Strategies Utilized during a Novel Rotary Task in Total Knee Replacement Subjects

By

Lauren A. Ferris

Submitted to the graduate degree program in Mechanical Engineering and the Graduate Faculty of the University of Kansas in partial fulfillment of the requirements for the degree of Master of Science.

____

Dr. Lorin Maletsky, Chairperson

____

Dr. Jessie Huisinga, Committee Member

____

Dr. Carl Luchies, Committee Member

Date Defended: May 13, 2014

The Thesis Committee for Lauren A. Ferris

certifies that this is the approved version of the following thesis:

Strategies Utilized during a Novel Rotary Task in Total Knee Replacement Subjects

Dr. Lorin Maletsky, Chairperson

Dr. Jessie Huisinga, Committee Member

Dr. Carl Luchies, Committee Member

Date Defended: May 13, 2014

Abstract

The ability to perform activities of daily living involving flexion/extension of the knee along with rotation are crucial for total knee replacement (TKR) patients to regain their independence post-surgery. The proposed research identified compensatory strategies used by TKR subjects during a novel rotary task. The task consisted of two activities, a high to low (H2L) and a low to high (L2H) button task where subjects utilized a crossover technique in order to press buttons located at shoulder and knee height by flexing and extending the knee. Ground reaction forces and kinematics were recorded for eleven TKR subjects and twelve healthy controls. Data were modeled in a musculo-skeletal modeling system to determine knee torque, center of mass displacement, and muscular activity. Each leg was categorized as affected (TKR knee), unaffected (non-TKR limb), or a healthy control. No statistical differences were found in the force transfer for the different groups, although differences in the variation of the loading within subjects were noted. Differences were found between healthy and unaffected knee angles and a strong trend between healthy and affected subject's knee angles in both tasks. L2H had the most variation where a significant difference was present predominantly between unaffected and healthy in the knee flexion, knee torque, and hip extensor muscles. Consistencies during both tasks in knee torque and muscle activation while knee angles varied suggests the kinematics during this type of motion is driven by the cross over. These outcomes suggest that individuals with a TKR may utilize strategies, such as keeping an extended knee and altered muscle activation, to achieve rotary tasks during knee flexion and extension, yet these strategies were not reported consistently from task to task. Early identification of these strategies could improve TKR success and the return to activities of daily living that involve flexion and rotation.

Acknowledgements

This thesis is the result of a collaborative effort from several people, and I owe a special thanks to them.

- To my advisor, Dr. Lorin Maletsky, thank you for allowing me to be a part of your EJBRL team. I am a very proud Jayhawk, mostly when I think back on all my years in the lab. You really pushed and encouraged me to be my best and without you I would not be in the engineer I am today. Thank you.

- To Dr. Carl Luchies, Dr. Jessie Huisinga, and the other faculty members at the School of Engineering and KU Medical Center for all their help and support.

- To Kevin Shelborne, Time Dorn, James Dunne, and the entire OpenSim group for their ongoing help and support.

- To my fellow graduate students in the lab, Kaity, Mark, Fallon, Sami, Adam, Max and Mariam. It has been quite the journey.

- To DePuy Synthes Orthopaedics, a Johnson & Johnson company, for funding my research experiences.

- To my family, thank you for always being there to support and love me. There have been a lot of mountains to climb and you never left my side.

- Most of all to Linda Denney, without you none of this would be possible. I couldn't have asked for a better co-researcher. "Always there, always cares"

Table of Contents

List of Abbreviations

ACL	Anterior Cruciate Ligament
ADL	Activity of Daily Living
ANOVA	Analysis of Variance
A/P	Anterior/ Posterior
BMI	Body Mass Index
BOS	Base of Support
BW	Body Weight
COM	Center of Mass
DOF	Degree of Freedom
EMG	Electromyography
GRF	Ground Reaction Force
H2L	High to Low
L2H	Low to High
LFR	Lead Force Ratio
OA	Osteoarthritis
ROM	Range of Motion
S/I	Superior/ Inferior
TA	Tibialis Anterior
TKR	Total Knee Replacement
TTT	Target Touch Task

List of Figures

List of Tables

Chapter 1: Introduction

Total knee replacements (TKR) are one of the top elective joint replacement surgery with over 600,000 performed annually in the United States [1]. This number is predicted to increase by 673% by the year 2030 [2]. With a growing population and life expectancy, it is inevitable the number of TKRs performed along with expectations of the implant will increase. Some of these expectations include longevity of the implant and functional ability of the patient post-TKR. Functional ability depends on the individual expectations of the patient [3], while activities of daily living, or ADLs, are necessary for the patient to be able to accomplish daily tasks without need of assisted living. Typical ADLs include lifting items, loading and unloading the dishwasher, bathing, maneuvering steps, and getting in and out of a car.

The ability to return to independent ADLs, along with reduced pain, are two subjective outcome measures related to satisfaction with TKR surgery [3-8]. Pain relief ranks higher than functional outcomes since patients are more willing to alter their behavior and level of activity to compensate for functional deficiencies [3]. In a study by Nilsdotter et al. it was reported that patient expectations are not met until five years post-TKR surgery [9], while Benedetti et al. reported TKR subjects to have significantly decreased function 24 months post-op compared to functional scores at 6 and 12 months post-op [11]. Other post-TKR deficiencies include decreased ability to walk and/or descend stairs without a hand rail [10], and slight pain in the contralateral, or unaffected knee [11]. By identifying factors that lead to these specific deficiencies, satisfaction of the patients and rehabilitation techniques could be improved.

Along with changes in functionality in the TKR knee, compensatory strategies of overloading the non-surgical knee could cause further knee problems. It has been reported that about 40% of patients with TKRs will have to have a TKR in the non-operated knee within ten years [12, 13]. Kinematics and muscle activation have been shown to be altered post-TKR in both the surgical and non-surgical limb [14-16]. Fallah-Yakhdani [17] found that during gait, the unaffected limb of TKR subjects had significantly less stability than healthy subjects pre-surgery and concluded that after heel strike muscular co-contraction of the unaffected limb can decrease the sagittal plane stability of the affected limb. Gage et al. reported TKR patients performing a stair ascend by reducing the work performed on the TRK knee by compensating with the hip and unaffected limb [14]. Knowledge of compensatory techniques in the unaffected limb could potentially reduce the likelihood of future surgery in the patient's non-TKR knee.

Many ADLs, such as gait, squatting, kneeling, and sit-to-stand, have been analyzed clinically, in vitro, and with a computational model, yet these activities frequently have little to no movement out of the sagittal plane. As younger, more active patients undergo TKR, the expectation or demand to return back to sports and pre-replacement activities becomes more of a priority [2, 18-20]. Although the kinematic and kinetic sagittal plane motions currently analyzed are important to the understanding of how TKR affect patients, little is known on activities in the transverse plane such as lifting a laundry basket, unloading the dishwasher, or reaching for an item, when the feet are stationary. ADLs incorporating rotation in the transverse plane, along with sports such as golf and tennis, requires the patient to transfer their weight while in double stance, flex/extend the knee, and also rotate. Due to these three dynamic requirements a greater axial load is applied to perform the rotary motion and there is a potential to increase the risk of the patient feeling unstable. Adding rotation and knee flexion to reaching tasks have proven difficult for post-TKR patients [21].

Initial studies into the relationship between compensatory muscle, kinematic, and kinetic strategies and functional activities have begun to show up in literature, yet little is known about how these factors change when out of plane rotation is added and when the feet are stationary. The purpose of this study was to investigate the strategies utilized by TKR subjects compared to healthy when performing a novel knee flexion/extension task including rotation and transfer of weight know as the Target Touch Task (TTT). The specific aims of this research were:

Specific Aim 1: Identify kinematic and kinetic strategies utilized by individuals with a TKR while in double-stance transferring load during rotational activities.

Specific Aim 2: Identify the compensatory strategies utilized during the TTT between the healthy, TKR affected knee, and the TKR subject's unaffected knee and to determine the relationship between musculature and biomechanical activity throughout the task.

The following chapters detail the research performed. Much collaboration was done during this study, thus the layout of this thesis is unique to account for co-authoring. Chapter 2 is a literature review that includes TKR and expected functionality post-TKR, musculature of the knee and how TKR affected it, currently identified strategies used by TKR patients, and stability of TKR patients. Chapter 3 details the work contributed to the publication attached in Appendix A. Appendix A paper describes the initial data collected from force plate data and calculated knee angles from trajectory data to address Aim 1. Chapter 4 details an in depth description of OpenSim and the model utilized in the data for the manuscript in Appendix B. Appendix B details the use of OpenSim to model subject specific data and address Aim 2. Chapter 5 is an overall summary, conclusions, and future work of the project and data.

Chapter 2: Literature Review

This literature review focuses on understanding the stabilizing structures of the knee, compensatory strategies of total knee replacement (TKR) patients already present in the collected works, and the various stability strategies utilized by TKR patients. A majority of current work focuses on motion and strategies in the sagittal plane, such as gait, squatting, and sit-to-stand activities, while few studies have reported on transverse plane or rotary motion. Twisting at the waist while transferring one's weight is a common daily activity and crucial for independence post-TKR. Therefore, the literature review highlights common strategies utilized by TKR patients during dynamic movement.

Musculature

Anatomy and knowledge of the musculature of the lower limb is crucial to the understanding of how TKRs affect a patient post-replacement. Key muscle groups often reported in TKR studies are the quadriceps and hamstrings. Other musculature important for gait include the gluteus maximum, gluteus medius, vastii, soleus, and gastrocnemius muscles that allow for forward progression in normal gait [22], while the hip extensors (gluteal muscles), quadriceps, hamstrings, gastrocnemius, and the soleus assist in squatting and rising from a deep squat [23, 24]. Since internal muscle forces are difficult to measure in human experiments, musculo-skeletal models have been used to predict individual muscle activation and activity [25, 26]. These types of models often incorporate the bones, muscles, joints, and passive structures with a variety of degrees of freedom to simulate dynamic movements [26].

The quadriceps are the anterior femoral muscle and are made up of four distinct muscles (vastus lateralis, vastus medialis, vastus intermedius, and the rectus femoris) that extend the leg.

The primary role of the rectus femoris is to assist the psoas and illiacus in supporting the pelvis and trunk on the femur [27]. The rectus femoris also aids in flexing the hip [23]. Quadriceps strength deficiency has been reported post-TKR by many researchers with significant weakness still occurring at one year post-op [14, 15, 25, 28-33]. It has been hypothesized this weakness is due to muscle atrophy and reduced muscle activation [34, 35]

The hamstrings are the posterior femoral muscles that flex the knee and consist of the long and short head of the bicep femoris, semitendinosus, and semimembranosus. The bicep femoris causes the leg to externally rotate while the semitendinosus and semimembranosus assist with internal rotation. Along with knee flexion, the hamstrings assist in supporting the pelvis and aid in hip extension [27]. Hamstring strength has been reported to be significantly lower in post-TKR patients in studies including reaching tasks [15] and leg strengthening exercises [32]. While hamstring strength recovers faster than the quadriceps, a deficiency is still present compared to healthy controls [36, 37]. Left untreated, hamstring deficiency could lead to hamstring tendinopathy and thus increased pain post-TKR [32].

The gastrocnemius is the major muscle of the calf and is made up of a medial and lateral head. The soleus is a broad, flat muscle that lies deep to the gastrocnemius. The gastroc-soleus complex is the main plantar flexor of the foot and ankle and are activated when standing, walking, and during most ADLs. The soleus' primary role is to support the body to prevent falling forward while the gastrocnemius flexes the femur upon the tibia and plantar flexes the ankle [27]. The gastrocnemius, soleus, and gluteus medius have been reported to aid in propelling the center of mass (COM) forward and provide vertical support during gait [25, 32, 38]. Benedetti et al. reported premature activation of the gastrocnemius at 6 months post-op [11] and prolonged co-contraction of the gastrocs and tibialis anterior (TA) during gait. Gage et al.

found gastrocnemius and TA muscle activity to be lower in TKR patients than healthy when examining COM response to a support surface perturbation. Gage speculated that this could alter knee range of motion (ROM) [28] while higher activation amplitudes during late stance of gait has been reported in TKR patients [29].

The muscles of the trunk also play a key role in stabilizing the pelvis during functional activities, gait, and perturbations post-TKR. In a gait study by Benedetti et al. the contribution of the ipsilateral and contralateral erector spinae were found to increase during the stance and swing phase, suggesting the trunk and spine muscles may be used as a strategy post-TKR [11, 39]. Abnormal muscle activation in the ipsilateral, erector spinae, and TA activation during mid-stance was also speculated as a compensatory strategy to reduce the adduction moment at the knee. Li et al. also found TKR subjects had a significantly increased back flexion during most phases of gait and tended to lean the trunk forward increasing the sagittal plane flexor moment [39].

Strategies

Strategies have been reported by many authors in relation to how TKR patients compensate for pain and functional outcomes post-TKR. Some of these strategies reported during specific tasks include reduced walking velocity and stride length during gait [14, 30, 39-42], reduced work at the knee while compensating with the hip [43] or trunk/back [39], and reduced external knee joint moments during stair climbing [44, 45]. Along with these tasks, quadriceps avoidance, co-contraction, and altered kinematics are the most commonly reported strategies employed by TKR patients reported in the literature.

Quadriceps Avoidance

In a study by Stevens et al., [46] subjects with osteoarthritis (OA) and age matched healthy subjects reported the affected group having 20% less quadriceps strength compared to the healthy. OA is most commonly treated by TKR surgery [29], thus most TKR patients already have some form of quadriceps weakness pre-TKR. TKR patients have a difficult time return back to pre-replacement quadriceps strength in the operated leg comparable to a healthy subject or their unaffected side [31]. Loss of quadriceps strength post-replacement can be as high as 64% deficiency 3-4 weeks following the procedure [47]. In a musculoskeletal modeling study by Li et al., TKR subjects 12 months post-op were compared to healthy controls during the stance phase of gait. Contributions from vastii and rectus femoris on the net knee extensor moment were significantly lower than the healthy controls [39]. When comparing long term quadriceps strength in TKR patients and healthy subjects, research has determined there will be a 30-48% deficiency between the two group's strength [31]. Due to such a high strength deficiency, quadriceps avoidance has become a recognized gait pattern for patients with TKR [11, 39, 48]. Andriacchi defined this gait pattern, "quadriceps avoidance gait", where the subject keeps an extended knee during the stance phase and avoids the use of the quadriceps [49]. This type of gait pattern decreases the mechanical advantage of the knee and is often employed when the subject has instability or weakness and has been observed in many OA [50] and TKR studies.

Co-Contraction

The quadriceps and hamstrings work together to stabilize the tibiofemoral joint by co-contracting. A common example of co-contraction is when the quadriceps and hamstrings fire together to reduce shear forces and strain on the knee joint [51]. While co-activation is important to stabilize the knee, increased activation can cause wear on TKRs and cartilage damage to the healthy knee [11, 51-53] by increasing the tibiofemoral contact forces [25]. This phenomena of excessive activation was present in Stevens-Lapsley study where the TKR limb hamstring co-activation increased compared to the unaffected and healthy legs [32]. Increased co-contraction can also lead to decreased motion efficiency and increased energy costs [25]. Hubley-Kozey et al. [29] found that patients with OA performed gait with increased lateral hamstring activation and a reduced medial gastrocnemius contribution [54, 55] and suggested that patient's with early stages of OA tend to have lateral co-contraction to compensate for medial complications [55].

Co-contraction has been reported as a strategy to compensate for quadriceps weakness in many studies [56]. Studies have also shown subject co-contracting longer, more, or have higher muscle activity during gait compared to healthy controls [54, 56-64]. These co-contraction strategies, along with increased joint loading, could also result in higher metabolic costs [65, 66]. It has been suggested that altered co-contraction is a compensatory strategy to control the kinematics at the knee [11, 65, 67, 68] and it has been hypothesized that longer co-contraction is associated with "stiff knee pattern" gait.

Kinematics and Kinetics

During gait, TKR patients demonstrate limited knee flexion during the swing and stance phase [11, 39] and demonstrate posting, or a "stiff knee gait pattern" [11], with the knee experiencing a greater knee flexion moment, quadriceps activation, and hamstring activity [69]. Knee flexion moments correlate to the forces acting across the knee and the demands of the musculature stabilizing the knee [70]. Compensatory strategies seen in TKR patients such as maintaining a knee flexor or knee extensor moment during gait instead of using a combination of both, as demonstrated in healthy gait, could lead to further TKR problems [71].

Dynamic task analysis provides the bulk of research data in lower extremity kinetics. Ground reaction force (GRF) is the most commonly measured loading profile during dynamic tasks with the use of force plates or force mats. Post-TKR loading configurations have been studied in both the TKR affected and unaffected. Yoshida et al. [72] found significant differences between the affected-unaffected load 3 months post-surgery. While differences were noted between the two group's GRF 15 months after surgery in a sit-to-stand study by Mundermann [73], the limbs performed similarly, leading to the conclusion that the unaffected and affected sides may become more symmetrical as soon as one year post-TKR [37, 74]. Various researchers (eg. Bergmann [75] and D'Lima[73]) have implanted tibial trays with pressure transducers built into the component during a TKR to collect loads of the artificial knee during dynamic tests. Through studies involving walking, squatting, stair activities, and rising from a chair it has been determined that TKR patients performing these activities load the knee at values that well exceed two-times body weight [19, 73].

Knee flexion excursion, or the knee flexion range of motion, is limited during gait thus contributing to the stiff knee gait [41, 74, 76]. Li et al. concluded that TKR subjects were able to reduce their knee flexion during gait by compensating with their back muscles and tilting the trunk forward. Significant decreases in knee angles have also been reported in chair rise and mild squatting activities, compensating with the unaffected limb as a pain avoidance strategy [30]. Gage et al. studied a sagittal moving platform and concluded that reduced knee flexion is due to compensation by the paraspinal and abdominal muscles and upper body movement to limit COM movement [14, 30].

Stability

Knee instability post-TKR is the number one reason for TKR revision [77-79]. Knee anatomy and muscle activation are used to stabilize the knee where instability results in the knee "giving way", loss of balance, or falls [43]. The anterior cruciate ligament (ACL) is the main ligament that stabilizes tibial rotation and anterior translation. Injuries to the ACL frequently result in knee instability. ACL reconstruction is often performed to repair ACL injuries and deemed successful when anterior stability is accomplished [80-83], while rotational instability is sometimes still present post-surgery [82, 84-88]. Anterior stability has been tested in many clinical settings using a KT-1000, a manual instrument used to measure anterior tibial motion relative to the femur [89]. Even though the KT-1000 has been used in many manual assessments, it has been determined that the KT-1000 is better as a comparison device between the unaffected and affected knee with a threshold of 2 or 3mm differences to determine deficiencies [89]. Compared to translating motions, tests such as the pivot-shift test [90], N-test [91], and magnetic resonance imaging [92] are also utilized to examine rotary stability of the knee.

Muscle activity such as co-contraction is a strategy to stabilize the knee [17, 93]. It has been reported that altered muscle contribution is due to patients feeling unstable during gait [94] and other dynamic tasks. Fallah-Yakhdani et al. [17] reported that instability in the sagittal plane caused a longer co-contraction to aid in stabilizing the knee during gait. Gage et al. reported the gastrocnemius and rectus femoris as the primary and secondary muscles that aid in balance and stability [14].

Control of COM aids in the ability to balance. Gage et al. found that TKR patients have a larger COM movement along with hip and pelvis rotation during a frontal plane platform rotation

[28] to aid in stabilizing their balance. Mandeville et al. [95] believed that TKR patients have already developed strategies pre-surgery to balance COM and reported smaller COM displacement in the TKR subjects compared to healthy controls during gait.

These findings highlight the need for more research regarding out-of-plane rotations to better understand how a TKR affect knee kinematics, musculature, strategies utilized during rotational tasks, and stability during the tasks. Specific compensatory strategies can be identified for dynamic movements such as gait, squatting, and sit-to-stand activities, yet little is known how flexing/extending the knee, rotation, and transferring of weight from one side to the other in double stance will affect a TKR patient. Since this type of activity is involved in every day activity, it is imperative to understand the strategies employed by the patient to further aid in post-surgery rehabilitation and/or TKR component expectations.

Chapter 3: Strategies Utilized to Transfer Weight during Knee Flexion and Extension with Rotation for Individuals with a Total Knee Replacement

I was lead author on the paper, "Strategies Utilized to Transfer Weight during Knee Flexion and Extension with Rotation for Individuals with a Total Knee Replacement" written in collaboration with Linda Denney PT, M.Appl.Sci (Manip), a doctoral student and clinical instructor in the KU Department of Physical Therapy and Rehabilitation Science. This manuscript was published in the Journal of Biomechanical Engineering. The full publication can be found in Appendix A. Linda and I collected all the subject data together where she ran the study and I was the engineer collecting the data. The data were processed and exported from the Vicon Workstation at the University of Kansas Medical Center by Linda. I performed all the data analysis once the Vicon data was constructed and exported. Linda predominantly focused on the introduction and discussion portion of the paper, along with running the statistics for Table 3.2 and 3.4 using SPSS. The methods and materials and results were my main contribution along with the data analysis, figure generation, and running the remaining statistics. Various first author duties such as formatting, collaborating with the authors, and editing were also my responsibility.

Chapter 4: Using a Musculo-Skeletal Model to Assess Muscle Activation and Biomechanical Strategies During a Rotational Task in Individuals with a Total Knee Replacement

Similar to Chapter 3, Linda Denney and I collaborated on the manuscript in Appendix B. Since the initial subject data was the same as Appendix A, those roles remain the same. I then utilized that data to create subject specific musculo-skeletal models in OpenSim to generate muscle activity. For the manuscript Linda focused on the introduction and contributed greatly to the discussion. She also performed the statistics in Table 1, 2, and 3 using SPSS. The remainder of the paper I wrote and produced. Linda also aided in editing the text and drawing conclusions from the presented data.

OpenSim is an open source software used to model dynamic simulations to study neuromuscular coordination and estimate muscular loads [96]. Gait 2392 Model and OpenSim 3.1 were utilized in the modeling of the Target Touch Task (TTT). This model, along with various types of models, are available as a download in OpenSim [97]. Model 2392 was chosen based on its dominates by lower extremity muscles. Various other lower limb models are available and listed in a comparison table [98] on the OpenSim website. Based on the accuracy and time to run the model the benefits of Gait 2392 outweighed the other models. Gait 2392 is a three dimensional, 23 DOF, 92 musculotendon actuator computer model of the musculoskeletal system created by Derryl Thelen at the University of Wisconsin-Madison and Ajay Seth, Frank C. Anderson, and Scott Delp at Stanford University [98]. The model is represented by 76 muscles of the torso and lower body defined by work by Delp et al. [99] on the lower extremity joints, Anderson and Pandy [100] on the low back joint, and Yamaguchi and Zajac [101] on the planar knee joint. Model 2392 was scaled for each subject that completed the TTT by indicating the subject's body weight in kilograms and height as a ratio of the subject's height over the model height of 1.8 meters.

Joint geometries of the pelvis, femur, patella, tibia/fibula, foot (calcaneus, navicular, cuboid, cuneiforms, metatarsals), and toes were all modeled as segments with individual Cartesian coordinates assigned to each body. Motions of these bodies were defined by the hip, knee, ankle, subtalar, and metatarsophalangeal models. The hip joint was modeled as a ball and socket joint where the femur reference frame rotated about the three axes fixed to the center of the femoral head. The ankle, subtalar, and metatarsophalangeal joints were modeled as frictionless revolute joints with axes located based on research by Inman [102]. Modification was made to the metatarsophalangeal axis to avoid dislocation of the joint during model movement.

The knee was modified from Yamaguchi's [101] one DOF model with both the kinematics of the tibio-femoral and patella-femoral joint in the sagittal plate. Delp et al. [99] then modeled the femoral condyles as ellipses, tibial plateau as a line segment, and specified the transformation matrices of the femur, tibia, and patella as a function of knee flexion. Tibio-femoral kinematics were modeled so the femoral condyles stay in contact with the tibial plateau, with the contact point depending on knee flexion [103]. Patellae were removed from the model due to kinematic constraints, thus the quadriceps insertion site was modeled as a moving point in the tibial reference frame.

Muscle-tendon actuators were represented as line segments in the lower extremities based on bony landmark insertion sites. For muscles that wrap over the bone, midway insertion points were employed to keep muscle line of action anatomically accurate and attempted to prevent the muscle from passing through the bone. Some muscles such as the hip flexors and extensors pass through the bone and deeper muscles in deep hip flexion ($> 60^o$) and must be considered when analyzing such muscles [103]. Inertial properties, cross sectional area, strength parameters, and optimal fiber lengths for the muscles were adapted from cadaveric work presented in the Gait 2392 and 2354 Models document. Motion capturing markers specific to the TTT protocol were lastly applied to the model using x,y,z coordinates and joint geometries.

Motion and force plate data collected at the Landon Center on Aging at the University of Kansas Medical Center was exported via Vicon Workstation (v4.5) as .c3d files and converted to OpenSim compatible files(.mot and .trc) using transformation matrices and the Gait-Extract toolbox [104] created in Matlab. Once the files were compatible for OpenSim, the Gait 2392 model was scaled to subject specific height and weight. Scaling factors of the subject's height over the models height (1.8 m) were calculated to make scaling factors for the torso, femurs, and tibiae to model approximately the same height and geometry of the subjects. Since the subjects were instructed to keep their feet planted on the ground the ankle, heel, and toe markers were weighted 5 times heavier in the scaling process and the subtalar joint was locked between -90^{o} and 90^{o}, preventing the foot from revolving about the heel to toe action line.

Inverse kinematics were ran on each trial to calculate the generalized coordinate trajectories. This was done by varying the joint angles to minimize the error between the model and experimental markers. Mathematically this is accomplished by using a weighted least square problem (Eqn. 1):

$$\min_q \left[\sum_{i\in markers} w_i \left\|x_i^{exp} - x_i(q)\right\|^2 + \sum_{j\in unprescribed\ coords} w_j\left(q_j^{exp} - q_j\right)^2\right] \qquad (1)$$

$$q_j = q_j^{exp} \text{ for all prescribed coordinates } j$$

q: vector or generalized coordinates being solved

w: marker weights

x_i^{exp}: experimental position of marker I

$x_i(q)$: position of the corresponding marker on the model

q_j^{exp} : the experimental value for coordinate j [105]

Once inverse kinematics were calculated at each time stamp a .mot file was generated and saved to be used on the following calculations.

Inverse dynamics used the calculated inverse kinematics and 4^{th} order low pass Butterworth filtered force plate analog data to calculate net joint torques and forces at each time stamp. This is accomplished through differential equations that take into account the generalized position, velocity, and acceleration of the model along with masses of the model bodies and gravitational forces. Lastly, muscular forces and activation times and amplitudes were calculated with the static optimization using residual actuators and outputs from inverse kinematics and dynamics. Residual actuators are necessary when performing static optimizations to compensate for dynamic inconsistencies between the model accelerations and clinically collected ground reaction force (GRF). Static optimizations are generated by solving for the inverse dynamics and using those results to solve the redundancy problem for muscles. Center of Mass (COM) data were also generated within OpenSim by the analysis tool body kinematics function. Application of these modeling techniques can be found in the manuscript in Appendix B.

Chapter 5: Conclusions

The objective of this research was to identify compensatory strategies TKR patients use during rotary tasks when flexion and extension of the knee is necessary. This research consisted of two specific aims. The first was to identify kinematic and kinetic strategies utilized by individuals with a TKR while in double-stance transferring load during rotational activities. This was done by analyzing ground reaction force parameters and knee angles. The second was to identify the compensatory strategies utilized during the TTT between the healthy, TKR affected knee, and the TKR subject's unaffected knee and to determine the relationship between musculature and biomechanical activity throughout the task with the aid of OpenSim.

Chapter 3 and Appendix A described how the subjects transferred their weight during both the H2L and L2H TTT along with how the lead knee angles changed during the two tasks. For this analysis no statistical differences were observed in the force parameters, regardless of task. A significantly greater knee angle during the low portion of the H2L task between the healthy subjects and unaffected limb of the TKR subject was reported. As for the L2H, the affected knee was more extended compared to the unaffected at the high button push. These data suggested the TKR patients used a compensatory strategy of keeping an extended knee during the TTT with little to no variation between the groups for weight transfer.

Chapter 4 and Appendix B built off the preliminary data presented in Chapter 3 and Appendix A to further identify compensatory strategies of the TKR subjects. This was done by utilizing OpenSim, a subject specific musculo-skeletal modeling system to analyze variables in both the lead and lag limbs for H2L and L2H. Variables analyzed were knee joint torques, knee angles, center of mass displacement, and muscle activity. Variations were mainly noted in the L2H task where the healthy subjects extended the lead knee more while activating the posterior/

hip extensor muscles to perform the L2H task. Consistencies were observed between the L2H and H2L knee torques, center of mass movement in the anterior/posterior and side/side, and muscle activation, independent of knee angle. This led us to believe the outcomes of the TTT were driven by the crossover portion of the task, not the flexion/extension of the knee.

Kinematic, kinetic, and muscular results from these data suggest that TTT identifies significant differences during a cross over task that may not present during symmetric, in plane tasks such as a squat. While flexion/ extension of the knee along with rotation was utilized in both H2L and L2H tasks, significant differences were mainly noted in the high button push of the L2H, not the H2L. This, along with consistent knee torques and muscle activity independent of the task, identifies the TTT variables analyzed are driven by the crossover, not the flexion/extension of the knee. Crossover activities are common in many ADLs, thus being able to identify strategies used during this type of task could further aid in the recovery and rehabilitation time of the TKR patient. Along with patient rehabilitation and functional satisfaction outcomes, wear on the TKR components is a significant cause for TKR revision. By identifying compensatory strategies TKR design could be altered for known kinematic anomalies utilized by TKR patients.

This study had a variety of limitations that have to be considered. The subjects in this study were very active and reported no knee instability. A power analysis was calculated for this study that reported 10 TKR subjects were necessary for 80% power. Although we met this criterion, an increase in number of subjects or altered recruitment criteria would have potentially captured TKR subjects that do have issues with their TKR. The study also had a BMI restriction on the subjects which may have not captured the main population receiving TKR.

When modeling in OpenSim a variety of limitations arose that has to be considered when interpreting the data. The knee in the model only had one degree of freedom, thus limiting our ability to fully analyze the rotational movement of the knee. The knee also did not have TKR components modeled into OpenSim, thus creating the same femoral and tibial geometry for all subjects. Lastly, EMG was not reported in these studies, although we believe the muscular predictions identified are accurate, EMG data would have had to be recorded to verify.

The TTT study was one task of two performed to better understand rotary instability of the knee after TKR. The second task was performed by descending a flight of stairs and then did a cross over step. Kinematic and kinetic analysis has been reported on these data [106], yet modeling the data in OpenSim may be useful to determine muscular compensatory components during that task. Data were also collected on six individuals with self-reported instability of the knee during both tasks. Further research into those subjects and how they compare to the healthy and TKR patients could be very interesting in understanding how rotary tasks are affected by muscle activity and knee pathologies. Lastly, EMG data on one TKR subject was collected and could be compared to muscle firing predicted in these tasks.

Chapter 6: References

1. Statistics, C.f.D.C.N.C.f.H. *FastStats: inpatient surgery.* 2012 September 28, 2012]; Available from: http://www.cdc.gov/nchs/fastats/insurg.htm
2. Healy, W.L., S. Sharma, B. Schwartz, and R. Iorio, *Athletic activity after total joint arthroplasty.* J Bone Joint Surg Am, 2008. **90**(10): p. 2245-52.
3. Baker, P.N., J.H. van der Meulen, J. Lewsey, P.J. Gregg, E. National Joint Registry for, and Wales, *The role of pain and function in determining patient satisfaction after total knee replacement. Data from the National Joint Registry for England and Wales.* J Bone Joint Surg Br, 2007. **89**(7): p. 893-900.
4. Weiss, J.M., P.C. Noble, M.A. Conditt, H.W. Kohl, S. Roberts, K.F. Cook, M.J. Gordon, and K.B. Mathis, *What functional activities are important to patients with knee replacements?* Clin Orthop Relat Res, 2002(404): p. 172-88.
5. Woolhead, G.M., J.L. Donovan, and P.A. Dieppe, *Outcomes of total knee replacement: a qualitative study.* Rheumatology (Oxford), 2005. **44**(8): p. 1032-7.
6. Eisler, T., O. Svensson, A. Tengstrom, and E. Elmstedt, *Patient expectation and satisfaction in revision total hip arthroplasty.* J Arthroplasty, 2002. **17**(4): p. 457-62.
7. Mahomed, N.N., M.H. Liang, E.F. Cook, L.H. Daltroy, P.R. Fortin, A.H. Fossel, and J.N. Katz, *The importance of patient expectations in predicting functional outcomes after total joint arthroplasty.* J Rheumatol, 2002. **29**(6): p. 1273-9.
8. Mancuso, C.A., T.P. Sculco, T.L. Wickiewicz, E.C. Jones, L. Robbins, R.F. Warren, and P. Williams-Russo, *Patients' expectations of knee surgery.* J Bone Joint Surg Am, 2001. **83-A**(7): p. 1005-12.
9. Nilsdotter, A.K., S. Toksvig-Larsen, and E.M. Roos, *Knee arthroplasty: are patients' expectations fulfilled? A prospective study of pain and function in 102 patients with 5-year follow-up.* Acta Orthop, 2009. **80**(1): p. 55-61.
10. Zeni, J.A., Jr. and L. Snyder-Mackler, *Preoperative predictors of persistent impairments during stair ascent and descent after total knee arthroplasty.* J Bone Joint Surg Am, 2010. **92**(5): p. 1130-6.
11. Benedetti, M.G., F. Catani, T.W. Bilotta, M. Marcacci, E. Mariani, and S. Giannini, *Muscle activation pattern and gait biomechanics after total knee replacement.* Clin Biomech (Bristol, Avon), 2003. **18**(9): p. 871-6.
12. Farquhar, S. and L. Snyder-Mackler, *The Chitranjan Ranawat Award: The nonoperated knee predicts function 3 years after unilateral total knee arthroplasty.* Clin Orthop Relat Res, 2010. **468**(1): p. 37-44.
13. McMahon, M. and J.A. Block, *The risk of contralateral total knee arthroplasty after knee replacement for osteoarthritis.* J Rheumatol, 2003. **30**(8): p. 1822-4.
14. Gage, W.H., J.S. Frank, S.D. Prentice, and P. Stevenson, *Organization of postural responses following a rotational support surface perturbation, after TKA: sagittal plane rotations.* Gait Posture, 2007. **25**(1): p. 112-20.
15. Venema, D.M. and G.M. Karst, *Individuals with total knee arthroplasty demonstrate altered anticipatory postural adjustments compared with healthy control subjects.* J Geriatr Phys Ther, 2012. **35**(2): p. 62-71.
16. Worsley, P., M. Stokes, D. Barrett, and M. Taylor, *Joint loading asymmetries in knee replacement patients observed both pre- and six months post-operation.* Clin Biomech (Bristol, Avon), 2013. **28**(8): p. 892-7.

17. Fallah-Yakhdani, H.R., H. Abbasi-Bafghi, O.G. Meijer, S.M. Bruijn, N. van den Dikkenberg, M.G. Benedetti, and J.H. van Dieen, *Determinants of co-contraction during walking before and after arthroplasty for knee osteoarthritis.* Clin Biomech (Bristol, Avon), 2012. **27**(5): p. 485-94.
18. Chatterji, U., M.J. Ashworth, P.L. Lewis, and P.J. Dobson, *Effect of total knee arthroplasty on recreational and sporting activity.* ANZ J Surg, 2005. **75**(6): p. 405-8.
19. D'Lima, D.D., N. Steklov, S. Patil, and C.W. Colwell, Jr., *The Mark Coventry Award: in vivo knee forces during recreation and exercise after knee arthroplasty.* Clin Orthop Relat Res, 2008. **466**(11): p. 2605-11.
20. Wylde, V., A. Blom, P. Dieppe, S. Hewlett, and I. Learmonth, *Return to sport after joint replacement.* J Bone Joint Surg Br, 2008. **90**(7): p. 920-3.
21. Rossi, M.D., S. Hasson, M. Kohia, E. Pineda, and W. Bryan, *Mobility and perceived function after total knee arthroplasty.* J Arthroplasty, 2006. **21**(1): p. 6-12.
22. Pandy, M.G. and T.P. Andriacchi, *Muscle and joint function in human locomotion.* Annu Rev Biomed Eng, 2010. **12**: p. 401-33.
23. Dahlkvist, N.J., P. Mayo, and B.B. Seedhom, *Forces during squatting and rising from a deep squat.* Eng Med, 1982. **11**(2): p. 69-76.
24. Dionisio, V.C., G.L. Almeida, M. Duarte, and R.P. Hirata, *Kinematic, kinetic and EMG patterns during downward squatting.* J Electromyogr Kinesiol, 2008. **18**(1): p. 134-43.
25. Liu, W. and M.E. Maitland, *The effect of hamstring muscle compensation for anterior laxity in the ACL-deficient knee during gait.* J Biomech, 2000. **33**(7): p. 871-9.
26. Shelburne, K.B., M.R. Torry, and M.G. Pandy, *Muscle, ligament, and joint-contact forces at the knee during walking.* Med Sci Sports Exerc, 2005. **37**(11): p. 1948-56.
27. Gray, H., *Anatomy of the Human Body*. 2000, Philadelphia: Lea & Febiger.
28. Gage, W.H., J.S. Frank, S.D. Prentice, and P. Stevenson, *Postural responses following a rotational support surface perturbation, following knee joint replacement: frontal plane rotations.* Gait Posture, 2008. **27**(2): p. 286-93.
29. Hubley-Kozey, C.L., G.L. Hatfield, J.L. Astephen Wilson, and M.J. Dunbar, *Alterations in neuromuscular patterns between pre and one-year post-total knee arthroplasty.* Clin Biomech (Bristol, Avon), 2010. **25**(10): p. 995-1002.
30. Leffler, J., L. Scheys, T. Plante-Bordeneuve, B. Callewaert, L. Labey, J. Bellemans, and A. Franz, *Joint kinematics following bi-compartmental knee replacement during daily life motor tasks.* Gait Posture, 2012. **36**(3): p. 454-60.
31. Meier, W., R.L. Mizner, R.L. Marcus, L.E. Dibble, C. Peters, and P.C. Lastayo, *Total knee arthroplasty: muscle impairments, functional limitations, and recommended rehabilitation approaches.* J Orthop Sports Phys Ther, 2008. **38**(5): p. 246-56.
32. Stevens-Lapsley, J.E., J.E. Balter, W.M. Kohrt, and D.G. Eckhoff, *Quadriceps and hamstrings muscle dysfunction after total knee arthroplasty.* Clin Orthop Relat Res, 2010. **468**(9): p. 2460-8.
33. Thompson, J.A., A.M. Chaudhari, L.C. Schmitt, T.M. Best, and R.A. Siston, *Gluteus maximus and soleus compensate for simulated quadriceps atrophy and activation failure during walking.* J Biomech, 2013. **46**(13): p. 2165-72.
34. Hurley, M.V. and D.J. Newham, *The influence of arthrogenous muscle inhibition on quadriceps rehabilitation of patients with early, unilateral osteoarthritic knees.* Br J Rheumatol, 1993. **32**(2): p. 127-31.

35. Mizner, R.L., S.C. Petterson, J.E. Stevens, M.J. Axe, and L. Snyder-Mackler, *Preoperative quadriceps strength predicts functional ability one year after total knee arthroplasty.* J Rheumatol, 2005. **32**(8): p. 1533-9.
36. Silva, M., E.F. Shepherd, W.O. Jackson, J.A. Pratt, C.D. McClung, and T.P. Schmalzried, *Knee strength after total knee arthroplasty.* J Arthroplasty, 2003. **18**(5): p. 605-11.
37. Walsh, M., L.J. Woodhouse, S.G. Thomas, and E. Finch, *Physical impairments and functional limitations: a comparison of individuals 1 year after total knee arthroplasty with control subjects.* Phys Ther, 1998. **78**(3): p. 248-58.
38. Neptune, R.R., F.E. Zajac, and S.A. Kautz, *Muscle force redistributes segmental power for body progression during walking.* Gait Posture, 2004. **19**(2): p. 194-205.
39. Li, K., D.C. Ackland, J.A. McClelland, K.E. Webster, J.A. Feller, R. de Steiger, and M.G. Pandy, *Trunk muscle action compensates for reduced quadriceps force during walking after total knee arthroplasty.* Gait Posture, 2013. **38**(1): p. 79-85.
40. Kramers-de Quervain, I.A., E. Stussi, R. Muller, T. Drobny, U. Munzinger, and N. Gschwend, *Quantitative gait analysis after bilateral total knee arthroplasty with two different systems within each subject.* J Arthroplasty, 1997. **12**(2): p. 168-79.
41. McClelland, J.A., K.E. Webster, J.A. Feller, and H.B. Menz, *Knee kinetics during walking at different speeds in people who have undergone total knee replacement.* Gait Posture, 2010. **32**(2): p. 205-10.
42. Simon, S.R., H.W. Trieshmann, R.G. Burdett, F.C. Ewald, and C.B. Sledge, *Quantitative gait analysis after total knee arthroplasty for monarticular degenerative arthritis.* J Bone Joint Surg Am, 1983. **65**(5): p. 605-13.
43. Byrne, J.M. and S.D. Prentice, *Swing phase kinetics and kinematics of knee replacement patients during obstacle avoidance.* Gait Posture, 2003. **18**(1): p. 95-104.
44. Heller, M.O., G. Bergmann, G. Deuretzbacher, L. Durselen, M. Pohl, L. Claes, N.P. Haas, and G.N. Duda, *Musculo-skeletal loading conditions at the hip during walking and stair climbing.* J Biomech, 2001. **34**(7): p. 883-93.
45. Draganich, L.F., J.B. Whitehurst, L.S. Chou, G.A. Piotrowski, L.A. Pottenger, and H.A. Finn, *The effects of the rotating-hinge total knee replacement on gait and stair stepping.* J Arthroplasty, 1999. **14**(6): p. 743-55.
46. Stevens, J.E., R.L. Mizner, and L. Snyder-Mackler, *Quadriceps strength and volitional activation before and after total knee arthroplasty for osteoarthritis.* J Orthop Res, 2003. **21**(5): p. 775-9.
47. Mizner, R.L., J.E. Stevens, and L. Snyder-Mackler, *Voluntary activation and decreased force production of the quadriceps femoris muscle after total knee arthroplasty.* Phys Ther, 2003. **83**(4): p. 359-65.
48. Andriacchi, T.P., *Functional analysis of pre and post-knee surgery: total knee arthroplasty and ACL reconstruction.* J Biomech Eng, 1993. **115**(4B): p. 575-81.
49. Natarajan, R.N., G.B. Andersson, A.G. Patwardhan, and T.P. Andriacchi, *Study on effect of graded facetectomy on change in lumbar motion segment torsional flexibility using three-dimensional continuum contact representation for facet joints.* J Biomech Eng, 1999. **121**(2): p. 215-21.
50. Crossley, K.M., T.W. Dorn, H. Ozturk, J. van den Noort, A.G. Schache, and M.G. Pandy, *Altered hip muscle forces during gait in people with patellofemoral osteoarthritis.* Osteoarthritis Cartilage, 2012. **20**(11): p. 1243-9.

51. Sharma, L., D.D. Dunlop, S. Cahue, J. Song, and K.W. Hayes, *Quadriceps strength and osteoarthritis progression in malaligned and lax knees.* Ann Intern Med, 2003. **138**(8): p. 613-9.
52. Hurley, M.V., *The role of muscle weakness in the pathogenesis of osteoarthritis.* Rheum Dis Clin North Am, 1999. **25**(2): p. 283-98, vi.
53. Lewek, M.D., K.S. Rudolph, and L. Snyder-Mackler, *Control of frontal plane knee laxity during gait in patients with medial compartment knee osteoarthritis.* Osteoarthritis Cartilage, 2004. **12**(9): p. 745-51.
54. Hubley-Kozey, C.L., K.J. Deluzio, S.C. Landry, J.S. McNutt, and W.D. Stanish, *Neuromuscular alterations during walking in persons with moderate knee osteoarthritis.* J Electromyogr Kinesiol, 2006. **16**(4): p. 365-78.
55. Hubley-Kozey, C.L., N.A. Hill, D.J. Rutherford, M.J. Dunbar, and W.D. Stanish, *Co-activation differences in lower limb muscles between asymptomatic controls and those with varying degrees of knee osteoarthritis during walking.* Clin Biomech (Bristol, Avon), 2009. **24**(5): p. 407-14.
56. Hortobagyi, T., L. Westerkamp, S. Beam, J. Moody, J. Garry, D. Holbert, and P. DeVita, *Altered hamstring-quadriceps muscle balance in patients with knee osteoarthritis.* Clin Biomech (Bristol, Avon), 2005. **20**(1): p. 97-104.
57. Benedetti, M.G., P. Bonato, F. Catani, T. D'Alessio, M. Knaflitz, M. Marcacci, and L. Simoncini, *Myoelectric activation pattern during gait in total knee replacement: relationship with kinematics, kinetics, and clinical outcome.* IEEE Trans Rehabil Eng, 1999. **7**(2): p. 140-9.
58. Briem, K., D.K. Ramsey, W. Newcomb, K.S. Rudolph, and L. Snyder-Mackler, *Effects of the amount of valgus correction for medial compartment knee osteoarthritis on clinical outcome, knee kinetics and muscle co-contraction after opening wedge high tibial osteotomy.* J Orthop Res, 2007. **25**(3): p. 311-8.
59. Childs, J.D., P.J. Sparto, G.K. Fitzgerald, M. Bizzini, and J.J. Irrgang, *Alterations in lower extremity movement and muscle activation patterns in individuals with knee osteoarthritis.* Clin Biomech (Bristol, Avon), 2004. **19**(1): p. 44-9.
60. Heiden, T.L., D.G. Lloyd, and T.R. Ackland, *Knee joint kinematics, kinetics and muscle co-contraction in knee osteoarthritis patient gait.* Clin Biomech (Bristol, Avon), 2009. **24**(10): p. 833-41.
61. Lewek, M.D., T.L. Chmielewski, M.A. Risberg, and L. Snyder-Mackler, *Dynamic knee stability after anterior cruciate ligament rupture.* Exerc Sport Sci Rev, 2003. **31**(4): p. 195-200.
62. Rudolph, K.S., M.J. Axe, T.S. Buchanan, J.P. Scholz, and L. Snyder-Mackler, *Dynamic stability in the anterior cruciate ligament deficient knee.* Knee Surg Sports Traumatol Arthrosc, 2001. **9**(2): p. 62-71.
63. Schmitt, L.C. and K.S. Rudolph, *Influences on knee movement strategies during walking in persons with medial knee osteoarthritis.* Arthritis Rheum, 2007. **57**(6): p. 1018-26.
64. Zeni, J.A., K. Rudolph, and J.S. Higginson, *Alterations in quadriceps and hamstrings coordination in persons with medial compartment knee osteoarthritis.* J Electromyogr Kinesiol, 2010. **20**(1): p. 148-54.
65. Pedersen, D.R., R.A. Brand, and D.T. Davy, *Pelvic muscle and acetabular contact forces during gait.* J Biomech, 1997. **30**(9): p. 959-65.

66. Winter, D.A., *Biomechanics and motor control of human movment*. 3 ed. 2005, Hoboken, NJ: John Wiley & Sons.
67. Banks, S.A., M.K. Harman, J. Bellemans, and W.A. Hodge, *Making sense of knee arthroplasty kinematics: news you can use.* J Bone Joint Surg Am, 2003. **85-A Suppl 4**: p. 64-72.
68. McGinty, G., J.J. Irrgang, and D. Pezzullo, *Biomechanical considerations for rehabilitation of the knee.* Clin Biomech (Bristol, Avon), 2000. **15**(3): p. 160-6.
69. Dorr, L.D., J.L. Ochsner, J. Gronley, and J. Perry, *Functional comparison of posterior cruciate-retained versus cruciate-sacrificed total knee arthroplasty.* Clin Orthop Relat Res, 1988(236): p. 36-43.
70. Minns, R.J., *The role of gait analysis in the management of the knee.* Knee, 2005. **12**(3): p. 157-62.
71. Andriacchi, T.P., J.O. Galante, and R.W. Fermier, *The influence of total knee-replacement design on walking and stair-climbing.* J Bone Joint Surg Am, 1982. **64**(9): p. 1328-35.
72. Yoshida, Y., R.L. Mizner, D.K. Ramsey, and L. Snyder-Mackler, *Examining outcomes from total knee arthroplasty and the relationship between quadriceps strength and knee function over time.* Clin Biomech (Bristol, Avon), 2008. **23**(3): p. 320-8.
73. Mundermann, A., C.O. Dyrby, D.D. D'Lima, C.W. Colwell, Jr., and T.P. Andriacchi, *In vivo knee loading characteristics during activities of daily living as measured by an instrumented total knee replacement.* J Orthop Res, 2008. **26**(9): p. 1167-72.
74. Mizner, R.L. and L. Snyder-Mackler, *Altered loading during walking and sit-to-stand is affected by quadriceps weakness after total knee arthroplasty.* J Orthop Res, 2005. **23**(5): p. 1083-90.
75. Kutzner, I., B. Heinlein, F. Graichen, A. Bender, A. Rohlmann, A. Halder, A. Beier, and G. Bergmann, *Loading of the knee joint during activities of daily living measured in vivo in five subjects.* J Biomech, 2010. **43**(11): p. 2164-73.
76. Milner, C.E., *Interlimb asymmetry during walking following unilateral total knee arthroplasty.* Gait Posture, 2008. **28**(1): p. 69-73.
77. Callaghan, J.J., R. O'Rourke M, and K.J. Saleh, *Why knees fail: lessons learned.* J Arthroplasty, 2004. **19**(4 Suppl 1): p. 31-4.
78. Parratte, S. and M.W. Pagnano, *Instability after total knee arthroplasty.* Instr Course Lect, 2008. **57**: p. 295-304.
79. Vince, K.G., A. Abdeen, and T. Sugimori, *The unstable total knee arthroplasty: causes and cures.* J Arthroplasty, 2006. **21**(4 Suppl 1): p. 44-9.
80. Bach, B.R., Jr., S. Tradonsky, J. Bojchuk, M.E. Levy, C.A. Bush-Joseph, and N.H. Khan, *Arthroscopically assisted anterior cruciate ligament reconstruction using patellar tendon autograft. Five- to nine-year follow-up evaluation.* Am J Sports Med, 1998. **26**(1): p. 20-9.
81. Daniel, D.M., M.L. Stone, B.E. Dobson, D.C. Fithian, D.J. Rossman, and K.R. Kaufman, *Fate of the ACL-injured patient. A prospective outcome study.* Am J Sports Med, 1994. **22**(5): p. 632-44.
82. Tashiro, Y., K. Okazaki, H. Miura, S. Matsuda, T. Yasunaga, M. Hashizume, Y. Nakanishi, and Y. Iwamoto, *Quantitative assessment of rotatory instability after anterior cruciate ligament reconstruction.* Am J Sports Med, 2009. **37**(5): p. 909-16.

83. Tashman, S., D. Collon, K. Anderson, P. Kolowich, and W. Anderst, *Abnormal rotational knee motion during running after anterior cruciate ligament reconstruction.* Am J Sports Med, 2004. **32**(4): p. 975-83.
84. Brandsson, S., J. Karlsson, L. Sward, J. Kartus, B.I. Eriksson, and J. Karrholm, *Kinematics and laxity of the knee joint after anterior cruciate ligament reconstruction: pre- and postoperative radiostereometric studies.* Am J Sports Med, 2002. **30**(3): p. 361-7.
85. Georgoulis, A.D., S. Ristanis, V. Chouliaras, C. Moraiti, and N. Stergiou, *Tibial rotation is not restored after ACL reconstruction with a hamstring graft.* Clin Orthop Relat Res, 2007. **454**: p. 89-94.
86. Loh, J.C., Y. Fukuda, E. Tsuda, R.J. Steadman, F.H. Fu, and S.L. Woo, *Knee stability and graft function following anterior cruciate ligament reconstruction: Comparison between 11 o'clock and 10 o'clock femoral tunnel placement. 2002 Richard O'Connor Award paper.* Arthroscopy, 2003. **19**(3): p. 297-304.
87. Ristanis, S., G. Giakas, C.D. Papageorgiou, T. Moraiti, N. Stergiou, and A.D. Georgoulis, *The effects of anterior cruciate ligament reconstruction on tibial rotation during pivoting after descending stairs.* Knee Surg Sports Traumatol Arthrosc, 2003. **11**(6): p. 360-5.
88. Woo, S.L., A. Kanamori, J. Zeminski, M. Yagi, C. Papageorgiou, and F.H. Fu, *The effectiveness of reconstruction of the anterior cruciate ligament with hamstrings and patellar tendon . A cadaveric study comparing anterior tibial and rotational loads.* J Bone Joint Surg Am, 2002. **84-A**(6): p. 907-14.
89. Arneja, S. and J. Leith, *Review article: Validity of the KT-1000 knee ligament arthrometer.* J Orthop Surg (Hong Kong), 2009. **17**(1): p. 77-9.
90. Kocher, M.S., J.R. Steadman, K.K. Briggs, W.I. Sterett, and R.J. Hawkins, *Relationships between objective assessment of ligament stability and subjective assessment of symptoms and function after anterior cruciate ligament reconstruction.* Am J Sports Med, 2004. **32**(3): p. 629-34.
91. Nakajima, H., M. Kondo, H. Kurosawa, and T. Fukubayashi, *Insufficiency of the anterior cruciate ligament. Review of our 118 cases.* Arch Orthop Trauma Surg, 1979. **95**(4): p. 233-40.
92. Okazaki, K., H. Miura, S. Matsuda, T. Yasunaga, H. Nakashima, K. Konishi, Y. Iwamoto, and M. Hashizume, *Assessment of anterolateral rotatory instability in the anterior cruciate ligament-deficient knee using an open magnetic resonance imaging system.* Am J Sports Med, 2007. **35**(7): p. 1091-7.
93. Gardner-Morse, M.G. and I.A. Stokes, *Trunk stiffness increases with steady-state effort.* J Biomech, 2001. **34**(4): p. 457-63.
94. Schmitt, L.C. and K.S. Rudolph, *Muscle stabilization strategies in people with medial knee osteoarthritis: the effect of instability.* J Orthop Res, 2008. **26**(9): p. 1180-5.
95. Mandeville, D., L.R. Osternig, and L.S. Chou, *The effect of total knee replacement surgery on gait stability.* Gait Posture, 2008. **27**(1): p. 103-9.
96. Delp, S.L., F.C. Anderson, A.S. Arnold, P. Loan, A. Habib, C.T. John, E. Guendelman, and D.G. Thelen, *OpenSim: open-source software to create and analyze dynamic simulations of movement.* IEEE Trans Biomed Eng, 2007. **54**(11): p. 1940-50.
97. Research, N.C.f.S.i.R. *OpenSim*. 2010 [cited 2013; David Delp:[Available from: http://opensim.stanford.edu/.
98. Au, C. *Musculoskeletal Models*. March 5, 2014; Available from: http://simtk-confluence.stanford.edu:8080/display/OpenSim/Musculoskeletal+Models.

99. Delp, S.L., J.P. Loan, M.G. Hoy, F.E. Zajac, E.L. Topp, and J.M. Rosen, *An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures.* IEEE Trans Biomed Eng, 1990. **37**(8): p. 757-67.
100. Anderson, F.C. and M.G. Pandy, *A Dynamic Optimization Solution for Vertical Jumping in Three Dimensions.* Comput Methods Biomech Biomed Engin, 1999. **2**(3): p. 201-231.
101. Yamaguchi, G.T. and F.E. Zajac, *A planar model of the knee joint to characterize the knee extensor mechanism.* J Biomech, 1989. **22**(1): p. 1-10.
102. Inman, V.T., *The Joints of the Ankle*. 1976, Baltimore: Williams & Wilkins.
103. Au, C. *Gait 2392 and 2354 Models*. [ ] August 9, 2013; Available from: http://simtk-confluence.stanford.edu:8080/display/OpenSim/Gait+2392+and+2354+Models.
104. Dorn, T., *Gait Extract Toolbox for Matlab*, 2008: simtk.org.
105. *OpenSim User's Guide*, in *Release 2.4*April 4, 2012, Sim TK.
106. Denney, L.M., L.A. Ferris, H. Dai, and L.P. Maletsky, *Analysis of a rotary task following total knee arthroplasty: Stair descent with a cross-over turn.* Proc Inst Mech Eng H, 2014.

Appendix A

This appendix contains the manuscript published for Chapter 3.

Strategies Utilized to Transfer Weight During Knee Flexion and Extension With Rotation for Individuals With a Total Knee Replacement

Lauren A. Ferris
Department of Mechanical Engineering,
University of Kansas,
Lawrence, Kansas 66045
e-mail: lferris@ku.edu

Linda M. Denney
Department of Physical Therapy
and Rehabilitation Science,
University of Kansas Medical Center,
Kansas City, Kansas 66105
e-mail: ldenney@kumc.edu

Lorin P. Maletsky
Department of Mechanical Engineering,
University of Kansas,
Lawrence, Kansas 66045
e-mail: maletsky@ku.edu

Functional activities in daily life can require squatting and shifting body weight during transverse plane rotations. Stability of the knee can be challenging for people with a total knee replacement (TKR) due to reduced proprioception, nonconforming articular geometry, muscle strength, and soft tissue weakness. The objective of this study was to identify strategies utilized by individuals with TKR in double-stance transferring load during rotation and flexion. Twenty-three subjects were recruited for this study: 11 TKR subjects (age: 65 ± 6 years; BMI 27.4 ± 4.1) and 12 healthy subjects (age: 63 ± 7; BMI 24.6 ± 3.8). Each subject completed a novel crossover button push task where rotation, flexion, and extension of the knee were utilized. Each subject performed two crossover reaching tasks where the subject used the opposite hand to cross over their body and press a button next to either their shoulder (high) or knee (low), then switched hands and rotated to press the opposite button, either low or high. The two tasks related to the order they pressed the buttons while crossing over, either low-to-high (L2H) or high-to-low (H2L). Force platforms measured ground reaction forces under each foot, which were then converted to lead force ratios (LFRs) based on the total force. Knee flexion angles were also measured. No statistical differences were found in the LFRs during the H2L and L2H tasks for the different groups, although differences in the variation of the loading within subjects were noted. A significant difference was found between healthy and unaffected knee angles and a strong trend between healthy and affected subject's knee angles in both H2L and L2H tasks. Large variations in the LFR at mid-task in the TKR subjects suggested possible difficulties in maintaining positional stability during these tasks. The TKR subjects maintained more of an extended knee, which is a consistent quadriceps avoidance strategy seen by other researchers in different tasks. These outcomes suggest that individuals with a TKR utilize strategies, such as keeping an extended knee, to achieve rotary tasks during knee flexion and extension. Repeated compensatory movements could result in forces that may cause difficulty over time in the hip joints or low back. Early identification of these strategies could improve TKR success and the return to activities of daily living that involve flexion and rotation. [DOI: 10.1115/1.4023385]

Keywords: rotary instability, total knee replacement, compensatory strategy, weight transfer

Introduction

Over 600,000 total knee replacement (TKR) surgeries are performed annually [1] and are predicted to rise 673% by the year 2030, with TKR revisions rising by 601% [2]. Due to the increasing physical activity of the aging population, total knee surgery has become common in individuals approaching sixty years old. This population has expectations of maintaining an active lifestyle into their retirement and, therefore, expects positive outcomes after surgery. Expectations will include activities of daily living (ADLs) such as doing laundry, loading the dishwasher, or lifting an item, or higher expectations such as returning to play sports; for example, golf or tennis. In order to meet these expectations, a high axial load needs to be transferred to both legs. Both ADLs and sports require a high axial load with rotary motion at the knee while transferring weight to one leg and maintaining double-stance. Therefore, it is important to explore the manner in which individuals with TKR perform rotary tasks that involve flexion and extension at the knee while transferring weight.

Recommendations for returning to activity post-TKR are varied, considering the risk of imbalanced or excessive implant loading, aseptic loosening, and risk of injury due to a feeling of instability [2]. Most individuals are satisfied with reduced pain and increased function following surgery; however, many assume movement patterns that produce asymmetrical loading to the prosthesis or transfer loading forces to other joints in order to compensate. Improper loading of the knee could lead to an imbalance in the medial and lateral soft tissue structures and wear on the prosthetic device. This imbalance may lead to malalignment or deformity and, eventually, failure of the replacement [3]. The patient may complain of a feeling of "instability" or "weakness" and limit the amount of weight placed on the surgical leg. Instability is one of the most common reasons for TKR revision [4–6] and yet, is poorly defined.

Contributed by the Bioengineering Division of ASME for publication in the Journal of Biomechanical Engineering. Manuscript received October 1, 2012; final manuscript received January 7, 2013; accepted manuscript posted January 18, 2013; published online February 7, 2013. Editor: Beth Winkelstein.

Clinical and functional stability are defined by different characteristics, but individuals that report a sensation of knee "buckling" or "giving way" are categorized as having knee instability. Clinical stability of the knee is measured by passive tests (both in the sagittal and transverse plane) associated with a subjective complaint, while functional stability may be described as a feeling of instability during an activity. Functional stability is difficult to measure due to the subjective nature of the definition but may be best characterized by measurements of the kinematics and the amount of excursion/translation at the knee joint while performing an activity [7–9]. Some researchers propose that the feeling of instability may be a result of altered kinematics of the knee; however, some individuals with knee instability do not realize compensatory movements at the knee and hip, even when challenged with negotiating obstacles. How is it that some respond well to challenging activities and others do not? Most studies analyze kinematic motion on level walking, ascending and descending stairs, and stepping over objects. The commonality of these activities is that they primarily occur in the sagittal plane. Byrne et al. [10] postulated that knee instability occurs in a medial and lateral direction, resulting in a "giving way" sensation, loss of balance, or fall. Therefore, it is difficult to determine if the feeling of instability is actually related to changes in kinematics or loading of the knee, based on the patient's expectation of a surgical outcome, or pain.

The bulk of the research for capturing loading and kinematics for individuals with TKR involve straight-forward gait, stairs, and sit to stand. Although these activities are important post-TKR, they lack movements in the transverse plane. Pregait strategies, pain, and/or gait velocity may influence knee loading post-TKR [11,12], while sit to stand performance is confounded by quadriceps strength resulting in weight bearing asymmetry [13]. Knee joint loading differs per activity and varies with change in the knee flexion angle. The load can be described over time, such as walking, or as a single point in time depicted in a squat or sit to stand. Although the load may vary with each of these activities, repetition affects the wear on the implant. It is important to identify strategies utilized by individuals with a TKR and relate those to the probable impact they have on the longevity of the implant. Knowledge of the forces on the knee joint during ADLs or recreational activities will also prove valuable for possible implant design and intervention in rehabilitation. Studies have utilized force transducers in the implant to collect specific knee loading [14–16]. Activities such as walking, squatting, stair ascent and descent, and rising from a chair exceed two times body weight (BW) [15]. Treadmill walking generated a lower force than normal level ground walking ($2.0 \times$ BW versus $2.6 \times$ BW, respectively) [17]. Although most studies recruit individuals closely matched to height and weight, it is recommended for any gait analysis to normalize for body weight in order to minimize confounding variables that may exist, including gender [18,19].

Vertical ground reaction force (GRF) is the most common load analysis utilizing force plates or a gait mat for the collection of data. Yoshida et al. [20] found a significant difference in the GRF at 3 months post-TKR. Individuals had less GRF on the operated leg compared to the contralateral leg while performing functional tasks, while Mundermann et al. [15] found a more symmetrical loading with the TKR limb compared to the unaffected leg during sit to stand. The amount of time post-surgery is the difference between these two studies. Mundermann [15] conducted the study 1.5 years after surgery, while Yoshida [20] found differences in the GRF between the surgical leg and the contralateral leg only at 3 months post-TKR but not at 12 months post-TKR. This indicates that if improvements are going to be made post-surgery, it may be reliant on muscle strength development, rehabilitation training, and the activity of the individual pre- and post-surgery. Mandeville et al. [21] supports the fact that individuals may have already developed a strategy presurgery with control of center of mass (COM) within base of support for level walking and over obstacles. Although the TKR individuals had a smaller COM displacement, slower gait, and shorter stride, the differences from the control group were unchanged when compared prior to surgery [21]. Hence, it is difficult to conclude that loading changes are a result of the TKR surgery; rather, the results may have been related to pain. The subjects may have demonstrated a more conservative movement due to pain at less than six months post-TKR.

Although most of the computational models and knee load analyses have been derived from gait cycles, knee loads during functional activities, such as squatting, should be considered since high-flexion is an area of concern for implant design. Squatting with rotation is a functional activity for someone getting in and out of the car or bathtub and transferring a load, such as picking up a child or a laundry basket. Despite the desire to return to recreational sports reported by TKR recipients [2,17,22,23], more consideration should be placed on ADLs that require movements in the transverse plane (flexion/extension with rotation). Our study utilizes a novel approach that facilitates the study of knee loading, rotation, and load transfer during transverse plane motion while in double stance. The objective of this study was to identify strategies utilized by individuals with a TKR while in double-stance transferring load during rotational activities.

Materials and Methods

Twenty-three subjects were recruited for this study: 11 TKR subjects (age: 65 ± 6 years; BMI 27.4 ± 4.1) and 12 healthy subjects (age: 63 ± 7; BMI 24.6 ± 3.8) (see Table 1). All subjects were within 50-75 years of age and excluded subjects with previous hip or ankle surgery, peripheral neuropathy, or a BMI greater than 29. The TKR subjects were further screened for individuals who could flex both knees at least 90 deg and had a unilateral TKR. Subjects were not screened based on the type of knee implant and the extent of participation in rehabilitation.

Each subject signed a consent form approved by the Human Subjects Committee and Institutional Review Board at the University of Kansas Medical Center and approval to be videotaped and photographed. Subjects wore tight-fitting athletic clothing with no reflective material on their clothes or shoes. Anthropometric measurements and bony landmark measurements were collected to configure the computer model and kinematic data using Vicon's Workstation (v 4.5) and the lower body Plug-in Gait Model (Oxford Metrics, Oxford, UK) (see Table 1). Goniometric measurements of the hip and knee's range of motion and manual muscle test of both leg's quadriceps strength (MicroFET2, Hoggan Health Industry, West Jordan, Utah) were collected and recorded for each subject (see Table 1). A KT-1000 measured the anterior translation of the tibio-femoral joint in both knees (see Table 1).

Twenty-four reflective markers (25 mm in diameter) were applied with adhesive to specific bony landmarks and secured with tape. The markers were captured at 120 Hz with an infrared six-camera motion analysis system (Vicon 512, Oxford Metrics,

Table 1 Average (standard deviation) of participant's measurements

	TKA		Healthy
Number of subjects	11		12
Age (years)	67 (6)		65 (8)
BMI	27.7 (4.0)		24.6 (3.8)
Gender	36% F		58% F
Post-op in months	19 (16)		N/A
	Unaffected	Affected	
Quad strength in % BW	0.17 (0.04)	0.19 (0.05)	0.19 (0.03)
KT-1000 (mm)	2.6 (1.0)	2.9 (1.0)	2.4 (0.9)
Knee RoM (deg)	137.9 (6.3)[a]	126.7 (7.6)[a,b]	141.2 (5.7)[b]
Hip RoM (deg)	111.6 (8.3)	107.0 (7.6)	109.1 (10.8)

[a]Affected statistically significant ($p < 0.05$) from unaffected.
[b]Affected statistically significant ($p < 0.05$) from healthy.

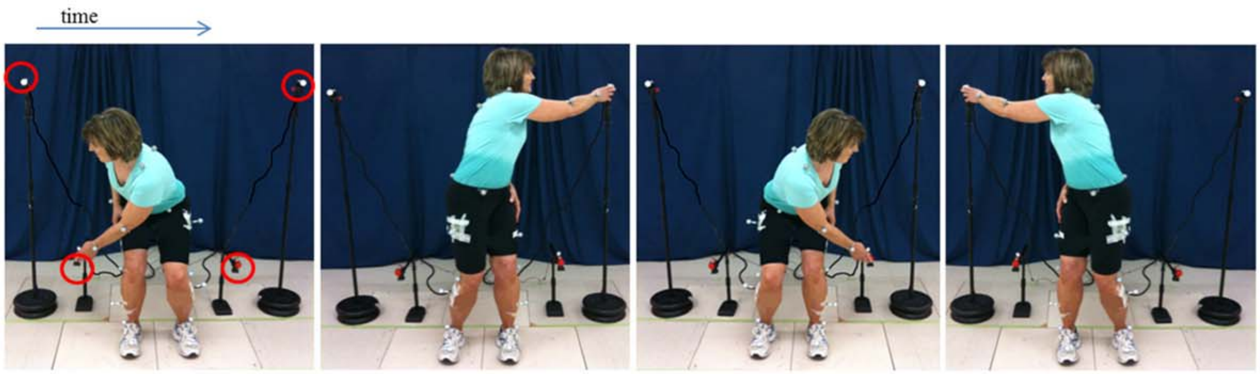

Fig. 1 Equipment set up. Subject performing a low to high (L2H) sequence. The subject performs this sequence three times. Button positions are indicated with circles.

Oxford, UK). The knee axis was determined by a knee alignment device (KAD) that identifies the *x*, *y*, and *z* configuration of the knee joint [24]. In an accuracy study for upper body range of motion by Henmi et al., the Vicon 512 motion tracker had a reliability of $\leq$3 deg [25]. Since the accuracy of the angle measurements are reliant on the placement of the KAD, it can be concluded that the KAD is as reliable as the motion tracker. The Vicon was calibrated according to the manufacturer's specifications. Two force plates (AMTI, Advanced Mechanical Technology, Inc., Watertown, MA) embedded in the floor were used to capture the GRF at 360 Hz. Force plate and analog devices were wired into the Vicon's A/D board and simultaneously triggered by the Workstation software.

A reach test was performed where the subject stood 20 cm in front of a white board and the centerline was referenced by the midline of the subject. The subject reached across his body with his right hand and placed a magnet at his maximum reach, then repeated the motion with the left hand. The distance from the centerline to the magnet was measured and recorded to configure the experimental set up.

For the target touch tasks (TTT) (see Fig. 1), two microphone stands were placed at the recorded reach test least maximum reach value from the centerline of the force plates in the frontal plane. The subject stood with one foot on each of the force plates. Two low buttons were positioned at knee height, just lateral to the knee. The distance from the knee was determined by the comfort level of the subject's ability to bend his knees, rotate to reach, and still press the button. Buttons were clipped to the microphone stands and wired so that when pressed, a confirmatory sound was produced. A static trial with KADs placed on the subject's medial and lateral femoral epicondyles was collected to configure the knee joints.

A description of one of the two tasks collected, either a high-to-low crossover sequence (H2L), or a low-to-high crossover sequence (L2H), was explained to the subject. The subject had the option to perform up to two practice trials. The subjects were instructed to keep their feet planted on the floor and use their legs while reaching to hit the low buttons. The H2L task was defined as when the subject used the opposite hand to cross over his body and hit the button next to his shoulder, switched hands, and crossed to hit the button next to the opposite knee (see Fig. 2). The subject then stood up, pressed the button on the same side by his shoulder with the opposite hand, and did a similar cross over to the other side. This sequence was performed three times at the subject's self-selected pace. Similarly, the L2H activities started with one hand crossing over and pressing the button near the opposite knee, then switched lead hands and pressed the button by the opposite shoulder. This cycle was also performed three times. At any time, if the subject pressed the wrong button, lifted his heels off the force plate, or performed the wrong sequence, a new trial was collected. After each task, the subject was questioned if anything was particularly difficult or challenging.

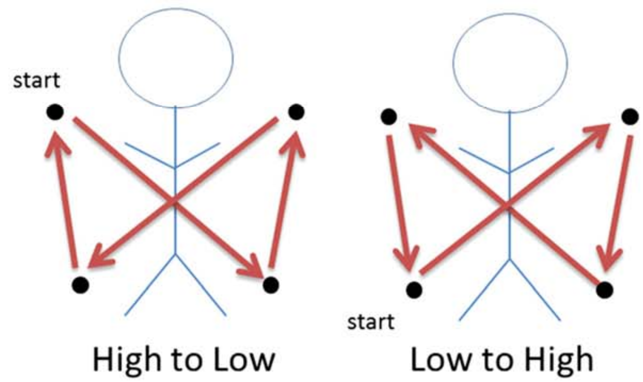

Fig. 2 Sequence of both high to low (H2L) and low to high (L2H) that the subjects performed

The subject's legs were categorized in one of three groups: (1) affected group, TKR leg; (2) unaffected group, the TKR subject's contralateral leg; or (3) the healthy group, the subjects who have had no previous knee injury or surgery. The GRFs were normalized to percent body weight and then according to the task, H2L or L2H. For each of the crossovers, a lead leg was defined as the leg the subject was leaning into on the crossover, or the leg closest to the second button of the sequence. Only five crossover trials were analyzed due to the data collection process; three with the left side leading, two with the right. The lead force ratio (LFR) was calculated by taking the lead leg's GRF divided by the total GRF, thus 0.5 LFR would be when the subject had equal distribution on the lead and lag legs. The trials were interpolated and normalized to a percent button-to-button movement: 0% being when the first button was pressed, 100% being when the second button was released. Any of the five cross over trials that lasted two times longer than the shortest one was eliminated from the data set. The LFR for the affected and unaffected, or healthy were averaged and the standard deviation was calculated for each subject at each percent button-to-button movement. The average and standard deviation of the LFR parameters for H2L and L2H was calculated for all subject groups and a single-factor analysis of variance (ANOVA) was performed ($p < 0.05$). The ANOVAs were performed at 10 and 90% of the button-to-button movement of the averaged data and at the time step of every trial when the subject crossed the 0.5 LFR. These positions (10%, 0.5 LFR, and 90%) were chosen to determine the difference between the three groups when the subject released the first button, the subject had an equal distribution of weight, and when the subject first hit the second button.

Knee angles for each subject were interpolated and cut into the five crossovers. The lead knee's angles were analyzed and time was normalized to percent button-to-button movement. Knee angles for the subjects' affected and unaffected, or healthy knees were averaged for all five trials at each percent of the button-to-button movement. The average and standard deviation of the knee angle parameters for H2L and L2H were calculated for all subject groups and a single factor ANOVA was performed ($p < 0.05$). A single-factor ANOVA was run on the knee angles of the subject at 0.5 LFR ($p < 0.05$). All ANOVAs were further analyzed using Tukey's procedure. To compare the maximum LFR and knee angles for the affected, unaffected, and healthy groups, an independent t-test and paired t-test were performed. Independent t-tests were performed using the affected leg of the TKR compared to the healthy control and then again using the unaffected leg of the TKR compared to the healthy control in order to compare the means between the two groups and separate knee conditions. A paired t-test was conducted when comparing the affected knee to the unaffected knee in the TKR group in order to compare the means of the knees within the same group.

Results

Loading patterns of the affected and unaffected legs during the H2L weight transfer were not statistically different throughout the task, with the unaffected leg transferring a slightly larger load where the maximum LFR for the unaffected leg was 0.71 and maximum LFR for the affected leg was 0.69 (see Table 2). For H2L, the healthy subjects started with less LFR on the lead leg (see Fig. 3), but finished the task with a similar loading to the TKR subjects. A large variation at early to mid-movement was evident with the affected knee (see Table 3). The healthy group also performed the H2L task by transferring their load later than the TKR patients since the healthy's 0.5 LFR occurred later in the button-to-button movement (see Fig. 3(*a*)). In contrast, the L2H LFR for the TKR subject's limbs generally were less than the healthy subject, but had the greatest variation (see Table 3). During mid-movement, the TKR affected leg had the least LFR and transferred their load more slowly than the unaffected and healthy legs (at 62% button-to-button movement compared to 54 and 53%, respectively) (see Fig. 3(*c*)). No statistical differences were found between the three groups at 0.5 LFR for both the H2L and L2H tasks (see Figs. 3(*a*) and 3(*c*)). No statistical difference was found between the groups for the LFR in both the H2L and L2H tasks, although there was a strong trend of statistical difference between the unaffected and healthy subjects during H2L ($p = 0.057$) (see Fig. 3(*a*)) and between the affected and unaffected subjects during L2H at maximum LFR ($p = 0.077$) (see Table 2).

Table 2 Unaffected, affected, and healthy subject's average and standard deviation for maximum lead force ratio and maximum knee angle across all knees. No significant differences were observed.

		Unaffected	Affected	Healthy
H2L	Max lead force ratio	0.71 (0.09)	0.69 (0.09)	0.64 (0.07)
	Max knee angle (deg)	28.03 (9.67)	26.69 (12.96)	38.18 (14.00)
L2H	Max lead force ratio	0.71 (0.09)	0.68 (0.09)	0.71 (0.06)
	Max knee angle (deg)	38.22 (12.98)	34.38 (19.15)	46.13 (16.01)

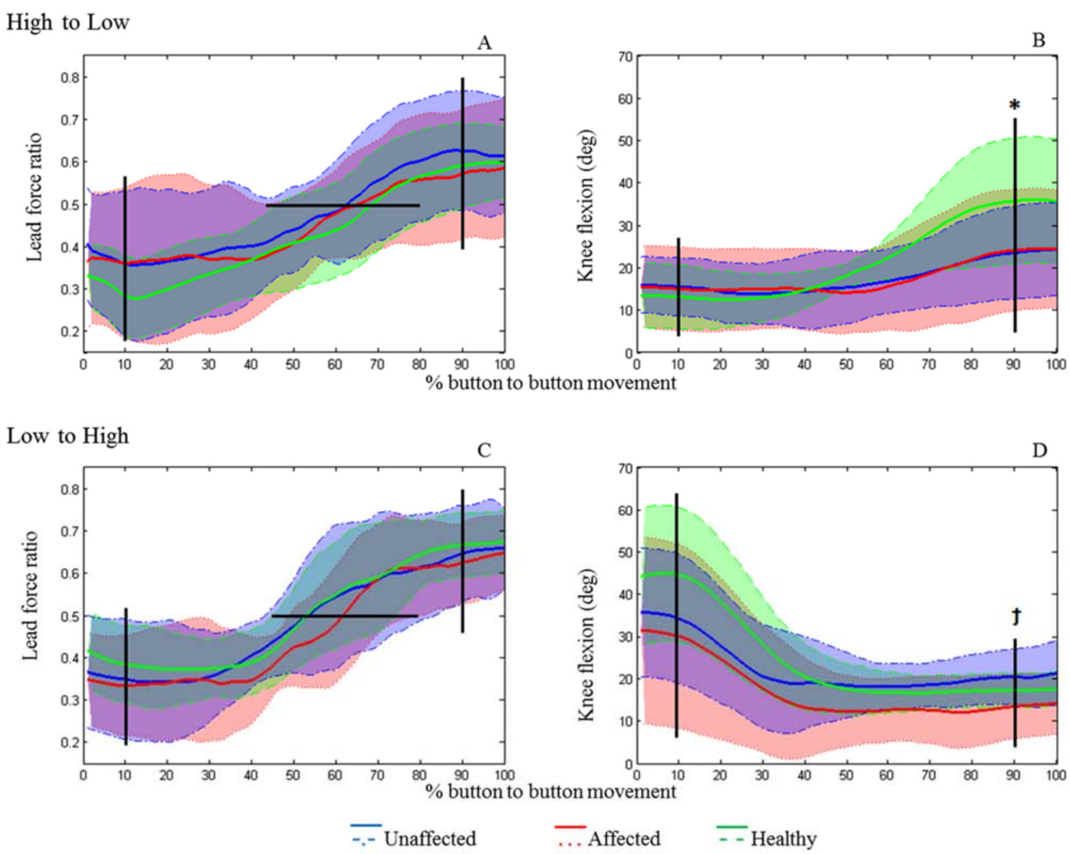

Fig. 3 **(*a*) Mean lead force ratio and (*b*) knee flexion angle throughout a H2L activity, and (*c*) mean lead force ratio and (*d*) knee flexion angle throughout a L2H activity. Shaded areas indicate a ±1 standard deviation. Black vertical lines indicate where the subject is at 10%, right after the first button push, and 90%, right before the second button push. The horizontal black line indicates when the subject has 0.5 LFR, or an equal distribution of weight. A single factor ANOVA was performed at each black line. (The * denotes unaffected statistically significant ($p < 0.05$) from healthy. The † denotes unaffected statistically significant ($p < 0.05$) from affected.)**

Table 3 Average standard deviations for unaffected, affected, and healthy subject groups while performing high to low (H2L) and low to high (L2H). Values reported for lead force ratio (LFR) and knee flexion angle at 10% and 90% button-to-button movement. The TKR subject values larger than the healthy are indicated in bold.

			Unaffected	Affected	Healthy
H2L	LFR	10% button-to-button	**0.045**	**0.039**	0.033
		90% button-to-button	**0.049**	0.031	0.045
	Angle	10% button-to-button	1.319	1.429	1.928
		90% button-to-button	**3.936**	2.847	3.082
L2H	LFR	10% button-to-button	0.028	0.023	0.050
		90% button-to-button	0.033	**0.040**	0.034
	Angle	10% button-to-button	2.382	2.423	3.234
		90% button-to-button	**1.855**	**2.526**	1.604

Table 4 Unaffected, affected, and healthy's averages (standard deviation) for all percent button-to-button movement that crossed 0.5 lead force ratio and knee angles at the percent the LFR crossed 0.5

		Unaffected	Affected	Healthy
H2L	Percent button-to-button cross 0.5 LFR	56.0 (13)	60.0 (11)	61.0 (18)
	Knee angle (deg) at percent button-to-button = 0.5 LFR	17.8 (8.8)[a]	14.7 (11.5)[a]	25.5 (9.3)
L2H	Percent button-to-button cross 0.5 LFR	60.0 (18)	61.0 (15)	57.0 (15)
	Knee angle (deg) at percent button-to-button = 0.5 LFR	16.5 (8.4)	13.6 (9.0)	17.3 (6.6)

[a]Statistically significant ($p < 0.05$) from healthy.

For both tasks the healthy subjects had a greater knee range of motion throughout the movement sequences (see Figs. 3(*b*) and 3(*d*)) and greater maximum knee flexion (see Table 2). The healthy and unaffected subjects also had a significantly greater goniometric premeasurement knee range of motion than the affected subjects (see Table 1). The TKR subjects tended to keep their legs more extended throughout the tasks, more specifically, during the H2L. While the healthy subjects had a total range of motion of 23.5 deg during the H2L, the unaffected and affected knees only moved a total of 10.5 deg and 10.3 deg, respectively. A significant difference was found between healthy and unaffected knee flexion angles at 90% button-to-button movement, or right before the second button push of the crossover (see Fig. 3(*b*)). Strong statistical trends were also observed at maximum knee flexion for the H2L task between the affected and healthy subjects ($p = 0.055$) and the unaffected and healthy subjects ($p = 0.058$) (see Table 2). A statistical difference in knee flexion was found between the healthy compared to the TKR where the subjects passed 0.5 LFR (Table 4). While the healthy subjects had an average knee angle of 25.5 deg (9.3) when they passed 0.5 LFR, the unaffected and affected knees were only flexed an average of 17.8 deg (8.8) and 14.7 deg (11.5), respectively.

The L2H tasks displayed the greatest amount of knee angle variation between the three groups (see Table 3). At 10% movement, there was a strong trend between the affected and healthy groups ($p = 0.07$), where the healthy subjects were very willing to bend knees to hit the lower buttons; the affected knees were more extended. A statistical difference was seen at 90% between the unaffected and affected limbs of the TKR subjects where the unaffected knee is more flexed throughout the L2H task.

Discussion

The ability to reach for objects within arm's reach while maintaining balance and stability is critical for performing ADLs in a safe manner. Added rotation with knee flexion and extension during this activity provides a challenge to individuals with a TKR [26]. Evidence of a large variation of load transfer for TKR individuals in our study suggests that positional stability of the knee is difficult when challenged during both flexion and extension activities involving rotation. The largest variation during mid-flexion supports previous studies showing instability at 30–60 deg of flexion without the added rotation [27,28]. It is also possible that cognitive planning is ongoing after movement initiation, evidenced by the large variation mid-task [29], a lower LFR, and slower transfer of force in the affected leg of the TKR participants. A delayed time to execute the task by loading the lead leg at a later point of time in the task was demonstrated by the healthy group during the H2L task and the TKR group during the L2H task. This is evident in Fig. 3 as the healthy group unloads the lead leg at the beginning of the H2L task (within the first 10% of movement) while the TKR participants during L2H maintain the same weight through the lead leg (regardless of affected or unaffected) before increasing the load in order to reach the next button. This weight shift for the TKR individuals began at approximately 40% of the total movement to complete this task. Although these differences occur during different cycles of the TTT (extension to flexion in the H2L and flexion to extension for L2H), there must be some reason for this altered timing in loading. A possible explanation is that the change in direction to touch the next button also incorporates a change in purposeful movement trajectory; therefore incorporating acceleration toward a target and the need for deceleration to complete the task. Corrections are needed during the movement in order to complete the task and the delay in loading the lead leg may be a result of cognitive preparation or the need for many corrections of trajectory motion [29]. The affected leg of the TKR group had a lower LFR and delayed loading during the L2H task mid-motion. This suggests the position of mid-flexion may create a need for trajectory correction or cognitive planning of how to achieve the requested task that includes knee flexion into extension while transitioning weight to the affected leg. Surprisingly, the TKR participants did not demonstrate much difference on loading during the H2L task, emphasizing equal distribution of leg loading during flexion tasks or the utilization of a different strategy to shift their center of mass to achieve a task moving into flexion with rotation.

The position of mid-flexion was avoided by the TKR subjects in our study who performed the tasks in more extended positions. The tendency to extend the knee may assist with a feeling of stability during loading and, therefore, the TKR group would be less likely to perform rotary tasks in a flexed knee position. Yoshida [20] found that individuals with a TKR demonstrated unloading of the affected limb up to 3–6 months before equally loading both extremities and continued to use a stiff knee gait pattern post-surgery despite an increase in quadriceps strength. This suggests that knee extension is a more "stable" position and the subjects utilize an alternate strategy to perform the task. Strategies may include loading or unloading the surgical leg. Loading differences could be a possible strategy since a difference was noted between the TKR and healthy group. No differences were found in hip flexion, hip tilt (in the frontal plane), and pelvis rotation between groups despite the fact that the TKR individuals achieved the low button push with a more extended knee than the healthy individuals. This may suggest that the movement to reach the lower button may be achieved by torso flexion (trunk on pelvis) with torso rotation, which would enable a more attainable reach with an extended knee. An analysis of data from the upper torso during the dynamic movements could be analyzed to determine if the TKR participants utilized their torso more than their lower extremities to perform dynamic rotational reaching tasks. These strategies need to be identified in order to alter rehabilitation intervention, avoid implant wear, and prevent future injuries. Compensation in movement contributes to implant wear and altered neuromuscular input (motor control). Early identification and correction of these strategies could improve TKR success and return to activities of daily living tasks that involve rotation.

Participants in this study with a TKR may represent a group that demonstrates alternate knee kinematic and kinetic strategies during rotational tasks. Maintaining a more extended knee and decreased loading of the surgical knee during rotational tasks are two examples of alternate knee strategies that were demonstrated in this study. This is consistent with the literature on obstacle avoidance and level-straight walking studies [7,10,30,31]. Due to the fact that no participant reported difficulty directly after each trial in each task, it is undetermined if any of these strategies are a result of knee instability. Furthermore, considering that conformity of the joint surfaces and anterior translation of the femur on the tibia were not directly measured in our study during the tasks, we can only determine altered knee kinematics, ground reaction, and predicted joint forces as probable strategies demonstrated in the rotary tasks. Although some would argue that the increased loading or altered knee angle may be due to the lack of proprioception and quadriceps control, studies are inconsistent in their findings to support either of these as single factors of activity performance [12,13,32]. Both weight acceptance and knee flexion excursion during gait were not significantly different in TKR participants at an average of 28 months post-surgery [33]. Furthermore, it is argued that the unaffected limb of the TKR participants should not be used as a comparison to the involved limb since both limbs are more symmetrical over time [33,34]. Although it is customary for clinicians and physicians to use the contralateral leg as a standard of "normal" for an individual when testing for range of motion and strength, the comparison of the surgical leg to the opposite leg on the individual should be avoided. Instead, the surgical leg should be compared to the mean of a healthy control group. This is due to the findings in previous studies along with our study that both knees of the individual who underwent surgery will display altered kinetics and kinematics. This may explain the similar performance of the participants in the current study during the load transfer in the H2L TTT and knee flexion during both tasks. Performance may be dependent on time post-surgery, with differences noted in knee flexion excursion at 3 months [35], yet progressing to symmetrical movements within the TKR subjects by 19 months in this current study and 28 months, as documented by Milner [33]. The possible abnormal symmetrical loading of both the involved and uninvolved limb may be the precursor to the progression of osteoarthritis that seems to be prevalent in the contralateral limb of TKR patients, requiring another TKR on the opposite limb [36]. Christiansen et al. [13] chose to examine weight bearing differences one month post-TKR and noted significant asymmetry; unloading the affected leg during a sit to stand activity. Follow-up of these same individuals demonstrated improved symmetry of weight bearing on both limbs at 3 months and was equal to the healthy control group by 6 months. In order to prevent abnormal loading or unloading of either leg following TKR surgery, early intervention addressing equal weight bearing should be a primary goal of post-operative rehabilitation, with the addition of an exercise including a rotary task.

The tendency to unload the affected leg was apparent in the TTT (see Table 2). The TTT may simulate movements such as reaching, getting out of a bathtub, or getting in or out of a car. Therefore, knowing that the TKR individual unloaded the affected extremity more than the unaffected leg provides a reason to incorporate single-leg weight bearing exercises or balance activities in rehabilitation. Rehabilitation intervention that includes biofeedback to promote symmetry in weight bearing during exercises has been suggested by McClelland et al. [37] and proved to produce outcomes similar to the healthy population. Training for symmetry was provided during sit to stand, gait, and balance activities; none of which included rotation as a primary motion. Recognizing this is a single case report, it is difficult to guarantee that practiced intervention will transfer to real life situations, especially due to the inconsistent patterns that are demonstrated in the literature. Individuals with a TKR demonstrated unloading of the affected limb up to 3–6 months before equally loading both extremities and continued to use a stiff knee gait pattern post-surgery despite an increase in quadriceps strength [20]. This is similar to our study, in which the participants demonstrate quick unloading at the beginning of the TTT and lack of knee flexion during squatting. The authors realize that the extent of post-operative rehabilitation with the TKR participants in this study could affect the willingness to put weight on the extremity (loading) and the ability to perform squatting techniques. However, that information was not available at the time of the study; therefore, it should be noted that large standard deviations in each task variable may be a result of individual activity level and/or a response to rehabilitation intervention or lack of intervention. It is also difficult to ascertain if our findings are an actual change due to surgical intervention since data (such as quadriceps strength and knee biomechanics) were not captured prior to surgery. Premeasurements would be beneficial to use as a comparison, however, this study was only conducted on individuals after surgery.

Compensatory movements have been documented as adjustments in kinetic and muscle activity that can affect other joints and influence the performance of functional activities [38]. In the current study, the healthy participants consistently flexed their knees more than the TKR participants throughout both the H2L and L2H tasks. Notably, the healthy group had twice as much knee flexion excursion when compared to both the affected and unaffected knees of the TKR group. At 0.5 of the LFR in H2L the healthy group had significantly greater knee flexion at approximately midline, or equal stance, on each leg. This suggests that the TKR group avoids the mid-flexion range in either knee, possibly due to a feeling of instability, even when standing with equal load on both legs. It is unknown if the TKR group avoided load on the affected leg when leading toward that leg or if they are unable to push off with the affected leg when leading toward the unaffected leg. During the movement from extension into flexion in the H2L task the healthy group flexed their knees 23 deg, while the TKR group flexed approximately 10 deg on either knee. Knee flexion range of motion is commonly referred to as knee excursion during gait analysis. Knee excursion can be compared between the surgical knee and contralateral knee in TKR individuals and may be reduced in the surgical knee for a period of time following surgery. Milner [33] found little difference between the involved knee (11 deg) and uninvolved knee (13 deg) for knee excursion during weight acceptance in individuals 28 months post-surgery, while Mizner and Snyder-Mackler [35] recorded 11 deg in the involved knee and 19 deg in the uninvolved knee at 3 months post-surgery. Although the calculations of knee excursion may be different for each of these studies, it demonstrates that the TKR knee remains in a more extended position up to and over 2 years post-op when performing functional tasks of ambulation and translational movements in the TTT. The added knowledge from the current study of knee biomechanics during a task that involves the transverse plane motion will help health professionals better understand the mechanics and response of the knee following a TKR. By utilizing a rotary task, a dynamic snapshot of both movement and loading of the knee were captured that would not be obtained by direct measurement of anterior-posterior translation in the sagittal plane, as reported by previous studies. The extended knee strategy leads to speculation of how the TKR participants in our study are able to reach the button near the knee in the H2L task where it has been shown that they demonstrate 10 deg knee excursion. Inevitably, torso motion has to be considered, despite the instruction given to each subject to use their legs while performing the crossover task. Collection of each participant's strategy to complete the TTT without cues from the investigator resulted in only one-time instruction at the beginning of the task and no correction of movement during the rest of the cycles. It appears that individuals in the TKR group selected to maintain a more extended knee on either the affected or unaffected knee and find a strategy to accomplish the lower button push. Shakoor et al. [39] recorded higher knee loading on the contralateral side of the individuals with a total hip replacement while walking and greater knee extension and abduction moments when compared to the ipsilateral side. Knowing that loading was transferred to another joint during ambulation 15 months post-surgery, it can be suggested that the TKR participants may have utilized upper torso rotation and altered joint loading to achieve the task of H2L.

Overall, it appears that TKR individuals utilize compensatory patterns, possibly at the torso, for lack of control at the knee. The difficulty of making such statements is that functional outcome measures have been correlated to presurgery status (such as quadriceps strength) [35]; therefore, caution must be taken in the interpretation of the study results due to the fact that we did not collect presurgical measurements. Cadaveric studies testing TKR implants may have difficulty simulating natural knee forces via a robotic or hydraulic device due to failure of the aged tissue. Although joint conformity is controlled, rotary stability in mid-knee flexion requires muscle control and this is difficult to simulate on a cadaveric knee without tissue failure [34–41]. The surgical technique is another factor that is critical for addressing mid-flexion instability of the TKR. Instability is identified as one of the most common reasons for TKR revision [3,42,43]. Therefore, to avoid further need for TKR revisions, a better understanding of weight shift and knee loading during ADLs is needed. Articular loading of the knee is dependent upon the knee angle and can be addressed in post-surgical rehabilitation. In order to decrease the progression toward improper loading of the primary knee replacement that may lead to the need for surgical revision, an early rehabilitation approach to address weight-bearing symmetry needs to be established. Instruction in equal weight bearing of the lower extremities should begin as early as the post-operative protocol permits, along with added balance and proprioception exercises to facilitate weight-shifting during practical tasks. Early intervention to assist in balanced lower extremity weight-shifting should be incorporated with total knee replacement patients to avoid compensatory movements that may create the potential for injury in other joints such as the back, hips, and contralateral knee. Further *in vivo* investigation is necessary to determine if compensatory movement patterns will guide the development for improved implant design, rehabilitation intervention, and a surgical approach for the TKR patient.

References

[1] Centers for Disease Control National Center for Health Statistics, "FastStats: Inpatient Surgery," accessed September 28, 2012, http://www.cdc.gov/nchs/fastats/insurg.htm

[2] Healy, W. L., Sanjeev, S., Schwartz, B., and Iorio, R., 2008, "Athletic Activity After Total Joint Arthroplasty," J. Bone Jt. Surg. Am. Vol., **90**, pp. 2245–2252.

[3] Del Gaizo, D. J., and Della Valle, C. J., 2011, "Instability in Primary Total Knee Arthroplasty," Orthopedics, **34**(9), pp. e519–e521.

[4] Callaghan, J. J., O'Rourke, M. R., and Saleh, K. J., 2004, "Why Knees Fail: Lesson Learned," J. Arthroplasty, **19**(4–1), pp. 31–34.

[5] Vince, K. G., Abdeen, A., and Sugimori, T., 2006, "The Unstable Total Knee Arthroplasty: Causes and Cures," J. Arthroplasty, **21**(4–1), pp. 44–49.

[6] Parratte, S., and Pagnano, M. W., 2008, "Instability After Total Knee Arthroplasty," J. Bone Jt. Surg. Am. Vol., **90**(1), pp. 184–194. Available at: http://jbjs.org/article.aspx?articleid=28421

[7] Segal, A. D., Orendurff, M. S., Czerniecki, J. M., Shofer, J. B., and Klute, G. K., 2008, "Local Dynamic Stability in Turning and Straight-Line Gait," J. Biomech., **41**, pp. 1486–1493.

[8] Hurd, W. J., and Snyder-Mackler, L., 2007, "Knee Instability After Acute ACL Rupture Affects Movement Patterns During the Mid-Stance Phase of Gait," J. Orthop. Res., **25**(10), pp. 1369–1377.

[9] Benoit, D. L., Ramsey, D. K., Lamontagne, M., Xu, L., Wrentenberg, P., and Renstrom, P., 2007, "*In Vivo* Knee Kinematics During Gait Reveals New Rotation Profiles and Smaller Translations," Clin. Orthop. Relat. Res., **454**, pp. 81–88.

[10] Byrne, J. M., and Prentice, S. D., 2003, "Swing Phase Kinetics and Kinematics of Knee Replacement Patients During Obstacle Avoidance," Gait and Posture, **18**, pp. 95–104.

[11] Smith, A. J., Lloyd, D. G., and Wood, D. J., 2004, "Presurgery Knee Joint Loading Patterns During Walking Predict the Presence and Severity of Anterior Knee Pain After Total Knee Arthroplasty," J. Orthop. Res., **22**, pp. 260–266.

[12] McClelland, J. A., Webster, K. E., and Feller, J. A., 2007, "Gait Analysis of Patients Following Total Knee Replacement: A Systematic Review," The Knee, **14**, pp. 253–263.

[13] Christiansen, C. L., Bade, M. J., Judd, D. L., and Stevens-Lapsley, J. E., 2011, "Weight-Bearing Asymmetry During Sit-Stand Transitions Related to Impairment and Functional Mobility After Total Knee Arthroplasty," Arch. Phys. Med. Rehabil., **92**, pp. 1624–1629.

[14] D'Lima, D. D., Patil, S., Steklov, N., Slamin, J. E., and Colwell, C. W., 2006, "Tibial Forces Measured *In Vivo* After Total Knee Arthroplasty," J. Arthroplasty, **21**, pp. 255–262.

[15] Mundermann, A., Dyrby, C. O., D'Lima, D. D., Colwell, C. W., and Andriacchi, T. P., 2008, "*In Vivo* Knee Loading Characteristics During Activities of Daily Living as Measured by an Instrumented Total Knee Replacement," J. Orthop. Res., **26**, pp. 1167–1172.

[16] Zhao, D., Banks, S. A., D'Lima, D. D., Colwell, C. W., and Fregly, B. J., 2007, "*In Vivo* Medial and Lateral Tibial Loads During Dynamic and High Flexion Activities," J. Orthop. Res., **25**, pp. 593–602.

[17] D'Lima, D. D., Steklov, B. S., Patil, S., and Colwell, C. W., 2008, "*In Vivo* Knee Forces During Recreation and Exercise After Knee Arthroplasty," Clin. Orthop. Relat. Res., **466**, pp. 2605–2611.

[18] Moisio, K. C., Sumner, D. R., Shott, S., and Hurwitz, D. E., 2003, "Normalization of Joint Moments During Gait: A Comparison of Two Techniques," J. Biomech., **36**(4), pp. 599–603.

[19] Rossi, M. D., Eberle, T., Roche, M., Wong, M. L., Waggoner, M. R., Blake, R., Burwell, B., and Baxter, A., 2011, "Squatting After Primary Unilateral Total Knee Arthroplasty: A Pilot Study Examining Differences Between Genders," PM+R, Jnl Injury, Function, and Rehabilitation, **3**, pp. 613–618.

[20] Yoshida, Y., Mizner, R., Ramsey, D. K., and Snyder-Mackler, L., 2008, "Examining Outcomes From Total Knee Arthroplasty and the Relationship Between Quadriceps Strength and Knee Function Over Time," Clin. Biomech., **23**, pp. 320–328.

[21] Mandeville, D., Osternig, L. R., and Chou, L. S., 2008, "The Effect of Total Knee Replacement Surgery on Gait Stability," Gait and Posture, **27**(1), pp. 103–109.

[22] Chatterji, U., Ashworth, M. J., Lewis, P., and Dobson, P., 2005, "Effect of Total Knee Arthroplasty on Recreational and Sporting Activity," Aust. N. Z. J. Surg., **75**, pp. 405–408.

[23] Wylde, V., Blom, A., Dieppe, P., Hewlett, S., and Learmonth, I., 2008, "Return to Sport After Joint Replacement," J. Bone Jt. Surg., **90-B**, pp. 920–923.

[24] Motion Lab System, Inc., "Knee Alignment Device: User Manual," Motion-labs.com, 1998, accessed 09/24/2012, http://www.motion-labs.com/pdf/kneealignment_ug.pdf

[25] Henmi, S., Yonenobu, K., Masatomi, T., and Oda, K., 2006, "A Biomechanical Study of Activities of Daily Living Using Neck and Upper Limb With an Optical Three-Dimensional Motion Analysis System," Mod. Trends Rheumatol., **16**, pp. 289–293.

[26] Rossi, M. D., Hasson, S., Kohia, M., Pineda, E., and Bryan, W., 2006, "Mobility and Perceived Function After Total Knee Arthroplasty," J. Arthroplasty, **21**, pp. 6–12.

[27] Oakeshott, R., Stiehl, J. B., Komistek, R. A., Anderson, D. T., and Haas, B. D., 2003, "Kinematic Analysis of a Posterior Cruciate Retaining Mobile-Bearing Total Knee Arthroplasty," J. Arthroplasty, **18**(8), pp. 1029–1037.

[28] Yercan, H. S., Ait Si Selmi, T., Sugun, T. S., and Neyret, P., 2001, "Tibiofemoral Instability in Primary Total Knee Replacement: A Review, Part 1: Basic Principles and Classification," The Knee, **12**, pp. 257–266.

[29] Wu, C., Wong, M., Lin, K., and Chen, H., 2001, "Effects of Task Goal and Personal Preference on Seated Reaching Kinematics After Stroke," Stroke, **32**, pp. 70–79.

[30] Houck, J. R., Duncan, A., and De Haven, K., 2006, "Comparison of Frontal Plane Trunk Kinematics and Hip and Knee Moments During Anticipated and Unanticipated Walking and Side Cutting Tasks," Gait and Posture, **24**, pp. 314–322.

[31] Zurcher, A. W., Wolterbeek, N., Harlaar, J., and Poll, R. G., 2008, "Knee Rotation During a Weight Bearing Activity: Influence of Turning," Gait and Posture, **28**, pp. 472–477.

[32] Harato, K., Nagura, T., Matsumoto, H., Otani, T., Toyama, Y., and Suda, Y., 2010, "Extension Limitation in Standing Affects Weight-Bearing Asymmetry After Unilateral Total Knee Arthroplasty," J. Arthroplasty, **25**, pp. 225–229.

[33] Milner, C. E., 2008, "Interlimb Asymmetry During Walking Following Unilateral Total Knee Arthroplasty," Gait and Posture, **28**, pp. 69–73.

[34] Walsh, M., Woodhouse, L. J., Thomas, S. G., and Finch, E., 1998, "Physical Impairments and Functional Limitations: A Comparison of Individuals 1 Year After Total Knee Arthroplasty With Control Subjects," Phys. Ther, **78**, pp. 248–258. Available at: http://ptjournal.apta.org/content/78/3/248.long

[35] Mizner, R. L., and Snyder-Mackler, L., 2005, "Altered Loading During Walking and Sit-to-Stand is Affected by Quadriceps Weakness After Total Knee Arthroplasty," J. Orthop. Res., **23**, pp. 1083–1090.

[36] Shakoor, N., Block, J. A., Shott, S., and Case, J. P., 2002, "Nonrandom Evolution of Endstage Osteoarthritis of the Lower Limbs," Arthritis Rheum., **46**, pp. 3185–3189.

[37] McClelland, J., Zeni, J., Haley, R. M., and Snyder-Mackler, L., 2012, "Functional and Biomechanical Outcomes After Using Biofeedback for Retraining Symmetrical Movement Patterns After Total Knee Arthroplasty: A Case Report," J. Orthop. Sports Phys. Ther., **42**(2), pp. 135–144.

[38] Wilson, S. A., McCann, P. D., Gotlin, R. S., Ramakrishnan, H. K., Wootten, M. E., and Insall, J. N., 1996, "Comprehensive Gait Analysis in Posterior-Stabilized Knee Arthroplasty," J. Arthroplasty, **11**(4), pp. 359–367.

[39] Shakoor, N., Hurwitz, D. E., Block, J. A., Shott, S., and Case, J. P., 2003, "Asymmetric Knee Loading in Advanced Unilateral Hip Osteoarthritis," Arthritis Rheum., **48**, pp. 1556–1561.

[40] Lo, J., Müller, O., Dilger, T., Wülker, N., and Wünschel, M., 2010, "Translational and Rotational Knee Joint Stability in Anterior and Posterior Cruciate-Retaining Knee Arthroplasty," The Knee, **18**, pp. 491–495.

[41] Stoddard, J. E., Deehan, D. J., Bull, A. M. J., ScCaskie, A. W., and Amis, A. A., 2012, "The Kinematics and Stability of Single-Radius versus Multi-Radius Femoral Components Related to Mid-Range Instability After TKA," J. Orthop. Res., **31**(1), pp. 1–6.

[42] Firestone, T. P., and Eberle, R. W., 2006, "Surgical Management of Symptomatic Instability Following Failed Primary Total Knee Replacement," J. Bone Jt. Surg. Am. Vol., **88**(4), pp. 80–84.

[43] McPherson, E. J., Cuckler, J., and Lombardi, A. V., 2008, "Midflexion Instability in Revision Total Knee Arthroplasty," Surg. Technol. Int., **17**, pp. 249–252.

Appendix B

This appendix contains the manuscript for Chapter 4.

Using a Musculo-Skeletal Model to Assess Muscle Activation and Biomechanical Strategies During a Rotational Task in Individuals with a Total Knee Replacement

Lauren A. Ferris[1], Linda M. Denney[2], Lorin P. Maletsky[1]

[1] University of Kansas, Department of Mechanical Engineering

[2] University of Kansas Medical Center, Department of Physical Therapy and Rehabilitation Science

Corresponding Author: Lorin P. Maletsky,

1530 W 15th St., Learned Hall, room 3138, Lawrence, KS 66045

Phone: (785) 864-2985 Fax: (785) 864-5152 Email: maletsky@ku.edu

Abstract:

Activities of daily living often require flexion and extension of the knee, rotation of the torso, and transferring weight from side to side that challenges stability of the knee, and more so in patients with a total knee replacement (TKR). Compensatory strategies utilized by patients with TKRs have been studied in a variety of sagittal plane activities, yet little is known for out-of-plane activities. The specific aims of this study were to identify the compensatory strategies utilized during the target touch tasks between the healthy and TKR subjects and to determine the relationship between musculature and biomechanical activity throughout the task. Nine TKR and eleven healthy subjects were analyzed using subject-specific musculo-skeletal models when performed two crossover tasks with coupled flexion or extension and rotation; high to low (H2L) and low to high (L2H). During H2L the affected knee showed significant difference to healthy range of side-to-side center of mass displacement. L2H had the most variation where the lead

unaffected knee was significantly more flexed than healthy, knee torque was greater in unaffected than affected, and hamstring, gastrocnemius, hip extensor and vastii activation parameters were greatest in healthy subjects. Strategies such as modified muscular contribution, center of mass movement, and limited knee flexion despite having full available ROM were determined as compensatory movements to achieve this novel rotary task. Significant differences were mainly noted in the high button push of the L2H task, while the high button push of the H2L showed no statistical differences. This result, along with differences in muscle activation and knee torque was mainly driven by the crossover, suggests that this novel rotary task would better assess strategies utilized by TKR subjects than symmetric, in plane activities such as squatting.

Introduction

Of the more than 500,000 individuals who undergo total knee replacement (TKR) surgery every year, most report satisfaction, while some continue to demonstrate difficulty with activities of daily living (ADLs) (Decade, 2008; Noble et al., 2005; Weiss et al., 2002). Difficulty with kneeling and stair descent are evident up to two years post-surgery (Weiss et al., 2002; Zeni and Snyder-Mackler, 2010), while rotational activities such as loading the dishwasher or doing laundry remain unscreened. Rotation at the knee occurs in most activities including getting in and out of the car and the bathtub and can be combined with a torso movement as in a golf swing. The combination of rotary forces and weight transfer on the TKR is dependent on muscle control for function and stability. Symmetrical loading of the lower extremities with appropriate muscular firing during functional activities is a standard rehabilitation goal following a TKR in order to maximize function and discourage compensatory movements.

Movement asymmetry or unequal loading of the lower extremities post-TKR has been attributed to patterns developed prior to surgery (Bade et al., 2010; Mandeville et al., 2008; Worsley et al., 2013; Zeni and Snyder-Mackler, 2010). Despite pain reduction in the surgical knee, weight is shifted toward the contralateral leg, thus unloading the TKR leg. Excessive loading of the non-operative knee has been associated with the need for a TKR on the opposite leg within 10 years of the initial TKR (Farquhar and Snyder-Mackler, 2010; McMahon and Block, 2003). Outcomes of rehabilitation intervention specifically designed to address unloading of the TKR leg are mixed (Mandeville et al., 2008; McClelland et al., 2012; Mizner and Snyder-Mackler, 2005). Despite direct supervision of rehabilitation intervention, some programs produce poor results of reducing impairments following TKR while others improved by incorporating balance or symmetry training (Pozzi et al., 2013). Knee flexion range of motion (ROM) and quadriceps strength was symmetrical at six months post-op in the symmetry training group when compared to those that received standard of care that included progressive strengthening exercises (Zeni et al., 2013).

The muscles of the lower extremities and trunk assist in controlling joint loading and motion of the center of mass (COM). Individuals with a TKR may demonstrate a strategy referred to as 'quadriceps avoidance' primarily during early stance in gait resulting in less contraction of the vastus lateralis and forward trunk flexion (Li et al., 2013). Altered biomechanics at the knee contribute to varying loading throughout the different knee angles depending on the activity. Subjects performing a novel rotary task, known as the Target Touch Task (TTT), described by Ferris et al. (Ferris et al., 2013) differed in knee flexion angle, with the TKR individuals performing the rotational task with less knee flexion when compared to the healthy group. The use of a compensatory strategy to complete the task may have included a shift in the torso, thus shifting the COM.

Individuals with a TKR may present compensatory strategies by using hip and trunk musculature rather than the quadriceps (Li et al., 2013). During gait, the gluteus maximus and soleus muscles are activated in the presence of weak quadriceps or poor activation of the quadriceps (Thompson et al., 2013) while the back musculature may assist in deceleration (Li et al., 2013). Most research is performed during gait to determine asymmetric muscle activation patterns (Hast and Piazza, 2013; Li et al., 2013; Nha et al., 2013; Thompson et al., 2013; Wilson et al., 1996), while other studies utilize stair ascent and descent or rise from a chair for kinetic data collection (Kutzner et al., 2010; Leffler et al., 2012; Worsley et al., 2013; Zeni et al., 2013). In order to consider a functional rotary task, Ferris et al. (2013) analyzed the double-stance reaching activity with rotation which incorporated the trunk and included a functional squatting activity. Squatting recruits most of the lower extremity musculature and torso muscles to maintain postural stability (Schoenfeld, 2010). The motion of squatting with rotation utilizes a combination of muscle forces, lower quarter joint mechanics and trunk stability which may reveal compensatory patterns with TKR individuals that relate to performance of ADLs.

The purpose of this study was to further investigate the strategies utilized by TKR subjects compared to healthy when performing the novel flexion/extension task including rotation and transfer of weight. By understanding these strategies TKR rehabilitation could be improved. Subjects with a TKR versus healthy controls were compared with regards to muscular function and COM control calculated by use of 3d musculoskeletal modeling. The specific aims of this study were to identify the compensatory strategies utilized during the TTT between the healthy, TKR affected knee, and the TKR subject's unaffected knee and to determine the relationship between musculature and biomechanical activity throughout the task.

Materials and Methods

Twenty subjects (9 TKR subjects avg. age 66 (SD 6) years; avg. BMI 26.7(3.0) and 11 healthy subjects avg. age: 64(8); avg. BMI 24.3(3.9)) volunteered to participate in this study and signed a consent form approved by the Human Subjects Committee and Institutional Review Board at the University of Kansas Medical Center. Healthy control subjects were included in the study if they were between the ages of 50-75, had a BMI less than 30, could bend both knees at least 90°, and had no previous hip or ankle surgery or peripheral neuropathy. TKR subjects were selected if the above criteria were met and they had a unilateral TKR. Anthropometric and bony landmark measurements were collected to configure the computer model using Vicon's Workstation (v 4.5) and the lower body Plug-in Gait Model (Oxford Metrics, Oxford, UK). Ranges of motion of the hips and knees were measured with a goniometer, anterior translation of the tibio-femoral joint using a KT-1000, and quadriceps strength using a dynamometer (MicroFET2, Hoggan Health Industry; West Jordan, Utah).

Twenty-four reflective markers (25 mm in diameter) were attached with double-sided adhesive to bony landmarks. Markers were captured at 120 Hz with an infrared six-camera motion analysis system (Vicon 512, Oxford Metrics, Oxford, UK) and calibrated according to the manufacturer's specifications. Two six-degree-of-freedom (dof) force plates (AMTI, Watertown, MA) embedded in the floor captured forces and moments at 360 Hz and were triggered simultaneously with an analog device.

The TTT was set up for each subject according to height and crossover reach width (Fig. 1A, 1B). Two microphone stands were placed the lateral width of the reach test and at the height of the subject's shoulder. Two lower buttons were placed just outside the knee where the subject was able to bend down and still press the button comfortably. Buttons were clipped to microphone stands and wired to make a sound upon activation. The subject stood with one foot on each force plate.

The subjects performed two TTT; a high-to-low (H2L) crossover sequence and a low-to-high (L2H) crossover sequence (Fig.1A, 1B). Both TTT were performed 5 times, 3 to the left, 2 to the right. Data collection began with a practice button push (button #4 in Fig. 1A, 1B) and ended once the subject hit the last button of the sequence. If the subject lifted his heels off the force plates or performed the wrong sequence, a retrial was collected.

A three-dimensional musculoskeletal computer model (Fig. 1C, 1D), utilized in OpenSim (Delp et al., 2007), was used to calculate muscle forces generated during the TTT. The Gait-Extract toolbox (Dorn, 2008) was utilized to extract and format kinematic and kinetic data exported from Vicon to OpenSim compatible files. The skeleton was primarily a lower extremity model with a lumped torso segment that included the head, 23 dof with 92 muscle-tendon actuators of the lower limb to represent 76 muscles (Au, 2012). The hips were modeled as ball-and-socket joints and the knees as single dof hinges. Each subject had their own model based on specific anthropometric measurements collected prior to testing. Joint angles were calculated using inverse kinematics based on the specific subject marker data and net joint torques were calculated via inverse dynamics from the force plate data. Force plate data were filtered using a 4th-order Butterworth filter prior to inverse dynamics to smooth results needed for static optimization. Static optimization was computed using results from the inverse kinematics and dynamics at each time step to determine individual muscle forces.

The six muscle groups analyzed were the vastii (consisting of the vastus lateralis, vastus medialis, and vastus intermedium), rectus femoris, hamstrings (long and short head of the biceps femoris, semitendinosus, and semimembranosus), gastrocnemius, hip flexors and hip extenders. The hip flexors were calculated by summing the adductor brevis, adductor longus, gluteus medius, gluteus minimus, gracilis, illiacus, pectineus, psoas, rectus femoris, sartorius, and tensor fasciae latae muscles. The hip extenders consisted of the adductor longus, adductor magnus, long head of the biceps femoris, gluteus maximum, gluteus medius, gluteus minimum, semimembranosus, and semitendinosus muscles. Muscle forces, knee angles, and knee torques were analyzed and compared between the three groups.

COM position parameters (anterior/posterior, superior/inferior, and side/side) were generated within the analyze tool in OpenSim at each time step of the TTT trials. The anterior/posterior values were normalized to the subject's foot size, superior/inferior to height, and the side/side or lag/lead to the width of the subject's stance. Measurements utilized for the normalization was calculated based off the recorded height prior to testing and markers on the heels and toes to represent foot size and stance width.

Each subject's task, H2L or L2H, was cut into the five crossovers performed during the TTT and each leg was labeled as lag or lead; lag representing the leg closest to the first button push and the lead being the leg the subject was leaning towards for the second button push. Each crossover trial was interpolated to represent 100 evenly spaced points from button push to button push, or percent cycle. The two right trials and three left trials were then averaged at each percent cycle and labeled based on the lead leg as either affected (TKR knee), unaffected (TKR contralateral knee), or either leg of the control subject (healthy knee). Lastly, all unaffected, affected, and healthy subjects' H2L and L2H lead trials and lag trials were averaged per event, per limb, at each percent cycle.

One-way ANOVAs ($p<0.05$) and Tukey's post processes were ran at every 10% of the button cycle for lead and lag knee angle, lead and lag knee torque, lead and lag muscle forces, and position anterior/posterior, superior/inferior, and side/side directions for the H2L and L2H task.

Results

The H2L lag and L2H lead legs of the healthy subjects had a greater knee ROM compared to the TKR subjects (Table 2). Although the healthy had significantly greater knee flexion goniometric measurement in the pre-measurement when compared to the affected, the affected knee consistently demonstrated a greater ROM in both H2L (lead) and L2H (lag) legs (Table 1). During the H2L task all subject groups had a greater knee ROM in the lag leg, while during the L2H the greater ROM was in the lead leg. No significant differences were observed between the three groups' knee angles during the H2L task, in either the lead or lag leg, while during the L2H a significant difference was observed between the lead unaffected and healthy in the later portion of the percent cycle (Fig. 2) where the healthy subjects extended the knee more to reach the high button. Throughout both tasks, lead or lag leg, the unaffected had less knee ROM compared to the affected knee (Table 2), but the unaffected knee ROM was significantly greater than the affected during pre-task goniometric measurements (Table 1).

When comparing the knee ROM (Table 2) and the knee flexion torques (Fig. 3) the knee angles differ from task to task, while the knee torque trends are consistent between H2L and L2H lead and similar with lag leg. There were no significant differences between the healthy and TKR subjects affected or unaffected leg during either task, but the healthy subjects tended to have a lower and smaller range of deviation compared to the TKR subjects. Significant differences were reported between the unaffected and affected lead knees during the L2H task at 80% ($p=0.03$) and 90% cycle ($p=0.02$) (Fig. 3), with a greater knee torque on the affected compared to the unaffected.

The COM position in the anterior/posterior and superior/inferior directions for either task showed no statistical significance between the knee conditions (Fig. 4) while the lag/lead direction indicated TKR subjects shifted COM more towards the lead leg. Statistical differences were recorded between the TKR subjects' (both unaffected and affected) and healthy controls' COM range in the lag/ lead direction (H2L: $p=0.024$, $p=0.030$, respectively; L2H: $p=0.010$, $p=0.017$, respectively) (Table 3), while the TKR subjects had a greater side-to-side position movement (Fig. 4) with a significant difference between the H2L unaffected and healthy at 80-100% of the cycle ($p<0.05$). While there was no significant difference during the L2H task lag/lead COM, trends were noted at 40% between affected and healthy ($p=0.053$), and between affected and unaffected at 50% ($p=0.054$).

There were no significant differences between the three groups on predicted muscle forces during the H2L task, yet distinct trends can be observed between the lead and lag forces for each given muscle group (Fig. 5). Consistently the H2L lag leg starts with the hamstrings, gastrocnemius, and hip extensors contributing most to the muscle activity (Fig. 5) and as the subject transfers their weight to the lead, the lag leg activated the quadriceps and hip flexors control the trailing leg's motion (Fig 5). These trends are inverted to the lead leg results with the quadriceps and hip flexors firing while the high button was pressed and the hamstrings, gastrocnemius, and hip extensors taking over for the low button.

Similarly to the H2L, the L2H lead and lag leg muscle forces followed the same activation trends per muscle group, yet significant differences between groups were observed during the later percentage of the L2H cycle (>70%), or when the subject was extended to hit the high button. As the subjects transferred to the high button the healthy had statistically greater gastrocnemius and hamstring (Fig. 5) lead leg contribution compared to the unaffected, where the unaffected had statistically greater lead vastii forces than the healthy. The only statistical difference between the affected and unaffected was observed in the hip extensors, with the affected recruiting the muscle group more for the high button push. The only statistical difference in the lag limb muscle forces was in the gastrocnemius during L2H where the healthy group had a greater force than the affected group during a high button push.

Discussion

The ability to return to functional activities such as ADLS, along with pain relief, are crucial components for TKR satisfaction (Baker et al., 2007). Compensatory strategies are frequently observed in TKR patients during ADLs such as gait, squatting, and sit-to-stand activities (Bade et al., 2010; Fitzgerald et al., 2004; Mizner and Snyder-Mackler, 2005). Frequently used compensations include abnormal co-contraction of the hamstrings and quadriceps, quadriceps avoidance techniques and/or quadriceps weakness, and altered knee kinematics. In the present study the researchers aimed to identify compensatory strategies utilized during the TTT and to determine the relationship between the biomechanical and muscular variables. Compensatory strategies were defined as statistical differences between the TKR subjects compared to the healthy controls. Compensatory strategies were also considered between the unaffected and affected leg of the TKR participants. Examples of this type of protective strategy or compensation for the surgical knee include the affected knee having a larger lead knee torque and hip extensor force during the L2H task. It has also been suggested that the contralateral leg (unaffected) demonstrates characteristics of instability post-TKR (Schmitt and Rudolph, 2008).

Although knee ROM was measured greater in healthy subjects pre-task, the affected knee had the greatest ROM in the limb closest to the low button during both tasks. A dynamic squat activity, such as the TTT, requires balance, ankle mobility, and control of the trunk and lower extremities to transfer the force. Individuals were not instructed in the amount of knee flexion to use in order to accomplish the task, therefore leaving the possibility of an altered technique or strategy to be displayed. Interestingly, the subjects, independent of H2L or L2H, kept the knee closest to the last button pushed more extended and the knee on the lag leg more flexed, thus the subject's right and left legs moved asymmetrically during the squatting portion of the task. A similar kinematic asymmetrical response was evident in the TKR individuals during frontal plane perturbation suggesting a central mediated response or use of the CNS to elicit a motor response (Gage et al., 2007). It is unclear if asymmetric movement is a protective response for the surgical knee in this study since the healthy subject had similar knee ROM throughout the tasks and all the subject's lead legs performed the tasks with task specific knee flexion profiles (Fig. 2) compared to their lag leg. Rather, the pattern may be a kinematic response to maintain balance centrally driven by the CNS due to COM displacement.

A majority of differences were prevalent in the L2H task, more specifically when the body was rotated to press the high button, or the extension portion. The TKR group significantly maintained a more flexed knee on the lead leg, a greater lag/lead COM range, and had increased use of the vastii musculature during the reach, whereas the healthy group kept a more extended knee and utilized the hamstrings to extend the hip. In a gait study by Benedetti et al. (Benedetti et al., 2003) a co-contraction strategy was demonstrated by the 'stiff-legged' stance phase on an extended knee; not a flexed knee. Although the TKR subjects had less hamstring activity compared to the healthy subjects (Fig. 5), the TKR subjects also kept the knee more flexed (Fig. 2) and had a smaller flexion ROM during the L2H task (Table 2), suggesting the TKR subjects did not have a co-contraction strategy to control knee motion. Both the knee laxity measurement and lack of reported knee instability suggest the TKR subjects in the current study did not have knee instability, thus the need for abnormal co-contraction. Even though there were signs of some diminished hamstring activation in the TKR group, consistent findings of decreased hamstring strength post-TKR (Stevens-Lapsley et al., 2010; Walsh et al., 1998) has been reported from other studies.

The ability to maintain balance decreases with age and is complicated by TKR surgery. Lateral stepping or corrective strategies are impaired in individuals with TKR and may be a result of pre-surgical motor patterns for pain avoidance or feeling of instability (Viton et al., 2002). A lateral change in direction is represented by the lag/lead COM data (Fig. 4). The healthy individuals show little change in COM while the TKR range of displacement was significantly increased in both activities (Table 3). This is consistent with both a forward and lateral perturbation as described in Gage et al. (Gage et al., 2007, 2008). Although no differences were found in the anterior/posterior COM between the groups, a shift in balance occurs with a reciprocal muscle co-activation to control the movement. In a study by Kuo et al. (Kuo et al., 2011) back extensors assist in the forward motion (COM displacement) during a forward reach, allowing the individual to balance and complete the task. Displacement of COM and postural responses were altered in TKR individuals (Gage et al., 2008) during frontal plane perturbation. The current study incorporated a squat with rotation further challenging frontal plane balance and the TKR unaffected displacement of COM in the lag/lead direction was larger when compared to the healthy (Fig. 4). The affected displacement had a noticeable change of COM during the H2L task as well but not significant. The shift observed in AP balance, along with the lag/lead displacement, surprisingly followed the same trends in both tasks, with the only differences being in the S/I direction, suggesting both the H2L and L2H task either had little effect on balance control or the subjects used a strategy that allowed for little movement of the COM.

Regardless of H2L or L2H task performed, the lead knee flexion torque increased as the lag decreased (Fig. 3), compared to the lead and lag knee flexion angles where knee angles altered depending on the task (Fig. 2). Similarly, regardless of H2L or L2H tasks, the COM in the A/P and lag/lead direction trended in the same direction. It can be concluded that the knee torques, and thus the muscle activity, was driven by the crossover, not the flexion or extension of the knee.

Why are there significant differences between the groups during the L2H and not the H2L, or vice versa, if the task strategies are driven by the crossover? Knee flexion angle, and thus muscle activation, must contribute to the compensation strategies. Ferris et al. (Ferris et al.,

2013) reported no difference in hip flexion during the TTT in an attempt to explain a possible strategy utilized during the low button push for the TKR subjects that maintained knee extension when compared to the healthy. This is evident in the current study where by further analyzing the knee angle ROM it becomes apparent that the lead unaffected knee has the least knee flexion ROM, along with the lag unaffected H2L. The healthy group had a greater muscular contribution from the hamstrings (assisting in hip extension) during the L2H task while the TKR group utilized the vastii (Fig. 5). It is possible that the TKR group, while maintaining some knee flexion, activated the trunk musculature to either assist in braking while extending to reach the high button or control posture. TKR individuals with decreased vastii activation to extend the knee during walking activated the back extensor muscles (erector spinae and obliques) to assist in braking before early stance in gait (Li et al., 2013). This is similar to the double stance activity of the TTT with the early activation of the RF, acting as a hip flexor and possible trunk musculature firing to assist with deceleration and postural control; specifically to avoid falling forward. Findings of increased vastii activation in L2H is consistent with a lower knee flexion torque of the unaffected when compared to healthy near the end of the cycle (90 to 100%) (Fig. 3). Li et al. (Li et al., 2013) noted increased knee extension moment in TKR individuals associated with a lower force of the vastii during early stance in gait. The RF seemed to have no effect on either TKR or healthy individuals in the study by Li et al. (Li et al., 2013) and the current study. It appears the early activation of the RF contributes to hip flexion rather than knee extension in the current study.

Pre-surgical compensation strategies are commonly used by TKR patients where they overload the contralateral or non-surgical limb. These strategies are commonly carried over post-TKR causing damage to the non-surgical limb (Farquhar and Snyder-Mackler, 2010;

McMahon and Block, 2003). This may be a strategy in the current study during the L2H at 90 and 100% of the cycle where the unaffected limb used the vastii more than the healthy to complete this task. The healthy group utilized the hamstring muscle group on the lead leg to achieve the last 80-100% of the L2H cycle. Although the hip extensors were not significantly different between the healthy and the TKR group during the L2H cycle, the hamstrings represent a portion of the hip extensors and could possibly be assisting the healthy group into hip extension to reach the higher button rather than using the quadriceps to control knee flexion into extension. This correlates with the knee angle results during L2H on the lead leg as the healthy group have significantly less knee flexion at 70-100% of the cycle. The unaffected lead leg demonstrates more knee flexion, hence the vastii or quadriceps would assist in extending the knee to enable the TKR individual to reach the high button and maintain control of the knee.

Overall, the majority of the compensation strategies were recorded during the extension portion of L2H, with no statistical differences during the extension portion of H2L. This could potentially be due to the subjects feeling more stable during the H2L; starting in a standing, extended knee position and using the vastii, rectus femoris, and hip flexors to bend over to reach the button. On the contrary, during the L2H the lead leg on average flexed to 50^{o} to perform the low button push and then extended to approximately $10\text{-}25^{o}$ lead knee flexion for the high button push. To perform the task the TKR subjects kept a more flexed lead knee where the healthy controls were able to extend the knee by recruiting the hamstrings, gastrocnemius, and hip extensors. During the L2H the lag knee angle for all three groups remained more or less constant, with the affected knee flexing as the subject shifted to hit the high button. This may suggest the TKR subjects felt more stable with some knee flexion, contrary to the theory of quadriceps avoidance, or where patients keep an extended leg during gait to avoid the quadriceps

(Andriacchi, 1993). Early identification of these strategies could improve TKR success and the return to activities of daily living that involve flexion and rotation.

It is important to recognize the limitations of this research. The TKR subjects in this study were an active group with lower BMI, and could flex the knee at least 90^{o}. These subjects may have not represented the general TKR community. By sampling a larger group that included those individuals with instability or difficulty with functional activities may display greater compensation. Further recruitment of subjects could focus on patients that have received the same TKR design and post-TKR rehabilitation. In the current study, the TTT did a decent job in demonstrating altered knee angle and some muscular differences during the crossover tasks, yet the H2L an L2H tasks did not result in similar outputs as expected. Further analysis into the TTT and/ or other crossover tasks may aid in understanding rotary strategies used by TKR patients.

Limitations are also present with the modeling. First, the knee was modeled in OpenSim as a single degree-of-freedom hinge which does not capture out-of-plane translations or rotations. While the model does take into account the moments recorded about the force plate and movement of the motion tracking markers, internal/external kinematics at the knee could have better aided in the understanding of the rotational aspect of the tasks. Knee flexion/extension has the largest magnitude of motion in the knee, although rotational variables at the knee could have potentially produced different muscular, torque, and COM results due to the crossover. Second, the TKR components were not represented in the model. Since the components were not reported, this made it so all subjects were modeled with the same geometry. By modelling all the subjects with the same geometry the contributing factors that altered the modeling outputs were subject height and weight, marker movement, and force plate data. By creating a model fitted with a TKR the analyses could more accurately represent the conformity of a replacement and thus the effects of the surgery to muscular and biomechanical outputs.

Conclusion

The findings in this study did not identify as many strategies utilized by TKR subjects during rotation as expected. Compensatory strategies where mainly observed in the L2H task and COM motion for the H2L. Surprisingly, the L2H and H2L tasks did not result in similar knee kinematics or statistical significant during the low button push of the L2H and H2L, as well as the high button push. The lead knee torques increased and lag knee torques decreased regardless of direction. This phenomenon was also observed between lag muscle and lead muscle activity where, regardless of H2L or L2H, the muscles activated similarly. This suggests the TTT kinematics are driven by the crossover portion, not the flexion/ extension aspect. These strategies are important to understand since this study is relative to everyday life tasks that include bending and reaching. It is possible that these movement patterns were present prior to surgery and now integrated in the central processing system for balance during movement. The CNS provides input to interpret information for proprioception, which may be altered post-TKR. Therefore the request to reach at maximum distance may alter stability and facilitate a compensatory reaction. The primary concern is the increase in loading and use of the non-surgical leg. Asymmetrical loading of the legs post-TKR may contribute to future degeneration of the contralateral leg and ultimately surgery. The combination of muscle weakness, altered neuromuscular timing during tasks and loading, and altered knee positioning contribute to adverse forces in both knees and areas such as the low back and hip. Physical therapy should be addressed immediately following a TKR with an emphasis on muscular strengthening and symmetrical balance training in order to decrease occurrence of pre-surgical movement patterns and future altered loading patterns.

References

Andriacchi, T.P., 1993. Functional analysis of pre and post-knee surgery: total knee arthroplasty and ACL reconstruction. Journal of biomechanical engineering 115, 575-581.

Au, C., 2012. Gait 2392 and 2354 Models, August 12, 2012 ed, OpenSim Docmuentation http://simtk-confluence.stanford.edu:8080/x/54Mz.

Bade, M.J., Kohrt, W.M., Stevens-Lapsley, J.E., 2010. Outcomes before and after total knee arthroplasty compared to healthy adults. The Journal of orthopaedic and sports physical therapy 40, 559-567.

Baker, P.N., van der Meulen, J.H., Lewsey, J., Gregg, P.J., National Joint Registry for, E., Wales, 2007. The role of pain and function in determining patient satisfaction after total knee replacement. Data from the National Joint Registry for England and Wales. The Journal of bone and joint surgery. British volume 89, 893-900.

Benedetti, M.G., Catani, F., Bilotta, T.W., Marcacci, M., Mariani, E., Giannini, S., 2003. Muscle activation pattern and gait biomechanics after total knee replacement. Clin Biomech (Bristol, Avon) 18, 871-876.

Decade, U.S.B.a.J., 2008. The Burden of Musculoskeltal Diseases in the United States: Executive Summary. American Academy of Orthopaedic Suregeons, Rosemont, IL.

Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E., Thelen, D.G., 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE transactions on bio-medical engineering 54, 1940-1950.

Dorn, T., 2008. Gait Extract Toolbox for Matlab, 1.71 ed, simtk.org.

Farquhar, S., Snyder-Mackler, L., 2010. The Chitranjan Ranawat Award: The nonoperated knee predicts function 3 years after unilateral total knee arthroplasty. Clinical orthopaedics and related research 468, 37-44.

Ferris, L.A., Denney, L.M., Maletsky, L.P., 2013. Strategies utilized to transfer weight during knee flexion and extension with rotation for individuals with a total knee replacement. Journal of biomechanical engineering 135, 021020.

Fitzgerald, G.K., Piva, S.R., Irrgang, J.J., 2004. Reports of joint instability in knee osteoarthritis: its prevalence and relationship to physical function. Arthritis and rheumatism 51, 941-946.

Gage, W.H., Frank, J.S., Prentice, S.D., Stevenson, P., 2007. Organization of postural responses following a rotational support surface perturbation, after TKA: sagittal plane rotations. Gait Posture 25, 112-120.

Gage, W.H., Frank, J.S., Prentice, S.D., Stevenson, P., 2008. Postural responses following a rotational support surface perturbation, following knee joint replacement: frontal plane rotations. Gait Posture 27, 286-293.

Hast, M.W., Piazza, S.J., 2013. Dual-joint modeling for estimation of total knee replacement contact forces during locomotion. Journal of biomechanical engineering 135, 021013.

Kuo, F.C., Kao, W.P., Chen, H.I., Hong, C.Z., 2011. Squat-to-reach task in older and young adults: kinematic and electromyographic analyses. Gait Posture 33, 124-129.

Kutzner, I., Heinlein, B., Graichen, F., Bender, A., Rohlmann, A., Halder, A., Beier, A., Bergmann, G., 2010. Loading of the knee joint during activities of daily living measured in vivo in five subjects. J Biomech 43, 2164-2173.

Leffler, J., Scheys, L., Plante-Bordeneuve, T., Callewaert, B., Labey, L., Bellemans, J., Franz, A., 2012. Joint kinematics following bi-compartmental knee replacement during daily life motor tasks. Gait Posture 36, 454-460.

Li, K., Ackland, D.C., McClelland, J.A., Webster, K.E., Feller, J.A., de Steiger, R., Pandy, M.G., 2013. Trunk muscle action compensates for reduced quadriceps force during walking after total knee arthroplasty. Gait Posture 38, 79-85.

Mandeville, D., Osternig, L.R., Chou, L.S., 2008. The effect of total knee replacement surgery on gait stability. Gait Posture 27, 103-109.

McClelland, J., Zeni, J., Haley, R.M., Snyder-Mackler, L., 2012. Functional and biomechanical outcomes after using biofeedback for retraining symmetrical movement patterns after total knee arthroplasty: a case report. The Journal of orthopaedic and sports physical therapy 42, 135-144.

McMahon, M., Block, J.A., 2003. The risk of contralateral total knee arthroplasty after knee replacement for osteoarthritis. The Journal of rheumatology 30, 1822-1824.

Mizner, R.L., Snyder-Mackler, L., 2005. Altered loading during walking and sit-to-stand is affected by quadriceps weakness after total knee arthroplasty. Journal of orthopaedic research : official publication of the Orthopaedic Research Society 23, 1083-1090.

Nha, K.W., Dorj, A., Feng, J., Shin, J.H., Kim, J.I., Kwon, J.H., Kim, K., Kim, Y.H., 2013. Application of computational lower extremity model to investigate different muscle activities and joint force patterns in knee osteoarthritis patients during walking. Computational and mathematical methods in medicine 2013, 314280.

Noble, P.C., Gordon, M.J., Weiss, J.M., Reddix, R.N., Conditt, M.A., Mathis, K.B., 2005. Does total knee replacement restore normal knee function? Clinical orthopaedics and related research, 157-165.

Pozzi, F., Snyder-Mackler, L., Zeni, J., 2013. Physical exercise after knee arthroplasty: a systematic review of controlled trials. European journal of physical and rehabilitation medicine 49, 877-892.

Schoenfeld, B.J., 2010. The mechanisms of muscle hypertrophy and their application to resistance training. Journal of strength and conditioning research / National Strength & Conditioning Association 24, 2857-2872.

Stevens-Lapsley, J.E., Balter, J.E., Kohrt, W.M., Eckhoff, D.G., 2010. Quadriceps and hamstrings muscle dysfunction after total knee arthroplasty. Clinical orthopaedics and related research 468, 2460-2468.

Thompson, J.A., Chaudhari, A.M., Schmitt, L.C., Best, T.M., Siston, R.A., 2013. Gluteus maximus and soleus compensate for simulated quadriceps atrophy and activation failure during walking. J Biomech 46, 2165-2172.

Viton, J.M., Atlani, L., Mesure, S., Massion, J., Franceschi, J.P., Delarque, A., Bardot, A., 2002. Reorganization of equilibrium and movement control strategies after total knee arthroplasty. Journal of rehabilitation medicine : official journal of the UEMS European Board of Physical and Rehabilitation Medicine 34, 12-19.

Walsh, M., Woodhouse, L.J., Thomas, S.G., Finch, E., 1998. Physical impairments and functional limitations: a comparison of individuals 1 year after total knee arthroplasty with control subjects. Physical therapy 78, 248-258.

Weiss, J.M., Noble, P.C., Conditt, M.A., Kohl, H.W., Roberts, S., Cook, K.F., Gordon, M.J., Mathis, K.B., 2002. What functional activities are important to patients with knee replacements? Clinical orthopaedics and related research, 172-188.

Wilson, S.A., McCann, P.D., Gotlin, R.S., Ramakrishnan, H.K., Wootten, M.E., Insall, J.N., 1996. Comprehensive gait analysis in posterior-stabilized knee arthroplasty. The Journal of arthroplasty 11, 359-367.

Worsley, P., Stokes, M., Barrett, D., Taylor, M., 2013. Joint loading asymmetries in knee replacement patients observed both pre- and six months post-operation. Clin Biomech (Bristol, Avon) 28, 892-897.

Zeni, J., Jr., Abujaber, S., Flowers, P., Pozzi, F., Snyder-Mackler, L., 2013. Biofeedback to promote movement symmetry after total knee arthroplasty: a feasibility study. The Journal of orthopaedic and sports physical therapy 43, 715-726.

Zeni, J.A., Jr., Snyder-Mackler, L., 2010. Preoperative predictors of persistent impairments during stair ascent and descent after total knee arthroplasty. The Journal of bone and joint surgery. American volume 92, 1130-1136.

Table 1: Average (standard deviation) of the subject demographics.

	n=	Age years	Height m	BMI	Gender	Post- op months		Quad strength %bw	KT-1000 mm	Knee RoM deg	Hip RoM deg
TKR	9	66 (6)	1.77 (0.12)	26.7 (3.0)	3 F	21 (17)	Unaffected	0.18 (0.04)	2.7 (1.0)	138.0 (5.7)	109.7 (8.7)
							Affected	0.19 (0.06)	2.9 (1.0)	125.3 (5.4) [x+]	106.8 (7.5)
Healthy	11	64 (8)	1.68 (0.08)	24.3 (3.9)	6 F	N/A		0.19 (0.03)	2.4 (1.0)	141.6 (5.5)	109.1 (11.2)

x Affected statistically significant ($p<0.05$) from unaffected
+Affected statistically significant ($p<0.05$) from healthy

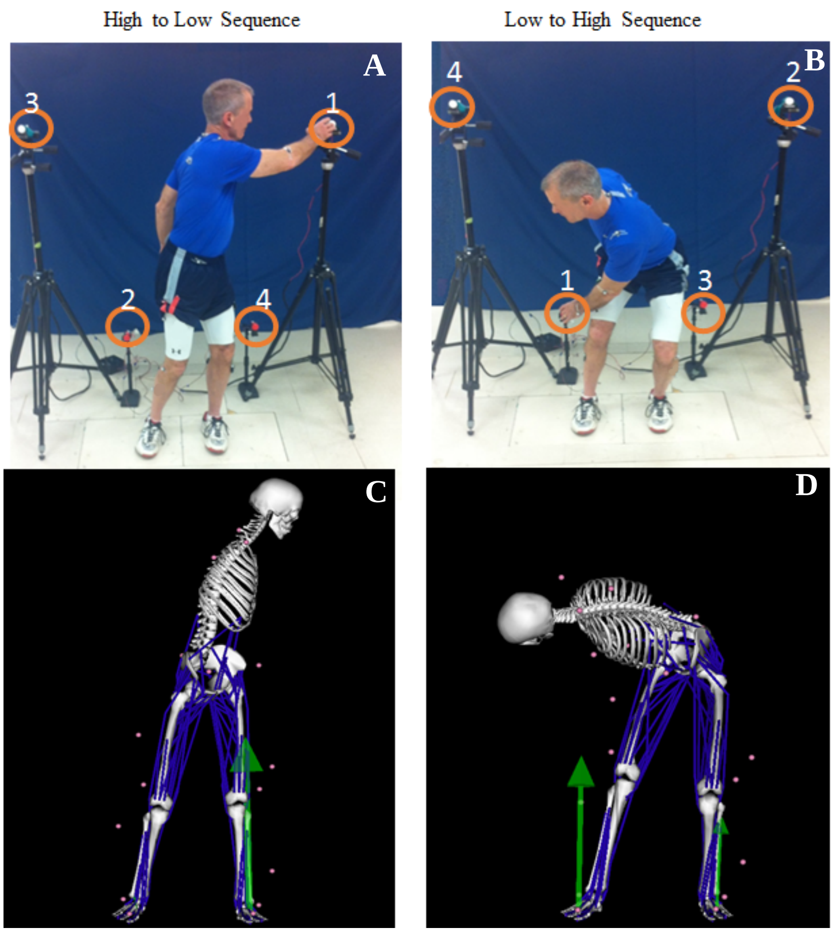

Figure 1: Equipment setup up along with a subject performing a H2L sequence (A) and L2H sequence (B). Numbers indicate the button sequence to perform the given task. Same subject performing the tasks in OpenSim (C, D). Arrows indicate forces recorded by the force plates.

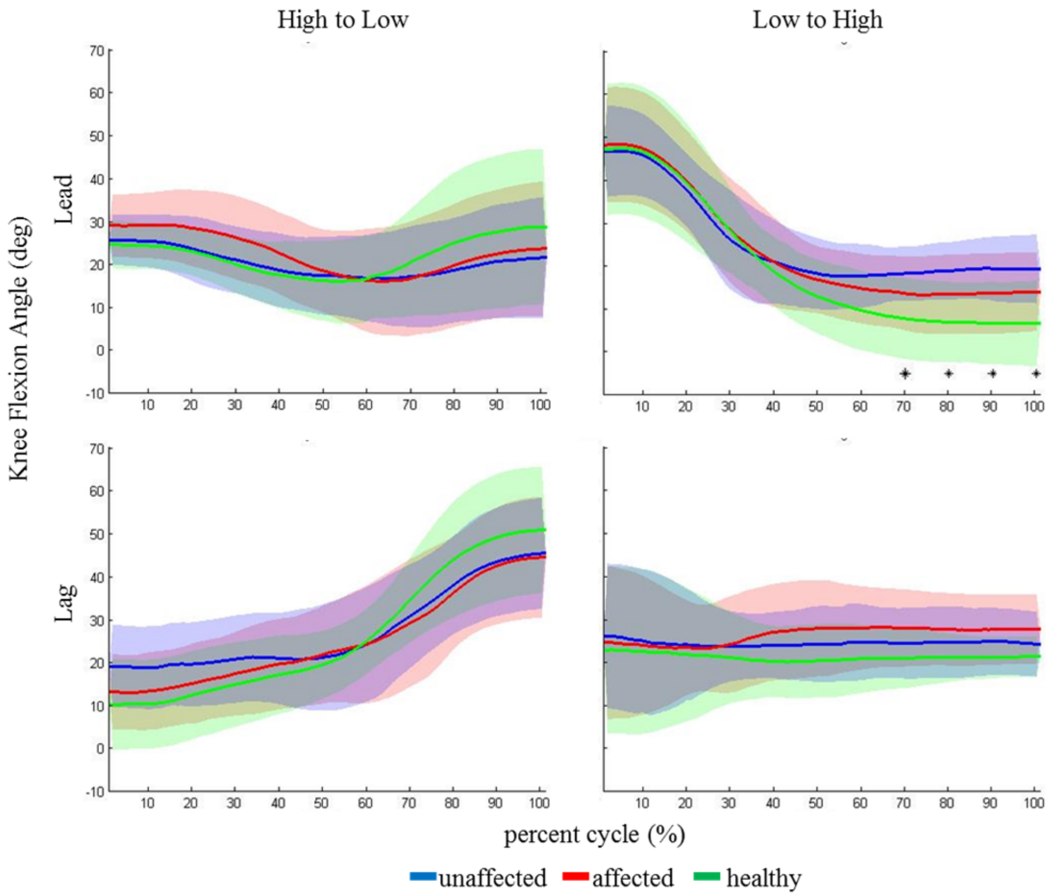

Figure 2: Average knee flexion angle for unaffected, affected and healthy subjects. Standard deviation ±1 indicated in the shading. Top row depicts the lead leg of the H2L and L2H task with the bottom row depicting the lag leg. A single factor ANOVA was performed at each 10% cycle. (The * denotes unaffected statistically significant ($p<0.05$) from healthy.)

Table 2: Average (standard deviation) Lead and Lag knee ROM in degrees for unaffected, affected, and healthy subjects performing the H2L and L2H task. There were no statistical differences between the three groups for either task.

		Lead	Lag
H2L	Unaffected	16.4 (7.1)	28.3 (10.5)
	Affected	21.5 (8.7)	32.1 (12.3)
	Healthy	20.0 (11.5)	41.0 (18.1)
L2H	Unaffected	30.1 (11.1)	17.2 (16.8)
	Affected	35.7 (14.1)	18.2 (14.4)
	Healthy	37.7 (20.6)	16.2 (10.6)

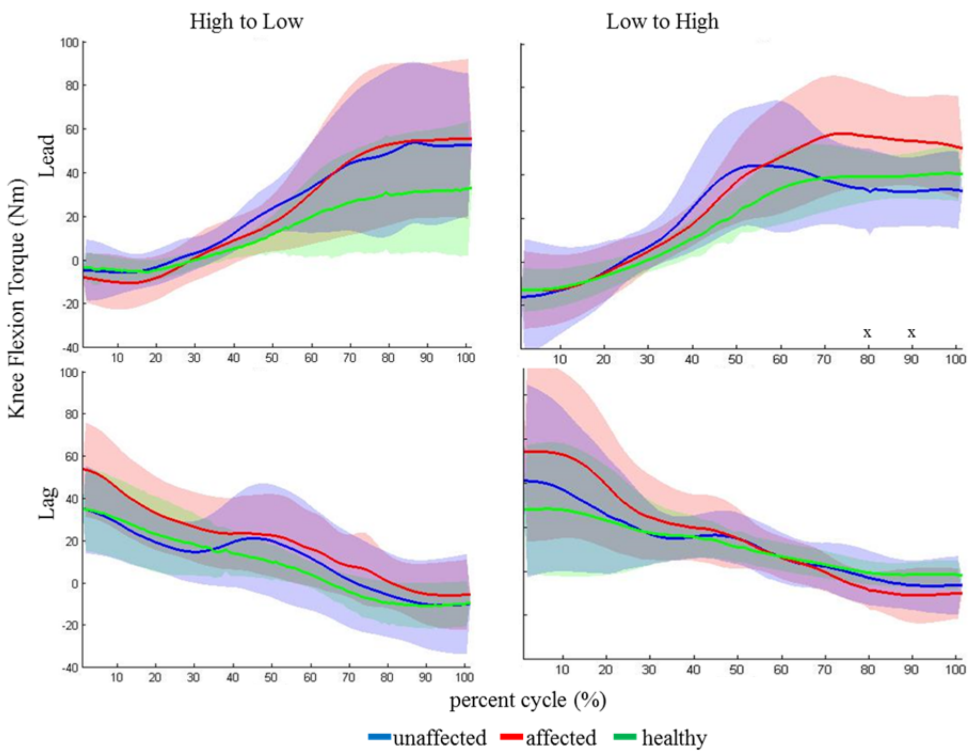

Figure 3: Mean knee flexion torque during H2L (column 1) and L2H (column 2) for unaffected, affected and healthy subjects. Standard deviation ±1 indicated in the shading. Top row depicts the lead leg with the bottom row depicting the lag leg. A single factor ANOVA was performed at each 10% cycle. (The x denotes affected statistically significant ($p<0.05$) from unaffected.)

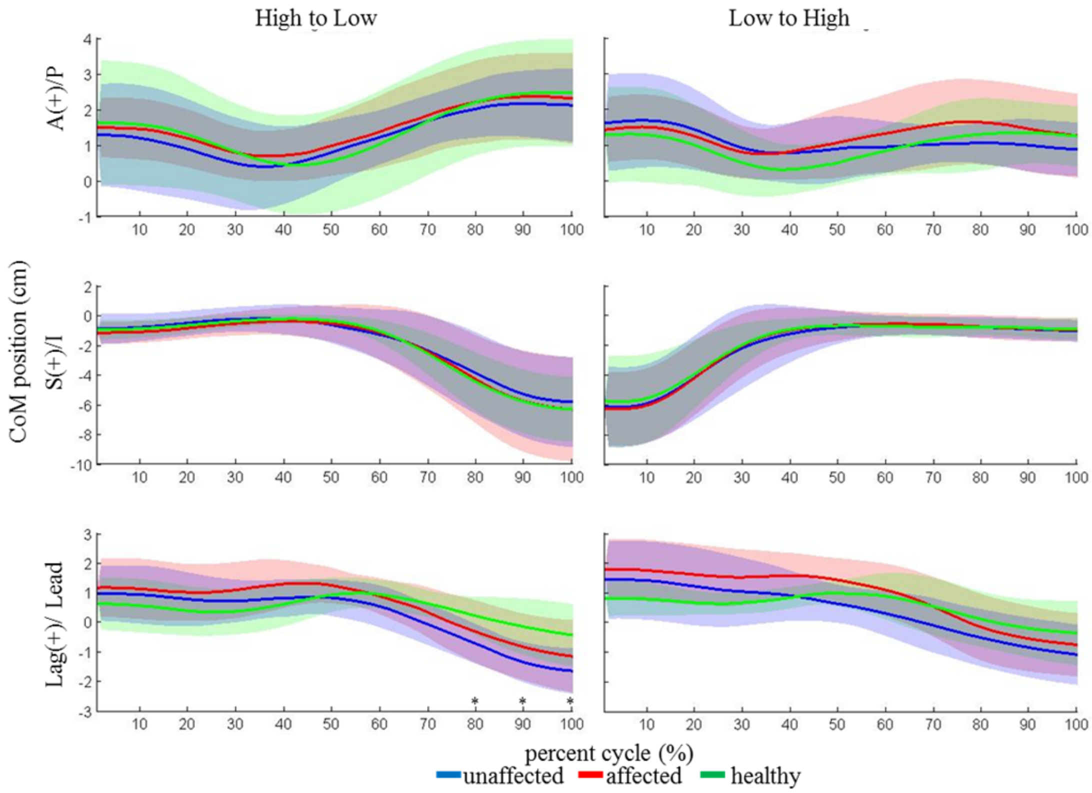

Figure 4: Mean center of mass deviation from static position in the anterior/ posterior (A/P) directions (top row), superior/ inferior(S/I) directions (middle row), and side-side position with the subject started towards the lag leg (+) then transferring to the lead leg (bottom row). Columns indicate activity (H2L column 1 and L2H column 2). Once the position offset from the static stance, A/P was normalized to foot length, S/I to height, and Lad/Lead to stance width. Standard deviation ±1 indicated in the shading. A single factor ANOVA was performed at each 10% cycle. (The * denotes unaffected statistically significant (p<0.05) from healthy.)

Table 3: Average (standard deviation) COM range in centimeters in the anterior-posterior, superior-inferior, and lag-lead directions for unaffected, affected, and healthy subjects performing the H2L and L2H task.

		A/P	S/I	Lag/Lead
H2L	Unaffected	2.13 (0.9)	5.65 (2.5)	3.12 (0.9)*
	Affected	2.00 (0.7)	6.03 (3.0)	3.03 (0.8)+
	Healthy	2.25 (0.6)	6.14 (2.0)	2.16 (0.9)
L2H	Unaffected	1.18 (0.7)	5.02 (3.0)	2.84 (1.5)*
	Affected	1.38 (0.7)	5.18 (3.0)	2.67 (1.5)+
	Healthy	1.37 (0.7)	5.08 (2.8)	1.74 (0.9)

*** Unaffected statistically significant ($p<0.05$) from healthy**
+Affected statistically significant ($p<0.05$) from healthy

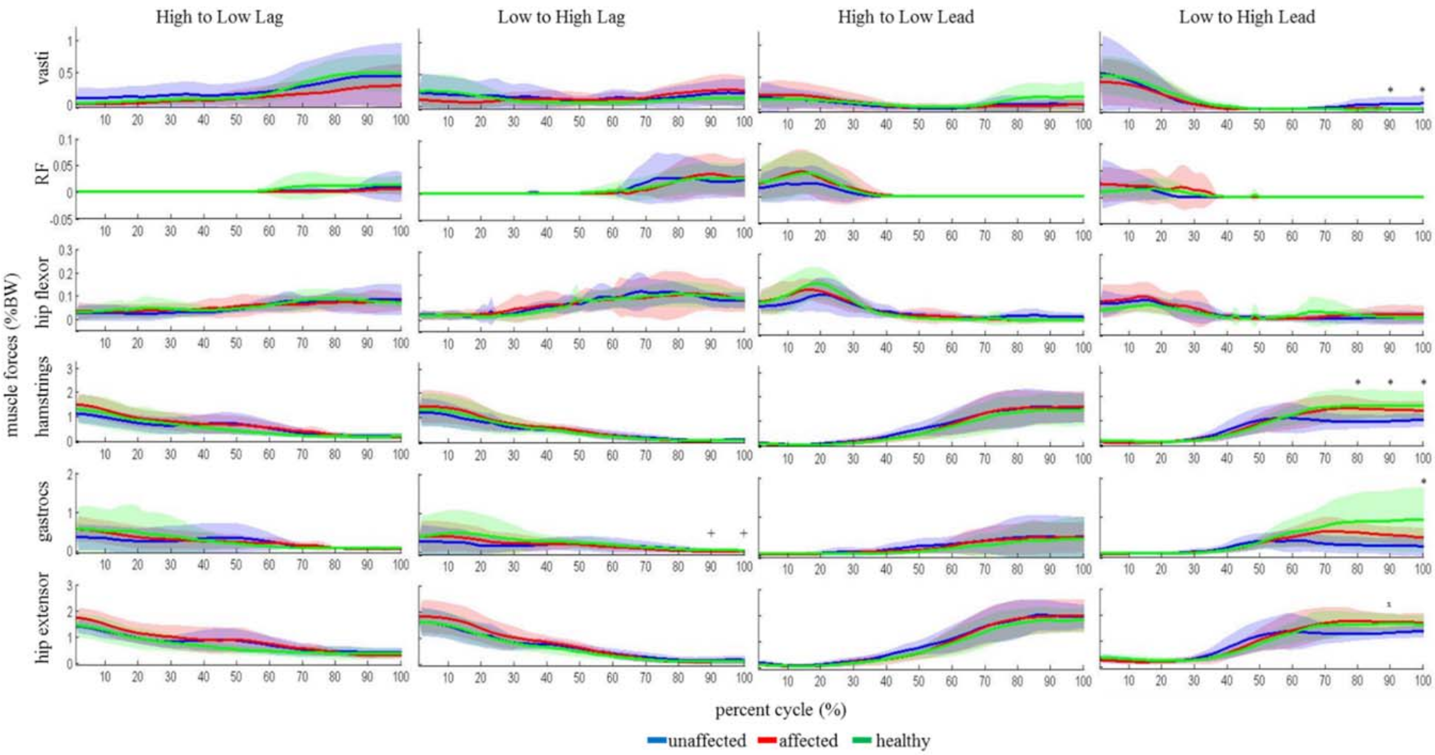

Figure 5: Mean muscle activation in the vastii, rectus femoris, hip flexors, hamstrings, gastrocnemius, and hip extensors during both H2L and L2H tasks. First two columns represent the lag legs, or the leg closest to the first button push. Last two columns represent the lead leg, or button closest to the second button push of the percent cycle. All data were normalized to percent body weight. Standard deviation ±1 indicated in the shading. A single factor ANOVA was performed at each 10% cycle. (The x denotes unaffected statistically significant ($p<0.05$) from affected. The * denotes unaffected statistically significant ($p<0.05$) from healthy. The + denotes affected statistically significant ($p<0.05$) from healthy.)