek, Medved, Tonković, & Ostojić, 2009; Perry & Bekey, 1981). Therefore, caution must be exercised when speculating about the association between changes in muscle EMG amplitude and joint loading.

As a muscle fatigues, slow-twitch motor units may be recruited to take over as quickly-fatiguing fast-twitch motor units are switched off (T. Moritani, Muro, & Nagata, 1986; T. Moritani, Nagata, & Muro, 1982). Conduction velocity also decreases along the muscle fiber, changing the shape of the muscle unit action potential waveforms, on which the EMG signal is based (Cifrek et al., 2009; Dimitrova & Dimitrov, 2003). Furthermore, time synchronization in the activity of motor units has been linked with post fatigue increases in EMG amplitude and a shift in the EMG power spectrum (De Luca, 1984; Person & Kudina, 1968). However, these post fatigue increases in EMG amplitude can be somewhat inconsistent, and have led to some disagreement around their reliability (Dimitrova & Dimitrov, 2003; Gerdle, Larsson, & Karlsson, 2000; Hultman & Sjoholm, 1983; Vøllestad, 1997). When a muscle experiences an increase in activation, it is difficult to determine whether it occurred due to an increase in absolute force production or an increase in neuromuscular effort. However, the decrease in vastus lateralis activation after fatigue, presented in Chapter 3, was certainly related to a decrease in absolute force from the muscle due to a combination of a decreased drive to an already weakened muscle.

6.4.4 Muscle Activation, Coactivation, and Muscle Strength

The relationship between muscle activation and muscle strength, as measured by torque production capacity, is complex. Several studies in this thesis assumed that during static activities, higher muscle activations might indicate that more neuromuscular effort is required to maintain a certain position or joint angle. It has then been assumed that a prolonged or repetitive increase in neuromuscular effort might stimulate physiological changes to increase torque generation capacity.

Several studies have demonstrated a positive relationship between increases in muscle strength and muscle activation measured by surface EMG (Folland & Williams, 2007; Hakkinen & Komi, 1983; Häkkinen et al., 1996; Komi, Viitasalo, Rauramaa, & Vihko, 1978; Narici, Roi, Landoni, Minetti, & Cerretelli, 1989; Reeves, Maganaris, & Narici, 2005), however this finding is inconsistent (Garfinkel & Cafarelli, 1992; Narici et al., 1996). For example, in a study of 14 men who participated in 16 weeks of knee extensor strength training followed by 8 weeks of detraining, the change in surface EMG of the quadriceps muscles followed the changes in muscle strength, leading the authors to conclude that increases and decreases in muscle force output are related to neural and muscular adaptations to activity (Hakkinen & Komi, 1983). More recently, Pietrosimone and Saliba demonstrated that 47% of the increase in maximal quadriceps torque after a 4 week therapeutic exercise program was explained by changes in maximal voluntary activation of the vastus medialis and vastus lateralis in 36 participants with knee OA (Pietrosimone & Saliba, 2012). Contrarily, after undergoing an 8-week program of isometric knee extensors strength training, knee extensor torque in 8 women measured by maximum voluntary contrary increased with no significant change in peak EMG (Garfinkel & Cafarelli, 1992). These conflicting reports may be explained by technical and processing challenges in reproducing EMG measurements from one testing session to another.

It is unclear what constitutes desirable or ideal muscle activation patterns for improving lower extremity torque production. Muscle activation patterns appear to change with the onset and progression of knee OA (Bennell et al., 2013), but it is unknown whether these changes expedite structural damage or represent an attempt to protect against it. Therefore, it is unclear whether these changes should be encouraged or deterred and whether it is even possible to do so. Continuing to gaining a clearer understanding of muscle activation response to muscle function impairments and improvements in healthy individuals will help to answer these questions.

From this thesis work, coactivation between the quadriceps and hamstrings muscle groups does not appear to play as crucial a role in knee loading during submaximal static and dynamic activities such as gait, squats, and lunges in young healthy women, as it does in individuals with knee pathologies (Bennell et al., 2013). Results from these thesis studies suggests that the torque and power production of the quadriceps muscles appear to be more related to knee loading, and quadriceps activation seems to be more affected by impairments in muscle function such as neuromuscular fatigue. This suggests that hamstrings function may be of secondary importance to quadriceps function when medial knee joint loading is a concern.

6.4.5 Sample Selection and Size

The participant groups analyzed in this thesis were comprised of convenience samples of young, healthy women from the university population, who volunteered as study participants. The participant sample sizes in these thesis studies ranged between 20-30 individuals. Samples of this size may be low, introducing the potential for the studies to be underpowered, and preventing confirmation of the study hypotheses. As this is a common issue in biomechanics research (Mullineaux, Bartlett, & Bennett, 2001), efforts were made to increase the statistical power by increasing the trial sizes of the static and dynamic activities. Bates et al. (1992) suggests that for a sample size of 20 participants, at least 3 trials of the same activity should be collected to achieve a statistical power of 90%. All studies within this thesis utilized a higher sample size and/or more trials than the recommendations by Bates et al. (1992).

6.5 Directions for Future Research

The work in this thesis has built a foundation for a comprehensive future research program focused on the contributions of muscle function to the knee loading environment. This thesis is a series of cross-sectional studies that build upon one another and naturally lead to further questions requiring longitudinal analysis. The dose-response relationship to isometric muscle strengthening exercises in healthy young women remains unclear. In particular, given the relationships found in this thesis between quadriceps torque and power with the KAM during gait, future studies should identify whether gradual changes in quadriceps torque and power alter the first and second peak KAM values.

The fifth chapter of this thesis introduced an outline for a yoga-based exercise program designed to minimize knee loads implicated in knee OA and ACL injury, while targeting lower extremity muscle activations with the goal of increasing torque production capacity in these muscles. Squat and lunge-based yoga postures that elicited muscle activations within the recommended guidelines for improving lower extremity torque were identified. However, it remains unclear how frequently or for what length of time this type of program would need to be undertaken in order to develop significant improvements in lower extremity torque. It is hypothesized that these exercise frequency and duration parameters will be lower for young healthy women than for women with knee pathology and injury.

Furthermore, if such a yoga-based exercise program is to be used as an intervention in people with pathology, changes in knee loading during the exercise program itself and during activities of daily living should be monitored. These changes will be important in evaluating whether a yoga-based strengthening program could elicit a mechanical knee loading environment implicated in knee OA and ACL injury. Finally, while this thesis focused on knee loading since the knee is most often affected by pathologies such as OA and injuries, it will be important to monitor loading changes in other joints such as the hip during activities of daily living after an exercise intervention to evaluate whether these changes may be detrimental to joint health.

An assumption throughout the studies in this thesis is that prolonged and repetitive increases in lower extremity muscle activation through muscle training exercises will lead to muscle hypertrophy and subsequent increases in voluntary torque production capability. While a rich source of rigorous studies on muscle strengthening supports this assumption (MacIntosh, Gardiner, & McComas, 2006), the individual contributions of neural adaptations and muscle hypertrophy to torque increases remain unclear. A better understanding of the time sequence from a heightened neural drive to muscle hypertrophy and larger magnitude maximum voluntary contractions may be useful in determining the ideal length of a muscle training exercise program.

Several other parameters aside from discrete measures from the KAM waveform may be important in evaluating the mechanical loading environment of the knee. The KAM has been a focus of studies of knee loading especially in knee OA since it best represents the load distribution in the medial compartment of the tibiofemoral joint. However, recently, attention has been given to other measures of knee loading that differentiate loading patterns between healthy and diseased joints. These include measures of the knee load impulse and cumulative knee load which account for the duration of load experienced by the knee during an activity (Foroughi et al., 2009; Maly, 2008; Thorp et al., 2006), the rate of loading as the knee takes on body weight during activities such as gait (Mundermann, Dyrby, & Andriacchi, 2005), and variations in the shape and magnitude of the full KAM waveform (Astephen, Deluzio, Caldwell, Dunbar, & Hubley-Kozey, 2008; Deluzio & Astephen, 2007). In future studies, analysis of these measures may provide more detailed insights into changes in the mechanical knee environment due to alterations in muscle function.

6.6 Conclusions and Reflections

In summary, the work in this thesis explored the effects and contributions of muscle function to the mechanical environment of the knee during static and dynamic tasks in young healthy women. The major findings from this work are that:

1) In healthy active women, short term, rapidly produced torque deficits of the quadriceps and hamstrings has some limited effects on knee loading. These effects included a reduction in the peak external knee extension moment during the terminal stance phase of gait, reduced quadriceps activation during squatting and lunging, and reductions in the mean KAM and mean knee flexion moment during lunging.
2) Quadriceps torque and power explain significant portions of the variance in the first and second KAM peaks, respectively, during gait. The positive nature of these relationships indicates that the KAM peaks are higher for those with greater knee extension torque and power.
3) Squatting and lunging-based isometric yoga postures produce quadriceps and hamstrings activations that are at or close to current recommended exercise guidelines, while producing minimal KAMs.

These results contribute to the foundational knowledge of healthy knee function and add crucial understanding to the relationships between muscle function and mechanical knee loading response. Insights from this thesis will drive the future creation of muscle training exercise programs for individuals with or at risk for knee OA or ACL injury, having highlighted activities which target lower extremity muscles while minimizing the risk for overloading the knee.

6.7 References

Alkner, B. A., Tesch, P. A., & Berg, H. E. (2000). Quadriceps EMG/force relationship in knee extension and leg press. Medicine and Science in Sports and Exercise, 32(2), 459-463.

American Geriatrics Society Panel on Exercise and Osteoarthritis. (2001). Exercise prescription for older adults with osteoarthritis pain: Consensus practice recommendations: A supplement to the AGS clinical practice guidelines on the management of chronic pain in older adults. Journal of the American Geriatric Society, 49(10), 1400.

Andrews, M., Noyes, F. R., Hewett, T. E., & Andriacchi, T. P. (1996). Lower limb alignment and foot angle are related to stance phase knee adduction in normal subjects: A critical analysis of the reliability of gait analysis data. Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society, 14(0736-0266; 2), 289-295.

Andriacchi, T. P., & Dyrby, C. O. (2005). Interactions between kinematics and loading during walking for the normal and ACL deficient knee. Journal of Biomechanics, 38(2), 293-298. doi:10.1016/j.jbiomech.2004.02.010

Astephen, J. L., Deluzio, K. J., Caldwell, G. E., Dunbar, M. J., & Hubley-Kozey, C. L. (2008). Gait and neuromuscular pattern changes are associated with differences in knee osteoarthritis severity levels. Journal of Biomechanics, 41(4), 868-876.

Bates, B., Dufek, J., & Davis, H. (1992). The effect of trial size on statistical power. Medicine & Science in Sports & Exercise, 24, 1059-1065.

Bennell, K. L., Bowles, K., Wang, Y., Cicuttini, F., Davies-Tuck, M., & Hinman, R. S. (2011). Higher dynamic medial knee load predicts greater cartilage loss over 12 months in medial knee osteoarthritis. Annals of the Rheumatic Diseases, 70(10), 1770-1774.

Bennell, K. L., Wrigley, T. V., Hunt, M. A., Lim, B., & Hinman, R. S. (2013). Update on the role of muscle in the genesis and management of knee osteoarthritis. Rheumatic Diseases Clinics of North America, 39(1), 145-176.

Bigland-Ritchie, B., & Woods, J. (1984). Changes in muscle contractile properties and neural control during human muscular fatigue. Muscle & Nerve, 7(9), 691-699.

Bilodeau, M., Schindler-Ivens, S., Williams, D. M., Chandran, R., & Sharma, S. S. (2003). EMG frequency content changes with increasing force and during fatigue in the quadriceps femoris muscle of men and women. Journal of Electromyography and Kinesiology, 13(1), 83-92. doi:10.1016/S1050-6411(02)00050-0

Charette, S. L., McEvoy, L., Pyka, G., Snow-Harter, C., Guido, D., Wiswell, R. A., & Marcus, R. (1991). Muscle hypertrophy response to resistance training in older women. Journal of Applied Physiology (Bethesda, Md.: 1985), 70(5), 1912-1916.

Cifrek, M., Medved, V., Tonković, S., & Ostojić, S. (2009). Surface EMG based muscle fatigue evaluation in biomechanics. Clinical Biomechanics, 24(4), 327-340.

Cram, J., & Kasman, G. (1998). The basics of surface electromyography. Introduction to Surface Electromyography.USA: Aspen Publishers, Inc, 1-8.

De Luca, C. J. (1984). Myoelectrical manifestations of localized muscular fatigue in humans. Critical Reviews in Biomedical Engineering, 11(4), 251-279.

Deluzio, K. J., & Astephen, J. L. (2007). Biomechanical features of gait waveform data associated with knee osteoarthritis: An application of principal component analysis. Gait & Posture, 25(1), 86-93.

Devan, M. R., Pescatello, L. S., Faghri, P., & Anderson, J. (2004). A prospective study of overuse knee injuries among female athletes with muscle imbalances and structural abnormalities. Journal of Athletic Training, 39(3), 263-267.

Dimitrova, N., & Dimitrov, G. (2003). Interpretation of EMG changes with fatigue: Facts, pitfalls, and fallacies. Journal of Electromyography and Kinesiology, 13(1), 13-36.

Folland, J. P., & Williams, A. G. (2007). Morphological and neurological contributions to increased strength. Sports Medicine, 37(2), 145-168.

Foroughi, N., Smith, R., & Vanwanseele, B. (2009). The association of external knee adduction moment with biomechanical variables in osteoarthritis: A systematic review. The Knee, 16(5), 303-309. doi:10.1016/j.knee.2008.12.007

Fransen, M., & McConnell, S. (2008). Exercise for osteoarthritis of the knee. Cochrane Database of Systematic Reviews (Online), (4)(4), CD004376. doi:10.1002/14651858.CD004376.pub2

Gabriel, D. A., Kamen, G., & Frost, G. (2006). Neural adaptations to resistive exercise. Sports Medicine, 36(2), 133-149.

Garfinkel, S., & Cafarelli, E. (1992). Relative changes in maximal force, EMG, and muscle cross-sectional area after isometric training. Medicine and Science in Sports and Exercise, 24(11), 1220-1227.

Gerdle, B., Larsson, B., & Karlsson, S. (2000). Criterion validation of surface EMG variables as fatigue indicators using peak torque: A study of repetitive maximum isokinetic knee extensions. Journal of Electromyography and Kinesiology, 10(4), 225-232.

Grace, T., Sweetser, E., Nelson, M., Ydens, L., & Skipper, B. (1984). Isokinetic muscle imbalance and knee-joint injuries. A prospective blind study. The Journal of Bone & Joint Surgery, 66(5), 734-740.

Häkkinen, K., Kallinen, M., Linnamo, V., PASTINEN, U., Newton, R., & Kraemer, W. (1996). Neuromuscular adaptations during bilateral versus unilateral strength training in middle-aged and elderly men and women. Acta Physiologica Scandinavica, 158(1), 77-88.

Hakkinen, K., & Komi, P. V. (1983). Electromyographic changes during strength training and detraining. Medicine and Science in Sports and Exercise, 15(6), 455-460.

Haskell, W. L., Lee, I., Pate, R. R., Powell, K. E., Blair, S. N., Franklin, B. A., Macera, C. A., Heath, G. W., Thompson, P. D., & Bauman, A. (2007). Physical activity and public health: Updated recommendation for adults from the american college of sports medicine and the american heart association. Circulation, 116(9), 1081.

Hultman, E., & Sjoholm, H. (1983). Electromyogram, force and relaxation time during and after continuous electrical stimulation of human skeletal muscle in situ. The Journal of Physiology, 339, 33-40.

Hurley, M. V., Rees, J., & Newham, D. J. (1998). Quadriceps function, proprioceptive acuity and functional performance in healthy young, middle-aged and elderly subjects. Age and Ageing, 27(1), 55-62.

Hurwitz, D. E., Ryals, A. B., Case, J. P., Block, J. A., & Andriacchi, T. P. (2002). The knee adduction moment during gait in subjects with knee osteoarthritis is more closely correlated with static alignment than radiographic disease severity, toe out angle and pain. Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society, 20(0736-0266; 1), 101-107.

Karlsson, S., & Gerdle, B. (2001). Mean frequency and signal amplitude of the surface EMG of the quadriceps muscles increase with increasing torque—a study using the continuous wavelet transform. Journal of Electromyography and Kinesiology, 11(2), 131-140.

Kim, H. J., Fernandez, J. W., Akbarshahi, M., Walter, J. P., Fregly, B. J., & Pandy, M. G. (2009). Evaluation of predicted knee-joint muscle forces during gait using an instrumented knee implant. Journal of Orthopaedic Research, 27(10), 1326-1331.

Komi, P., Viitasalo, J., Rauramaa, R., & Vihko, V. (1978). Effect of isometric strength training on mechanical, electrical, and metabolic aspects of muscle function. European Journal of Applied Physiology and Occupational Physiology, 40(1), 45-55.

Kraemer, W. J., Adams, K., Cafarelli, E., Dudley, G. A., Dooly, C., Feigenbaum, M. S., Fleck, S. J., Franklin, B., Fry, A. C., Hoffman, J. R., Newton, R. U., Potteiger, J., Stone, M. H., Ratamess, N. A., & Triplett-McBride, T. (2002). American college of sports medicine position stand. Progression models in resistance training for healthy adults. Medicine and Science in Sports and Exercise, 34(2), 364-380.

Lewek, M. D., Rudolph, K. S., & Snyder-Mackler, L. (2004). Quadriceps femoris muscle weakness and activation failure in patients with symptomatic knee osteoarthritis. Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society, 22(1), 110-115.

MacIntosh, B. R., Gardiner, P. F., & McComas, A. J. (2006). Skeletal muscle: Form and function. Human kinetics.

Maly, M. R. (2008). Abnormal and cumulative loading in knee osteoarthritis. Current Opinion in Rheumatology, 20(1531-6963; 5), 547-552.

Moritani, T., Muro, M., & Nagata, A. (1986). Intramuscular and surface electromyogram changes during muscle fatigue. Journal of Applied Physiology, 60(4), 1179-1185.

Moritani, T., Nagata, A., & Muro, M. (1982). Electromyographic manifestations of muscular fatigue. Medicine and Science in Sports and Exercise, 14(3), 198-202.

Mullineaux, D., Bartlett, R., & Bennett, S. (2001). Research design and statistics in biomechanics and motor control. Journal of Sports Sciences, 19(10), 739-760.

Mundermann, A., Dyrby, C. O., & Andriacchi, T. P. (2005). Secondary gait changes in patients with medial compartment knee osteoarthritis: Increased load at the ankle, knee, and hip during walking. Arthritis and Rheumatism, 52(9), 2835-2844.

Narici, M., Hoppeler, H., Kayser, B., Landoni, L., Claassen, H., Gavardi, C., Conti, M., Cerretelli, P. (1996). Human quadriceps cross-sectional area, torque and neural activation during 6 months strength training. Acta Physiologica Scandinavica, 157(2), 175-186.

Narici, M., Roi, G., Landoni, L., Minetti, A., & Cerretelli, P. (1989). Changes in force, cross-sectional area and neural activation during strength training and detraining of the human quadriceps. European Journal of Applied Physiology and Occupational Physiology, 59(4), 310-319.

Palmieri-Smith, R. M., Thomas, A. C., & Wojtys, E. M. (2008). Maximizing quadriceps strength after ACL reconstruction. Clinics in Sports Medicine, 27(3), 405-424. doi:10.1016/j.csm.2008.02.001

Perry, J., & Bekey, G. A. (1981). EMG-force relationships in skeletal muscle. Critical Reviews in Biomedical Engineering, 7(1), 1-22.

Person, R., & Kudina, L. (1968). Cross-correlation of electromyograms showing interference patterns. Electroencephalography and Clinical Neurophysiology, 25(1), 58-68.

Pietrosimone, B. G., & Saliba, S. A. (2012). Changes in voluntary quadriceps activation predict changes in quadriceps strength after therapeutic exercise in patients with knee osteoarthritis. The Knee, 19(6), 939-943.

Pisters, M. F., Veenhof, C., Van Meeteren, N. L., Ostelo, R. W., De Bakker, D. H., Schellevis, F. G., & Dekker, J. (2007). Long-Term effectiveness of exercise therapy in patients with osteoarthritis of the hip or knee: A systematic review. Arthritis Care & Research, 57(7), 1245-1253.

Ratamess, N. A., Alvar, B. A., Evetoch, T. E., Housh, T. J., Ben Kibler, W., Kraemer, W. J., & Triplett, N. T. (2009). Progression models in resistance training for healthy adults. Medicine and Science in Sports and Exercise, 41(3), 687-708.

Reeves, N. D., Maganaris, C. N., & Narici, M. V. (2005). Plasticity of dynamic muscle performance with strength training in elderly humans. Muscle & Nerve, 31(3), 355-364.

Rice, D. A., McNair, P. J., & Lewis, G. N. (2011). Mechanisms of quadriceps muscle weakness in knee joint osteoarthritis: The effects of prolonged vibration on torque and muscle activation in osteoarthritic and healthy control subjects. Arthritis Research & Therapy, 13(5), R151. doi:10.1186/ar3467

Sale, D. G. (1988). Neural adaptation to resistance training. Medicine and Science in Sports and Exercise, 20(5 Suppl), S135-45.

Segal, N. A., & Glass, N. A. (2011). Is quadriceps muscle weakness a risk factor for incident or progressive knee osteoarthritis? The Physician and Sportsmedicine, 39(4), 44-50. doi:10.3810/psm.2011.11.1938

Segal, N. A., Yack, H. J., & Khole, P. (2009). Weight, rather than obesity distribution, explains peak external knee adduction moment during level gait. American Journal of Physical Medicine & Rehabilitation / Association of Academic Physiatrists, 88(3), 180-8; quiz 189-91, 246. doi:10.1097/PHM.0b013e318198b51b; 10.1097/PHM.0b013e318198b51b

Shelbourne, K. D., & Gray, T. (1997). Anterior cruciate ligament reconstruction with autogenous patellar tendon graft followed by accelerated rehabilitation. A two- to nine-year followup. The American Journal of Sports Medicine, 25(6), 786-795.

Shelburne, K. B., Torry, M. R., & Pandy, M. G. (2006). Contributions of muscles, ligaments, and the ground-reaction force to tibiofemoral joint loading during normal gait. Journal of Orthopaedic Research, 24(10), 1983-1990.

Slemenda, C., Brandt, K. D., Heilman, D. K., Mazzuca, S., Braunstein, E. M., Katz, B. P., & Wolinsky, F. D. (1997). Quadriceps weakness and osteoarthritis of the knee. Annals of Internal Medicine, 127(0003-4819; 0003-4819; 2), 97-104.

Thorp, L. E., Sumner, D. R., Block, J. A., Moisio, K. C., Shott, S., & Wimmer, M. A. (2006). Knee joint loading differs in individuals with mild compared with moderate medial knee osteoarthritis. Arthritis & Rheumatism, 54(12), 3842-3849.

Tremblay, M. S., Warburton, D. E., Janssen, I., Paterson, D. H., Latimer, A. E., Rhodes, R. E., Kho, M. E., Hicks, A., Leblanc, A. G., Zehr, L., Murumets, K., & Duggan, M. (2011). New Canadian physical activity guidelines. Applied Physiology, Nutrition, and Metabolism, 36(1), 36-46.

Vøllestad, N. K. (1997). Measurement of human muscle fatigue. Journal of Neuroscience Methods, 74(2), 219-227.

Walter, J. P., D'Lima, D. D., Colwell, C. W.,Jr, & Fregly, B. J. (2010). Decreased knee adduction moment does not guarantee decreased medial contact force during gait. Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society, 28(10), 1348-1354. doi:10.1002/jor.21142

Young, A., Stokes, M., & Crowe, M. (1984). Size and strength of the quadriceps muscles of old and young women. European Journal of Clinical Investigation, 14(4), 282-287.

Zhao, D., Banks, S. A., Mitchell, K. H., D'Lima, D. D., Colwell, C. W., & Fregly, B. J. (2007). Correlation between the knee adduction torque and medial contact force for a variety of gait patterns. Journal of Orthopaedic Research, 25(6), 789-797.