Rotary Task Kinematic and Kinetic Analysis of the Lower Extremity After Total Knee Arthroplasty: Stability and Strategies

BY

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ABSTRACT

Knee instability has been identified as a contributor to the need for total knee replacement (TKR) revision surgery. As TKR revisions increase, the importance of understanding the mechanisms of knee instability becomes a priority to the surgeons, rehabilitation specialists and designers in industry. Objective measurements of knee instability have been recorded by instruments such as the knee arthrometer or with the leg encased in a boot driven by a dial test tool for rotation. Both of these tests are performed in the open-chain position and the knee arthrometer only measures laxity in the sagittal plane. Despite capturing total range of motion (ROM) excursion or laxity measurement, these tests do not reproduce what is described as the working definition for knee instability. Functional knee instability, defined as the subjective report of the knee 'buckling' or 'giving way', correlates to a dynamic event rather than an open-chain occurrence. Therefore, a weight-bearing, dynamic test would be required in order to identify knee instability.

In order to test dynamic knee instability, rotation should be included to add a dimension of out-of-plane movement that correlates to pivoting, change of direction or rotating with bending and extending or reaching while loading and unloading the knee. Twenty-eight individuals (10 TKR, 12 Healthy, 6 Unstable) performed two tasks, Stair Task and Target Touch Task (TTT), and kinetic and kinematic data were captured by motion analysis and force platforms. The Stair Task included a pivot turn after stair descent and the TTT required a series of button pushes while squatting and extending, with rotation or crossing mid-line. Variables from both tasks were imputed to a Principle Components Analysis (PCA) in order to identify differences of performance across the groups.

In the Stair Task, the TKR group had less Ground Reaction Force (GRF) on initial impact after stair descent compared to the unaffected leg (p=0.021) and during mid-stance compared to the healthy group (p=0.049). The affected stance leg had less knee flexion during mid-stance

in both the straight trial (p=0.002) and turn (p=0.010). Similarly, the TKR individuals maintained a more extended knee position for both affected (14.7°) and unaffected (17.8˚) during the TTT when compared to the healthy (25.5˚) when approaching mid-line to transfer weight in order to push the low button when squatting (p<.05). Further, a difference was noted between the TKR legs during the motion to push the high button. At 90% of this cycle of movement, the unaffected knee was more flexed when compared to the affected knee (p<.05). A large variation of loading during mid-cycle was noted with the TKR group suggesting difficulty with stabilization during the transition of side to side motion with rotation.

The PCA model was utilized to compile 29 variables selected from the two tasks in order to identify related performance variables for all groups. Interestingly, no dominant concepts were derived from the analysis; rather, the percentage of variables loaded up to 12 PCs to achieve 80% of the explained variation. Temporal variables dominated PC1 when comparing Unstable to Healthy as well as TKR to Healthy. Both the Unstable and TKR groups performed the tasks slower than the Healthy group. While delay or slowed response may be related to a protective response to avoid fall, the hesitation may be associated with information response from the Central Nervous System (CNS). In addition, compensatory movements from adjacent joints were noted in the TKR group such as utilizing torso movements to achieve the TTT rather than bending the knee. Altered mechanics that may be a result of pre-operative habits or a result of surgery, could lead to injuries to other joints. Rehabilitation interventions should address symmetrical movements with weight transfer and rotation in order to avoid future injuries.

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Chapter I

Introduction

1. Significance and Overview

Over 700,000 individuals in the United States receive a total knee replacement (TKR) each year[1] with a 673% predicted increase by the year 2030[2]. A majority of knee recipients report improved function and reduction of pain[3], however, some individuals are not able to return to a desired activity such as gardening, golf and kneeling[4]. Contrary to belief that certain activities are restricted by doctor's orders[2], the decision to not return to an activity is self-selected and (related to a feeling of instability in the knee) determined by a perception of an abnormal feeling in the knee[5]. Stability of the knee is difficult to define and therefore difficult to measure. However, when associated with function, the subjective complaints are directed toward an inability to perform an activity. Furthermore, kinematics of the knee and hip are altered when instability is combined with an unexpected change of direction[6-8]. Poor recruitment of the quadriceps may also contribute to the altered kinematics[6,9,10]. Therefore, in order to analyze rotary instability of the knee, it is pertinent to explore the medial/lateral and rotational planes of movement related to function.

Instability of the knee is difficult to define and has been described as either clinical or functional 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 defined by measurements of kinematics and amount of excursion/translation at the knee joint while performing an activity[8,9,11]. Some researchers propose that the feeling of instability may be a result of altered kinematics of the knee; yet, kinematic analysis was measured during activities primarily performed in a sagittal plane. Byrne et al (2003)[12] postulated that the risk of knee instability occurs in a medial and lateral direction, resulting in a 'giving way' sensation, loss of

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balance or fall. This side to side motion may be similar to a cutting motion that would be involved in pivoting or in Activities of Daily Living (ADL's) such as bending, transferring weight side to side, and rotating while doing laundry. Measuring kinematics during functional activities that involve rotation would provide an increased understanding of rotary instability of the knee and the possible compensatory motions that result while transferring weight to the lower extremities.

2. Stability

Knee instability, defined by the subjective report of 'buckling' or 'giving way' occurs in TKR individuals and is the leading factor for TKR revision surgery[13-16]. Although repeatable kinematic measurements are difficult to capture to define knee instability, the relationship of excessive rotation or translation of the tibial-femoral joint may contribute to a feeling of instability especially during pivoting. Response to rotary tasks following ACL reconstruction constitutes the bulk of research due to the correlation of pivoting and sport-related injuries in this population. Ristanis et al (2003)[17] investigated functional dynamic stability in individuals with ACL reconstruction compared to healthy individuals. Participants descended 3 steps and performed a pivot turn at the bottom of the stairs described as a cross over step technique. Both the ipsilateral and contralateral legs were analyzed. The reconstructed knee had significantly more internal-external tibial rotation (21.68±5.7) during the pivoting maneuver when compared to their contralateral leg (18.63±5.6) and both (left and right respectively) legs (19.01±6.7,19.14±6.8) of the healthy individuals. The intact (contralateral) knee was similar in tibial rotation to the healthy individuals. Although the ACL reconstructed group had a greater tibial rotation, the clinical tests (Lachman's, pivot-shift) were negative. They also reported no feeling of instability with activity and the anterior displacement measurement with the KT-1000 was not suggestive of laxity (less than 3mm side to side comparison). Tashiro et al (2009)[18] found no correlation between the knee arthrometer measurements and rotary instability and supported the use of the pivot-shift

test. Despite the findings in each of these studies, neither of these assessments are utilized when assessing a TKR patient. It is possible that reliance of mechanical stability of the TKR is on the implant design; specifically the geometry, and functional stability may be associated with pre-operative activity and quadriceps strength.

3. Function After a Total Knee Replacement (TKR)

Due to the increasing physical activity of the aging population, total knee surgery has become common in individuals approaching sixty years old. This population may have expectations of maintaining an active lifestyle into their retirement and therefore expect positive outcomes after surgery. It has been shown that high expectations result in positive outcomes[19], however most patients' expectations are not met until five years after surgery[20]. Benedetti et al (2003)[21] reported TKR individuals had significantly decreased function 24 months post-op compared to functional scores at 6 and 12 months post-op. Limitation of activity may be viewed as an unsuccessful outcome of the TKR surgery if the patient's expectations are not met[4,20,22,23]. Expectations important to the patient include return to ADLs and sport. These particular activities include high load rotary tasks. Individuals that undergo TKR expect to return to high load rotary tasks; however some are unable to perform these tasks without a feeling of instability. It is unknown why some individuals are able to cope, despite the possibility of mechanical instability; while others cannot perform these tasks.

Stair climbing has been utilized as a functional outcome measurement post-TKR as measured in the Stair Climbing Task (SCT)[24]. Stair climbing, both ascent and descent, requires quadriceps control and knee stability in order to avoid falls. A pre-operative predictive measurement of utilizing the handrail during stair ascent and descent showed those individuals still required the handrail to perform stairs up to two years after surgery[24]. Persistent decreased quadriceps strength was found in those that continued to use the handrail, while those who

increased quadriceps strength from three months to two years post-TKR did not require the handrail.

Pain in the contralateral leg up to three years post-TKR has proven to limit functional activities as well[25]. Alteration in kinematic and muscle activity has been found in both the affected and unaffected legs post-TKR[26-28]. Increased dependence in the opposite leg could result in future TKR for the contralateral knee. Approximately 40% of those with a TKR will have a TKR on the on the non-operated knee within 10 years[25,29]. Reliance on the opposite limb during functional activities is one strategy that TKR individuals utilize that may result in arthritic changes to the opposite knee or adjacent hip joints[30]. This transfer of weight may be due to quadriceps weakness on the affected leg or pre-operative habits. For example, during stair descent, TKR individuals reduced the work for the affected knee by compensating with the hip and unaffected leg[26]. Increased awareness of this compensatory pattern by both the patient and rehabilitation specialist could reduce the possibility of surgery on the unaffected knee.

4. Strategies

Strategies have been identified in TKR individuals during functional activities; however it is uncertain if these movements are a result of the surgery or a continuation of pre-surgical patterns. Obvious strategies have been noted such as slower gait and decreased stride length[31,32], while subtle changes in the trunk and hips may impact the transfer of forces to the lower extremities[33]. Anticipatory Postural Adjustments (APAs) describe movements of the trunk and muscle activation as a reaction to predictive balance changes, while Compensatory Postural Adjustments (CPAs) deals with the perturbation and activation of movement strategies to restore balance[34]. In addition to these postural adjustments, quadriceps avoidance and changes in kinematics and kinetics of the lower extremities are the most commonly employed strategies utilized by TKR individuals.

4.1 Postural Adjustments

The combination of electromyographic (EMG) activity and center of pressure (COP) displacement provides a measurement of postural adjustment, or APA, during a functional activity that involves perturbation. Age related changes have been found that elderly individuals had a larger EMG magnitude and larger COP displacement with CPAs during a predictable lateral perturbation when compared to younger individuals[34]. This may be due to anxiety of falling or overreacting to decreased balance control. Despite this increase in EMG activity, improved performance of the task was not observed[34]. The elevated muscle activity may be a feedback response and used as a preparation technique for the next movement. Further, postural adjustments assist in proprioception which is driven by the Central Nervous System (CNS). This internal processing provides feedback for positioning and prevention of falls. While difficult to collect data on CNS processing and reaction, both EMG and motion analysis provide data on kinematic changes at the trunk and lower extremities (Appendix)[26,33,35]. Mandeville et al (2008)[36] identified a smaller shift in center of mass and COP for individuals with a TKR when compared to healthy matched controls during level walking and obstacle avoidance. Alteration in movement patterns or a shift of COP can lead to compensatory adjustments in other joints such as the hip and ankle.

4.2 Quadriceps Avoidance

The term, quadriceps avoidance, describes a gait pattern wherein an individual maintains and extended knee during weight acceptance; therefore, avoiding the use of the quadriceps to stabilize the knee[37]. This gait pattern has been observed in individuals post-TKR and those suffering from osteoarthritis (OA)[21,28,38]. Pre-operative quadriceps strength has been correlated with function outcomes following TKR[30]. Individuals with poor quadriceps strength prior to surgery demonstrated increased difficulty with stair climbing and descending 1 month after surgery. The loss of quadriceps strength after TKR has been reported as high as 64%

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deficiency 3-4 weeks post-surgery[39] and the strength may never reach pre-surgical status or comparable to healthy or the unaffected leg[40]. Zeni et al (2010)[24] reported quadriceps weakness of both the affected and contralateral leg at 3 months and two years post-TKR. Persistent quadriceps weakness increases the likelihood for the use of the quadriceps avoidance strategy during functional activities. It is unclear if weak quadriceps formulate the need to develop a strategy or if the lack of loading the involved leg results in poor muscular strength of the quadriceps to assist in knee control.

4.3 Kinematics and Kinetics

Quadriceps strength and previous activity level of the individual may contribute to achieving flexion activities post-surgery such as stair climbing and squatting. Knee flexion angle on initial weight bearing for loading will affect the force production of the quadriceps and also impact the load on the implant itself. Consensus of the literature is that individuals with a TKR perform differently than the controls in regards to knee flexion angle in gait[38]. Overall, TKR individuals walk with less knee flexion during gait and demonstrated reduced flexion during the loading phase of gait when compared to controls. The collective review[38] summarized that TKR individuals also demonstrate a more sagittal moment pattern as opposed to a biphasic moment pattern in gait. This means that TKR individuals either show a pattern of quadriceps avoidance or quadriceps overuse. In other words, these individuals either use a flexion moment (quadriceps overuse) or an extension moment (quadriceps avoidance) pattern throughout the stance phase of gait. Hilding et al (1996)[41] utilized roentgen stereophotogrammetry to analyze migration of tibial and femoral components of TKR individuals in a retrospective study. Individuals were divided into 2 groups 'stable' and 'unstable' with use of the roentgen stereophotogrammetry analysis (RSA) and gait analysis was compared across groups. Those in the 'unstable' (predicted risk of component loosening) group only differed from the 'stable' group in sagittal plane peak flexion moment. The 'unstable' group demonstrated a larger peak flexion

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moment at three different time periods: pre-TKR surgery, six months post-TKR and two years post-TKR. This was statistically significant (p=0.011) when all three time periods were analyzed together as repeated measures compared to the 'stable' group. Due to the fact that the findings were significant both before and after TKR surgery, conclusions cannot be made that this alteration in kinematics is a direct result of the TKR surgery. Although full range of motion (ROM) is highly desirable post-surgery for TKR individuals, evidence has yet to be established if this will carry over to improved gait or lack of compensatory patterns, as just mentioned. Other considerations must be included in the equation such as quadriceps use/strength and neuromuscular firing for proper timing of muscle contraction.

Knee joint loading differs per activity and varies with change in knee flexion angle. The load can be described over time (frequency), such as walking, or as a single point in time as you would depict in a squat or sit to stand. Although the load may vary with each of these activities, frequency impacts the wear on the implant. Therefore, it is important to identify strategies utilized by individuals with a TKR and relate those to the probable impact it has 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 intervention in rehabilitation and implant design. Studies have utilized force transducers in the actual implant to collect specific knee loading[42-44]. Activities such as walking, squatting, stair ascent and descent and rising from a chair exceed two times body weight[43]. Vertical GRF is the most common load analysis utilizing force plates or a gait mat for collection of data. Yoshida et al (2008)[10] found a significant difference in 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 (2008)[43] 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 (2008)[43] conducted the study 1.5 years after surgery, while Yoshida (2008)[10] found differences in 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.

Due to the large number of variables that could possible effect performance in a rotary task, we elected to utilize the Principle Components Analysis (PCA) in order to combine variables together, including kinematic and kinetic variables, and analyze their relationships. Correlative variables would provide insight on how they either move together or work in the opposite direction; thus, providing a mechanism for explaining compensatory movements or strategies.

5. Relationship of Stability and Strategies

Recommendations for return to activity post-TKR are varied considering the risk of improper 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, may assume movement patterns that produce improper loading to the prosthesis or transfer loading forces to other joints in order to compensate. The culmination of kinematic data and kinetic data in this study will provide a profile of movement patterns that identify strategies utilized during rotary tasks. This information will be valuable for implant design, surgical approach and rehabilitation intervention. Specific rehabilitation, as well as proper implant design may increase the longevity of functional use for the TKR recipient and possibly reduce the need for TKR revision surgery.

6. Purpose of Study

The purpose of this study is to better understand the concept of rotary instability in TKR individuals. In order to do this, we will identify strategies utilized by individuals with TKR, and

individuals with non-surgical knee pathology that complain of 'instability', while performing two rotary tasks and compare them to a healthy group. The variables will be analyzed as characteristics of these strategies. Further analysis of these variables will differentiate the performance of the groups per task and lastly, analysis of the culmination of these variables via PCA will identify compensatory movements that are correlated and explain a particular strategy. This data will assist in component design and rehabilitation intervention.

Chapter II

Analysis of rotary task following total knee arthroplasty: Stair descent with a cross-over turn

Linda M Denney, Lauren A Ferris, Hongying Dai and Lorin P Maletsky

Abstract

Leg loading and knee angle differences have been reported in total knee replacement individuals during straight gait, however, little is known about the impact on the knee during turning. Rotary motions may be difficult following total knee replacement surgery; therefore, some individuals may develop strategies, or utilize pre-surgical strategies in order to maintain function. The primary aim of this study was to identify differences in individuals with a total knee replacement as compared to their healthy counterparts during stair descent followed by a cross-over turn. Ground reaction force, knee angle and moments were recorded on 10 total knee replacement and 12 healthy individuals during stair descent followed by a turn and compared to walking straight. Variables were analyzed for the affected, unaffected and healthy knees during the gait cycle. On initial contact, the total knee replacement group had less ground reaction force on the affected leg compared to the unaffected leg ($p = 0.021$) and had delayed contact ($p = 0.044$) and a slower loading rate ($p = 0.020$) compared to healthy group. During mid-stance, the affected leg had less ground reaction force compared to the healthy ($p = 0.049$). The affected stance leg had less knee flexion during mid-stance in both the straight trial ($p = 0.002$) and turn ($p = 0.010$). Moment differed between straight and turn trials but not between groups. Stair descent with or without a turn was approached in a precautionary manner by individuals with a total knee replacement. Slow approach, reduced impact and weight-bearing with a more extended knee on the affected leg may suggest a protective strategy to avoid risk of fall.

Introduction

Functional activities frequently require turning and pivoting in order to change direction, negotiate obstacles and/or react to a perturbation[45]. Various turning and pivoting maneuvers are documented in the literature and are not diagnosis-specific[46]. Further, measurement of knee rotation varies in the literature which leads to inconsistent conclusions. This is due to the lack of accurate method to capture knee rotation with external markers. Zurcher et al. (2008)[46] utilized a Femoral Tracking Device developed by Houck et al. (2004)[7] in order to lower the incidence of error. Zurcher et al. (2008)[46] analyzed healthy individuals rising from a chair and either walking forward, sidestepping to turn or performing a cross-over turn. A greater combined axial knee rotation was recorded with both turning tasks when compared to rising from the chair and walking forward. Similarly, greater knee rotation was documented on the operated knee when compared to the contralateral knee and healthy group in individuals with anterior cruciate ligament reconstruction when performing a cross-over turn after stair descent[17]. The combination of rotary forces and weight bearing affect the knee in propulsive activities, such as a turn, but may also occur while standing with the feet in a fixed position. A novel reaching task described by Ferris et al. (2013)[47] incorporated knee rotation while weight bearing for individuals with a TKR. The TKR individuals utilized a more extended knee during this task and unloaded the affected knee when compared to the contralateral knee and healthy group. The task involved rotation yet loading was not as great as the compressive forces and acceleration involved in stair descent[48]. Moreover, knee rotation, in combination with the GRF, produces a moment at the knee. Moment itself is not a sole determinant of how the knee will perform during turns as it is influenced by other aspects such as muscle control, knee geometry, pain response and possibly joint laxity.

Stair climbing may not be an everyday activity for individuals, yet training how to negotiate stairs is included in TKR post-surgical rehabilitation. Stair descent may be a greater

concern for both the patient and care provider, since this requires greater quadriceps control[49] and GRF can exceed two times the body weight[50]. Individuals with poor quadriceps strength prior to TKR surgery demonstrate increased difficulty with stair climbing and descending one month after surgery[30].

The Stair Climbing Test (SCT) has been highlighted as a functional test for individuals following TKR and can be utilized as an objective measurement when associated with time[10,49,51,52]. Improved time is recognized as a positive functional outcome, however, an added cross-over turn after the stair descent could reveal an individual's willingness to stand-pivot on their surgical leg. Furthermore, this additional rotational task may reveal compensatory strategies that an individual may not demonstrate when asked to simply descend the stairs. Stair descent followed by a turn can be especially challenging as it involves the combination of high joint forces, acceleration coupled with rotation.

The purpose of the study was to investigate the performance of a cross-over turn after stair descent in individuals with a TKR. This task requires stability during single stance due to the combination of deceleration and rotation at the knee. An objective understanding of how individuals with a TKR control a rotary task when combined with compression forces could assist rehabilitation intervention and possibly alter implant design. The objective of this study was to identify differences in individuals with a TKR as compared to their healthy counterparts during stair descent followed by a cross-over turn.

Methods

Twenty-two subjects participated in this study: 10 TKR and 12 healthy subjects (Table 1). Subjects 50-75 years old were pre-screened for previous hip or ankle surgery, peripheral neuropathy, and a BMI greater than 30. The TKR subjects were screened to confirm descent of

stairs without use of a hand rail, bilateral knee flexion at least 90°, no previous surgery on the opposite knee, and no unicompartmental knee implant or a bilateral knee implant.

Subjects signed a consent form approved by the Human Subjects Committee and Institutional Review Board at the University of Kansas Medical Center (KUMC) and an approval to be videotaped and photographed. All procedures were in accordance with the Declaration of Helsinki. The average of three trials of quadriceps strength of each leg was recorded by a handheld dynamometer (MicroFET 2, Hoggan Health Industries; West Jordan, Utah). 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).

Twenty-four reflective markers (25 mm) were applied with double sided adhesive to specific landmarks. The markers were captured at 120 Hz with an infrared six-camera motion analysis system (Vicon 512, Oxford Metrics, Oxford, UK). The knee axis was determined by a Knee Alignment Device that identified the x, y, and z configuration of the knee joint[53]. The Vicon was calibrated according to the manufacture's specifications. A force plate (AMTI, Advanced Mechanical Technology, Inc., Watertown, MA) embedded in the floor was used to capture GRF at 360 Hz.

A single static calibration trial was obtained before testing. A set of stairs with four steps (15 cm in height) were configured so that the front of the stairs were flush with the force plate (Figure 1). A total of eight stair descent trials were performed; two trials leading with alternate feet, descending then walking straight, and three trials descending, landing on the force plate with their right foot, then crossing their left foot over to turn right. Three additional trials were collected landing on their left foot and crossing over with their right foot performing a left turn (Figure 2). All trials were performed without use of the handrail and at the subject's own

selected pace. The subjects were allowed to perform as many practice trials as necessary.

Trials were rerun if the subject missed the force plate, crossed in the wrong direction, or did not

keep their planted foot straight upon landing.

GRF, KA, and moment upon ground contact were analyzed as functions of percent

stance. A 4th order low-pass Butterworth filter was applied to all trials. Contact with the force

plate was defined using a threshold method where data was truncated if less than 2% of the

max GRF. GRF was normalized to percent body weight and moment was normalized to body

weight times percent height. The remaining data were scaled to percent stance: 0% preliminary

contact to 100% lift off.

GRF was analyzed against temporal parameters to examine differences between

groups. Time, in seconds, was recorded to achieve passive peak, active peak, and the

minimum between the two peaks in regards to vertical force (Figure 3). Loading rate to passive

peak and the vertical forces observed at these points were also evaluated for differences

between the groups.

Statistical Analysis

Data were analyzed with SPSS Statistical Software (v. 20) and SAS 9.2 (Cary, NC).

Independent t-tests were used to compare GRF, KA and moment for the TKR participants

affected and unaffected limb compared to the healthy controls. Paired t-tests were performed for

these same variables for the affected and unaffected knee of the TKR participants. Non-

parametric tests were used if a lack of linear relationship or unequal variances was found.

A linear mixed model (LMM) was utilized to analyze the trends of GRF, KA and moment

during the stance phase of stair descent before the subject turned or walked straight. Group

main effect (unaffected vs. affected vs. healthy), percent stance effect (0-100%), and group by

percent stance interaction were analyzed. In the LMM, a compound symmetric correlation

matrix was used to model the correlation within subjects over percent stance. If the group or group by percent was significant, a post hoc comparison was performed to compare the three groups at each percent stance. An α = 0.05 determined level of significance.

The effect size was determined with use of 10˚ difference in KA, thus determining the need for at least 10 participants in each group to achieve a power of at least 0.80. Statistical power was achieved for this study with the recruitment of 10 TKR and 12 healthy participants.

Results

All 22 participants successfully completed the stair task. The groups were similar with respect to age, BMI and quadriceps strength (Table 1).

Ground Reaction Force

During the turn, the GRF showed significant differences overall (p=0.049) for the TKR (affected) stance leg when compared to healthy (Figure 4b). The TKR (affected) had a lower GRF during the turn as compared to healthy (72.6±1.2, 75.9±1.1 respectively, p=.049) and significant differences were noted at 10% (82.9±3.7, 96.1±3.4, p= .009) and 20% (106.0±3.7, 122.4±3.4, p=.001) stance for the post hoc analysis. The GRF for the TKR (unaffected) stance leg were lower than the healthy at 10% (83.7±4.1, 96.1±2.3, p=.014) and 20% (110.0±4.1, 122.4±2.8, p=.014).

As shown in Figure 4a, the GRF for the TKR (affected) and healthy had significant group by stance interaction during the straight trials (p=.005), which indicates a lower GRF in the affected group than healthy at 20% (125.7±4.1, 142.5±3.8, p=. 003) and 30% (106.3±4.1, 126.1±3.8, p=.0005) stance. Post hoc analysis revealed significance at 20% and 30% stance for all groups (p=.008, p=.002 respectively) and between TKR (unaffected) and healthy (p=.027, p=.018 respectively).

Temporal Findings

While either walking straight or performing a turn, the healthy group performed the overall task in a shorter cycle time when compared to the TKR group (Table 2). The TKR group (affected) approached foot contact in a delayed manner when compared to the healthy group (p=.044) depicted by the Time to Passive Peak (Table 2). The loading rate (slope) for the affected leg of the TKR group was significantly less than the healthy individuals (p=.020). The TKR group impacted the ground (Passive Peak) with significantly less GRF on the affected leg with the initial foot contact compared to the unaffected leg (p=.021); however, there were no significant differences between the TKR group and healthy group for GRF on Passive Peak. There was a large amount of variation with both the affected and the unaffected Passive Peak when compared to the healthy participants. The TKR group loaded the unaffected limb during single support stance for a longer period of time compared to the healthy during the turn (p=.032). Weight-acceptance was consistently longer than the healthy controls up to the beginning of the push-off (Active Peak) phase (p=.008).

As shown in Figure 5a, the three groups were significantly different with respect to KA of the stance leg while walking straight after descent (p=.002). The TKR (affected) stance leg had significantly less knee flexion when compared to the healthy group at both 30% (17°±2°, 23°±2° respectively, p=.043) and 40% stance (15°±2°, 22°±2°; p=.03). The groups were also significantly different with respect to KA of the stance leg during the turn (p=.01) (Figure 5b). The KA means of the affected group were lower than the KA means of the healthy group for a majority of the cycle. The affected group had a lower KA then the healthy at 30% (16°±3°, 22°±2°, p=.047) stance. The affected group had a significantly lower KA then the unaffected group at 20% (16°±2°, 20°±2°, p=.03), 30% (17°±2°, 21°±2°, p=0.02) and 40% (16°±2°, 20°±2°, p=0.04) stance. Further post hoc analyses reveal significance at 30% stance for all groups (p=.02).

Moment during the stance cycle was not significantly different between the groups while walking straight (p=.61) or turning (p=.66) after stair descent (Figure 6). There was a large variation noted with the TKR group during the push off phase of the cycle on the turn.

Comparison of Straight versus Turn

When comparing the turn trials to the straight trials across stance time, the healthy participants differed more than the TKR group (Table 3). The healthy group had significantly different GRF at various points throughout the cycle, while the TKR group differed at 20- 70% stance. Initial impact was less for the TKR group during the straight trial, while the GRF increased during the turn at 40% for the TKR group and 50% for the healthy.

The affected KA did not differ during the trials, however, both the unaffected and healthy groups showed differences from 70-100%. While the affected group performed the turn or straight trial within 1-3° difference, the unaffected and healthy groups had greater knee flexion with the healthy over twice the amount of knee flexion during the turn.

Although no differences were found between groups for moment, there was a difference when the straight and turn trials were compared. The TKR group had difference in internal moment at the beginning of the cycle (20-40%) changing over to external moment at 30-40% stance cycle while walking straight. The healthy group had differences throughout the entire cycle switching to external moment as early as 30% of the cycle for straight walking and consistently maintained external moment between 60-90% with both straight walking and turning.

Discussion

Turning activities can be difficult following a TKR[4,31], however it is unclear if this is a result of the surgery or strategies developed prior to surgery[28]. The investigators hypothesized that individuals with a TKR would perform the stair task with a change in direction differently

than their healthy counterparts. The TKR participants performed the task at a slower cadence which is consistent with recent literature[21,31,54,55], and, further investigation of the temporal cycle showed a difference in the loading aspect; specifically the GRF.

Hesitation to initial impact and less GRF on the affected leg after stair descent are consistent with kinetic findings during gait, stairs and functional activities. Functional testing performed three months after TKR showed significantly less GRF of the affected leg compared to the unaffected leg, but not at 12 months[10]. Yet, during sit to stand TKR participants loaded the legs symmetrically 1.5 years after surgery[43]. Interestingly, Stacoff et al. (2007)[56] recorded almost twice the body weight on the affected leg during stair descent on individuals 25 months post-surgery without range of motion (ROM) restriction, while the group with restricted knee flexion (< 90°) had less GRF on the affected leg. The researchers attributed the differences in GRF to possible quadriceps weakness and/or the restricted knee ROM. These findings were not consistent with our current study as our participants were, on average, 20 months post-surgery, had at least 90˚ knee flexion and were still demonstrating differences in loading the affected leg. Therefore, GRF differences may be due to the type of task, quadriceps strength or available knee motion rather than the post-surgical time.

The initial loading delay for the TKR participants was similar to TKR participants in the Ferris et al. (2013)[47] study when load transfer to the affected leg occurred later compared to the healthy group during a weight-shifting task. Furthermore, weight acceptance time and load was higher on the unaffected leg similar to our study. Although the weight shift task in the Ferris et al. (2013)[47] study differs from the forward momentum of stair descent; both tasks involve a rotational component of twisting or a turn. The controversy of post-surgical time and task indicates that if improvements are going to be made post-surgery, it may rely on muscle strength development, rehabilitation training and the activity of the individual pre and post-surgery.

Quadriceps avoidance and knee instability have been mentioned as probable causes of variable performance in individuals with a TKR[41,57,58]. Byrne and Prentice (2003)[12] postulated that the risk of knee instability occurs in a medial and lateral direction, resulting in a 'giving way' sensation, loss of balance or fall. Participants in our study did not report symptoms of instability or a sensation of falling despite a request to perform a turn that included alterations in a medial and lateral motion. Knee laxity or lack of stability is normally related to a sagittal plane activity such as ascending, descending stairs or forward walking but could also occur during a rotary task. Rotational instability is difficult to quantify during activities and the clinical measurement is not correlated to laxity in the sagittal plane[59]. Anterior tibial laxity is less controlled by the individual and can be enhanced by a strong quadriceps contraction. This may be the reason why individuals avoid the use of the quadriceps and maintain a flexed knee to avoid a large translation toward full extension. Conversely, individuals post-TKR tend to use a 'stiff-knee' gait in order to assist propulsion[55]. Knee laxity, objectively measured in the sagittal plane, varies with knee flexion angles and is difficult to correlate with rotary tasks.

All participants in our study performed the initial weight acceptance with a normal (0-15°) KA yet the affected group impacted with less flexion, then gradually increased in the stance cycle. Weight acceptance occurs as early as 20% of the gait cycle after initial foot contact and loading (0-10%) and continues through single leg support up to 50%; or when double stance occurs when the contralateral limb makes contact after swing phase[60]. Both GRF (10%-30% stance) and KA [31,55] (30% stance) differed between knees within groups (TKR) and between groups (TKR and healthy) during weight acceptance in straight gait. Differences in KA for straight gait analysis are well documented in the literature[31,38], however, an unexpected finding of no differences in KA of the affected knee during straight and turn may be explained by the protocol. Participants were asked to land with the stance foot pointing straight ahead, leading to consistency in performance regardless of the task to walk straight or turn. The protocol may

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have limited the natural performance of the turn, thus altering any compensatory motion of the foot or hip in anticipation of a turn. Therefore, if the weight acceptance phase of this task was truly observed as a sagittal measurement (prior to the turn), the results could not be interpreted as alteration in the rotary function as described by Markolf et al. (2010)[59].

Pre-operative quadriceps strength has been correlated with function outcomes following TKR[30]. Although Mizner et al. (2005)[30] reported minimal difference of the TKR subjects and healthy subjects at 6 months in the SCT, Dawson et al. (1998)[61] reported 36% of the subjects interviewed at 6 months stated descending stairs was still 'moderately difficult'. Zeni and Snyder-Mackler (2010)[24] highlighted that quadriceps strength in the unaffected (contralateral leg) improves functional status such as stair climbing and descending up to two years post-TKR. This was proven by the SCT which is a calculation of the total time ascending and descending stairs without the use of the handrail. In the current study, ascending was not recorded and the subjects descended the stairs at their own pace. For this reason, results may not coincide with a timed stair test; rather the quadriceps influence may be more obvious in the initial foot contact or GRF results. Initial thoughts in the design of the current study was that the GRF would be greater on the unaffected leg during foot contact due to the possible quadriceps weakness of the affected leg trailing on the last step assisting in deceleration. In fact, the unaffected leg GRF was greater compared to the affected leg during initial contact (Passive Peak) but there were no side to side differences noted with quadriceps strength (Table1). Additionally, pre-surgical measurements were not collected so it cannot be assumed that the pre-testing measurements reflect the status of quadriceps strength pre-TKR. Pre-surgery performance has been correlated to post-surgery function[52] so it is unclear if the current participants developed a pattern or strategy prior to surgery[36,62] or if our results are truly the outcome of the surgery.

It is unclear if weak quadriceps formulates a need to develop a strategy or if the lack of loading the involved leg results in poor muscular strength of the quadriceps to assist in knee

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control. If knee control is compromised in the sagittal plane, then most likely adjustments may need to be made by the individual during rotary motions such as turning. Therefore, it is likely that an individual performing a turn may demonstrate a strategy in order to complete the task with the least amount of effort to stabilize the knee. Strategies have been identified in individuals with TKR during level walking and obstacle avoidance[36], however, this only explains straight-forward walking and little is known about how TKR individuals cope with rotary tasks.

Studies of the impact of rotary forces on the knee are limited with variable conclusions specifically on moment about the knee[21,31,50,55,57]. Moment is a result of the combination of muscular control influenced by center of mass (COM) and is reliant on the KA and GRF. Although the average maximum moment values were not significantly different between the groups, moment as related to stance cycle for straight and turn mirrored findings in other studies[50,57]. The TKR group incorporated external moment early in the gait cycle when compared to the healthy group. This suggests the possibility of a quadriceps avoidance gait which could be due to pain[62], weak quadriceps[57] or a pre-surgical gait pattern[62]. Further analysis will need to be conducted regarding moment that includes muscle involvement and COM in order to make an accurate assessment. A computational model that includes the ankle and hip joint may better represent moment at the knee; specifically during the static weight acceptance or loading response. Our findings of altered GRF, variable KA and changes in moment during the stance cycle represent what is currently found in the literature and support the hypothesis that TKR individuals perform stair descent with a cross-over turn differently than healthy individuals.

Limitations

There are limitations of this study that the reader should be aware in order to avoid generalization to a larger TKR population. Recruitment included individuals who descended

stairs without a handrail, which may have attracted an active population that was not representative of all individuals that undergo TKR. Although elective surgery is becoming more common to maintain an active lifestyle, our intent was to analyze performance during stair descent followed by a turn. The population represented in this study did not complain of difficulty with the task, whereas individuals who require use of a handrail may have reported a feeling of instability or demonstrated a compensatory strategy in order to achieve the task. The BMI restriction of less than 30 may not represent the TKR population, especially those that may have difficulty with stair descent and rotary motions. However, a recent study[63] compared clinical and radiographic outcomes of TKR patients who were obese and non-obese. Overall there were no significant differences in implant survivorship and Knee Society objective and functional scores. Further, Stickles et al. (2001)[64] reported obese patients had difficulty with stair ascent and stair descent when compared with non-obese TKR patients. Both studies agree that overall satisfaction was not different between the groups but function may be impaired. Neither of these studies performed a pivot with stairs so it is difficult to conclude that obese TKR patients would have the same performance as non-obese TKR patients during a biomechanical analysis of descending stairs with a turn. Therefore, it is unknown how the obese TKR patient may perform a rotary task and should be considered as future research for information regarding risk of aseptic loosening. Again, active individuals are seeking elective surgery earlier rather than giving up an active lifestyle. Analysis of the active population is beneficial to evaluate the response to rotary demands without other comorbidities. Lastly, the use of external markers for 3-D analysis introduces a greater chance for error when assessing knee rotation, thus moment was calculated via the force plate.

Conclusion

This study concludes that individuals that undergo TKR perform a rotary task, such as a turn after stair descent, differently when compared to healthy individuals. Individuals with a TKR

have an overall slower rate of performance in turning after descending stairs and demonstrate a lower initial loading rate on foot contact when compared to the contralateral leg and a their healthy counterparts. Individuals with a TKR step down with a lower GRF following stair descent and tend to maintain a more extended knee when compared to their contralateral leg and healthy controls. It is difficult to conclude if the differences are a result of a habitual pattern developed prior to surgery or adapted due to post-surgical variables such as quadriceps function, ROM, or an independent feeling of instability.

Author contributions

Linda Denney contributed to project conception, data collection, data analysis, data interpretation, and draft and submission of the manuscript. Lauren Ferris contributed to data collection, data analysis, data interpretation, drafting and editing the manuscript. Hongying Dai contributed to statistical analysis of data, data interpretation, and drafting and editing the manuscript. Lorin Maletsky contributed to project conception, data analysis, data interpretation and editing the manuscript.

Tables and Figures

Table 1: Demographics of participants

	Number of Subjects	Age	BMI	Gender	Post-Op in months	Quad Strength in %BW	
						Affected:	Unaffected:
TKR	10	66(6)	26.9(3.0)	30% F	20 (16)	0.18 (0.03)	0.19 (0.05)
Healthy	12	65(8)	24.6(3.8)	58% F	N/A	0.19 (0.03)	

BMI: body mass index; BW: body weight; TKR: total knee replacement; Aff: affected; Un: unaffected F: female.

Table 2: Average (SD) of total time of tasks after stair descent, temporal parameters and loads applied at various events during turning.

		Healthy	Unaffected	Affected	p value
Cycle time (s)	Straight:	0.63 (0.06)	0.72 (0.06)	0.73 (0.08)	0.002‡ 0.003†
	Turn:	0.78 (0.08)	0.91 (0.12)	0.94 (0.19)	0.007‡ 0.021†
Passive peak (%BW)		137.10 (10.11)	124.72 (19.82)	119.17 (21.68)	0.021*
Time to passive peak (s)		0.15 (0.03)	0.18 (0.05)	0.19 (0.06)	0.044†
Loading rate (%BW/s)		13.89 (2.60)	11.90 (3.43)	10.69 (3.18)	0.020†
Minimum between peaks (%BW)		82.45 (4.14)	82.84 (6.98)	83.96 (7.09)	0.877‡ 0.562†
Time to min between peaks (s)		0.39 (0.06)	0.46 (0.08)	0.42 (0.11)	0.032‡
Active peak (%BW)		102.28 (6.85)	102.85 (7.24)	102.33 (8.77)	0.439† 1.000‡
Time to active peak (s)		0.06 (0.06)	0.70 (0.09)	0.67 (0.10)	0.008†

s: seconds; BW: body weight; min: minimum.
* Indicates a difference between the unaffected vs affected legs
† Indicates a difference between the affected vs healthy legs

‡ Indicates a difference between the unaffected vs healthy legs

Table 3: Mean differences in ground reaction force (%BW), knee angle (degrees) and moment (%BW x %height) during the stance cycle between the groups: affected, unaffected and healthy when comparing turn versus straight.

Group	Task	0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Affected	GRF	0.31	8.57	**-19.75**	**-10.40**	**5.93**	**8.13**	1.71	**-6.13**	-3.18	2.03	-0.26
	KA	-0.61	0.76	1.41	0.20	0.76	1.35	1.14	1.31	1.88	2.00	2.44
	Mz	-0.15	4.83	**9.24**	**10.97**	**9.60**	3.74	0.97	2.65	8.71	10.74	0.31
Unaffected	GRF	0.34	8.65	**-20.80**	**-14.09**	**6.14**	**9.99**	**4.23**	**-3.82**	-0.98	5.56	-0.30
	KA	-2.09	-1.65	0.94	-0.62	-0.22	1.00	1.71	**3.67**	**4.94**	**6.01**	**4.66**
	Mz	0.71	1.74	**7.40**	**10.01**	**8.45**	3.14	1.28	1.49.	4.25	4.50	0.43
Healthy	GRF	**-0.56**	**11.89**	**-20.14**	**-22.42**	-4.92	**12.65**	**10.22**	-0.08	-2.75	-2.13	**-0.46**
	KA	0.00	0.22	2.09	0.54	-0.66	1.37	4.02	**6.48**	**7.70**	**8.07**	7.07
	Mz	-0.12	-1.33	**5.84**	**8.89**	**10.93**	**8.11**	**7.11**	**10.59**	**16.17**	**11.54**	0.32

BW: body weight, GRF: ground reaction force; KA: knee angle; Mz: moment. The values in **bold** indicates significance.

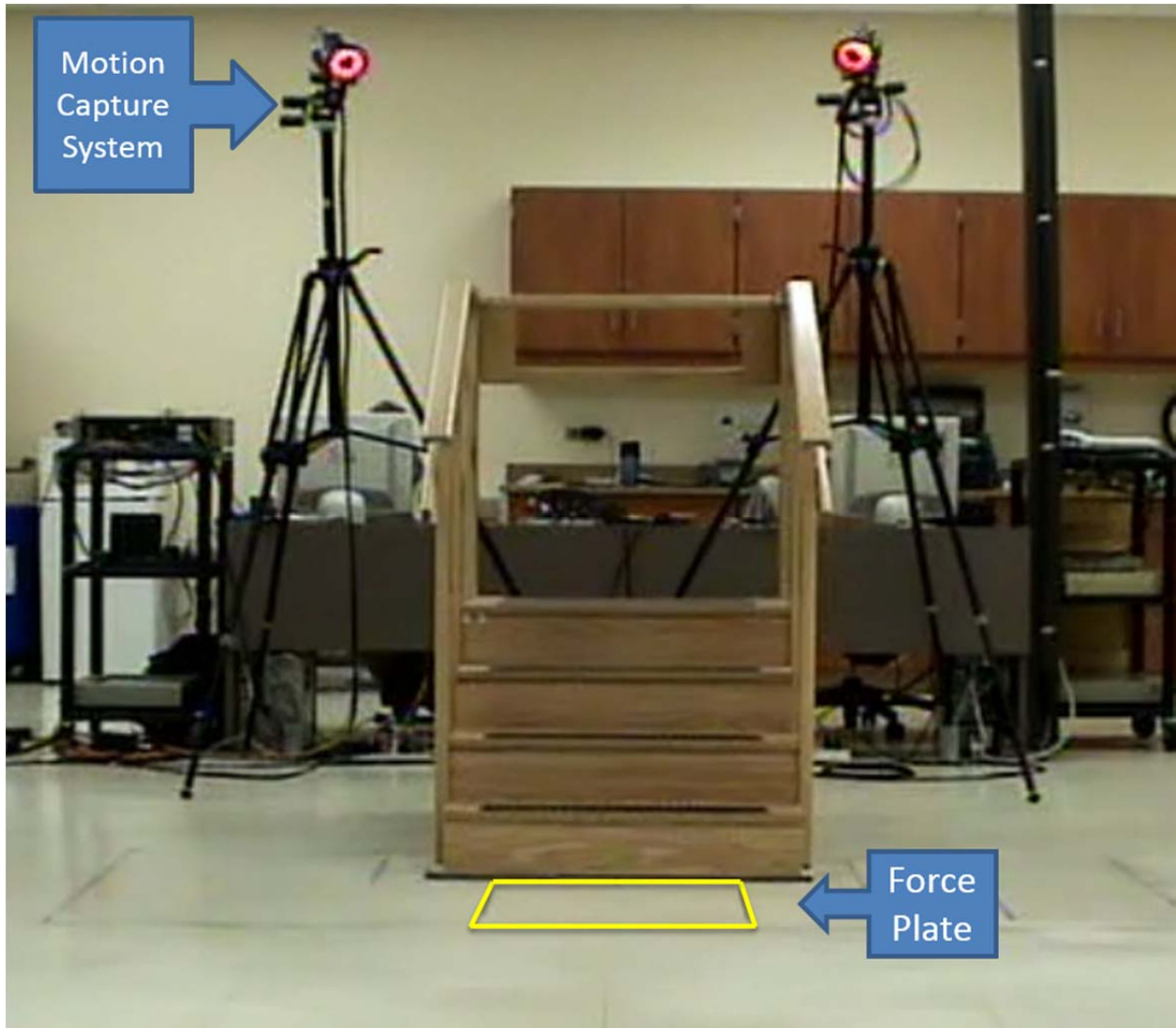**Figure 1: Stair task setup with force plate at the bottom of the stairs.**
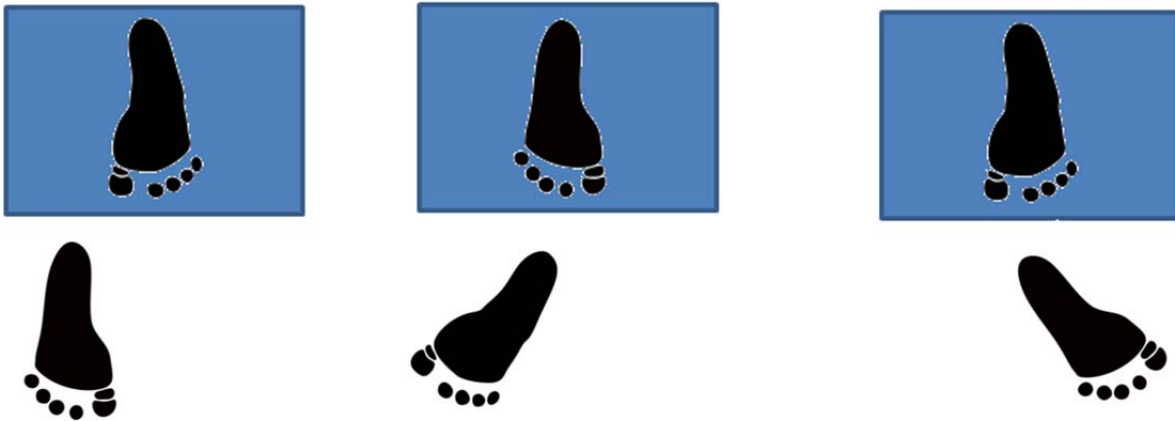
Figure 2: Footprints indicate stepping on the force platform at the bottom of the stairs followed by walking straight (left image), right turn (middle image), and left turn (right image).
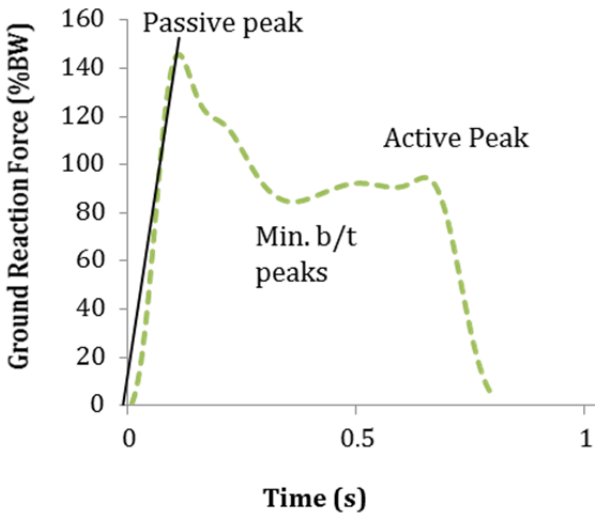
Figure 3: Ground reaction force of a healthy subject who performed stair descent with a pivot at the bottom of the stairs.
BW: body weight.

Figure 4: Average GRF for TKR and healthy groups: affected (red), unaffected (blue), and healthy (green) during the stance phase after stair descent followed by (a) walking straight and (b) turning.
BW: body weight
*Significant difference between affected and healthy groups.
+Significant difference between unaffected and healthy groups.

Figure 5: Average knee angle for TKR and healthy groups: affected (red), unaffected (blue) and healthy (green) during the stance phase after stair descent followed by (a) walking straight and (b) turning.
*Significant difference between affected and healthy groups.
‡Signifcant difference between affected and unaffected groups.

Figure 6: Average knee moment for the TKR and healthy groups: affected (red), unaffected (blue) and healthy (green) during stance phase after stair descent followed by (a) walking straight and (b) turning.
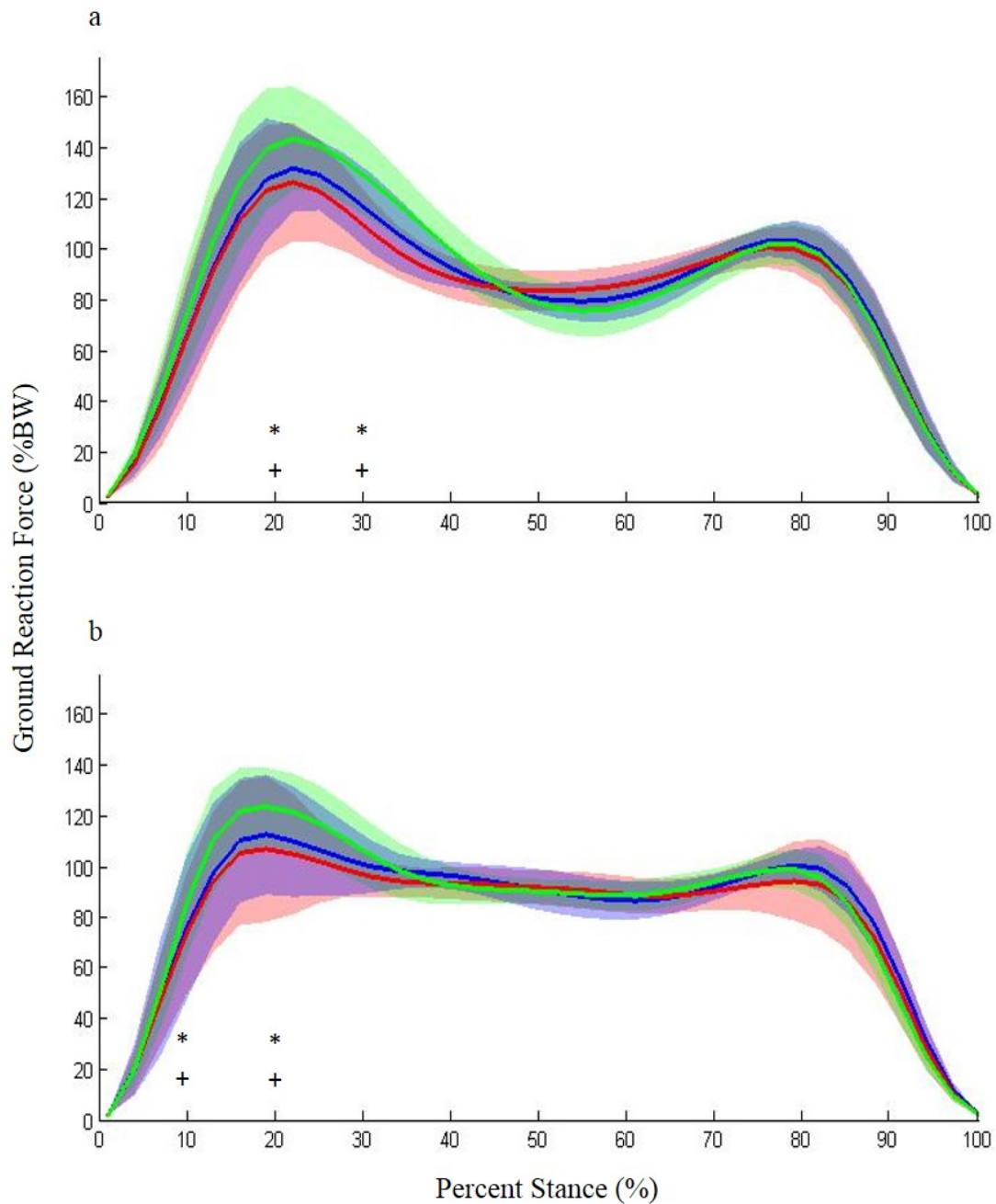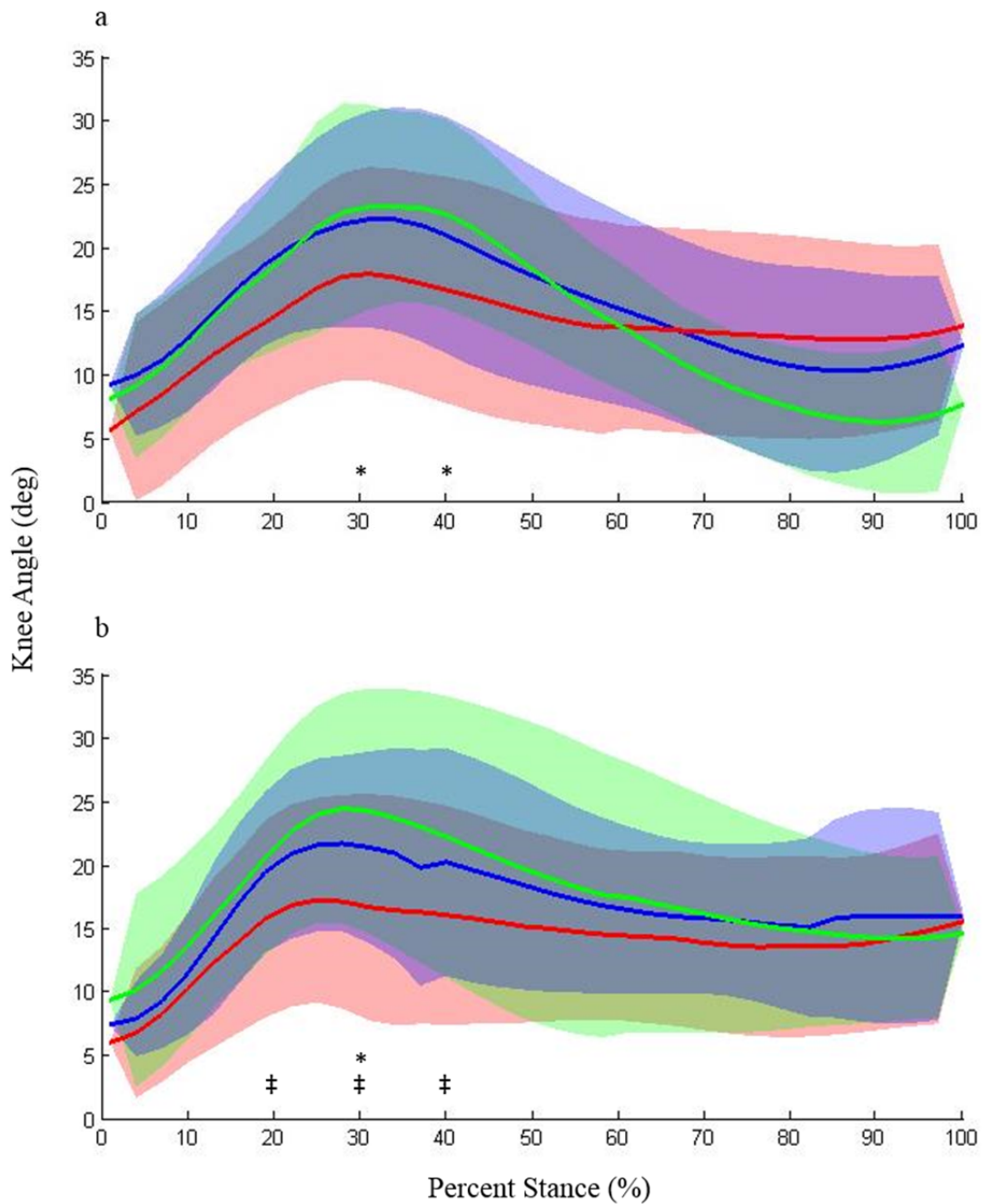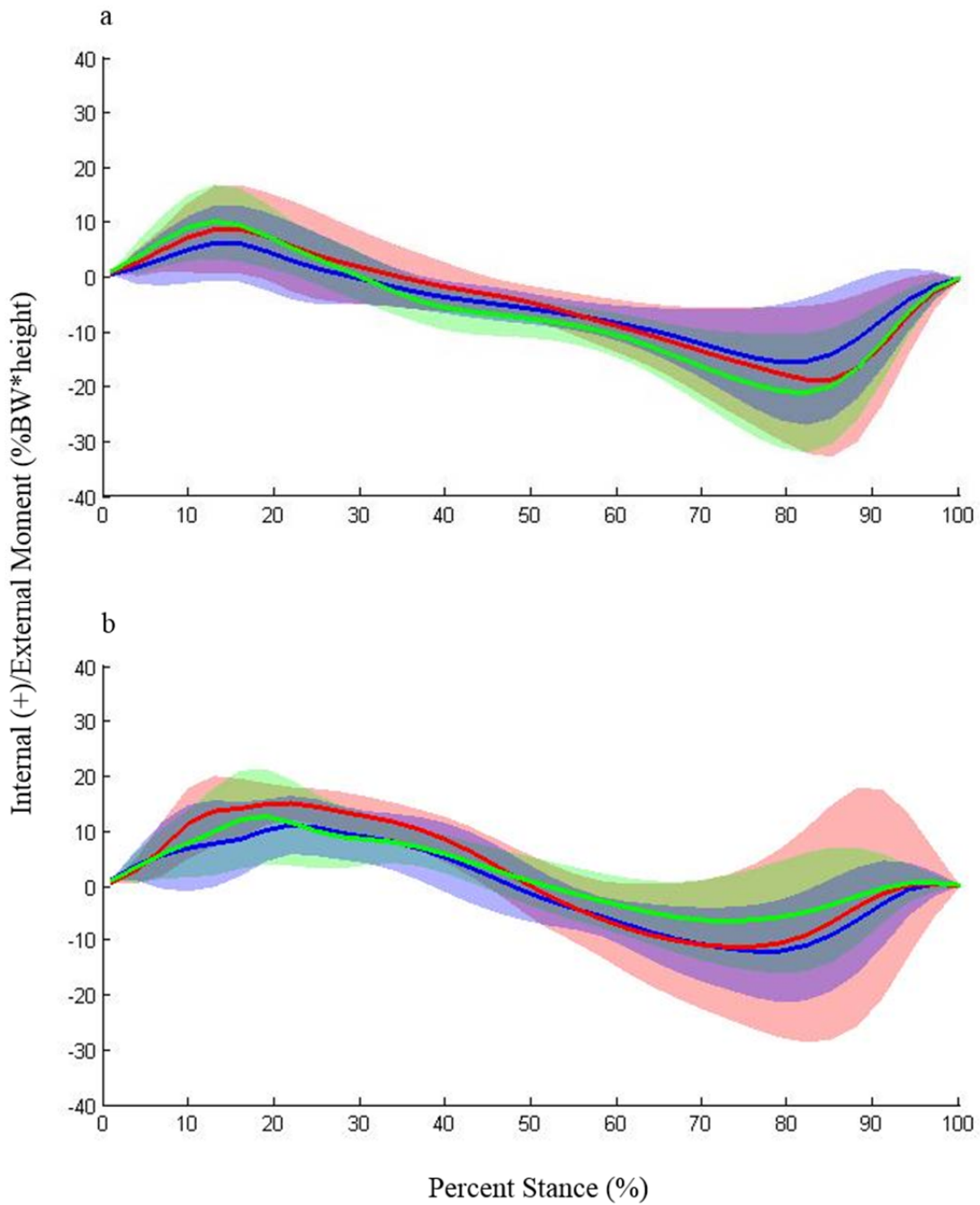**BW: body weight.**

Chapter III

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

Lauren A Ferris, Linda M Denney and Lorin P Maletsky

Abstract

Background: 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, non-conforming 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.

Method of Approach: 23 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 (LFR) based on the total force. Knee flexion angles were also measured. **Results:** 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. **Conclusions:** Large variations in LFR at mid-task in TKR subjects suggested possible difficulties in maintaining positional stability during these tasks. The TKR subjects maintained more of an extended knee, a consistent quadriceps avoidance strategy seen by other researchers in different tasks. These outcomes suggest 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 return to activities of daily living that involve flexion and rotation.

Introduction

Over 600,000 total knee replacement (TKR) surgeries are performed annually[65] 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[66]. 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[67-69] and yet, is poorly defined.

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 kinematics and amount of excursion/translation at the knee joint while performing an activity[8,9,11]. 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 occur primarily in the sagittal plane. Byrne et al. (2003)[12] 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. Pre-gait strategies, pain and/or gait velocity may influence knee loading post-TKR[38,62], while sit to stand performance is confounded by quadriceps strength resulting in weight bearing asymmetry[70]. Knee joint loading differs per activity and varies with change in 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 effects the wear on the implant. It is important

to identify strategies utilized by individuals with a TKR and relate those to the probable impact it has 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[42-44]. Activities such as walking, squatting, stair ascent and descent and rising from a chair exceed two times body weight[43]. Treadmill walking generated a lower force than normal level ground walking (2.0 x BW versus 2.6 x BW, respectively)[71]. 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[72,73].

Vertical ground reaction force (GRF) is the most common load analysis utilizing force plates or a gait mat for collection of data. Yoshida et al. (2008)[10] found a significant difference in 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. (2008)[43] 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 (2008)[43] conducted the study 1.5 years after surgery, while Yoshida (2008)[10] found differences in GRF between the surgical leg and the contralateral leg only at 3 months post-TKR but not at 12 months post-TKR. This indicates 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. (2008)[36] supports the fact that individuals may have already developed a strategy pre-surgery with control of center of mass (COM) within base of support (BOS) 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[36]. 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 analysis 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 as well as transferring a load such as picking up a child or laundry basket. Despite the desire to return to recreational sports reported by TKR recipients[2,71,74,75], 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) (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. TKR subjects were further screened for individuals who could flex both knees at least 90° and had a unilateral TKR. Subjects were not screened based on type of knee implant and the extent of participation in rehabilitation.

Each subject signed a consent form approved by the Human Subjects Committee (HSC) and Institutional Review Board (IRB) 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)

(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 (Table 1). A KT-1000 measured the

anterior translation of the tibio-femoral joint in both knees (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, 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[53]. In an accuracy study for upper body range of motion by Henmi

et al., Vicon 512 motion tracker had a reliability of $< 3°$[76]. Since the accuracy of the angle

measurements are reliant on the placement of the KAD, it can be concluded 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 GRF at 360 Hz. Force plate and analog

devices were wired into the Vicon's A/D board and triggered simultaneously by 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 midline of the subject. The subject reached across his body

with his right hand and placed a magnet at his maximum reach, then repeated 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) (Fig. 1) two microphone stands were placed at the recorded white board 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 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/her body and hit the button next to his/her shoulder, switched hands and crossed to hit the button next to the opposite knee (Fig. 2). The subject then stood up, pressed the button on the same side by his/her 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/her 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.

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. 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. 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 leg. 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. LFR for the affected and unaffected, or healthy were averaged and standard deviation was calculated for each subject at each percent button-to-button movement. Average and standard deviation of the LFR parameters for H2L and L2H was calculated for all subject groups and a single-factor ANOVA was performed ($p<0.05$). 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 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. Average and standard deviation of the knee angle parameters for H2L and L2H was calculated for all subject groups and a single factor ANOVA was performed ($p<0.05$). A single-factor ANOVA was ran 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 was 0.71 and maximum LFR for the affected was 0.69 (Table 2). For H2L the healthy subjects started with less LFR on the lead leg (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 (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 (Fig. 3A). In contrast, the L2H LFR for the TKR subject's limbs generally were less than the healthy subject, but had the greatest variation (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) (Fig. 3C). No statistical differences were found between the three groups at 0.5 LFR for both H2L and L2H tasks (Figs. 3A, 3C). No statistical difference was found between the groups for the LFR in both H2L and L2H tasks, although there was a strong trend of statistical difference between the unaffected and healthy subjects during H2L (p=0.057) (Fig. 3A) and between the affected and unaffected subjects during L2H at maximum LFR (p=0.077) (Table 2).

For both tasks the healthy subjects had a greater knee range of motion throughout the movement sequences (Figs. 3B, 3D) and greater maximum knee flexion (Table 2). The healthy and unaffected also had a significantly greater goniometric pre-measurement knee range of motion than the affected (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° during the H2L, the unaffected and affected knees only moved a total of 10.5° and 10.3°, respectively. 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 (Fig. 3B). Strong statistical trends were also observed at maximum knee flexion for the H2L task between the affected and healthy (p=0.055) and the unaffected and healthy (p=0.058) (Table 2). 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° (9.3) when they passed 0.5 LFR, the unaffected and affected knees were only flexed an average of 17.8° (8.8) and 14.7° (11.5), respectively.

The L2H tasks displayed the greatest amount of knee angle variation between the three groups (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. 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[77]. 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° of flexion without the added rotation[78,79]. It is also possible that cognitive planning is ongoing after movement initiation evidenced by the large variation mid-task[80], 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 his 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[80]. 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 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 (2008)[10] 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 as 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. Analysis of data from the upper torso during the dynamic movements could be analyzed to determine if the TKR participants utilize 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[7,8,12,46]. 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[38,70,81]. Both weight acceptance and knee

flexion excursion during gait were not significantly different in TKR participants at an average of

28 months post-surgery[82]. Further, it is argued that the unaffected limb of the TKR participants

should not be used as a comparison to the involved limb as both limbs are more symmetrical

over time[82,83]. 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 as well as 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[30], yet progressing to symmetrical movements within the

TKR subjects by 19 months in this current study and 28 months as documented by Milner

(2008)[82]. 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[84]. Christiansen et

al. (2011)[70] 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 (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 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. (2012)[85] 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[10]. 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 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 as data (such as quadriceps strength and knee biomechanics) were not captured prior to surgery. Pre-measurements would be beneficial to use as 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[86]. In the current study, the healthy participants consistently flexed their knees more than the TKR participants throughout both H2L and L2H tasks. Notably, the healthy group had twice as much knee flexion excursion when compared to both 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 are they 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° while the TKR group flexed approximately 10° 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 (2008)[82] found little difference between the involved knee (11˚) and uninvolved knee (13˚) for knee excursion during weight acceptance in individuals 28 months post-surgery, while Mizner and Snyder-Mackler (2005)[30] recorded 11˚ in the involved knee and 19˚ 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

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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 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° 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. (2003)[87] 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 after 15 months post-surgery, it can be suggested that 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 hip, for lack of control at the knee. The difficulty of making such statements is that functional outcome measures have been correlated to pre-surgery status (such as quadriceps strength)[30], therefore caution must be taken in interpretation of the study results due to the fact that we did not collect pre-surgical 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[30,83-89]. 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[90,91]. 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 on knee angle and can be addressed in post-surgical rehabilitation. In order to decrease 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 as well as 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 development for improved implant design, rehabilitation intervention and surgical approach for the TKR patient.

Author contributions

Linda Denney and Lorin Maletsky contributed to project conception. The TTT task was developed as one of two tasks to be performed for Linda Denney's Dissertation. Lauren Ferris assisted with this project and presented the material at the ASME Summer Bioengineering Conference; Fajardo, Puerto Rico, USA and was awarded second place for the poster. She was then asked to publish in the Journal of Biomechanical Engineering. Therefore, she is first author on this paper. The manuscript was completed via collaboration of all three authors. More specifically Lauren Ferris contributed to data collection, data analysis, data interpretation,

statistical analysis and draft and submission of the manuscript. Linda Denney contributed to data collection, data analysis, data interpretation, statistical analysis, and drafting and editing the manuscript. Lorin Maletsky contributed to project conception, data analysis, data interpretation and editing the manuscript.

Tables and Figures

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

	TKA		HEALTHY
Number of subjects	11		12
Age in years	67 (6)		65 (8)
BMI	27.7 (4.0)		24.6 (3.8)
Gender	36% F		58% F
Post-Op in mos.	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 in degrees	137.9 (6.3)*	126.7 (7.6) *†	141.2 (5.7) †
Hip RoM in degrees	111.6 (8.3)	107.0 (7.6)	109.1 (10.8)

*Affected statistically significant (p<0.05) from Unaffected

†Affected statistically significant (p<0.05) from Healthy

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)

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. TKR subject values larger than the healthy 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 % button-to-button movement that crossed 0.5 lead force ratio and knee angles at the % the LFR crossed 0.5.

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

*Statistically significant ($p<0.05$) from Healthy

Figure 1: Equipmental set up. Subject performing a Low to High (L2H) sequence. The subject performs this sequence three times. Button positions indicated with circles.
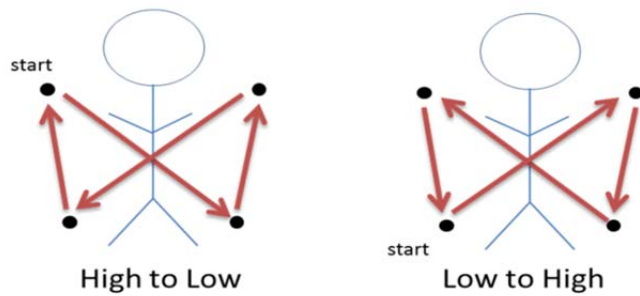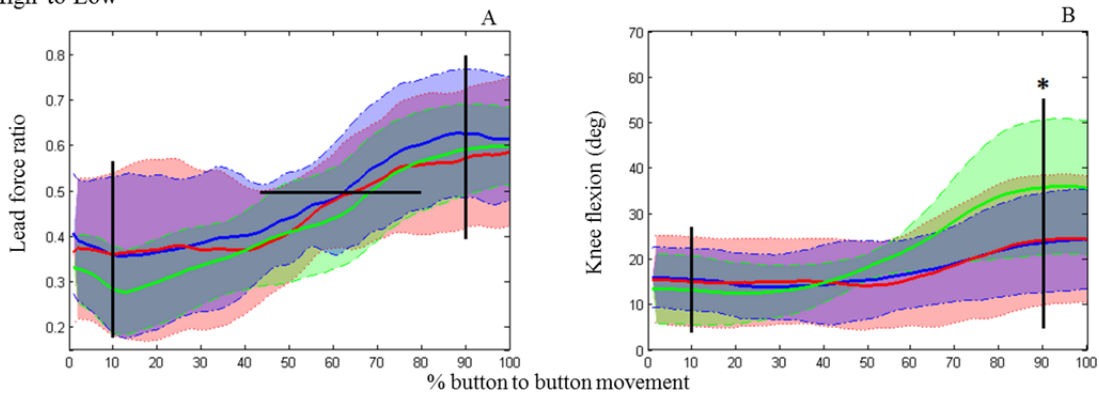

Figure 2: Sequence of both High to Low (H2L) and Low to High (L2H) that the subjects performed
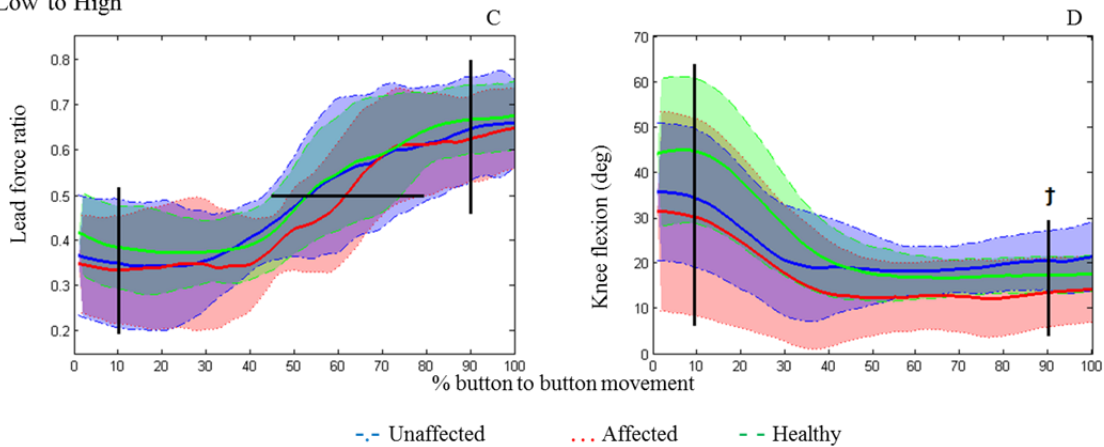
Figure 3: Mean lead force ratio (A) and knee flexion angle (B) throughout a H2L activity and mean lead force ratio (C) and knee flexion angle (D) throughout a L2H activity. Shaded areas ±1 standard deviation. Black vertical lines indicate where the subject is at 10%, right after the first button push, and 90%, right before the second button. Horizontal black line indicates when the subject has 0.5 LFR, or equal distribution of weight. A single factor ANOVA was performed at each black line.

     * Unaffected statistically significant (p<0.05) from Healthy
     † Unaffected statistically significant (p<0.05) from Affected

Chapter IV

Biomechanical strategies of individuals with self-reported knee instability and healthy controls during two dynamic rotary tasks.

Linda M Denney, Lauren A Ferris, Hongying Dai, Jason Herold, Sarah Williamson, and Lorin P Maletsky

Abstract

Background: Rotary instability of the knee facilitates the use of strategies such as altered posture in the torso, altered kinematics in the hip and opposite knee, and hesitation during dynamic activities. While a majority of the research has been found in excessive knee translation in the sagittal plane, the rotary component during a pivot or turn incorporates the trunk and stabilization of other joints to assist control. Knee instability continues to be defined by individual self-reporting of 'knee buckling' or 'giving way' despite attempts to relate clinical tests and biomechanical analyses to functional outcomes. The inability to link testing to function suggests a lack of understanding of the mechanism of knee instability, specifically dynamic rotary instability.

Hypothesis/Purpose: Determine how individuals with self-reported knee instability perform rotary tasks differently than healthy controls. Further, we hypothesize that the unstable group would complete the tasks with hesitation and with altered mechanics, commonly referred to as compensatory movements, to avoid additional rotary forces on the knee. The hesitation may be due to quadriceps weakness or altered neuromuscular timing.

Study Design: Descriptive laboratory study.

Methods: Eighteen individuals from two groups: self-reported knee instability (n=6) and healthy controls (n=12) performed two rotary tasks that included flexion and extension of the knee and weight shifting during a one-time visit. Each individual completed a novel rotational reaching task, Target Touch Task (TTT) and descending stairs followed by a cross-over turn. A Principle Components Analysis (PCA) was used to correlate variables between the groups during the tasks. Variables compared across the groups identified temporal, kinetic, and kinematic patterns during the rotary tasks.

Results: Five principle components (PC) explained 53% of the variance of performance of the rotary tasks and 10 were required to explain 80%. Components in PC1 represent temporal aspects of both tasks such as time to impact, stance time and single stance time and accounted

for 15% of the total variation. The unstable group performed tasks at a slower cadence when compared to the healthy controls. Components of the Stair Task loaded in PC1-3 while the TTT was represented across all five components (PC1-5). Knee, hip and spine flexion range of motion (ROM), and quadriceps strength correlated with loading factors such as ground reaction force (GRF), stance time, knee moment (Mz) and center of pressure (COP).

Conclusion: Individuals with self-reported knee instability performed rotary tasks differently than healthy controls and utilized compensatory strategies to complete the tasks. The unstable group performed the tasks slower and utilized trunk movements to compensate for maintaining knee extension rather than squatting. A higher transfer of load during weight shift was demonstrated by the unstable group leading to altered mechanics of the lower extremities. Rotary tasks that include use of the torso may influence lower extremity loading and mechanics and result in use of compensatory movements.

Clinical Relevance: Individuals with knee instability utilize altered mechanics when loading the lower extremities during a rotary task; specifically if it involves trunk positioning. Spine flexion is utilized in order to execute reaching or transfer of load and may be a result of poor neuromuscular control and decreased proprioception. Rehabilitation training should focus on accurate positioning, timing and activation of the musculature for postural control in order to avoid substitution with an extended knee and lack of hip flexion to gain stability.

Introduction

Knee instability is defined by subjective reporting of 'buckling' or 'giving way' with little

evidence that subsequent objective tests correlate with actual functional performance[92,93]. While

feasible to find laxity measurements of the knee, these measures do not necessarily equate with

functional instability, clinical tests, or rotary instability[18,59,95,96]. Rotary instability is the most

difficult to measure, yet the most important aspect of return to sport and return to Activities of

Daily Living (ADLs) that involve pivoting or twisting.

Knee instability can prevent return to sport even with surgical intervention[97]. Anterior

cruciate ligament (ACL) injury is one of many causative factors of rotary knee instability and

provides the bulk of the research. Other injuries, such as meniscus tear and posterior capsule

laxity, could also cause rotary knee instability[98]. Pivoting and cutting maneuvers in sports

impart forces on the knee which may result in an ACL tear or rupture. Most athletes consent to

surgical repair in the hopes of returning to play. The primary goal for any knee rehabilitation,

including post-ACL reconstruction, is to restore proper knee mechanics which includes

regaining knee flexion and extension range of motion (ROM); however, rarely is the amount of

tibial-femoral rotation measured or even assessed dynamically. Multiple devices and tests have

been utilized to measure knee rotation, yet lack the ability to capture an accurate clinical

measure of rotary instability or measurement that relates to dynamic functional stability[18,99-102].

The pivot shift test has been associated with clinical outcome measures[98,103] but is difficult to

reproduce consistently and lacks a standard of procedure for testing.

Although the ACL prevents anterior translation of the femur on the tibia, deceleration

during internal rotation with knee flexion may be a bigger factor in considering pivoting or

change of direction. Anterior shear during loading is the culprit to ACL injury and is the result of

a combination of movements[104]. Altered anterior-posterior translation and internal-external

rotation of the knee has been captured by motion analysis during gait and step up/down in

subjects who are ACL-deficient and in those that have osteoarthritis (OA) compared to healthy controls[105,106]. Therefore, it is appropriate to consider that instability of the knee may cause an alteration in the kinematics during dynamic activities. Further, habitual compensation may be more prominent when individuals with knee instability are asked to do a rotary task.

Kinetics and kinematics during gait have been analyzed with motion capture systems primarily with straight-gait performance[106,107]. Wave form analysis identifying common constructs of gait and group differences has been shown with the use of Principle Components Analysis (PCA)[105,107]. Individuals post-ACL reconstruction were compared to healthy age-matched individuals finding differences in adduction moment during early stance time. Further analysis revealed a trend that differences were apparent between the contralateral leg of the surgical group and the healthy[105]. These finding demonstrate the benefits of using PCA to identify differences in all groups for all constructs.

In order to assess dynamic knee instability, individuals with self-reported knee instability were tested as part of a larger study investigating total knee replacement (TKR) recipients and healthy individuals performing two rotary tasks. The objective of this study was to measure how individuals with self-reported knee instability perform rotary tasks differently than healthy controls and to identify compensatory movement patterns.

Material and Methods

Two tasks were designed that incorporated both loading and transfer of weight while flexing and extending the knee, and loading during a pivot turn. The Stair Task included a pivot turn at the bottom of the stairs[108] and is a stability-demanding activity that is commonly performed daily, while the novel Target Touch Task (TTT) challenged the rotational movement during knee flexion and extension while reaching[47].

Two groups, experimental (unstable) and control (healthy), participated in the two rotary tasks on a one-time visit. The experimental group (n=6) had self-reported knee instability (age 37±13, BMI 25.6±2.1) while the control (n=12) had no report of knee or lower extremity problems (age 65±8 BMI, 24.6±3.8). Despite the age difference, all data collected except the quadriceps strength (laxity, hip, and knee ROM) were similar between groups (Table 1). Individuals in the experimental group were recruited based on self-reported occasional knee instability within the last two months during activity. Subjects were included if they were able to navigate up and down stairs (reciprocating) without the use of a handrail or assistive device. Individuals were excluded from this study if they had any of the following: previous hip or ankle surgery, previous surgery on the opposite knee, peripheral neuropathy, Body Mass Index (BMI) greater than 30, unicompartmental or total knee implant.

Testing Procedure

Consent to participate in the study was received and approval for video and photographs to be taken were obtained. Once the subject consented to participate, pre-measurements were recorded. Participants wore tight fitting clothes and shoes with non-reflective material. Anthropometric measurements for height and weight (to configure BMI), and bony landmark measurements were taken to configure/develop the computer model. Goniometric measurements for the lower extremities and bilateral quadriceps strength were recorded (MicroFET 2, Hoggan Health Industries; West Jordan, Utah). A knee arthrometer (KT-1000) was utilized to measure anterior translation of the tibial-femoral joint for both knees. Lastly, the subject maximum rotational reach was recorded which measured lateral reaching distance of each hand across the body.

Twenty-four reflective markers (25 mm) were applied to bony landmarks by using adhesive and then secured with tape. Markers were applied to the skin and on clothing where

exposing the skin was not possible or comfortable for the participant. A Knee Alignment Device was utilized to identify the knee axis[53] and the x,y,z configuration was captured.

Each participant performed two tasks: a TTT and Stair Task. Data were collected with a Vicon 512 (Oxford Metrics, Oxford, UK) infared 6-camera motion analysis system at 120 Hz and two AMTI force plates (Advanced Mechanical Technology, Inc., Watertown, MA) embedded in the testing laboratory's floor collected at 360 Hz.

Target Touch Task: The button configuration was set up for the TTT in accordance to the subject's maximum rotational reach and the force platforms were calibrated prior to the testing according to the manufacturer's recommendations. The participant stood with one foot on each force plate, feet approximately shoulder width apart, and performed a series of rotational movements and button pushes as directed by the researcher (Figure 1). Two targets were placed on both the right and left side of the participant: two at shoulder height and two level with the knee. A quiet stance or static trial was recorded prior to the sequence button pushes with the subject looking straight ahead and arms at their side. The participant was asked to touch the targets (confirmed by a sound on button push) in a requested series of movements while keeping their feet flat on the floor. The cross-over reach of High to Low (H2L) combined the motions of knee flexion and extension with rotation. H2L started with the left arm/hand crossing over to push the right high button. The subject continued the series by using their right arm/hand to cross over to press the left low button followed by pressing the left high button, finishing with the left hand on the right low button. This series was continued for 2 more cycles. One practice trial was allowed before data collection began. If the participant demonstrated this practice trial without bending their knees, the investigator requested that the participant perform the lower button pushes by "using their legs" during the motion to reach the buttons. The task required repetition if the data collection was interrupted, the task was stopped for any reason, or if the participant did not keep full contact of their feet on the floor (heels came off the floor).

Stair Task: Participants were instructed to descend 4 steps (15 cm in height) in a

reciprocating pattern without use of the handrail. The stairs were positioned flush behind the

force plate. Calibration of the force platform was repeated for the Stair Task once the staircase

was set in place. Instructions for which foot was to lead were provided before each trial by the

investigator. The sequence started with a straight gait after descent, followed by trials of a turn

after descent, either right or left. The participant completed a turn at the bottom of the stairs by

using a cross-over technique (Figure 2). For example, a right turn was performed by planting the

right foot on the floor and the left foot crossed over the right foot in order to take the next step

(Figure 2). A cone was placed at approximately 45° angle from the anticipated foot placement at

the bottom of the stairs as a guide for the participant. The participant was instructed to step on

the floor with their foot pointing straight ahead on all trials. Participants performed the task at a

self-selected pace and were given one practice trial prior to collection of data. The participants

performed this task two times straight and three times for each direction of turn. Trials were

repeated when data collection was interrupted or the participant performed the task incorrectly.

Examples of reasons for repeat included the subject overstepping the force plate, positioning

their foot on the force plate to the right or the left anticipating the turn, wrong sequence of foot

lead descending the stairs, or use of the handrail.

Data Collection and Reduction

Kinematic data were obtained using Vicon's Workstation (v4.5) software and the lower

body Plug-in Gait model (Oxford Metrics, Oxford, UK). Based on the computer model (pre-

measurements) and location of the markers, the coordinate system calculated joint centers of

rotation and angles. The static trial was used to record neutral joint angles in all planes.

Target Touch Task Data Processing: Each series of this task was cut according to the

direction of the button pushes (H2L). Transfer of force was identified by the lead leg (the leg that

the subject was leaning toward in the task) which was "affected" or "unaffected" in the unstable group. Force, knee angle and moment were examined and data were interpolated to 100 points evenly spaced in time between button pushes. GRF data were normalized to Lead Force Ratio, or LFR, where the lead legs force was divided by the overall force. Averages (for each crossover and for across subjects) and standard deviations were calculated for all trials at each percent button push cycle.

Stair Task Data Processing: GRF, moment (Mz) calculated by rotation about the vertical axis of the force plate, knee angles, and pelvis/hip orientation were analyzed. Force and moment data were filtered (4th- order low-pass Butterworth filter). All data were interpolated to reflect 100 evenly spaced in time data points for each trial. A threshold of 2% of the maximum vertical force was used to identify 0% (heel strike) to 100% (toe off) stance phase. Forces were normalized to the participant's body weight and moments to percent body weight times height. Hip orientation was used to determine rotation of the pelvis in the direction of the turn prior to the pivot.

Initial GRF was documented as the max force across all trials, and within the first 40% (or initial weight acceptance on the force plate) of the stance time. Maximum internal and external moments about the force plate and stance knee angle were calculated. Data obtained from the average of turn trials were compared to the straight walk after stair descent.

Statistical Analysis

Independent and paired t-tests were conducted for maximum LFR and knee angle for H2L during the TTT; and for GRF, knee angle and moment during the Stair Task. Independent t-tests were conducted when comparing the unaffected and affected to healthy and paired t-tests comparing differences between the unaffected and affected.

Principal component (PC) analysis explains data sets by reducing multiple variables while keeping the variation present in the data[109]. Data consists of p variables that are transformed into p orthogonal and independent PCs. These PCs are ranked based on how much they explain the variation in the data, first being the largest explained variation, second being the second largest, and so on until 100% total variation is explained.   Data sets are constructed into a matrix that is N x n; N (N=36) being the number of subjects times two to account for both legs of the subject and n (n=29) the data points, or PC variables. Matrix [A] is the N x n matrix and [X] is calculated as an n x n covariance matrix. Matrix [E] is made up of the eigenvalues and eigenvectors of X, with each column holding the eigenvectors, or the PCs of the data. A, a 1 x n vector, is calculated by averaging each variable. P, or the PC scores, are calculated as the dot product of the PC matrix, P = AE and utilized in reconstructing the PC.

Variables used to construct the PC analysis consisted of pre-measurements of the subject and outcomes of the two functional tasks (Table 2). The stair temporal data consisted of stance time, time to the first peak of the gait cycle (or the passive peak), time from the second peak of the gait cycle (active peak) to the end of the trial, and the amount of time the subject was in single stance. GRF of the passive peak and active peak were variables, as well as the moment during the pivot about the force plate. Knee angle of the stance leg during the passive peak and active peak were considered, along with the anticipatory rotation of the pelvis during the pivot. TTT temporal data were utilized for the search time (or when the subject's wrist marker was within 15 cm of the button), cycle time, and the time it took the subject to transfer weight from the first button push to equal LFR. GRF range, maximum moment (Mz) on the lead leg during the first button push, and maximum moment (Mz) of the lag leg during the search time, and the range of the center of pressure acceleration in both the medial-lateral and anterior-posterior direction were applied. Kinematics consisted of knee and hip flexion of the lead leg, and the range of motion of the pelvis, shoulder, and spine during the task. The model

was created with all subjects' individual legs. PC analysis was used to understand correlating relationships between variables. To determine if the two groups were different along these PC directions, student t-tests (p=0.05) were performed on the group scores. A discriminatory analysis was performed on all 29 variables in order to determine which variables represented group separation.

Results

The unstable group loaded both the affected and unaffected leg similar to the healthy group during the transfer of weight in the TTT, yet reduced ground impact during the Stair Task. The LFR for both the affected (.706±.095) and unaffected (.709±.116) knees while reaching for the low button during the H2L task for the unstable group were not significantly different (p=.128, p=.145 respectively) when compared to the LFR of the healthy group (.639±.078). During the Stair Task, the unstable group had less maximum knee flexion on the affected (32.8±9°) and unaffected (37.4±11°) knee at initial impact compared to the healthy group (37.7±13°) (p=.437, p=.962 respectively). Initial GRF after stair descent for the affected (111.8± 12.4) and unaffected (111.5± 9.3) leg compared to the healthy controls (134.0±14) was significantly less (p=.005, p=.003 respectively) using independent t-tests. The differences between the groups in each task predicated the analysis of comparing constructs as both tasks involve a rotary motion and could be grouped. Therefore, in order to understand the relationship between these trends, further analysis using PCA was conducted.

PCA was utilized as a means to identify correlations in the measures and try to explain differences between individuals with knee instability and healthy controls during rotary tasks. Further, these variables identify specific strategies utilized by the individuals and the PCA model identifies the differences of these strategies between the groups. Variables weighted heavily in a specific PC identified variables that move together. Individual PC scores were identified for each subject and group differences were tested by using an independent t-test for PCs 1-5

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which were identified by the cumulative variation explained by PCs greater than 50%. Component loading PC 1-5 explained 53% of the overall variance with the highest explained variance of 15% for PC1. The features of PC 1-5 identified in Table 3 indicate a positive (+) or negative/inverse (-) correlation and the mean and the standard deviation (sd) of each group's PC score in the subsequent column.

The first PC, which accounted for 15.1% of the variation between the legs, was largely influenced by eight variables (Table 3) which moved together. Four of the five variables that were positively correlated relate to temporal aspects of the Stair Task, meaning that the time duration for all variables was either longer or shorter; always in the same direction. GRF moved in the same manner (i.e. as GRF increased, time was longer). As these variables moved in one way, knee extension, pelvis rotation and GRF PP moved in the opposite direction (i.e. as temporal measures increased, knee extension and pelvic rotation decreased; along with GRF PP). When the two groups are clustered and their PC scores compared, there was a significant difference along this direction between the two groups (p=.008), with the unstable group consistently scoring higher (i.e. taking a longer time, greater GRF range, lower knee extension, pelvic rotation and GRF PP). The healthy group had a lower PC score corresponding to a shorter time to complete the tasks.

PC2 accounted for 12.4% of the variance and included eleven variables. Three of the seven components that were positively correlated represent dynamic movement during the TTT (COP ML and AP; Mz on the lag leg); two components were from the Stair Task (single stance time, GRF AP), while the remaining two components were pre-measurement (quadriceps MMT and knee extension ROM). These variables moved in the same direction irrespective of the task or source of measurement. As COP acceleration increased, the Mz on the lag leg increased; while an increase in single stance time, increased GRF AP. Four variables moved in the opposite direction with a decrease in laxity, H2L cycle time and KFA (PP and AP) during the

stairs relative to an increase in the positively correlated variables. There were no significant differences between the groups when clustered (p=0.61), therefore, no inferences of causative correlations can be made with positive and inverse direction differentiating performance of the tasks between the healthy and unstable, yet this identifies a consistent strategy among all subjects.

PC3 explained 10% of the variance with the five variables that were positively correlated all related to the TTT including two temporal, one GRF range, one ROM (spine flexion) and one Mz (lead leg). These variables moved in the same direction (i.e. as time duration increased, spine flexion and lead Mz and GRF range increased). As these variables moved in one direction, three variables (hip and knee flex, pelvis rot change from straight PP) moved in the opposite direction (i.e. as spine flexion increased, hip and knee flexion decreased). Differences of the clustered variables were not significant between the groups for PC3 (p=0.33) but did suggest a difference in how the TTT was completed.

PC4 accounted for 8.5% of the variance and included five variables that were positively correlated while only one negatively correlated. The positive variables included three pre-measurements (hip flex, knee flex and ext) and two TTT measures (H2L cycle time and Max KFA lead H2L). The negative variable was Pelvis Rot Range H2L from the TTT. So, while hip and knee ROM, TTT cycle time and maximum knee flexion on the lead leg during H2L increased; pelvis rotation range during H2L decreased.

Lastly, PC5 accounted for 7.3% of the variance and included three positively correlated variables and two negatively correlated variables. As hip extension ROM, maximum hip flexion angle on the lead leg during H2L, and maximum spine flexion on the H2L increased; quad MMT and COP range AP Accel during the TTT decreased. No Stair Task components loaded in either PC4 or PC5.

Overall the differences between the groups were highlighted in PC1 with temporal aspects dominating the correlations. The unstable group performed the tasks at a slower rate when compared to the healthy group. The hesitation or slower performance was correlated to other variables such as quadriceps strength, spine flexion and knee flexion angle in PC2-4 suggesting compensatory torso and lower limb strategies were utilized to achieve the task which may contribute to the increased time to complete the task. Additionally, the slow rate of performance and quadriceps strength suggests a possible neuromuscular delay or slow timing issue associated with the performance of the two rotary tasks.

A discriminatory analysis was performed on the 29 variables in order to determine which variables were the best predictors between the groups. The seven variables listed in Table 4 were the most discriminatory variables or best predictors between the unstable group and healthy individuals. On cross-validation, there was no misclassification of these variables. Each of these seven variables indicated how the measure contributed to group separation.

Time to impact had the largest discriminatory factor followed by kinematic measures during the TTT for discriminatory variables two and three. Further analysis divided these variables by sub-groups (pre-measurements, TTT, and Stair Task) indicating 37% misclassification on the pre-measurements, 7% on TTT and 0% on Stair Task. The Stair Task was consistent with classification based on 'stable' as imputed in the discriminatory analysis and follow-up with sub-groups; both 0% misclassification. The temporal aspect of the Stair Task was the most discriminatory contributing factor with the least variance to establish group separation.

Discussion

The ubiquitous difficulty of defining knee instability is due to the number of variables that are related to the altered mechanics at the knee. The multiple variables and additional challenge of measuring dynamic rotary knee function creates an optimal area of research that identifies

strategies presented in trunk movements and altered kinematics at adjacent joints during rotary

tasks. Thus, the mechanism of how individuals move with self-reported knee instability provides

impetus to consider all variables that may impact the knee in a rotary motion in order to further

understand knee instability. PCA is a powerful tool to compare multiple variables and was

utilized to group the variables from two rotary tasks in order to summarize commonly extracted

data with the follow up discriminatory analysis to characterize group separation.

The loading vector of PC1 was driven by the temporal components in both tasks. Our

healthy controls were older than the unstable group and demonstrated a quicker self-selected

speed to complete the tasks and less time in stance phase when compared to the unstable

group. Thus, findings in the Lark et al. (2004)[110] study of the elderly individuals, when compared

to younger, maintaining a longer 'foot flat' and stance time during a step down is not consistent

with our results, considering only age difference. However, those in the younger group in the

current study did have report of an unstable knee which explains why there were differences

between groups during performance of a rotary task. A longer stance time may be related to

increased base of support or need for stability which would then agree with why the unstable

group performed the tasks at a slower speed and with a longer stance time. Knee flexion angle

also changes with stance. According to Deluzio et al. (2007)[107] individuals with osteoarthritis

(OA) utilized less knee range of motion during the gait cycle when compared to healthy controls.

Late stance (30-50%) demonstrated the largest magnitude of difference between the groups for

knee flexion. This would be similar to the knee flexion angle on initial impact to push off during

the Stair Task in the current study. The unstable group in our study performed the Stair Task at

a slower pace and with less knee flexion on initial impact.

Greater moment about the hip suggests the need for stability when descending stairs

with the elderly population[111]. Greater moment at the hip was complimented by a large moment

at the knee whether descending stairs or walking on a level surface. Group differences during a

normal gait cycle in individuals with OA compared to healthy demonstrated a large adduction moment at the knee during early stance (10-20%) decreasing towards mid-stance (30%)[107]. Similarly, Sanford et al. (2012)[105] reported the same findings with individuals post-ACL reconstruction; however, differences were found in the surgical leg (affected), the contralateral leg (unaffected) and healthy legs. Although Mz about the force plate during the crossover was not a loading component for the Stair Task in the first five PCs, Mz was identified in the lag leg during the latter phase of H2L during the TTT increasing along with COP range of AP and ML acceleration. More importantly, clinicians should take caution when using the contralateral leg for comparison in the clinic when conducting side-to-side clinical tests. Sanford et al. (2012)[105] reported differences in the contralateral leg compared to the healthy group; therefore suggesting some adaptation.

While most of the literature is reported on gait, the novel rotational task used in this study demonstrated differences between the groups. The maximum knee flexion angle on the lead leg for the unstable group during the TTT was less than the healthy group; however, not significant and not a loading component in the first five PCs. Loading components in PC3, GRF on the lead leg, overall cycle time and time to search for the button push, suggest a preparatory strategy that is commonly discussed in the literature with individuals post-stroke and with Parkinson's disease[80,112]. The anticipation of pushing the button may be more of a planning strategy rather than a stability strategy[26,27,35,80,113]. Planning to execute the task without neuromuscular training could cause delay or hesitation, which has been linked to injury in sport[95,114]. Moreover, movement at the spine (Spine Flex H2L Max) with inverse hip and knee flexion suggest compensatory motion at the torso while limiting use of the lower extremities to complete the task. Subtle trunk movements drive action in the lower extremities during weight shift or pivoting in sport; therefore altering mechanics at the knee[115-117].

High GRF and a small knee flexion angle were identified as two factors predisposing female athletes to ACL injury[114]. The unstable group utilized both of these strategies during the TTT when shifting weight to the lead leg to push the high button. Speculation of why there was increased loading may be explained by the lack of proprioception or lack of neuromuscular input to control the knee[118]. The altered control could be supported by the loading of the quadriceps strength (Quad MMT) component during the TTT as the unstable group had considerably less quadriceps strength than the healthy controls. Alternatively, the transfer of load during the TTT may be due to inaccuracy of trunk positioning during the rotary task, thus excessive weight transfer to the lead leg with altered positioning of the knee. Error in repositioning the trunk, whether by poor neuromuscular control or due to low back pain has contributed to an increase in lower extremity injury in female athletes[116,119]. Both directions of COP acceleration (AP and ML) and maximum spine flexion for the H2L task were loaded in PC2 and PC3 and would explain the altered control of the trunk during the reaching task while hip flexion was loaded as a negative component, possibly due to the fact that the individuals were utilizing spine flexion to complete the task rather than flexing at the hip.

Conclusion

We hypothesized that individuals with unstable knees would perform rotary tasks differently than healthy individuals. Using the PCA identified the strategies of all individuals that were then compared between groups. The PCA model supports that differences can be detected in temporal aspects as well as GRF, Mz, KA and COP. Higher magnitude of loading was significant (p=.008) in the strategy involved with temporal aspects with the unstable group performing the tasks at a much slower pace (PC1). However, discriminant differences were apparent in the loading of the leg depending on use of the torso to complete the movement or an alteration of the hip and knee angle. Although stair activity is commonly utilized as both a research tool and objective measure, it appears the novel TTT identifies more differences with

the efforts to utilize the trunk rotation and its impact on the lower extremities; specifically the knees.

Limitations

The small sample size of the unstable group may limit the ability to report robust conclusions between the unstable and healthy group. However, each leg provided individual variables, therefore doubling the data; relative to the n size. The age difference of the two groups is considerable but arguably not a limitation as found in literature. Minimal biomechanical differences were reported between a group of older individuals when compared to younger individuals during a step down and stair descent[110,111]. Exclusion criteria did not allow for truly unstable individuals who require use of handrail and knee instability was self-reported. Although, self-report is concurrent with the literature of 'buckling' or 'giving way', a clear diagnosis was unknown with the individuals in the unstable group. Lastly, L2H cycles in the TTT were not included in the PCA due to the fact that two subject data sets were missing leaving the average of L2H cycles different than the H2L cycles. Components of the H2L cycle were identified as loading components of the PCA model and therefore the choice to include more cycles, H2L, in the TTT was a decision to provide the most accurate average of each individual performance during the TTT.

Author Contributions

Linda Denney contributed to project conception, data collection, data analysis, data interpretation, and drafting the manuscript. Lauren Ferris contributed to data collection, data analysis, data interpretation, drafting and editing the manuscript. Hongying Dai contributed to statistical analysis of data, and data interpretation of the manuscript. Jason Herold and Sarah Williamson assisted in subject recruitment, data collection, data analysis, data interpretation,

and drafting the manuscript. Lorin Maletsky contributed to project conception, data analysis,

data interpretation and editing the manuscript.

Tables and Figures

Table 1: Demographics of the subjects

	Unstable		Healthy
Number of subjects	6		12
Age	37 (13)		65 (8)
Gender	33% F		58% F
BMI	25.6 (2.1)		24.6 (3.8)
	Unaffected	Affected	
Quadriceps Strength (N)	141.9 (28.0)	132.6 (29.4)	130.3 (20.5)
Knee Laxity (mm)	2.1 (0.6)	2.2 (0.7)	2.4 (0.9)
Knee range of motion (deg)	141.2 (4.0)	139.8 (5.3)	141.2 (5.7)
Hip range of motion in (deg)	109.2 (7.6)	108.7 (7.8)	109.1 (10.8)

Table 2: Variables used to construct PCA matrix; pre-measurements and items from each task followed by type of variable.

Pre-Measurements	TTT		Stairs	
Quadriceps strength	Cycle Time	Temporal	Stance Time	Temporal
Anterior-Posterior Laxity	Search Time	Temporal	Time to Impact	Temporal
Hip Flexion	Time to .5 LFR	Temporal	Time from Active Peak to toe off	Temporal
Hip Extension	GRF range	Kinetic	Single Stance Time	Temporal
Knee Flexion	Max lead KFA	Kinematic	Stance KFA at Active Peak	Kinematic
Knee Extension	Max lead HFA	Kinematic	Stance KFA at Passive Peak	Kinematic
	Range of Pelvis Rotation	Kinematic	Pelvis rotation from neutral at Passive Peak	Kinematic
	Range of Shoulder Rotation	Kinematic	GRF at Passive Peak	Kinetic
	Max Spinal Flexion	Kinematic	GRF at Active Peak	Kinetic
	Max Moment from 0-20% on lead leg	Kinetic	Moment at crossover	Kinetic
	Lag leg Max Moment from search time-100%	Kinetic		
	Medial-Lateral CoP range	Kinetic		
	Anterior-Posterior CoP range	Kinetic		

Table 3: Principle Components Analysis (PC 1-5) loading vectors for the Target Touch Task and Stair Task.

	% Variation Explained	Features		Mean (s.d.)		p-Value
		+	-	Unstable	Healthy	
PC1	15.1	Stance time Time to impact GRF range LFR Time to AP to end Single Stance time	Knee Ext Pelvis rot Range H2L GRF PP	2.21 (2.00)	-1.13 (0.99)	.008
PC2	12.4	Quad MMT Knee Ext Single Stance time GRF AP H2L lag Mz to end COP range ML Accel TTT COP Range AP Accel TTT	Laxity H2L cycle time Stance KFA AP Stance KFA PP	0.27 (0.98)	-0.20 (2.21)	0.61
PC3	10.0	Search Time H2L cycle time GRF range LFR Spine Flex H2L Max H2L lead Mz 0-20%	Hip Flex Knee Flex Pelvis rot change from straight PP	0.47 (1.83)	-0.29 (1.60)	0.33
PC4	8.5	Hip Flex Knee Flex Knee Ext H2L cycle time Max KFA lead H2L	Pelvis Rot Range H2L	0.34 (0.94)	-0.06 (1.76)	0.61
PC5	7.3	Hip Ext Max HFA lead H2L Spine Flex H2L Max	Quad MMT COP Range AP Accel TTT	-0.10 (2.03)	0.04 (1.42)	0.94

Table 4: Discriminatory Analysis results for all variables of pre-measurement, Target Touch Task and Stair Task and R^2 value.

	Variable		R^2
1	Time to Impact	Stair Temporal	0.48
2	Max H2L lead Knee Flexion Angle	TTT Kinematics	0.59
3	H2L Range of Pelvis Rotation	TTT Kinematics	0.69
4	Knee Extension	Pre-Measurement	0.74
5	Knee Flexion Angle of stance leg at Active Peak	Stair Kinematics	0.78
6	Range of LFR	TTT Kinetics	0.80
7	H2L Range of Shoulder Rotation	TTT Kinematics	0.82

Figure 1: Experimental set up. H2L sequence was performed three times.
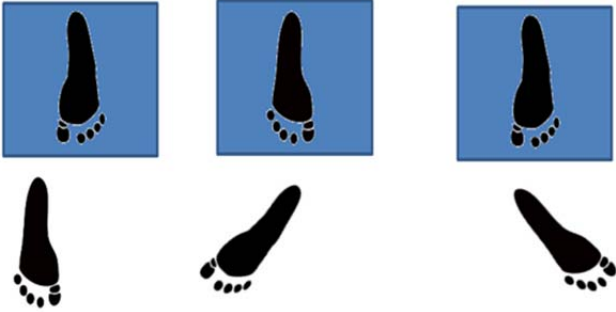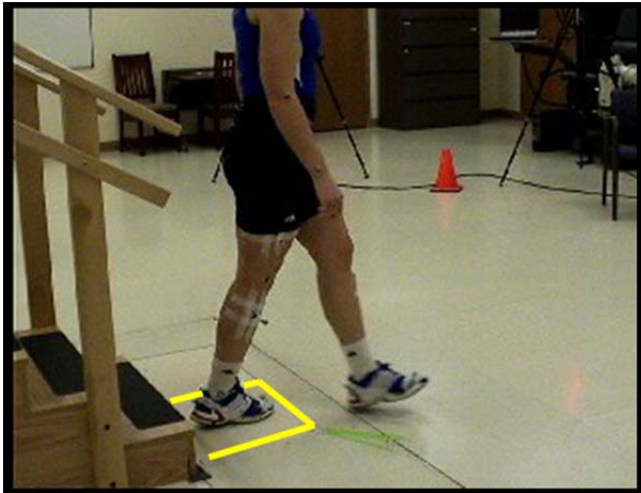

Figure 2: Experimental set up (top) and stepping sequence for the eight trials performed. Force plate used in the collection process highlighted in yellow.

Chapter V

Analysis of two rotary tasks comparing the performance of individuals with a total knee replacement, non-surgical self-reported knee instability and healthy controls.

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(Manuscript in preparation for publication in The Knee. This journal allows a combination of Results and Discussion)

Abstract

Background: Rotary tasks can involve flexion and extension of a knee while moving in a transverse plane. Individuals with knee pathology utilize compensatory mechanisms to complete tasks that may be contributing to arthritic changes at adjacent joints.

Hypothesis/Purpose: The purpose of this study was to better understand how individuals with TKR, individuals with self-reported knee instability and their healthy controls perform rotary tasks that involve flexion and extension of the knee during dynamic loading of the lower extremities in both a stair task and reaching task through the use of a Principle Component Analysis (PCA).

Basic Procedures/Methods: Twenty-eight subjects from three groups participated in two rotary tasks; 10 TKR, 12 healthy, and 6 unstable subjects. The two rotary tasks included stair descent followed by a cross-over turn and a novel rotational reaching task, Target Touch Task (TTT), involving flexion and extension of the knee. PCA was used to correlate the variables for the three groups. Variables compared across the groups were used to identify temporal, kinetic and kinematic patterns during the rotary tasks and pre-measurement variables taken before the tasks.

Main Findings/Results: Twelve principle components (PC) explained 80% of the variance of performance of the rotary tasks. Components in PC1 related temporal aspects of both tasks such as time to impact, stance time, time to push off and pivot turn after stair decent and account for 12.8% of the total variation. The TKR and unstable groups performed tasks at a significantly slower pace when compared to the healthy controls ($p < .001$). Components of the TTT were represented in PC1-4 and included all items of temporal, kinetic and kinematic variables. Both time and kinematic variables largely influenced the significant difference between the TKR group and the healthy in PC4 ($p = .017$).

Principle Conclusions: Individuals with knee pathology performed rotary tasks differently than

healthy controls and utilized compensatory strategies to complete the tasks. Tasks were

performed slower and trunk movements were utilized to compensate for maintaining knee

extension rather than squatting. The unstable group loaded the affected leg more than when

compared to the TKR during the out-of-plane squatting task. This may be due to coping

mechanism to control knee instability or that the TKR group demonstrated postural adjustments

and habits acquired pre-surgery. The PCA model was effective in identifying a range of

variables utilized by all three groups in order to differentiate performance of task. A testing

assessment and rehabilitation intervention to address rotary movement in the transverse plane

with the use of the upper extremities and torso should be incorporated as the TTT variables

largely influenced each PC.

Introduction

Knee instability is reported as a complication after total knee replacement (TKR) surgery and defined by a subjective report of buckling or 'giving way' which interferes with an individual's ability to participate in weight-bearing functional activities[16,66,92,120]. Knee instability is usually defined in terms related to laxity in the sagittal plane, specifically excessive anterior tibial-femoral excursion; however, individuals complain of knee instability with many activities including out-of-plane motions. In order to avoid the feeling of instability, individuals with a TKR will utilize compensatory strategies such as quadriceps avoidance or movement at adjacent joints to complete a functional task[28,38,47,108]. Although not performed deliberately, subtle compensatory movements contribute to altered mechanics at the knee when compared to healthy controls[32,38]. Altered knee mechanics are also identified in rotary tasks utilized in activities of daily life (ADLs) such as doing laundry, picking up a child, and reaching[31,47]. The use of the trunk and upper extremities during the rotary tasks identify additional variables that contribute to the overall compensatory strategy.

Gait analysis, stair ascent and descent, and Timed Up and Go are functional tasks that have been analyzed pre- and post-TKR in order to identify functional deficits[24,28,121,122]. The dynamic variables captured post-TKR mimic those pre-TKR suggesting that movement patterns may develop before surgery; possibly in response to pain. Unfortunately, these movement patterns continue post-TKR despite significant pain relief[4,21]. Further, these tasks do not include a rotary component and therefore do not represent performance of essential tasks required in sustaining independent living.

In addition to the lower extremity variables that contribute to rotary function, studies have also included trunk motion, use of the upper extremities, transfer of weight and joint positioning that influence knee position and loading[7,31,117]. While most studies utilized the straight-plane tasks, such as gait, differences have been found between TKR groups, unstable knee groups,

and controls during rotary tasks[46,47,108]. Movement of the upper extremities and trunk displacement drives positioning and weight transfer in the lower extremities[33,123,124]. By identifying these differences, alteration in methods of rehabilitation intervention and surgical approach may be considered.

In order to differentiate movement patterns in TKR and healthy individuals for a large number of variables, multi-dimensional statistical analyses have been utilized[125]. Principle component analysis (PCA) has been applied to gait analysis deriving principle components (PCs) of loading and kinematic differences between groups with knee pathology and healthy controls[107,126-129]. Interpretation of the PC score allows the statistical model to discriminate among groups for multiple dimensions. The purpose of this study was to use the PCA statistical model to understand how individuals with TKR, individuals with self-reported knee instability, and their healthy controls perform rotary tasks that involve flexion and extension of the knee during dynamic loading of the lower extremities in both a stair task and reaching task.

Materials and Methods

Twenty-eight subjects were recruited for this study; 10 TKR subjects [post-op mean 20.4 (16.1standard deviation) months], 12 healthy subjects, 6 unstable subjects (Table 1). All subjects signed a consent form approved by the University of Kansas Medical Center Human Subjects Committee and Institution Review Board. Subjects were screened prior to testing to verify they met the testing criteria; BMI less than 30, could bend both knees at least 90 degrees, no previous hip or ankle surgery, no peripheral neuropathy, and could descend stairs without use of a handrail. The TKR and healthy controls were screened to be between the ages of 50-75. TKR subjects were selected if they had a unilateral TKR and no previous surgery on the other knee. The unstable group was based on self-reported knee instability, regardless of age. 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). Range of motion for hip and knee flexion and extension was measured with use of a standard goniometer, anterior translation of the tibio-femoral joint measurements via knee arthrometer (KT-1000), and quadriceps strength via a handheld dynamometer (MicroFET2, Hoggan Health Industry, West Jordan, Utah) were collected before completion of the two tasks.

Twenty-four reflective markers (25mm diameter) were attached with double-sided adhesive tape to bony landmarks and further secured with tape. Subjects were instructed to wear tight fitting clothing and athletic shoes with little to no reflective material. Markers were captured via an infrared six-camera motion analysis system (Vicon 512, Oxford Metics, Oxford, UK) at 120 Hz. Two force plates (AMTI, Watertown, MA) embedded in the floor collected the forces and moments about the plates at 360 Hz. Both data-capturing devices were simultaneously triggered with an analog device upon start of the trials. All subjects completed two tasks which incorporated flexion/ extension of the knee along with rotation: the Target Touch Task (TTT) and stair descent.

Target Touch Task

The TTT was set up per the subject's height and maximum reach width when extending their arm across their body (Figure 1). Two microphone stands were set at shoulder height and a width of the subject's maximum crossover reach. Two more microphone stands were placed at the knee joint level, lateral to each knee. Buttons were attached to the four microphone stands and wired so that a noise was produced when pressed. The subject stood with their feet shoulder-width apart, one on each of the force plates. Static and Knee Alignment Device[53] trials were recorded to configure the Vicon model and to record the subject's natural stance. A high-to-low (H2L) task was performed by the subject. The task was performed by first using the left hand to hit the right shoulder height button, and then with the right hand the subject would crossover the body to press the left low button. The subject would then stand and press the left

shoulder high button with the right hand and switch over to press the right low. This sequence was repeated three times.

Each crossover was interpolated such that there were 100 points between opposite button pushes, totaling five cross overs (three to the left and two to the right). The limbs of the subjects were categorized as either affected (TKR limb), unaffected (TKR unaffected knee), unstable (unstable group's affected knee), stable (unstable's unaffected knee) or healthy. The subject's sides were also categorized as lag (the limb closest to the first button push) and lead (the limb closest to the second button push). These crossovers were then averaged based on if the subject was leading into the second button with their affected (TKR limb) or unaffected side. All healthy subjects' trials were averaged, regardless if they were leading into the right or left side. The subjects completed this task at self-selected pace and trials were recollected if the subject lifted their heels off the ground or performed the wrong button sequence.

Stair Task

The Stair Task was set up such that a set of stairs with four steps (0.15 m in height) were positioned flush with the back of the force plate (Figure 2). The subjects were directed to descend the steps in a reciprocating pattern without using the handrail, and land with their foot pointing straight ahead upon contact with the force plate. Eight total trials were collected: 2 straight trials and 6 crossover trials (3 to each side). The straight trials were performed by the subject descending the stairs and walking straight ahead; one with the right foot leading, one with the left. Three crossover trials were recorded by having the subject descend the stairs, land on their right foot, and swing the left leg over the right to walk at approximately 45 degree angle towards a cone. Three other crossover trials were completed where the subject landed on the left foot and took a crossover step with the right foot. The subject's limbs were also categorized as affected, unaffected, unstable, stable, or healthy. The subjects completed this

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task at self-selected pace and trials were recollected if the subject used the handrail, did not keep their foot pointing forward, crossed over in the wrong direction, did not take the stairs in a reciprocating motion, or if they hesitated on the crossover.

Principle Component Analysis

The variables analyzed in this study can be sorted into three categories: pre-measurements, TTT variables, and stair descent variables, as well as temporal, kinematic, and kinetic for the tasks (Table 2). These variables were entered into a PCA matrix to determine the relationship between these variables and how they contributed to the variation of the results. In total, there were 56 limbs, 28 subjects; 2 legs, (rows of data) and 29 variables for each subject (columns of data). All subjects were entered into the PCA data set without association to group. Average PC scores for the different groups were then compared in order to differentiate group performance. Independent t-tests were conducted for each PC comparing the groups. Significance was determined at an alpha level of $p < .05$. A discriminatory analysis was performed on all 29 variables in order to determine which variables represented group separation.

Results and Discussion

The first four principle components explained 40% of the variation in the data set and included a broad range of variables (Table 3). For each of the first four principle components generated from data, average PC scores for the three groups were calculated. Variables listed identify those components that were correlated or that moved together, and explained greater than 50% of the PC variation. The variables in the two columns (+ and -) indicate the relative direction the variables moved, with variables in the same column moving together, but opposite from variables in the other column. PC1 had the highest explained variation with 12.8%, but the first four PCs had relatively similar values for explained variation (ranging from 12.8% for PC1 to

8.5% for PC4). The mean and standard deviation (sd) of each group were calculated and group differences were identified by conducting independent t-tests for each PC using the significance of $\alpha < .05$ (p-value) (Table 3).

PC1 accounted for 12.8% of the variation in the data and was driven by six variables. The four variables that moved in the same direction were temporal aspects of the Stair Task; while GRF on impact after descending stairs and pelvis rotation range during the TTT moved in the opposite direction. More specifically, stance time (including single stance time), time to impact and time from push off to the end of the pivot turn after stair descent moved in the same direction while GRF at impact after stair descent and pelvis rotation during the TTT was less comparatively. There was a significant difference between both the TKR and Unstable groups when compared to Healthy for PC1 (p=.003, p=.007 respectively).

PC1 was largely influenced by temporal components indicating that the TKR and unstable groups demonstrated delay in the performance of task similar to previous studies[47,108]. It has been shown in previous studies that increased search time and a longer stance time are both found in individuals with either central processing delay or in the elderly at risk for fall[110,130]. A delay in task performance was consistent due to the fact there were significant differences in PC1 between the knee pathology groups and healthy. Adaptation to time includes a slower time to impact which individuals demonstrated descending stairs. However, slow to impact after stair descent may be associated with a safety issue rather than categorized as a compensatory mechanism. Trunk extension is another compensatory mechanism, not found in healthy controls, utilized in gait as an alternative to deceleration when loading the leg with knee pathology[33]. Although trunk positioning was not measured in the stair task, time to impact was delayed in the two groups with knee pathology suggesting they could have altered trunk posture to perform the task.

The second PC, which accounted for 10.4% of the variance between groups included 9 main variables. Seven of the variables move in the same direction and were largely influenced by the TTT; while the two inverse variables include one pre-measurement (laxity) and pelvis rotation range during the TTT. Only one stair task variable, GRF at push off during the pivot after stair descent, coincides with all the positive variables that loaded in PC2. There were no significant differences between the groups for PC2.

While excessive knee laxity or instability is a concern for the unstable group, the pre-measurement of laxity only loaded once in the total variance and was inversely related to variables mostly involved in the TTT. This suggests that the TTT may be a true test of knee stability if laxity was decreased with movement in the transverse plane that involved reaching while squatting and extending. Consequently, the laxity measurement in this study provides information on the anterior translation of the knee and may not correlate to actual rotary instability[59]. A clinical rotary test of stability, such as the pivot-shift, may be a better indicator of stability[18,98] with this type of task. Pelvis rotation range during the TTT moved in the same direction as laxity suggesting if the pelvis rotation is limited, the subjects utilized the upper torso to accomplish the task such as spine flexion. COP acceleration in both the AP and ML directions were inversely correlated to pelvis rotation range, again suggesting the task was accomplished with trunk movements rather than pelvic rotation. Moreover, GRF on the lead leg moved with COP supporting a larger shift in weight transfer to push the high button with a noted delay to move to midline.

Most of the variables loaded in PC3 were from the TTT and accounted for 8.7% of the variation. Seven of the 11 variables were positive features with 4 inversely correlated. Six of the seven positive features were TTT variables and represented a variety of constructs. Temporal items in the TTT correlated with movement of the subject in the sagittal plane and frontal plane to complete the task as well as the movement of the shoulders/torso in order to push the

buttons. Turning the pelvis in anticipation of the pivot after stair descent moved in the opposite direction as the torso movement in the TTT. The lack of pelvis compensatory movement in anticipation of a turn compared to TTT may be due to the need for upper extremity movement or reaching in the TTT while little to no arm movement is needed in the stair task. No significance was noted between these groups for PC3.

In double stance, such as in the TTT, subjects needed to maintain a stable posture in order to accurately press the high and low buttons. A change in posture in anticipation of using the upper extremities is referred to as Anticipatory Postural Adjustments (APAs). Alteration in posture during a dynamic movement could influence COP and the use of alternate muscles to adjust to the task[35,131]. Although muscle firing was not collected in this study, the COP acceleration, shoulder rotation and knee kinematics all moved in the same direction as cycle time and search time. This suggests that anticipatory postural adjustments (APAs) become apparent in individuals that require increased time to process the information to perform a task[124]. Therefore, the possibility of knee and trunk positioning via a central processing mechanism could be responsible for altered performance in a novel rotary task shown by individuals with knee pathology. Although both groups with knee pathology were able to complete the tasks, altered mechanics in the trunk and the lower extremities were utilized in order to achieve them. Conceivably, continued use of compensatory movements may produce abnormal mechanics or overuse to adjacent joints, leading to arthritic changes. The PCA was utilized as a discriminatory tool to understand these mechanisms.

PC4 accounted for 8.5% of the total variance and was influenced by 11 variables; 6 from the TTT, 2 from the Stair Task and 3 from pre-measurements. Knee angle loaded for both the TTT and Stair Task which also relates to hip flexion angle and knee moment. The pre-measurement, quadriceps strength, loaded a single time in the first four PCs and was inversely correlated in PC4. Temporal variables were also inversely associated with quadriceps strength

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suggesting an added component of neuromuscular timing. Significant difference (p=.017) was found between TKR and Healthy groups.

Quadriceps avoidance is a common protective mechanism utilized by individuals with knee pathology[21,37,38]. This term, coined by Andriacchi in the early 1990's, was used to describe the extended position of the knee during gait; with little to no use of flexion for both weight acceptance and push off. Similarly, in the rotary tasks of TTT and stair descent with a pivot; the TKR individuals maintained a more extended knee and demonstrated less weight acceptance on the affected leg[47,108]. Both tasks included an increased loading phase; transferring the load onto a single leg to pivot and turn after the stairs or to transfer weight to reach a button. For example, variables that correlated in PC4 were knee angle on the stance leg at impact and push off in the stair task and knee, hip angle on the lead leg during the TTT. The PC scores were significant only between the TKR group and healthy, despite previous findings that the unstable group demonstrated significantly decreased GRF on initial impact and a trend to maintain and extended knee on the stance leg after stair descent[129].

Loading differences on the lead leg during the TTT H2L task showed the unstable group loaded the unstable leg more when compared to the healthy group[129], while the TKA group loaded the lead leg less when compared to the healthy group[47]. The difference between the unstable group and the TKR group may be due to a concept utilized for individuals with compromised structures of the knee who are able to perform every day activities without difficulty. These individuals, defined as 'copers', may even return to sport and use proper neuromuscular timing during high demand activities to overcome their deficiencies[132,133]. Neuromuscular timing was not calculated in this study, however quadriceps strength and laxity were measured prior to task performance. Although quadriceps strength has been correlated with improved functional outcomes[24,49,134], the variable only loaded once with the 40% total variation in both tasks. Quadriceps strength was inversely correlated with knee angle

suggesting the need to maintain knee extension if the quadriceps is weak. Concurrently, strength was correlated with temporal variables suggesting possible delay or altered muscle firing, which could lead to injury[97,135]. Therefore, weakness of the quadriceps suggests use of the quadriceps avoidance mechanism, complicated by a neuromuscular timing deficit. The combination of these mechanisms could increase the risk of injury to the knee joint and adjacent joints.

PC9 and 11, although further along in the explained variation (4.1% and 3.5% respectively), shows some differences between the TKR and healthy group. While PC9 was trending toward significance (p=.056), PC11 was significantly different (p=.019) between the TKR and healthy. The theme of knee angle and time is revealed in PC11 with added shoulder rotation or torso movement to accomplish the task. This suggests that the torso is utilized while moving toward the target to push the button influencing the search time to find the button. Interestingly, while laxity is listed as an inversely correlated variable, significance was found only between the TKR and healthy rather than the Unstable group. Consideration of the possible decreased knee angle for the TKR individuals in both PC9 and 11 and the use of shoulder rotation and hip flexion to achieve the task is supported in the literature as a strategy used to perform a turning or pivoting task[47,108,117].

A total of 12 PCs was required to explain 80% of the total variance (Figure 3) suggesting there are many variables that occur during these two rotary tasks that explain the variation of subject performance. The dispersed variance reveals there are no strong correlations amongst the groups identifying one dominant factor and significance of group differences allows variables to be grouped showing individuals with knee pathology perform rotary tasks differently than healthy controls.

The PCA model established correlative variables during two rotary tasks for TKR individuals, individuals with non-surgical self-reported knee instability and healthy individuals. While other researchers have used the PC method to reduce the data and develop common constructs that identify various gait deviations[126-128], this study used PCA across two different tasks with multiple groups to determine if there were identifiable relationships between variables that might correlate patterns of movement. A wide range of variables were identified in the first four PCs that accounted for 40% of the total variance of how these groups performed both the TTT and stair task. Interestingly, compared with many PCA studies which have a few dominate PCs, this model did not identify a strong dominate correlation that explained a disproportionately high amount of variation. Instead, the first four PCs explained similar variation and suggest four different, but similarly important relationships between the variables. The variables of the four PCs themselves suggest interesting relationships, some of which distinguish between groups of interest.

A discriminatory analysis was conducted to identify the differences between the TKR and healthy groups. Previous analysis was conducted on the 29 variables between the Unstable group and healthy finding time to impact or the temporal variable was the best predictive variable between these groups[129]. The five variables that were the most predictive between the TKR and healthy group agree with the findings that knee flexion angle differentiates the TKR and health groups (Table 4).

Of the five variables, there was a misclassification of 3% and these five variables were then divided into sub-groups of pre-measurement, TTT and Stair Task for another discriminatory analysis. The misclassification of the pre-measurements was 12%, TTT 6% and Stair Task 12% concluding that the TTT may be the best tool for discriminating between the TKR and healthy group.

Conclusions

The objective of this study was accomplished by discussing correlative variables for individuals with a knee pathology compared to healthy individuals while performing two rotary tasks. Use of the PCA provided an account of components that could largely be discussed as themes: temporal, kinetic and kinematic and pre-measurements. It is evident that individuals that have knee pathology utilize compensatory mechanisms that allow functional movement to achieve tasks that include rotary demands with loading. Compensatory mechanisms that loaded in the PC were use of the torso, pelvis and upper extremities to limit movement about the knee as well as positioning the knee in a more extended position to avoid use of the quadriceps muscle possibly due to weakness. Lastly, a central processing mechanism for task planning was required altering the timing and speed of the task, yet not affecting the ability to complete the task. Limited loading and weight shift on to the affected leg could affect adjacent joints and produce altered positioning. Therefore, rehabilitation interventions that address task planning, avoidance of compensatory movements should be instituted early in order to avoid further injury; primarily to the contralateral knee, hips and low back.

Limitations

The unstable sample size was comparatively smaller than the TKR and healthy groups, however both legs were utilized in the data collection therefore, doubling the n size. The unstable group was characterized by a subjective report of frequent episodes of knee 'buckling' without knowledge of specific pathology. Although subjective report is a key clinical determination of knee instability, the addition of a clinical test to confirm rotary instability may have added to the robust categorization of 'instability' for this group. The age range of the unstable group contains a younger population when compared to the TKR and healthy group. However, only minimal biomechanical differences were reported during stair descent when

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comparing older individuals to a younger group[110,111]. Lastly, the criteria to exclude individuals that required use of a handrail to descend stairs may have limited the ability to recruit those individuals with knee pathology that were unstable. The added variable of using the handrail would have added confounding variables to the stair descent task while the emphasis of this research was on the rotary portion of the task; not specifically stair descent.

Author Contributions

Linda Denney contributed to project conception, data collection, data analysis, data interpretation, and drafting the manuscript. Lauren Ferris contributed to data collection, data analysis, data interpretation, drafting and editing the manuscript. Hongying Dai contributed to statistical analysis of data, and data interpretation of the manuscript. Lorin Maletsky contributed to project conception, data analysis, data interpretation and editing the manuscript.

Tables and Figures

Table 1: The mean (standard deviation) of the demographics and pre-measurements of all participants

Groups		Age	BMI	Hip ROM (degree)	Knee ROM (degree)	K-T 1000 (mm)	Quad MMT (%BW)
TKR (n=10)	Affected	66 (6)	26.9 (3.0)	106.1 (7.4)	127.2 (7.8)	3.0 (1.0)	0.19 (0.05)
	Unaffected			111.6 (8.7)	138.8 (5.9)	2.8 (0.9)	0.18 (0.03)
Unstable (n=6)	Unstable	37 (13)	25.6 (2.1)	108.7 (7.8)	139.8 (5.3)	2.2 (0.7)	0.17 (0.04)
	Stable			109.2 (7.6)	141.2 (4.0)	2.1 (0.6)	0.18 (0.03)
Healthy (n=12)		65 (8)	24.6 (3.8)	109.1 (10.8)	141.2 (5.7)	2.4 (0.9)	0.19 (0.03)

Abbrev: TKA: Total Knee Replacement, BMI: Body Mass Index, ROM: Range of Motion, MMT: Manual Muscle Test

Table 2: Abbreviation, description and type of variables collected for both tasks.

	Variable Abbreviation	Variable Description	Type
Pre-test	QuadMMT	Quadriceps strength (in %BW) of the subject via a dynamometer	Pre-measurement
	Laxity	Anterior-posterior translation of tibia relative to the femur via a KT-1000	Pre-measurement
	HipFlex	Hip flexion angle via goniometer	Pre-measurement
	HipExt	Hip extension angle via goniometer	Pre-measurement
	KneeFlex	Knee flexion angle via goniometer	Pre-measurement
	KneeExt	Knee extension angle via goniometer	Pre-measurement

TTT	SearchTime	Time the subject took to press the second button once their wrist marker was within a magnitude of .15 m of the button	Temporal
	H2Lcycletime	Time the subject took to complete a H2L crossover	Temporal
	Timeto.5H2L	Time from the first button push to when the subject had equal weight distribution on the force plates	Temporal
	GRFrangeLFR	Range of weight distribution (in LFR)	Kinetic
	MaxKFAleadH2L	Maximum knee flexion angle of the lead leg	Kinematic
	MaxHFAleadH2L	Maximum hip flexion angle of the lead leg	Kinematic
	PelvisRotRangeH2L	Range of the pelvis rotation via the markers on the subject ASIS during the crossover	Kinematic
	SHRotH2Lrange	Range of the shoulder rotation via the markers on the subjects shoulder during the crossover	Kinematic
	SpineFlexH2LMax	Maximum flexion of the spine via the markers located on PSIS and C7	Kinematic
	H2LleadMz020	Maximum moment (in %BW*%height) about the force plate of the lead limb from 0-20% cycle	Kinetic
	H2LlagMztoend	Maximum moment (in %BW*%height) about the force plate of the lag limb once the subject crossed over 0.5 LFR to the end of the cycle	Kinetic
	COPrangeMLAccel	Range of Center of Pressure acceleration in the medial-lateral direction	Kinetic
	COPRangeAPAccel	Range of Center of Pressure acceleration in the anterior-posterior direction	Kinetic
Stair	Stancetime	The amount of time the subject was in contact with the force plate after stair descent	Temporal
	Timetoimpact	Time after leaving last step to impact ground	Temporal
	TimetoAPtoend	Time from active peak of the gait cycle to the end of the stance time	Temporal
	SingleStancetime	Time the subject was in single stance during the crossover	Temporal

	StanceKFAAP	Knee flexion angle at the active peak			Kinematic		
	StanceKFAPP	Knee flexion angle at the passive peak			Kinematic		
	Pelvisrotchangefrom straightPP	Rotation of the pelvis after stair descent compared to the straight stair trials.			Kinematic		
	GRFPP	GRF at the passive peak			Kinetic		
	GRFAP	GRF at the active peak			Kinetic		
	Mzatturn50100	Maximum moment (in %BW*%height) about the force plate from during the second half of the stair activity			Kinetic		

Table 3: Principle Components Analysis (PC 1-5) loading vectors for the Target Touch Task and Stair Task.

	% Variation Explained	Variables		TKR	Unstable	Healthy	p- value
		+	-				
PC 1	12.8	Stance time TimetoAPtoend Timetoimpact SingleStancetime	GRFPP PelvisRotRangeH2L	0.82(1.75)	1.83(1.86)	-1.39(0.94)	.003[*] .007[†]
PC 2	10.4	GRFAP SpineFlexH2LMax GRFrangeLFR COPRangeAPAccelTTT COPrangeMLAccelTTT Timeto_5H2Lsec H2LlagMztoend	PelvisRotRangeH2L Laxity	-0.57(2.21)	0.81(0.83)	-.34(1.67)	
PC 3	8.7	H2Lcycletime SearchTime COPrangeMLAccelTTT, COPRangeAPAccelTTT STanceKFAPP SHRotH2Lrange	HipFlex, KneeFlex HipExt Pelvisrotchangefro mstraightPP	0.36(2.16)	-0.28(1.38)	-0.23(1.43)	

PC							
		MaxKFAleadH2L					
PC 4	8.5	SHRotH2Lrange StanceKFAAP StanceKFAPP MaxHFAleadH2L MaxKFAleadH2L H2LleadMz020	SearchTime KneeExt H2Lcycletime HipExt, QuadMMT	-1.01(1.73)	0.19(1.14)	0.40(1.38)	.017*
PC 9	4.1	Mzatturn50100 MaxHFAleadH2L HipExt KneeExt StanceKFAPP	KneeFlex COPrangeMLAccelTTT Timeto_5H2Lsec	0.82(1.30)	-0.32(1.14)	0.01(0.99)	.056‡
PC 11	3.5	KneeFlex SearchTime SHRotH2Lrange	PelvisrotchangefromstraightPP Laxity StanceKFAAP	-0.75(1.13)	0.25(1.23)	0.18(0.95)	.019*

* significant difference between TKR and Healthy
† significant difference between Unstable and Healthy
‡ trending significance between TKR and Healthy

Table 4: Five variables between the TKR group and healthy from discriminatory analysis of 29 variables of both tasks.

	Variable		R^2
1	Knee Flexion	Pre-measurement	0.49
2	Max H2L lead Knee Moment 0-20%	TTT Kinetics	0.63
3	Time to Impact	Stair Temporal	0.69
4	Max H2L lead Knee Flexion Angle	TTT Kinematics	0.73
5	Knee Extension	Pre-measurement	0.76

Figure 1: Subject performing TTT H2L cycle; red circles around buttons; yellow outline of force platforms.
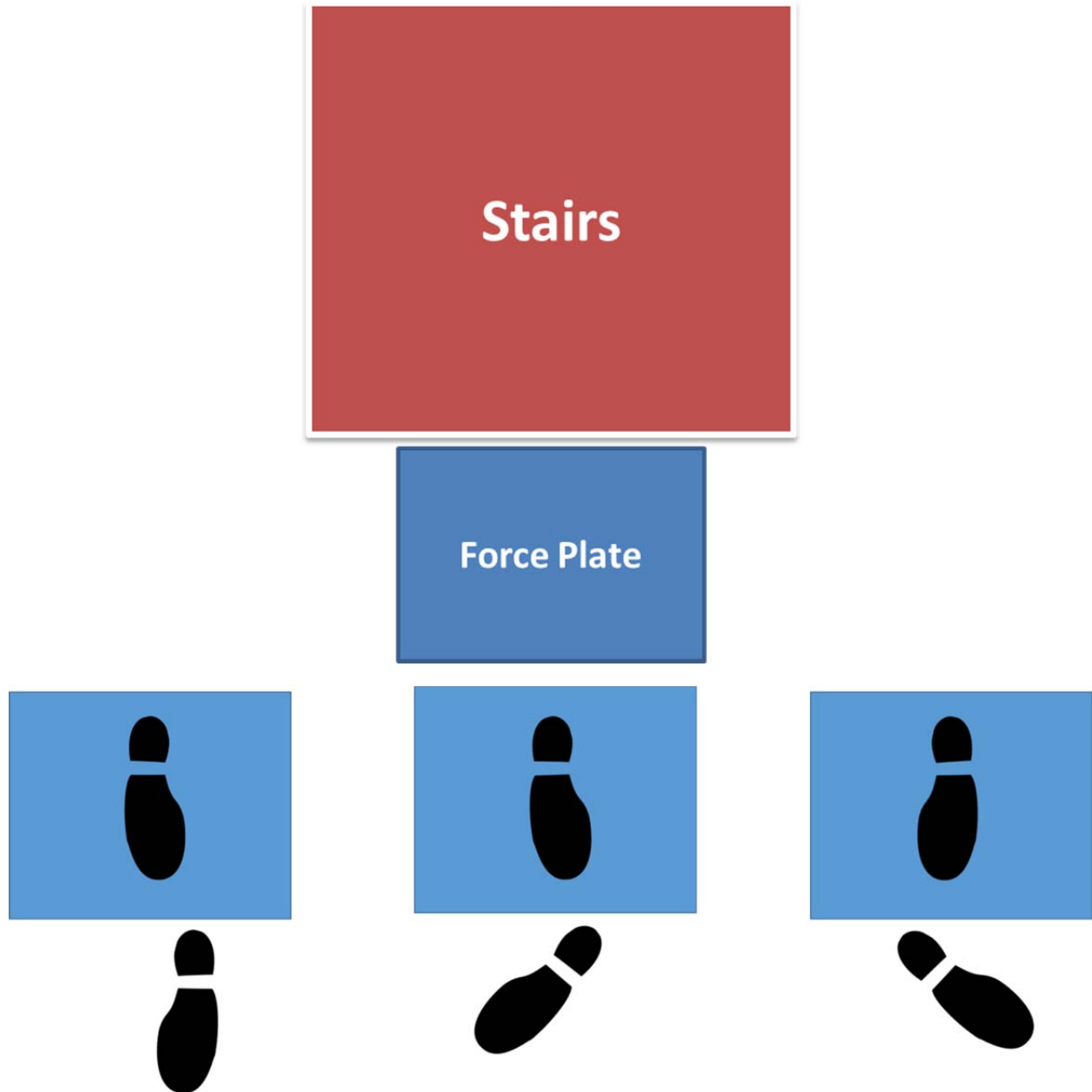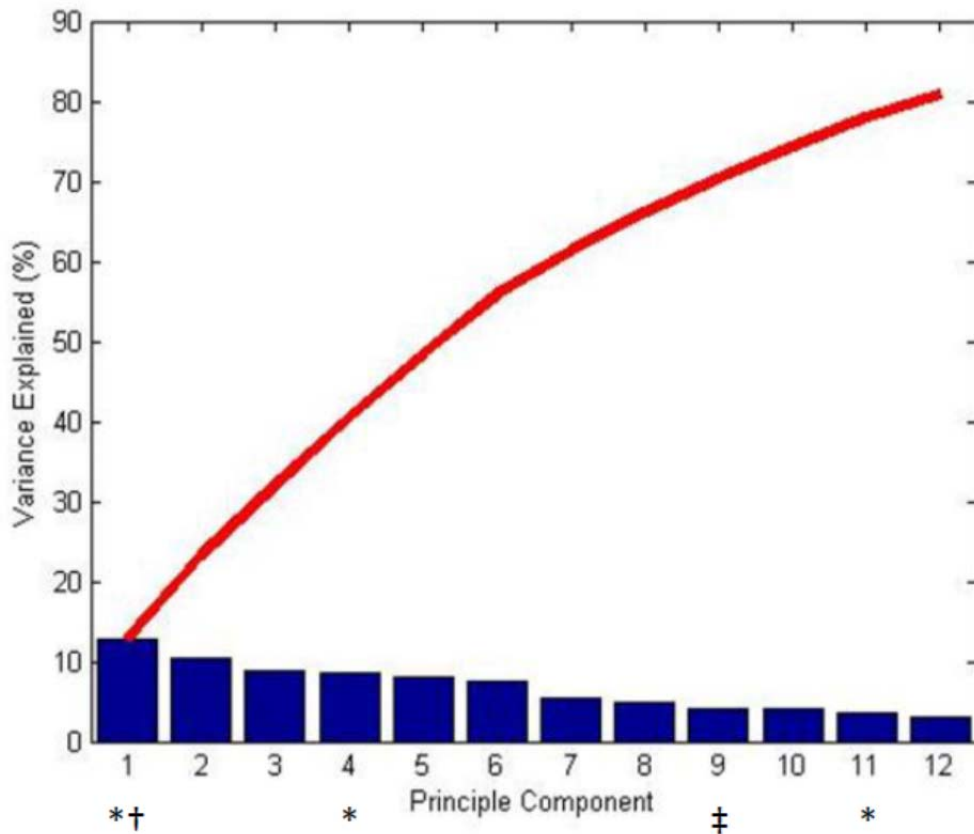
Figure 2: Stair set up with force plate and step sequence for straight, right turn and left turn

* significant difference between TKR and Healthy
† significant difference between Unstable and Healthy
‡ trending difference between TKR and Healthy

Figure 3: Percent of variance explained in the 12 principle components

Chapter VI

Discussion

This body of work explored the performance of individuals with a TKR and those with non-surgical self-reported knee instability during two rotary tasks and compared these groups with a healthy group. Overall, the outcomes are in agreement with previous literature that individuals with knee pathology utilize strategies to perform rotary tasks that include loading, weight-shifting as well as transfer of weight with squatting and reaching. Individuals with knee pathology not only performed rotary tasks slower, they also demonstrated use of compensatory movements such as torso rotation, trunk flexion and maintaining an extended knee in order to complete the tasks. The correlative findings between the TKR and unstable individuals suggest that the TKR group may feel less 'stable' or secure with the use of the surgical leg and therefore use strategies to decrease the impact of rotary forces on their surgical leg. It is possible that these strategies were established prior to TKR surgery as the unstable group demonstrated some of the same strategies. The findings of this work may impact rehabilitative interventions to prevent the use of strategies post-TKR and encourage symmetrical loading of the lower extremities during ADLs.

Chapter II: Analysis of rotary task following total knee arthroplasty: Stair descent with a cross-over turn

The purpose of this work was to determine the difference of TKR and healthy individuals in the performance of stair descent followed by a pivot turn. Our results indicate that individuals with a TKR impact the ground at a slower rate, with less weighted force after stair descent. TKR individuals maintain a more extended knee on the affected leg at both initial impact and on the stance leg at pivot turn when compared to healthy individuals. Although this may suggest a precautionary manner to avoid falls, the extended knee may also be a compensatory mechanism or strategy during a pivoting maneuver to avoid rotary forces at the knee and increase stability.

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

The purpose of this work was to determine if TKR individuals performed a rotary task differently than healthy controls and identify strategies utilized to complete the task. Our results indicate that TKR individuals maintained a more extended knee position during the rotary task of squatting and reaching. The TKR individuals loaded the unaffected leg more than the healthy group suggesting an unloading of their affected leg. Both considerations of knee position and loading suggest that the TKR individuals utilize strategies to complete the rotary task that may impact other joints such as the low back, hips and contralateral knee.

Chapter IV: Biomechanical strategies of individuals with self-reported knee instability and healthy controls during two dynamic rotary tasks

The purpose of this work was to determine how individuals with self-reported knee instability perform rotary tasks differently than healthy controls. The unstable group utilized a more extended knee when impacting the floor after stair descent and had less ground force on both the affected and unaffected leg when compared to the healthy group. During the TTT both groups loaded the legs similarly, however, the PCA correlations indicated spine flexion was utilized as a possible strategy to accomplish the TTT. Overall the differences between the groups were highlighted in PC1 with temporal aspects dominating the correlations. The unstable group performed the tasks at a slower rate when compared to the healthy group. The hesitation or slower performance was correlated to other variables such as quadriceps strength, spine flexion and knee flexion angle in PC2-4. This suggests compensatory torso and lower limb strategies were utilized to achieve the task which may contribute to the increased time to complete the task. Additionally, the slow rate of performance and quadriceps strength suggests

a possible neuromuscular delay or slow timing issue associated with the performance of the two rotary tasks.

Chapter V: Analysis of two rotary tasks comparing the performance of individuals with a total knee replacement, non-surgical self-reported knee instability and healthy controls

The purpose of this work was to understand how individuals with TKR, individuals with self-reported knee instability and their healthy controls perform rotary tasks that involve flexion and extension of the knee during dynamic loading of the lower extremities in both a stair task and reaching task through the use of a Principle Component Analysis (PCA). The results indicate that individuals that have knee pathology perform the tasks slower than the healthy group and utilized compensatory mechanisms such as movement in the torso, pelvis and upper extremities to limit movement about the knee as well as positioning the knee in a more extended position to avoid use of the quadriceps muscle possibly due to weakness. It is possible that a central processing mechanism for task planning was required altering the timing and speed of the task, yet not affecting the ability to complete the task.

Discussion

Individuals with a TKR present with the likelihood of demonstrating kinematic and kinetic strategies during rotational tasks. This is consistent with the literature on obstacle avoidance and level-straight walking[7,8,12,46]. These strategies may be defined by use of an extended knee during gait, known as 'quadriceps avoidance'[37], overuse or overloading of the contralateral knee[28], altered positioning of the trunk[33] and pelvis or slower performance that may be related to safety and prevention of falls[31].

Rotary task performance is relevant to TKR individuals in order to perform ADLs while a younger population with knee pathology may need to consider return to sport. The bulk of knee research during pivoting has been in sport-related activities and little is known about how the post-TKR individual performs rotary tasks. Recipients of a TKR are younger and more active and have expectations of return to activities such gardening, tennis, golf and running[5]. Some of these activities involve squatting and reaching while twisting and pivoting. Unfortunately, despite rehabilitation efforts, some individuals continue to unload the surgical leg, consequently overloading the contralateral leg[28]. Return to the high activity level with asymmetrical loading may lead to arthritic changes in the opposite knee resulting in a TKR on the non-surgical knee[25,29] and could lead to compensatory movements at the hip and trunk. Altered mechanics at the trunk have shown to improperly load the lower extremities. This is evident in both TKR and individuals with knee pathology[33,117]. The torso and hip position have a large impact on the lower limb mechanics especially during pivoting or change in direction and therefore could lead to further injury if not corrected.

Change in trunk position or use of the torso to alter the loading of the lower extremities was consistent in the knee pathology groups during both rotary tasks in the current study. Reaction to perturbation during pivoting or change in direction would be considered an anticipatory response (APA) as the individuals were instructed in the task prior to performance. However, the only adjustment that would be categorized as a true APA would be the change in pelvic position prior to the pivot at the bottom of the stairs. Despite the cues to maintain foot forward placement on the stance leg when impacting the floor, individuals started a rotation of the pelvis in the direction of the turn prior to the pivot. The remaining variables that define compensatory movements, or CPAs, were adjustments made during the task in response to perturbation. For example, spine flexion and torso rotation was utilized during the TTT in order to press the buttons rather than flexing the knee or transferring weight to the lead leg. Also,

knee extension was maintained to gain stability on the stance leg on the pivot turn after stair descent. Overall, both knee pathology groups performed the tasks at a slower rate than the healthy group and had increased search time when transferring weight to press the high button. Both Li et al. (2013)[33] and Jamison et al. (2012)[117] reported deceleration or a hesitation in task during walking and playing sport. The TKR individuals utilized back extension to lessen the impact of force at the knee, while individuals post-ACL reconstruction, reported kinesiophobia when returning to play. Further, torso movement away from the direction of the cutting maneuver in the Jamison et al. (2012)[117] study was related to peak knee abduction moment which predisposes the knee to ACL injury. Therefore, the trunk and torso position could impact the kinematic position of the knee as well as predispose the individual for further injury.

It is argued that quadriceps weakness largely influences knee positioning, potential future reinjury and predicts estimated return to sport. As a modifiable factor for return to sport, quadriceps strengthening has gained popularity in post-ACL reconstruction rehabilitation. Similarly, institution of neuromuscular electrical stimulation (NMES) to the quadriceps has proven to gain functional status for individuals post-TKR[136]. Quadriceps testing includes one maximum recorded contraction or an average of three with the use of an isometric testing device or it may be quantified functionally by the performance of stair descent without a handrail, a step down or single-leg squat or hop test. Unfortunately, these measures do not predict the function of a rotary task, pivot or cutting maneuver. Both knee pathology groups in our study demonstrated altered kinetics and kinematics on rotary tasks despite no significant differences in quadriceps strength from the healthy group. Therefore, quadriceps strength may not be a predictor of altered mechanics but when associated with timing as found in the PCA could predispose the individuals to adopt a strategy.

A low quadriceps-hamstring strength ratio has been reported to affect return to sport[118] while a large co-contraction of quadriceps-hamstring may reduce shear forces on the TKR yet

increase the wear on the implant[21]. The coordinated coactivation and force generation must be perfectly timed in order to stabilize the knee during vigorous activity. This involves proper neuromuscular timing along with proprioception of all adjacent muscles for a safe orchestration of movement. Neuromuscular timing can be measured with the use of EMG; however it is difficult to capture efficient joint mechanics with associated motor control in order to predict safe return to sport[118]. Output from EMG only provides a portion of what is needed during movement analysis. Both proprioception and motor control are part of the central processing system driven by the CNS. Therefore, only inferences can be made about the quadriceps, trunk and hip positions, and temporal variables listed as correlative factors in the PCA during the rotary tasks that include both TKR individuals and those with non-surgical knee pathology. It is possible that both knee pathology groups had neuromuscular timing deficits resulting in a slower performance of tasks with related strategies utilizing trunk, hip and extended knee positions to perform the tasks. Error in trunk repositioning, trunk lean and trunk flexion displacement has been related to ACL reinjury in athletes[116,119]. Further studies are needed to predict if the postural adjustments associated with neuromuscular timing deficits are promoting improper loading of the lower extremities in individuals with a TKR.

Multiple strategies have been reported in relation to how TKR individuals function post-surgery. Most strategies are related to gait that includes stride and gait velocity[31,33]. Our study incorporated loading and transfer of weight with a rotary task and also included the torso for reaching. Various strategies were revealed in the PCA without one dominating factor. This concludes that during these rotary tasks there were many variables that influenced the performance and further research is needed on postural adjustments, neuromuscular timing and the effect of knee positioning and loading.

Clinical Implications

Over 700,000 individuals receive a TKR annually in the United States[1] and most of these individuals would like to return to recreational activity. The longevity of the implant has become a priority due to the younger age of the recipients. Therefore, it is important to address the factors that may impact adverse outcomes of TKR individuals adopting strategies that will predict functional performance. More specifically, this research provides insight for out-of-plane activities that include loading and transfer of weight while flexing and extending the knee.

Symmetrical training has been incorporated into rehabilitation post-TKR in order to ensure proper loading of both lower extremities during functional activities[85]. The findings in our study conclude that symmetrical training of the lower extremities is necessary in order to avoid overloading of the contralateral leg. Balance training may assist with neuromuscular timing deficits if provided in a functional manner that includes rotation. Rehabilitation to address proper loading and balance may decrease the tendency for individuals post-TKR to adopt strategies.

It is uncertain if quadriceps strengthening alone will assist in increased function as reported in the literature[24,136], as our participants had comparable quadriceps strength and were able to complete both tasks. Although some work has been completed on postural adjustments with TKR individuals[27] with findings of decreased quadriceps EMG during a standing reaching task, further research is needed to incorporate rotation. The TTT proved that the torso impacts the loading of the lower extremities and could relate to activities that include squatting such as getting in and out of a car or bathtub. Rehabilitation interventions should incorporate transfer of weight activities that simulate everyday activities as well as return to sport activities such as golf.

Limitations

The TKR individuals who participated in this study were high-level functioning individuals with a low BMI. The inclusion criteria for the Stair study required that they descend stairs without

the use of the handrail, therefore it is possible that those recruited do not represent the typical TKR individuals with knee instability. If the criteria were to include individuals requiring the handrail, a harness/safety mechanism would have to be administered as well as further approval through the Human Subject Committee. Additional confounding variables would also have to be considered such as using the hand-rail during the Stair Task.

The small sample size (n=6) of individuals with non-surgical knee instability may have impacted results of the study, however, each leg was considered when data was entered for affected and unaffected resulting in 2 variables per subject. Also, these individuals were deemed 'unstable' only by frequency of 'giving way' or 'buckling' reports in the last three months. No clinical tests were performed to determine knee instability besides the laxity test (KT-1000).

External markers utilized in the 6-camera Vicon motion analysis were attached to the skin or clothing and may have been subject to movement during the study. This may have compromised the accuracy of true kinematics as it is not a depiction of bone movement. Benoit et al. (2007)[11] compared skin markers to actual bone pin markers for data collection during gait and revealed two kinematic measurements that are not interchangeable. Knee internal-external rotation and anterior-posterior tibial translation were not recorded as the same kinematic motion when the two types of measurement were compared, however the use of bone pins to collect data in this study is not an option. Therefore, use of the force platform data was used to establish knee rotation. The use of fluoroscopy is a more effective option however, very expensive and difficult to conduct with patients actually traversing on stairs and changing direction. Therefore, the 6-camera Vicon system with reflective markers attached to the skin was the best approach.

Patients were fitted with various types of prosthetic implants; each functions in a different capacity for individual patients. Collective assumptions made from data collection of knee kinematics may be driven by the type of prosthetic implant and not the individual differences of the human motion: or vice versa. Implant type could be included as a future analysis for comparison of performance during rotary tasks. This paper is not intended to compare total knee prostheses, however may be of interest to surgeons and industry and eventually beneficial to the patient population.

Future Direction

In summary, the large amount of data collected revealed trends that depict difference in groups of TKR individuals, those with non-surgical knee pathology and healthy individuals. The large amount of data collected could assist in calculating predictive gait patterns as well as predictive muscle forces. Further correlations can be calculated with use of the pre-measurements and task results such as hip extension availability and negotiating stairs as well as hip moment with stair descent. Video analysis for postural adjustments during rotational tasks could be correlated to transfer of weight and an intervention could be incorporated to determine effectiveness of balance training. A computational model could be made to analyze muscle forces during the stair task. Lastly, a comparison of implant type and task performance could be conducted. Further analyses may prove to be beneficial to surgeons, therapists and for future implant design. Future use of the data may be ideal for simulation loading models and development of computational models.

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Appendix

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, Linda M Denney and Lorin P Maletsky

Author contributions

Lauren Ferris presented this material for her Master thesis in Mechanical Engineering at the University of Kansas, Lawrence, KS. Ferris created a subject specific musculo-skeletal model using an open source software, OpenSim. The manuscript was completed via collaboration of all three authors. More specifically Lauren Ferris created the model using previous data collection, performed data analysis, data interpretation, statistical analysis and drafting of the manuscript. Linda Denney contributed to original data collection, data interpretation, statistical analysis, and drafting and editing the manuscript. Lorin Maletsky contributed to project conception, data analysis, data interpretation and editing the manuscript.

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.

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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 maximus, 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 50o to perform the low button

push and then extended to approximately 10-25o 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 90o. 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

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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.

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Tables and Figures

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

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)

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
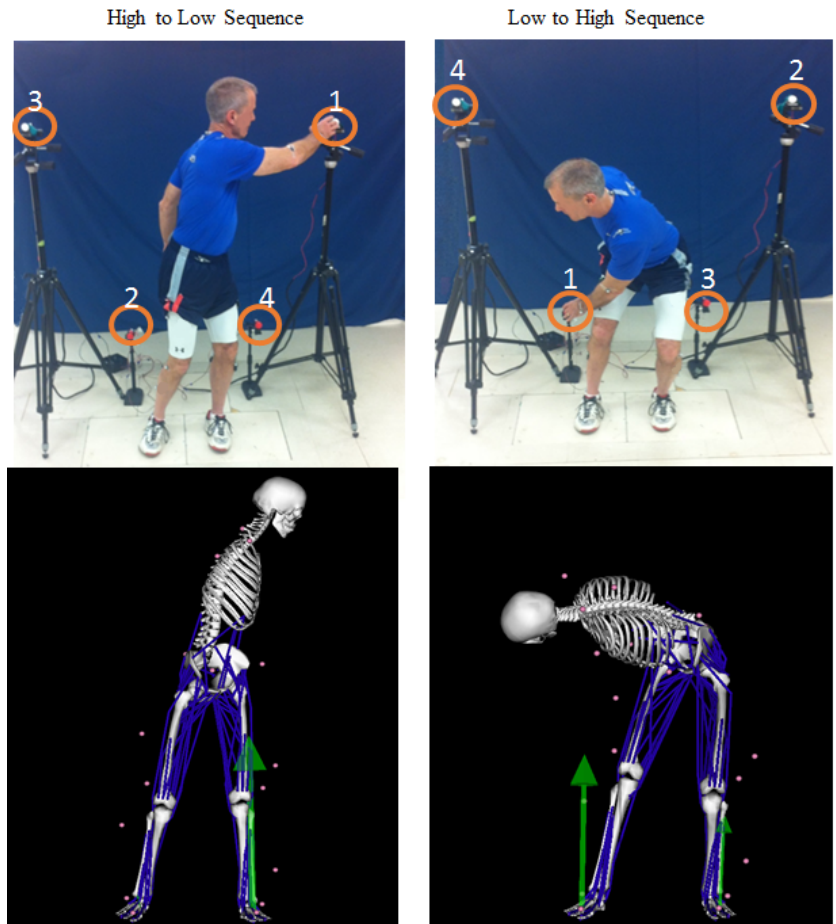
Figure 1: Equipment setup up along with a subject performing a H2L sequence (A) and L2H sequence (B). Numbers indicate the button sequence to perform the given task. Same subject performing the tasks in OpenSim (C, D). Arrows indicate forces recorded by the force plates.
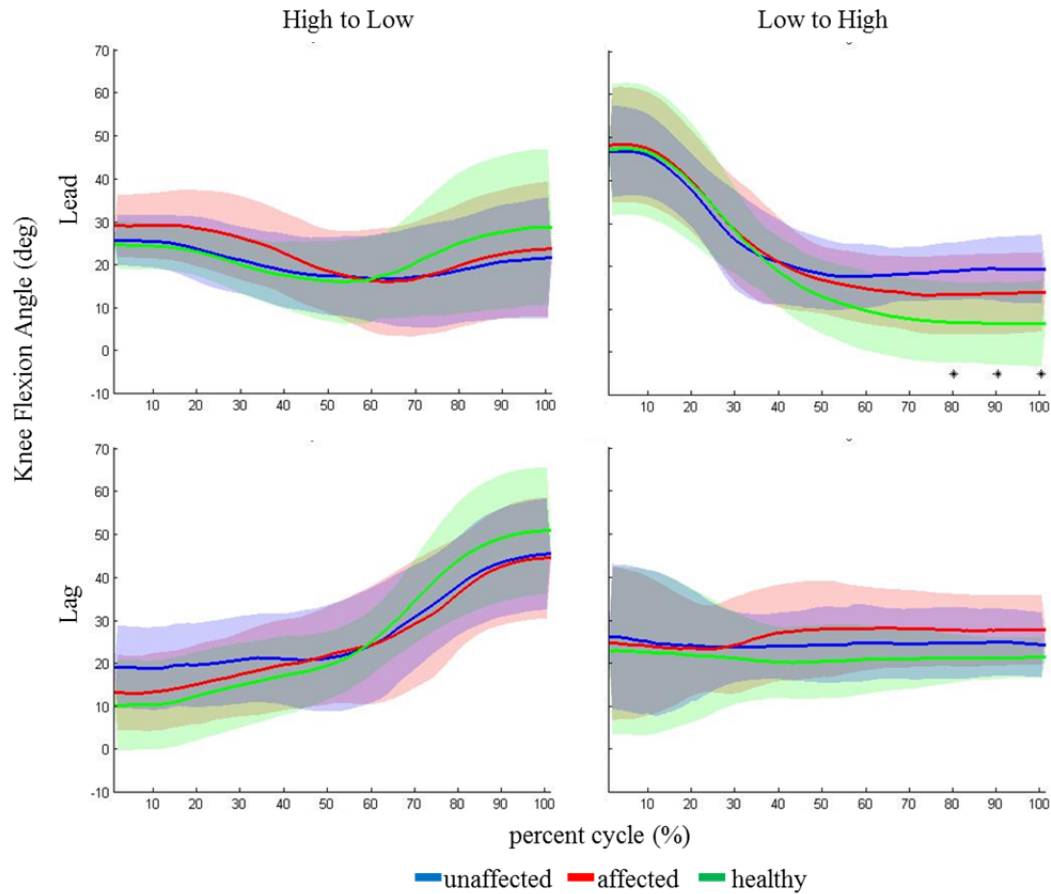
Figure 2: Average knee flexion angle for unaffected, affected and healthy subjects. Standard deviation ±1 indicated in the shading. Top row depicts the lead leg of the H2L and L2H task with the bottom row depicting the lag leg. A single factor ANOVA was performed at each 10% cycle. (The * denotes unaffected statistically significant ($p < 0.05$) from healthy.)
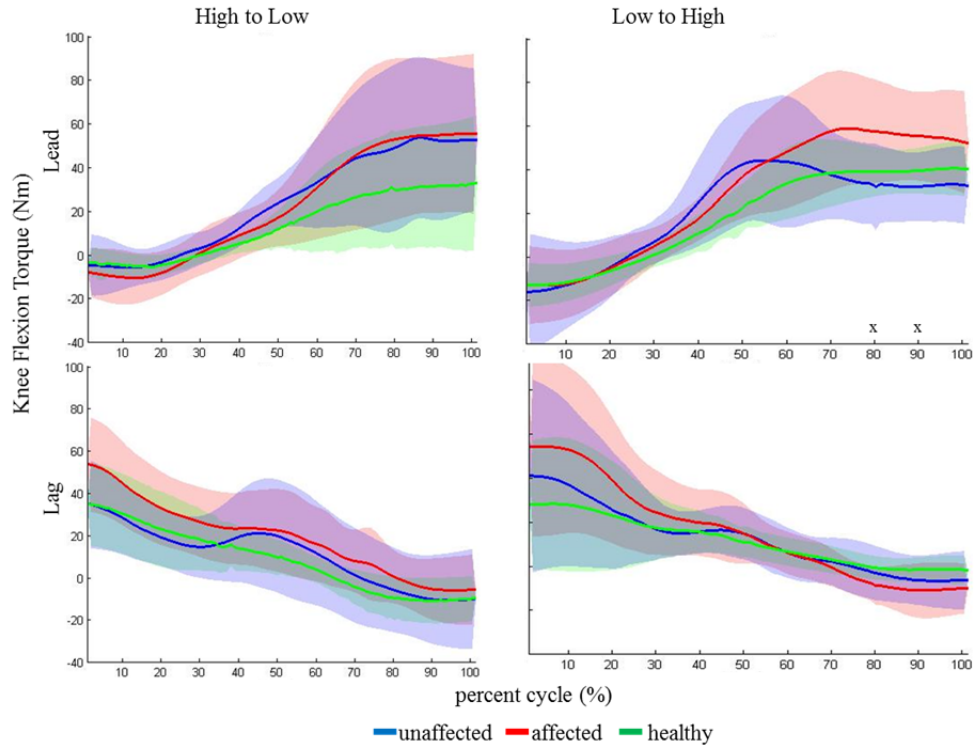
Figure 3: Mean knee flexion torque during H2L (column 1) and L2H (column 2) for unaffected, affected and healthy subjects. Standard deviation ±1 indicated in the shading. Top row depicts the lead leg with the bottom row depicting the lag leg. A single factor ANOVA was performed at each 10% cycle. (The x denotes affected statistically significant ($p<0.05$) from unaffected.)
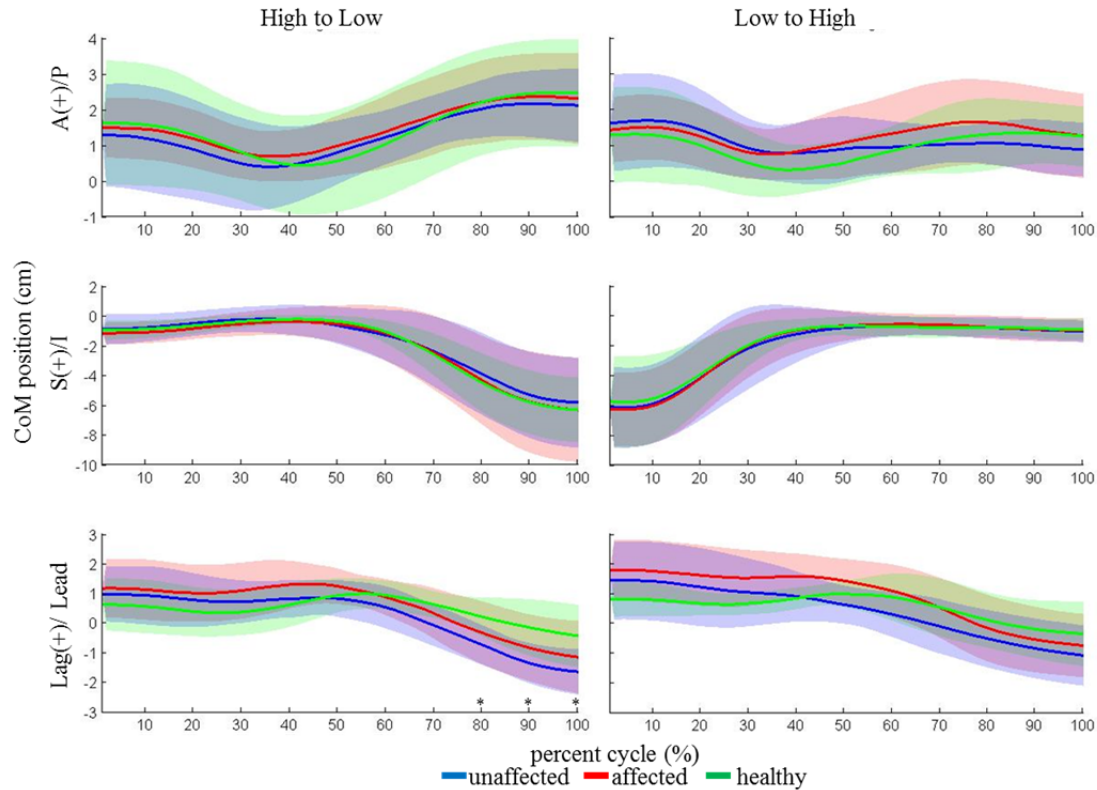
Figure 4: Mean center of mass deviation from static position in the anterior/ posterior (A/P) directions (top row), superior/ inferior(S/I) directions (middle row), and side-side position with the subject started towards the lag leg (+) then transferring to the lead leg (bottom row). Columns indicate 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.)
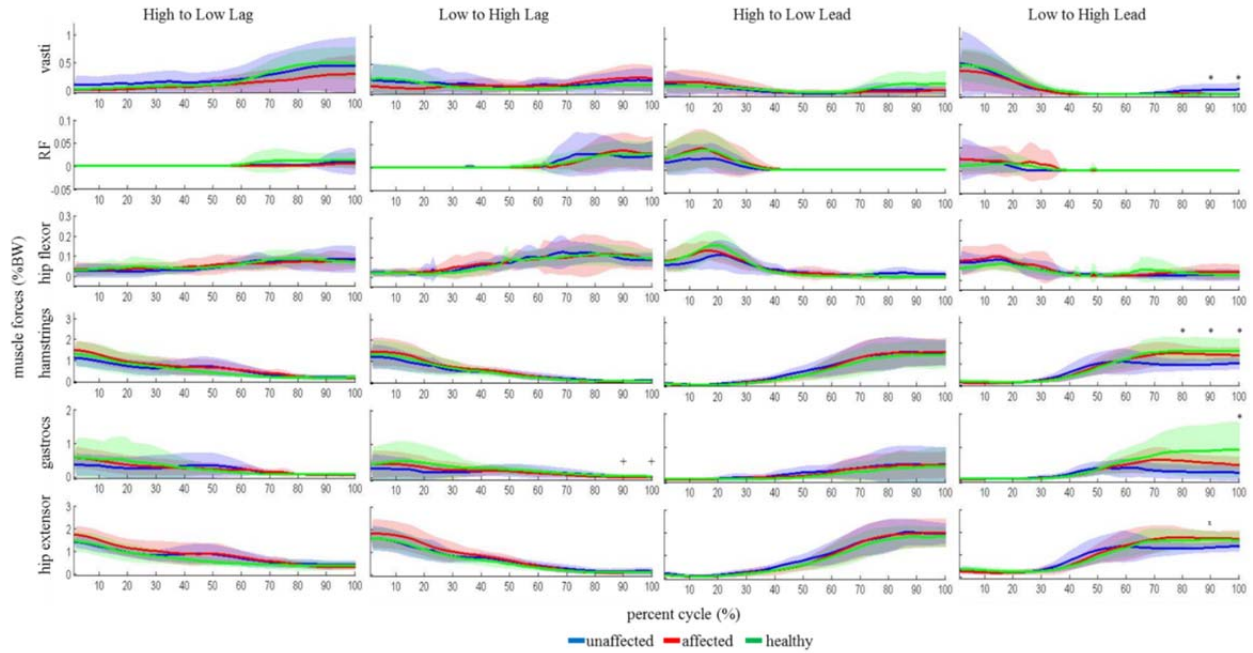
Figure 5: Mean muscle activation in the vastii, rectus femoris, hip flexors, hamstrings, gastrocnemius, and hip extensors during both H2L and L2H tasks. First two columns represent the lag legs, or the leg closest to the first button push. Last two columns represent the lead leg, or button closest to 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.)