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Joint Reaction Force and Contributions of Surrounding Muscles to Knee Joint Load during Stair Ascent in Total Knee Replacement Patients and Healthy Individuals

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To the Graduate Council:

I am submitting herewith a thesis written by Robert Jacob Rasnick entitled "Joint Reaction Force and Contributions of Surrounding Muscles to Knee Joint Load during Stair Ascent in Total Knee Replacement Patients and Healthy Individuals." I have examined the final electronic copy of this thesis for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Master of Science, with a major in Kinesiology.

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Joint Reaction Force and Contributions of Surrounding Muscles to Knee Joint Load during Stair Ascent in Total Knee Replacement Patients and Healthy Individuals

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ABSTRACT

Total knee replacement (TKR) is commonly used to correct end stage knee osteoarthritis (OA) of the knee joint. Unfortunately, difficulty with stair climbing has been seen to exist, prolonging the challenges of TKR patents. Complete understanding of loading at the knee is of great interest in order to aid patient populations, implant manufacturers, rehabilitation, and future research. The outcome of a TKR is intended to reestablish normal motion and loading of the knee. Musculoskeletal modeling provides a means to accurately approximate joint loading and the corresponding muscle contributions during a movement.

The purpose of the present study was to examine if the knee joint loading are recovered to the level of healthy individuals following a TKR, and determine the contribution of the muscles to knee joint loading. Data from five healthy and five TKR patients were selected for musculoskeletal simulation using Opensim. Variables of interest included knee joint reaction forces and the corresponding muscle forces. A paired samples t-test was used to detect difference between groups for each variable of interest with an alpha level set at 0.05 a priori.

TKR patients showed a trend of having higher 2nd peak compressive JRF than healthy individuals. Some muscle compensatory strategies appear to be present in the push-off phase; however the differences in muscles do not clearly explain the trend present in compressive JRF during the 2nd 50% of stance. Evidence from knee extension moment and muscle force contributions during the loading response phase indicates reduced muscle strength in the knee extensors of TKR patients. This result combined

with greater flexor muscle force resulted in similar compressive JRF during loading response between groups.

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CHAPTER I
BACKGROUND

Background

Total knee replacement (TKR) is commonly used to correct end stage knee osteoarthritis within the knee joint. The frequency with which TKRs are performed is expected to double in the US alone by 2015, reaching nearly 3.5 million by 2030 (Kurtz et al., 2007). The primary purposes of a TKR are to alleviate pain, restore normal range of motion (ROM), and restore the ability to perform activities of daily living. Several studies have reported reductions in the pain after a TKR (Bruyere et al., 2012; Hawker et al., 1998; Ko et al., 2011), however another study found disappointment of patients due to post-surgery pain and difficulties with stair negotiation (Dickstein et al., 1998). Difficulty with stair climbing prolongs challenges of TKR patents which a TKR is intended to correct. Stair climbing is a common activity of daily living, with older adults utilizing stairs as frequently as younger adults (Startzell et al., 2000). Because of the regularity of its use, as well as the potential for continued difficulty and pain post TKR surgery, a holistic understanding of what occurs biomechanically during stair climbing is essential.

Traditional biomechanics are utilized to provide an understanding of how well a TKR actually restores a healthy gait in end-stage osteoarthritis (OA) patients. There have been consistent findings that TKR patients have greater amounts of knee flexion at foot contact than healthy controls during stair negotiation (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Mandeville et al., 2007). Peak knee flexion has been shown to have mixed results in several studies (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Joglekar et al., 2012; Ouellet and Moffet, 2002; Saari et al., 2004). Some have found no differences between healthy and TRK patients in peak knee flexion (Berti et al., 2006; Joglekar et al., 2012), while others have shown TKR patients to have increased knee flexion during stair climbing (Catani et al., 2003; Fantozzi et al., 2003; Ouellet and Moffet, 2002; Saari et al., 2004). In the case of sagittal plane knee ROM, some have reported reductions in TKR patients compared to their healthy counterparts in climbing stairs (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003) while others reported that no differences existed between groups(Berti et al., 2006; Kelman et al., 1989; Wilson et al., 1996).

Mixed results were also seen in the kinetics in stair ascent. Saari et al. (2004) found no differences between healthy subjects and TKR patients in peak internal extension moment, while Mandeville et al. (2007) found reductions. Frontal plane variables are often associated biomechanically with the onset and progression of knee OA and are frequently measured in studies to determine the effectiveness of a TKR in restoring healthy frontal plane motion and loading. Conflicting results were also found among studies in frontal plane moments (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Mandeville et al., 2008; Saari et al., 2004). Two studies found a reduction in peak knee external adduction moment during stair climbing when using the mobile bearing design compared to healthy controls but no differences with the posterior stabilizing design (Catani et al., 2003; Fantozzi et al., 2003). Conversely, Berti et al. (2006) found the non-resurfaced patella with the posterior stabilized design to result in increases of the peak external adduction moment while patellar resurfacing TKR produced no differences compared to healthy counterparts. Saari et al. (2004) found no differences in peak internal abduction moment. Mandeville et al. (2008) found decreases in the peak internal adduction moment when utilizing a posterior stabilized design compared to healthy individuals. The effects of a TKR remain unclear due to the disagreement in the majority of findings provided via kinematics and kinetics. The only aspects that can be agreed upon with some confidence are that the knee contact angle and velocity during stair ascent are reduced in TKR patients compared to healthy individuals. As a result further research and different methods of research are needed to better understand the level to which a TKR restores healthy biomechanics.

Also, the information provided by joint moments and joint reaction forces via inverse dynamics does not provide a true bone on bone loading at the knee joint. Additional loading at the joint results from the contraction of muscles not just the GRF propagated up through the body. Several studies have provided a good understanding of these effects by utilizing instrumented TKR implants while climbing stairs (Catani et al., 2009; D'Lima et al., 2007; D'Lima et al., 2005; D'Lima et al., 2006; Heinlein et al., 2009; Kim et al., 2009; Kutzner et al., 2010; Mundermann et al., 2008). Instrumented implants offer a more detailed representation of joint loading within the knee joint during stair climbing. While instrumented implants do provide additional information to the understanding of the kinetic requirements of stair ascent, there are limitations to consider such as the limited subject populations, inability to examine the muscle contributions to loading, and high costs.

Musculoskeletal modeling provides a means to accurately approximate joint loading associated with a movement in a cost effective manner. Musculoskeletal modeling utilizes data collected using a motion capture system before performing the simulation. There have been several studies which the researchers chose to develop their own mathematical program to run simulations of movement data (Anderson and Pandy, 2003; Ghafari et al., 2009; Kim et al., 2009; Liu et al., 2006; Piazza and Delp, 2001). However, software programs have also been developed to ease the process of simulation and allow simulation based research to be performed more freely. SIMM, LifeMod, AnyBody, and OpenSim are four of the software programs commonly used in research.

Musculoskeletal modeling has been utilized in applications for over-ground walking, running, crouch gait, and stair negotiation in both healthy and patient subject groups. For over-ground walking common analyses investigate the joint loading (Lerner et al., 2013; Richards and Higginson, 2010; Sasaki and Neptune, 2010), individual muscle contributions (Anderson and Pandy, 2003; Kim et al., 2009; Li et al., 2013; Richards and Higginson, 2010; Sasaki and Neptune, 2010), net muscle moments (Lerner et al., 2013; Li et al., 2013), effects of speed on muscle contribution and joint loading (Kim et al., 2009; Lerner et al., 2013; Liu et al., 2008), and the consistency of simulation muscle activations with electromyography (EMG) (Kim et al., 2009; Lerner et al., 2013; Richards and Higginson, 2010; Sasaki and Neptune, 2010).

Only a limited number of studies have utilized musculoskeletal simulations to investigate stair negotiation (Ghafari et al., 2009; Piazza and Delp, 2001; Rouston, 2010; Taylor et al., 2004). Piazza and Delp (2001) investigated a single step-up in a healthy subject. Flexion angles were shown to correlate well between the simulation and experimentally collected data. Simulated anterior/posterior knee forces during the step-up task matched well with experimental data. However, axial JRFs were nearly half of those measured experimentally in a single step-up (Piazza and Delp, 2001). Taylor et al. (2004) compared over-ground walking to a single step-up in total hip arthoplasty patients. Peak knee compressive force was found to range between 4.9 and 5.6 BWs. This was notably larger than the average peak of 2.0 to 4.45 BWs typically found during over-ground walking (Lerner et al., 2013; Richards and Higginson, 2010; Sasaki and Neptune, 2010). Peak anterior-posterior shear forces were found to range between 1.1 and 1.5 BW during a step-up, while only 0.5 while walking. Additionally, shear loading was shown to be largest when the knee was in greater than 15° of flexion (Taylor et al., 2004).

Complete understanding of loading at the knee is of great interest in order to aid patient populations, implant manufacturers, rehabilitation, future research, and the advancement of the medical community. The outcome of a TKR is intended to reestablish normal motion and loading of TKR patients. Unfortunately this is not the case. Musculoskeletal simulation can provide an excellent means of determining what true knee loading is in subject specific models. The increase understanding provided by these analyses can aid in improving TKR design and restoring TKR patients to normal loading and movement patterns.

Statement of the Problems

To date no studies have investigated the joint reaction loading and the contributions of muscle forces in TKR patients during stair ascent via musculoskeletal modeling. All simulation studies of stair ascent have also only utilized one step in their analysis failing to reflect the actual movement while climbing stairs. It is not clear if the contributions of knee joint related muscles to the knee joint loading in stair ascent for TKR patients. Additionally, traditionally biomechanics has failed to provide a strong consensus in the literature as to the effects of a TKR in end-stage OA patients. Therefore, the purpose of the present study was to examine if the knee joint loading and kinematics are recovered to the level of healthy individuals following a TKR, and determine the contribution of the muscles to knee joint load.

Research Hypotheses

It was hypothesized that knee joint compressive and shear loading would be different between TKR patients and their healthy counterparts. In addition, contributions of knee muscle forces would be different between groups.

Delimitations

Healthy Adults

Exclusion criteria included:

- Knee pain for at least 6 months during daily activities.
- Diagnosed with any type of lower extremity joint osteoarthritis (self-reported).
- Any lower extremity joint replacement.
- Any lower extremity joint arthroscopic surgery or intra-articular injection within past 3 months.
- Systemic inflammatory arthritis (rheumatoid arthritis, psoriatic arthritis) (self-reported).
- BMI greater than 35.
- Inability to ascend/descend stairs without the use of a handrail.
- Inability to walk without a walking aid.
- Neurologic disease (e.g. Parkinson's Disease, stroke patients) (self-reported).
- Any major lower extremity injuries/surgeries.
- Any visual conditions affecting gait or balance.
- Women who are pregnant or nursing.
- Any cardiovascular disease or primary risk factor which precludes participation in aerobic exercise as indicated by the Physical Activity Readiness Survey. If any participant marks "yes" on the survey they will be required to obtain written consent from their doctor indicating they are healthy enough to participate in the study.

Inclusion criteria included:

- Men and women between the ages of 35 and 80

TKR Patients

Exclusion criteria included:

- Any additional lower extremity joint replacement.
- Any lower extremity joint arthroscopic surgery or intra-articular injection within the past month.
- Systemic inflammatory arthritis (rheumatoid arthritis, psoriatic arthritis) (self-reported).
- BMI greater than 35.
- Inability to ascend/descend stairs without the use of a handrail.
- Neurologic disease (e.g. Parkinson's Disease, stroke patients) (self-reported).
- Any major lower extremity injuries/surgeries.
- Inability to walk without a walking aid.
- Any visual conditions affecting gait or balance.
- Women who are pregnant or nursing.
- Any cardiovascular disease or primary risk factor which precludes participation in aerobic exercise as indicated by the Physical Activity Readiness Survey. If any participant marks "yes" on the survey they will be required to obtain written consent from their doctor indicating they are healthy enough to participate in the study.

Inclusion criteria included:

- Men and women between the ages of 35 and 80.
- Total knee replacement in one knee.
- At least 6-months from TKR.
- No more than 5-years from TKR

Limitations

- All tests were conducted in a laboratory setting.
- Skin marker placement in obese participants may not reflect accurate bony landmark location.
- Reflective markers used to track the feet during motion trials were placed on the shoe. Thus, foot motions within the shoe may not have been accurately captured.
- It was assumed that the healthy older adults do not have radiographic knee osteoarthritis.

CHAPTER II
LITERATURE REVIEW

Introduction

The purpose of the present study was to examine if the knee joint loading and kinematics are recovered to the level of healthy individuals following a TKR, and determine the contribution of the muscles to knee joint load. This chapter emphasizes the review of previous literature on the prevalence and purpose of a total knee replacement (TKR) and knee kinematic and kinetic variables associated with 1) traditional kinematics and kinetics and 2) musculoskeletal modeling of stair ascent. In addition, variables associate with the musculoskeletal modeling of over-ground walking are included to introduce the expected differences between traditional and simulation based techniques.

Prevalence and Purpose of a Total Knee Replacement

A TKR is commonly used to correct end stage knee osteoarthritis. The frequency with which TKRs are performed is expected to double in the US alone by 2015, reaching nearly a total of 3.5 million by 2030 (Kurtz et al., 2007). The primary purposes of a TKR are to alleviate pain, restore normal ROM, and restore the ability to perform activities of daily living. Several studies have reported reductions in the pain after a TKR (Bruyere et al., 2012; Hawker et al., 1998; Ko et al., 2011), however another study (Dickstein et al., 1998) found disappointment of patients due to post-surgery pain and difficulties with stair negotiation. Difficulty with stair climbing prolongs challenges of TKR patients which a TKR is intended to correct. Stair climbing is a common activity of daily living with older adults utilizing stairs as frequently as younger adults (Startzell et al., 2000). Because of the regularity of its use, as well as the potential for continued difficulty and pain post TKR surgery, a holistic understanding of what occurs biomechanically during stair climbing is essential.

A complete understanding of the biomechanics of stair climbing consists of not only the TKR patients but also a baseline of healthy individuals. The reasoning for this is to provide a means to determine if the TKR surgery and recovery were successful. A successful TKR surgery would alleviate pain, restore normal ROM, and restore the ability to perform activities of daily living. Ideally any discrepancies existing between the TKR patients and healthy individuals would be identified. Design corrections could then be made to further minimize the differences between groups. Several studies have attempted to provide this insight using analysis via gait biomechanics (Andriacchi et al., 1980; Catani et al., 2003; Costigan et al., 2002; Mandeville et al., 2007; Mandeville et al., 2008; Ouellet and Moffet, 2002; Saari et al., 2004; Stacoff et al., 2007; Yu et al., 1997).

Kinematics and Kinetics of Stair Ascent

Kinematic data are performed using a motion capture system often with infrared cameras while kinetic data are collected using force platforms. Analysis of kinetic data is then done using an inverse dynamics approach. In the case of stair climbing, collection of kinetics becomes more complicated. Some studies elected to use only a few steps, often two or fewer (Andriacchi et al., 1980; Costigan et al., 2002; Mandeville et al., 2007; Mandeville et al., 2008; Saari et al., 2004), and often only collect kinetics for the first step (Andriacchi et al., 1980; Costigan et al., 2002; Mandeville et al., 2007; Mandeville et al., 2008; Ouellet and Moffet, 2002; Saari et al., 2004). Yu et al. (1997) showed that consideration should be made for the differences between the first step of stair ascent as compared to the remainder of the steps. In their study, healthy subjects were asked to ascend and descend a series of four steps. Kinetics was collected for the first and second step of the staircase and kinematics for all steps. Joint angles and moments of each step were correlated to determine which steps had exhibited the largest variability. It was found that the second step demonstrated the greater reproducibility than that of the first step (Yu et al., 1997). The first step is a transition from level walking to stair climbing, while on the remaining steps the body must be raised from the step below to the step above. These are clearly two separate tasks each deserving their own investigation. Yu et al. (1997) suggested that angle and moment variability could be reduced even further by performing analyses on steps after the second. Two studies collected kinematics and kinetics for a greater number of steps to allow for a more accurate replication of what movements and loadings are required during stair climbing (Catani et al., 2003; Stacoff et al., 2007).

Reduction in the variability of the measurement of kinematics and kinetics allows for greater accuracy identifying differences between two groups during stair climbing. In the case of TKR patients, variables of interest have been shown to be: knee flexion angles(Catani et al., 2003; Mandeville et al., 2007; Mandeville et al., 2008; Ouellet and Moffet, 2002; Saari et al., 2004), ROM (Catani et al., 2003; Saari et al., 2004), ground reaction forces (GRFs) (Stacoff et al., 2007), and moments (Catani et al., 2003; Mandeville et al., 2007; Mandeville et al., 2008; Ouellet and Moffet, 2002; Saari et al., 2004) primarily in the sagittal and frontal planes as well as the speed of stair climbing (Catani et al., 2003; Mandeville et al., 2007; Ouellet and Moffet, 2002). Differences have been found in knee angle at contact (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Joglekar et al., 2012; Mandeville et al., 2007; Ouellet and Moffet, 2002; Saari et al., 2004), knee ROM (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Kelman et al., 1989; Wilson et al., 1996), knee extension moment (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Joglekar et al., 2012; Mandeville et al., 2007; Saari et al., 2004;

Wilson et al., 1996), and self-selected speeds (Berti et al., 2006; Catani et al., 2009; Fantozzi et al., 2003; Mandeville et al., 2007) of TKR patients when compared to their healthy counterparts.

Kinematics

There have been consistent findings that TKR patients have greater amounts of knee flexion at foot contact than healthy controls during stair negotiation (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Mandeville et al., 2007). Differences in knee flexion angle at foot contact were shown to range from 8.7° to 22.8°. Peak knee flexion has been shown to have mixed results in several studies. Some have found no differences between healthy and TRK patients in peak knee flexion (Berti et al., 2006; Joglekar et al., 2012), while others have shown TKR patients to have increased knee flexion during stair climbing (Catani et al., 2003; Fantozzi et al., 2003; Ouellet and Moffet, 2002; Saari et al., 2004). Direct comparison of these studies is not appropriate as some chose to investigate peak knee flexion during stance (Joglekar et al., 2012; Ouellet and Moffet, 2002) and others during swing phase (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Joglekar et al., 2012; Saari et al., 2004). The stance phase is often of more interest because of the difficulty often associated with weight bearing in TKR patients. Peak knee flexion angle during stance phase of stair ascent was reported by two studies (Joglekar et al., 2012; Ouellet and Moffet, 2002). Of the two studies, only Ouellet et al. (2002) found reductions in peak knee flexion angle during stance of stair climbing. Joglekar et al. (2012) found no differences in peak knee flexion angle between the cruciate retaining and the posterior stabilizing designs.

In the case of sagittal plane knee ROM, some have reported differences to exist between TKR patients and their healthy counterparts in climbing stairs (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003) while others reported that no differences existed between groups(Berti et al., 2006; Kelman et al., 1989; Wilson et al., 1996). These studies have shown implant design to influence the ROM while ascending stairs. The posterior stabilizing TKR design resulted in a decreased sagittal ROM by 5° to 11° when compared to healthy individuals. The same studies also found that the mobile bearing design TKR reduced flexion ROM (Catani et al., 2003; Fantozzi et al., 2003). Another study found reductions as large as 16.6° for TKR designs with non-resurfaced patella (Berti et al., 2006). This same study found no differences in the knee flexion ROM when the patella was resurfaced (Berti et al., 2006). It was also found no differences in flexion ROM with the use of the posterior stabilizing TKR design compared to age-matched healthy individuals (Wilson et al., 1996).

Knee angle at contact, peak knee angle, and knee ROM are very interrelated and should be analyzed together in order to better understand what true effects exist. The effects of a TKR remain unclear due to the disagreement in the majority of findings. The only aspect that can be agreed upon with some confidence is that the knee contact angle during stair ascent is reduced in TKR patients compared to healthy individuals.

Kinetics

GRFs were recorded in ten studies, however were not directly reported in any of them (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Joglekar et al., 2012; Kelman et al., 1989; Mandeville et al., 2007; Mandeville et al., 2008; Ouellet and Moffet, 2002; Saari et al., 2004; Wilson et al., 1996). Seven of these studies collected GRF data for the first step only(Fantozzi et al., 2003; Joglekar et al., 2012; Mandeville et al., 2007; Mandeville et al., 2008; Ouellet and Moffet, 2002; Saari et al., 2004; Wilson et al., 1996), one study collected kinetics on the first and second steps (Kelman et al., 1989), and two studies collected data on the second and third steps (Berti et al., 2006; Catani et al., 2003). As previously mentioned, interpretation of results should take into consideration the effect different step's kinetics have on the overall analysis.

These studies instead elected to examine differences of joint moments between TKR patients and healthy controls. Peak sagittal moment was reported by eight studies (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Joglekar et al., 2012; Mandeville et al., 2007; Ouellet and Moffet, 2002; Saari et al., 2004; Wilson et al., 1996). Only three of the studies reported those moments as internal moments (Joglekar et al., 2012; Mandeville et al., 2007; Saari et al., 2004) while four of the studies reported external moments (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Wilson et al., 1996). One of the studies did not report whether the moment was internal or external (Ouellet and Moffet, 2002). Peak external flexion moment was found to be reduced in TKR patients compared to healthy controls (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Mandeville et al., 2007). However, Wilson et al. (1996) found no differences in peak external flexion moment between TKR subjects and controls. Mixed results were also seen in the studies reporting peak internal extension moments. Saari et al. (2004) found no differences between healthy subjects and TKR patients in peak internal extension moment. Contrasting these findings, Mandeville et al. (2007) found that the peak internal extension moment was reduced in TKR subjects compared to healthy individuals. Joglekar et al. (2012) only made comparisons between the cruciate retaining and posterior stabilizing TKR designs and found no differences between the two.

Peak frontal plane moments were commonly reported in studies analyzing the effects of a TKR (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Mandeville et al., 2008; Saari et al., 2004). Frontal plane variables are often associated biomechanically with the onset and progression of knee osteoarthritis and are frequently measured in studies to determine the effectiveness of a TKR in restoring healthy frontal plane motion and loading. As seen in the reporting of sagittal plane variables, studies often elected to report peak external (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003) or internal (Mandeville et al., 2008; Saari et al., 2004) frontal plane moments and so considerations need to be made when making comparisons.

Conflicting results were found among studies in frontal plane moments. Two studies found a reduction in peak external adduction moment during stair climbing when using the mobile bearing design compared to healthy controls but no differences with the posterior stabilizing design (Catani et al., 2003; Fantozzi et al., 2003). Conversely, Berti et al. (2006) found the non-resurfaced patella with the posterior stabilized design to result in increases to the peak external adduction moment while patellar resurfacing produced no differences. Saari et al. (2004) found no differences in peak internal abduction moment. Mandeville et al. (2008) found decreases in the peak internal adduction moment when utilizing a posterior stabilized design.

Velocity of stair ascent was commonly reported in the literature (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Mandeville et al., 2007). Velocity was found to be reduced in TKR patients compared to healthy individuals when climbing stairs in all four studies. Three studies reported velocities ranging from 0.275 to 0.37 m/s for TKR patients and ranges of 0.39 to 0.439 m/s for healthy individuals (Berti et al., 2006; Catani et al., 2009; Fantozzi et al., 2003). However, the velocities reported in the study by Mandeville et al. (2007) were significantly larger with TKR patients ascending stairs at 0.52 m/s and healthy subjects ranging between 0.66 and 0.71 m/s.

Conflicting results are seen with sagittal and frontal plane knee moments during stair negotiation. Both reductions and increases were seen to result in TKR patients compared to healthy individuals. This level of disagreement within the literature leaves the true effects of a

TKR on sagittal and frontal plane knee moments in question. Consistency was only seen in the velocity of stair ascent. The velocity of TKR patients was seen to be reduced as compared to healthy subjects in majority of studies. (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Mandeville et al., 2007).

Kinetics via Instrumented Implants

Unfortunately, the current literature on kinetics of stair ascent falls short of presenting a complete picture of loading conditions of knee joint in stair climbing. Additional loading at the joints results from the activation of muscles surrounding a joint not just force propagated up through the body from the GRF. Because of this the true bone on bone contact forces are significantly underestimated. Some studies have sought to provide a greater understanding of muscle's effects by utilizing instrumented TKR implants (Catani et al., 2009; D'Lima et al., 2007; D'Lima et al., 2005; D'Lima et al., 2006; Heinlein et al., 2009; Kim et al., 2009; Kutzner et al., 2010; Mundermann et al., 2008). Instrumented implants provide a more realistic representation of joint loading within the knee joint during stair climbing. The variables commonly reported by studies utilizing instrumented TKR implants are joint reaction forces and moments.

Six studies report on the effects of a TKR on joint loading during stair ascent (D'Lima et al., 2007; D'Lima et al., 2005; D'Lima et al., 2006; Heinlein et al., 2009; Kutzner et al., 2010; Mundermann et al., 2008). Compressive loads at the knee during stair ascent were found to range between 2.5 and 3.16 times bodyweight (BW) with the majority reporting approximately 3 BW (D'Lima et al., 2007; D'Lima et al., 2005; D'Lima et al., 2006; Heinlein et al., 2009; Kutzner et al., 2010). Only three studies reported findings on the shear forces occurring at the knee during stair climbing (D'Lima et al., 2007; Heinlein et al., 2009; Kutzner et al., 2010). D'lima et al. (2007) found the anterior shear forces to be as high as 0.26 BW while Kutzner et al. (2010) reported peak posterior shear as 0.32 BW when climbing stairs. Heinlein et al. (2009) reported a peak anterior and posterior shears of 0.30 BW and 0.23 BW respectively for one subject. The second subject had a peak anterior shear of 0.12 BW and a peak posterior shear of 0.32 BW. These large differences show that there is a high variability in the shear forces between subjects.

Three studies reported findings on joint moments using instrumented implants (Heinlein et al., 2009; Kutzner et al., 2010; Mundermann et al., 2008). Peak extension moments during stair ascent were found to range between 1.7 and 2.4 % BW·m. Heinlein et al. (2009) found the peak flexion moment to be 0.2 % BW·m for one subject while the other subject in the study had an extension moment through the entire movement. The peak internal adduction moment was found to range between 0.1 and 1.26 % BW·m, and the peak internal abduction moment ranged from 2.2 to 4.2 % BW·m. The range for peak external rotation moment was found to be 0.5 to 0.92 % BW·m (Heinlein et al., 2009; Kutzner et al., 2010). Heinlein et al. (2009) reported peak internal rotation moments of 0.0 and 0.3 % BW·m for their two subjects. The study performed by Kutzner et al. (2010) only mentioned that peak internal rotation moments were small for stair climbing.

While instrumented implants do provide additional information to the understanding of the joint loading of stair ascent, there are limitations to consider. Because of the substantial cost of the instrumented implant only a few subjects are typically utilized. Four studies only used one subject (D'Lima et al., 2007; D'Lima et al., 2005; D'Lima et al., 2006; Mundermann et al., 2008) while one study used two (Heinlein et al., 2009). Kutzner et al. (2010) used the largest number of subjects at five. This limit on the number of available subjects hinders the ability of research to identify differences common to all TKR patients not just the individual subjects. Instrumented implants and inverse dynamics are limited in that they do not account for muscle contributions to joint loading. While compressive loading was consistent among instrumented implant studies, shear loading and moments were found to be variable. An alternative method is needed that allows for the increased level of detail that the instrumented implant provides while not being limited by subject population, cost, or the ability to account for muscle contributions.

Overview of Musculoskeletal Modeling

Musculoskeletal modeling provides a means to accurately approximate kinetics and kinematics associated with a movement in a cost effective manner. Musculoskeletal modeling utilizes data collected using a motion capture system before performing the simulation. There have been several studies which the researchers chose to develop their own mathematical program to run simulations of movement data (Anderson and Pandy, 2003; Ghafari et al., 2009; Kim et al., 2009; Liu et al., 2006; Piazza and Delp, 2001). However, software programs have also been developed to ease the process of simulation and allow simulation based research to be performed more freely. SIMM, LifeMod, AnyBody, and OpenSim are four of the software programs commonly used in research.

Delp et al. (2007) developed OpenSim as an open-source software allowing researchers the ability to develop their own models and analyses to use within the program. This functionality allows for additional precision based on the needs of a given research study such as the calculation of a joint reaction force. The goal of the musculoskeletal simulation is to take experimental movement data to drive the entire motion using muscle-tendon actuators. Producing a muscle driven simulation requires four steps: creation of a dynamic musculoskeletal model, solving of an inverse kinematics problem, a residual reduction algorithm (RRA), and computed muscle control (CMC) (Delp et al., 2007).

The creation of a dynamic musculoskeletal model begins with the scaling of a general generic model to the proportions of the specific subject to which the experimental data corresponds. The general model consists of the skeletal structure and the applied muscle-tendons with standard physiological properties. Scaling modifies the model to match the anthropometric and physiological features (e.g. segment mass, muscle fiber length, tendon slack length) of the model to the subject. The movement from the experimental data is then applied to the scaled model to create a completed dynamic musculoskeletal model (Delp et al., 2007).

Next, inverse kinematics is computed on the experimental data to identify the joint angles and translations that correlate most accurately with the collected data. The foundation for this inverse kinematics problem is to minimize the differences between the experimental movement and the models reproduction. Those differences can be expressed as a least squared problem in which the weighted squared error is minimized:

$$\text{Squared Error} = \sum_{i=1}^{\text{markers}} w_i(\vec{x}_i^{\text{subject}} - \vec{x}_i^{\text{model}})^2 + \sum_{j=1}^{\text{joint angles}} w_j(\theta_j^{\text{subject}} - \theta_j^{\text{model}})^2$$

$\vec{x}_i^{subject}$ and $\vec{x}_i^{model}$ correspond to the ith marker or joint center position in three-dimensions for the subject and model, respectively. $\theta_j^{subject}$ and θ_j^{model} corresponds to the jth joint angle for the subject and model. w_i and $w_j$are weighting factors to allow marker and joint angle data to be weighted separately (Delp et al., 2007).

The third step consists of performing a RRA. This process alters the joint angles and translations as computed by the inverse kinematics to increase their consistency with the GRFs and moments. Because of experimental errors and modeling assumptions, kinematics and kinetics are not dynamically consistent. Residual forces are applied to a model in order to account for these discrepancies between the kinematics and kinetics. Therefore as the name implies, RRA is intended to minimize the amount of residual forces added to the model by making small adjustments to the motion trajectory and mass parameters. Residual forces go to zero as the amount of experimental and modeling error also approach zero. The level to which changes are made can be controlled by creating limits for the acceptable magnitude of the residuals. After each RRA, new kinematics is outputted that consistently relates to the kinetics. The results of the RRA are then evaluated to ensure changes to kinematics do not drastically alter the movement. Table 1 below shows threshold values recommended by OpenSim for the evaluation of RRA results for over-ground walking. Table 1 provides ranges for maximum and root mean square (RMS) values for residual forces and moments and translational and rotational errors (pErr) (Delp et al., 2007). Values for stair climbing are expected to be higher due to the increased difficulty of stair climbing therefore Table 1 will be utilized as a guideline for interpretation of results.

Table 1. Recommended Threshold Values for Evaluation of RRA Results (Delp et al., 2007).

Thresholds:	GOOD	OKAY	BAD
MAX Residual Force (N)	0-10 N	10-25N	> 25 N
RMS Residual Force (N)	0-5 N	5-10 N	> 10 N
MAX Residual Moment (Nm)	0-50 Nm	50-75 Nm	>75 Nm
RMS Residual Moment (Nm)	0-30 Nm	30-50 Nm	>50 Nm
MAX pErr (trans, cm)	0-2 cm	2-5 cm	>5 cm
RMS pErr (trans, cm)	0-2 cm	2-4 cm	>4 cm
MAX pErr (rot, deg)	0-2 deg	2-5 deg	> 5 deg
RMS pErr (rot, deg)	0-2 deg	2-5 deg	> 5 deg

The following equation describes the relationship of residual forces with the experimental forces and segment accelerations:

$$\vec{F}_{external} = \sum_{i=1}^{segments} m_i \vec{a}_i - \vec{F}_{residual}$$

$\vec{F}_{external}$ corresponds to the GRF measured without the body weight vector. $\vec{a}_i$ is the center of mass translational acceleration for the ith segment and m_i is the mass of that segment. $\vec{F}_{residual}$ is the residual force that has been applied to initially account for the dynamic inconsistencies (Delp et al., 2007).

Lastly, muscle forces and excitations calculated via CMC drive a simulation of the movement originally collected experimentally. The combination of a static optimization and proportional-derivative control allow for the forward dynamics simulation to nearly mirror the movements produced by the RRA. Activation and contraction dynamics also play a role in determining the output simulation motion. Activation dynamics takes into account magnitude of muscle excitation and activation as well as the rate of change in muscle activation. Force-length-velocity of the muscles and elastic properties of the tendons are accounted for by the contraction dynamics aspect of the simulation (Delp et al., 2007). This simulation technique allows for simulations to be performed much faster than traditional techniques with a model containing 23 degrees of freedom and 92 muscles requiring approximately 10 minutes of computation time depending on the task and duration of movement (Delp et al., 2007).

Analysis tools exist within OpenSim for additional calculations to be performed after completion of the simulation. Joint reaction forces and muscle-induced accelerations are among the several available analyses. The joint reaction forces analysis allows for the calculation of any joint's three dimensional loading over the entire movement duration. Muscle-induced acceleration analysis allows for quantifying of the effect of moments generated by muscle contractions which result in acceleration at a joint not crossed by the activated muscle (Challis, 2011; Hamner et al., 2013).

Musculoskeletal modeling has been shown to be an excellent answer to the problems associated with the accuracy of in vivo loading measurements without the aid of instrumented implants. Simulations can be performed on a subject specific basis, analyzed for several subjects at a fraction of the costs associated with instrumented implants, and calculate muscle forces contributing to the overall joint loading. Further understanding of all that musculoskeletal modeling makes available to researchers is needed before applications can be made.

Musculoskeletal Modeling in Over-ground Walking

Musculoskeletal modeling has been utilized in applications for over-ground walking, running, crouch gait, and stair negotiation in both healthy and patient subject groups. For over-ground walking common analyses investigate the joint loading (Lerner et al., 2013; Richards and Higginson, 2010; Sasaki and Neptune, 2010), individual muscle contributions (Anderson and Pandy, 2003; Kim et al., 2009; Li et al., 2013; Richards and Higginson, 2010; Sasaki and Neptune, 2010), net muscle moments (Lerner et al., 2013; Li et al., 2013), effects of speed on muscle contribution and joint loading (Kim et al., 2009; Lerner et al., 2013; Liu et al., 2008), and the consistency of simulated muscle activations with electromyography (EMG) (Kim et al., 2009; Lerner et al., 2013; Richards and Higginson, 2010; Sasaki and Neptune, 2010).

Compressive or axial joint reaction forces during over-ground walking have been shown to range between 2.00 and 4.40 BW in healthy individuals (Lerner et al., 2013; Sasaki and Neptune, 2010) and as high at 4.45 in TKR patients (Richards and Higginson, 2010). One study analyzed joint reaction forces in the anterior-posterior direction. Peak anterior-posterior loading in healthy subjects was found to range between 0.16 BWs posteriorly and 0.40 BWs anteriorly

(Lerner et al., 2013). Kim et al. (2009) made comparisons between musculoskeletal simulation and instrumented implant data finding good agreement between the two. Average root mean square error of the total knee contact force between simulation and the instrumented implant was found to be 11%.

Understanding individual muscle contributions during over-ground walking allows for identification of the key muscles and any deficiencies in patient populations. The vasti and gastrocnemius muscles were shown to generate the majority of knee loading (Kim et al., 2009; Sasaki and Neptune, 2010). Two studies investigated differences between healthy and patient populations (Li et al., 2013; Richards and Higginson, 2010). Richards et al. (2010) found an increase in co-contraction between the hamstring and quadriceps muscles with increasing osteoarthritis severity. Li et al. (2013) reported reductions in force output by the vasti muscles during early stance in TKR patients.

Two studies investigated net muscle moments at the knee during over-ground walking (Lerner et al., 2013; Li et al., 2013). Li et al. (2013) found reduced net knee extensor moments in TKR patients during early stance. Another study showed that 75% of the variance in compressive knee joint loading could be explained by net muscle moments (Lerner et al., 2013).

Speed has been shown to have a great influence over the outcome of musculoskeletal simulation (Kim et al., 2009; Lerner et al., 2013; Liu et al., 2008). Two studies showed that the contributed muscle forces increase with speed (Lerner et al., 2013; Liu et al., 2008). In turn, tibiofemoral loading will also increase with speed. Lerner et al. (2013) found that compressive loading rates at the knee increase 300% with increasing speed from 0.75 to 1.5 m/s. However, agreement between simulation and data from an instrumented implant was shown to decrease with increasing speed (Kim et al., 2009). Other studies showed the output muscle activation from musculoskeletal simulations to be consistent with EMG collected for the same movements (Kim et al., 2009; Lerner et al., 2013; Richards and Higginson, 2010; Sasaki and Neptune, 2010).

While musculoskeletal modeling does show variability in the compressive loads of up to 2 BWs in healthy individuals during over-ground walking, speed was shown to play a major role in the outcome with muscle contribution, loading, and loading rate all increasing with increasing speed. Special attention is needed for the speed of subjects when attempting to make comparisons between subject groups and TKR designs.

Musculoskeletal Modeling in Stair Negotiation

Only four studies analyze musculoskeletal simulations of stair negotiation (Ghafari et al., 2009; Piazza and Delp, 2001; Rouston, 2010; Taylor et al., 2004). Three of those studies utilize musculoskeletal simulation to analyze stair ascent in healthy individuals (Ghafari et al., 2009; Piazza and Delp, 2001; Rouston, 2010). Piazza and Delp (2001) developed a model in order to investigate a single step-up in a healthy subjects. Flexion angles were shown to correlate well between the simulation and experimentally collected data. Two additional studies performed analyses on healthy individuals during stair ascent and provided only qualitative comparisons of simulation results. However neither study provided detailed quantitative results of their findings (Ghafari et al., 2009; Rouston, 2010). Taylor et al. (2004) compared over-ground walking to a single step-up in total hip arthroplasty patients. Peak knee compressive force was found to range between 4.9 and 5.6 BWs during stair climbing. This was notably larger than the average peak of 2.0 to 4.45 BWs typically found during over-ground walking (Lerner et al., 2013; Richards and Higginson, 2010; Sasaki and Neptune, 2010). Peak anterior-posterior shear forces were found to range between 1.1 and 1.5 BW during a step-up, while only 0.5 BW while walking.

Additionally, shear loading was shown to be largest when the knee was in greater than 15° of flexion (Taylor et al., 2004).

Stair climbing has been shown to require increased loading over that of over-ground walking. This increased loading adds to the difficulty for the TKR patient populations. To date, no studies have investigated the muscle activations and joint loading during stair ascent of TKR patients via musculoskeletal modeling. Complete understanding of loading at the knee is of great interest in order to aid patient populations, implant manufacturers, rehabilitation, future research, and the advancement of the medical community. The outcome of a TKR is intended to reestablish normal motion and loading in patients as mentioned above, but unfortunately this is not the case. Musculoskeletal simulation can provide an excellent means of determining what true knee loading is in subject specific models and the muscle forces that contribute to that loading. The increase understanding provided by these analyses can aid in improving TKR design, restoring TKR patients to normal loading and movement patterns.

CHAPTER III
METHODS

Participants

All participants were recruited for a larger study currently underway in the UTK Biomechanics/Sports Medicine Lab. Twelve healthy control subjects and seven TKR patients were recruited for biomechanical analysis of stair climbing. TKR patients were recruited through a local clinic and all surgeries were performed by the same surgeon. A minimum of 6 months and a maximum of 5 years post-surgery were required for all TRK patients. A complete list of inclusion and exclusion criteria for the healthy and TKR patients can be seen in Table 1 and 2, respectively. All participants signed an informed consent approved by the University of Tennessee Institutional Review Board. Data from four healthy and four TKR patients were selected randomly from the total participant population for analysis via musculoskeletal simulation.

Instrumentation

3-D kinematics for the trunk, pelvis, thighs, shanks, and feet of each subject was collected experimentally using a nine-camera motion analysis system (240 Hz, VICON Motion Analysis Inc., Oxford, UK). Reflective anatomical markers were place bi-laterally on the following anatomical landmarks: acromion processes, iliac crests, greater trochanters, medial and lateral femoral epicondyles, medial and lateral malleoli, 1st and 5th metatarsal heads, and toes (i.e. the most anterior aspect of the shoes). Reflective tracking markers in sets of four were connected to semi-rigid thermoplastic shells and secured to the trunk, pelvis, thighs, and shanks.

Table 2. Inclusion and Exclusion Criteria for the Healthy Subjects.

Exclusion Criteria:	Inclusion Criteria:
- Knee pain for at least 6 months during daily activities. - Diagnosed with any type of lower extremity joint osteoarthritis (self-reported). - Any lower extremity joint replacement. - Any lower extremity joint arthroscopic surgery or intra-articular injection within past 3 months. - Systemic inflammatory arthritis (rheumatoid arthritis, psoriatic arthritis) (self-reported). - BMI greater than 35. - Inability to ascend/descend stairs without the use of a handrail. - Inability to walk without a walking aid. - Neurologic disease (e.g. Parkinson's Disease, stroke patients) (self-reported). - Any major lower extremity injuries/surgeries. - Any visual conditions affecting gait or balance. - Women who are pregnant or nursing. - Any cardiovascular disease or primary risk factor which precludes participation in aerobic exercise as indicated by the Physical Activity Readiness Survey.	- Men and women between the ages of 35 and 80.

Table 3. Inclusion and Exclusion Criteria for the TKR Subjects.

Exclusion Criteria:	Inclusion Criteria:
- Any additional lower extremity joint replacement. - Any lower extremity joint arthroscopic surgery or intra-articular injection within the past month. - Systemic inflammatory arthritis (rheumatoid arthritis, psoriatic arthritis) (self-reported). - BMI greater than 35. - Inability to ascend/descend stairs without the use of a handrail. - Neurologic disease (e.g. Parkinson's Disease, stroke patients) (self-reported). - Any major lower extremity injuries/surgeries. - Inability to walk without a walking aid. - Any visual conditions affecting gait or balance. - Women who are pregnant or nursing. - Any cardiovascular disease or primary risk factor which precludes participation in aerobic exercise as indicated by the Physical Activity Readiness Survey.	- Men and women between the ages of 35 and 80. - Total knee replacement in one knee. - At least 6-months from TKR. - No more than 5-years from TKR

Four discrete reflective tracking markers were placed on the posterior and lateral heel counter of the lab shoes. After the collection of a static trial, all anatomical markers were removed for the dynamic trials. A three-step staircase (FP-Stairs, American Mechanical Technology Inc., Watertown, MA, USA) was securely bolted to two force platforms (1200 Hz, BP600600 and OR-6-7, American Mechanical Technology Inc., Watertown, MA, USA) in order to measure kinetics during stair negotiation (Figure 1). An additional two steps and a platform were also included. Each step had a rise of 17.8 cm, width of 60.0 cm, and depth of 29.9 cm. A wooden rail on the right of the staircase was available to prevent any risk of falling. Speed was monitored between the 1st and 4th steps using two photo cells (63501 IR, Lafayette Instrument Inc., IN, USA) and two electronic timers (54035A, Lafayette Instrument Inc., IN, USA). Participants were provided a standardized laboratory cushioned running shoe (Noveto, Adidas, USA) and athletic clothes to wear during the experiment.

Experimental Procedures

As this study is part of an ongoing research project, only the procedures relevant to the current study were presented here. Before the reflective markers were placed on the subjects completed a 3-minute warm-up on a treadmill at a self-selected pace. Step 2 was the step of interest, so data were collected for each leg contacting the 2nd step. Subjects were asked to perform 3-5 trials per leg of stair ascent at a self-selected speed. A minimum of three practice trials were used to determine the subject's self-selected speed. Speed was then monitored using two photo cells and electronic times and controlled for the experimental trials within ±5% of the average identified self-selected speed.

Figure 1. Complete setup of 5-step staircase for experimental data collections.

Data Analyses

Visual 3D (C-Motion, Inc., Germantown, MD, USA), a biomechanical analysis software suite was used to filter both kinematic and ground reaction force data at 8 Hz (Kristianslund et al., 2012), respectively, using a fourth-order Butterworth low-pass filter. An X-y-z Cardan rotational sequence was used in joint calculations and the right hand rule for determining the conventions for joint angles and moments. All joint moments were computed as internal moments. The processed individual trials were exported for use in OpenSim (3.0.1, SimTK, Stanford, CA, USA) an open-source musculoskeletal simulation software package. A generic 12-segment, 19-degree of freedom, and 92 muscle OpenSim musculoskeletal model (Gait 2392 Model), originally developed by Delp et al. (1990) was used and scaled to the height and weight of each individual subject before simulations.

Scaling modifies the model to match the anthropometric and physiological features (e.g. segment mass, muscle fiber length, tendon slack length) of the model to the subject. The movement from the experimental data was then applied to the scaled model to create a completed dynamic musculoskeletal model (Delp et al., 2007). Next, inverse kinematics is computed on the experimental data to identify the joint angles and translations that correlate most accurately with the collected data. The foundation for this inverse kinematics problem is to minimize the differences between the experimental movement and the models reproduction. Those differences can be expressed as a least squared problem in which the weighted squared error is minimized:

$$\text{Squared Error} = \sum_{i=1}^{\text{markers}} w_i(\vec{x}_i^{\text{subject}} - \vec{x}_i^{\text{model}})^2 + \sum_{j=1}^{\text{joint angles}} w_j(\theta_j^{\text{subject}} - \theta_j^{\text{model}})^2$$

$\vec{x}_i^{subject}$ and $\vec{x}_i^{model}$ correspond to the ith marker or joint center position in three-dimensions for the subject and model, respectively. $\theta_j^{subject}$ and θ_j^{model} corresponds to the jth joint angle for the subject and model. w_i and w_j are weighting factors to allow marker and joint angle data to be weighted separately (Delp et al., 2007).

Inverse dynamics was then run to calculate internal joint forces and torques for each trial. Unfortunately, experimental kinematics is not dynamically consistent with the kinetics so

OpenSim applies residual forces to account for differences. The following equation describes the relationship of residual forces with the experimental forces and segment accelerations:

$$\vec{F}_{external} = \sum_{i=1}^{segments} m_i \vec{a}_i - \vec{F}_{residual}$$

Where $\vec{F}_{external}$ corresponds to the ground reaction force measured without the body weight vector. $\vec{a}_i$ is the center of mass translational acceleration for the ith segment and m_i is the mass of that segment. $\vec{F}_{residual}$ is the residual force that has been applied to initially account for the dynamic inconsistencies (Delp et al., 2007). In order to improve the accuracy of the simulation a residual reduction algorithm (RRA) was run. As the name implies, RRA is intended to minimize the amount of residual forces ($\vec{F}_{residual}$) added to the model by making small adjustments to the motion trajectory and mass parameters. Residual forces go to zero as the amount of experimental and modeling errors also approaches zero. The level to which changes are made can be controlled by creating limits for the acceptable magnitude of the residuals. The changes to the motion trajectories were checked using the RRA best practices recommended thresholds for evaluating results to ensure the changes made are not too large that the movement has drastically changed. Table 1 below shows threshold values recommended by OpenSim for the evaluation of RRA results for over-ground walking. Table 1 provides ranges for maximum and root mean square (RMS) values for residual forces and moments and translational and rotational errors (pErr) (Delp et al., 2007). Values for stair climbing are expected to be higher due to the increased difficulty of stair climbing therefore Table 1 will be utilized as a guideline for interpretation of results.

Individual muscle forces and excitations were calculated via computed muscle control (CMC) drive a simulation of the movement originally collected experimentally. The

Table 4. Recommended Threshold Values for Evaluation of RRA Results (Delp et al., 2007).

Thresholds:	GOOD	OKAY	BAD
MAX Residual Force (N)	0-10 N	10-25N	> 25 N
RMS Residual Force (N)	0-5 N	5-10 N	> 10 N
MAX Residual Moment (Nm)	0-50 Nm	50-75 Nm	>75 Nm
RMS Residual Moment (Nm)	0-30 Nm	30-50 Nm	>50 Nm
MAX pErr (trans, cm)	0-2 cm	2-5 cm	>5 cm
RMS pErr (trans, cm)	0-2 cm	2-4 cm	>4 cm
MAX pErr (rot, deg)	0-2 deg	2-5 deg	> 5 deg
RMS pErr (rot, deg)	0-2 deg	2-5 deg	> 5 deg

combination of a static optimization and proportional-derivative control allow for the forward dynamics simulation to nearly mirror the movements produced by the RRA. Activation and contraction dynamics also play a role in determining the output simulation motion. Activation dynamics takes into account magnitude of muscle excitation and activation as well as the rate of change in muscle activation. Force-length-velocity of the muscles and elastic properties of the muscles are accounted for by the contraction dynamics aspect of the simulation (Delp et al., 2007). Further analyses were performed using the OpenSim tools to calculate joint reaction forces (JRF). The JRF analysis allows for the calculation of three dimensional loading (compressive and shear forces) of joints over the entire movement duration.

The dependent variables included: peak vertical GRF, peak knee extension moment, peak knee abduction moment, peak knee compressive force, peak knee anterior/posterior forces, peak knee extensor and flexor muscle forces, peak knee flexion angle, knee flexion ROM, and peak adduction angle. In order to compare differences between TKR and healthy individuals an independent samples t-test was run for each variable of interest (21.0, IBM SPSS, Chicago, IL). Additionally, a qualitative analysis was performed to assess the differences in the timing of peak muscle forces with respect peak JRFs.

CHAPTER IV
JOINT REACTION FORCE AND CONTRIBUTIONS OF SURROUNDING MUSCLES TO KNEE JOINT LOAD DURING STAIR ASCENT IN TOTAL KNEE REPLACEMENT PATIENTS AND HEALTHY INDIVIDUALS

Abstract

Total knee replacement (TKR) is commonly used to correct end stage knee osteoarthritis (OA) of the knee joint. Unfortunately, difficulty with stair climbing has been seen to exist, prolonging the challenges of TKR patents. Complete understanding of loading at the knee is of great interest in order to aid patient populations, implant manufacturers, rehabilitation, and future research. Musculoskeletal modeling provides a means to accurately approximate joint loading and the corresponding muscle contributions during a movement. The purpose of the present study was to examine if the knee joint loading are recovered to the level of healthy individuals following a TKR, and determine the contribution of the muscles to knee joint loading. Data from five healthy and five TKR patients were selected for musculoskeletal simulation. Variables of interest included knee joint reaction forces and the corresponding muscle forces. A paired samples t-test was used to detect difference between groups for each variable of interest with an alpha level set at 0.05 a priori. TKR patients showed a trend of having higher 2nd peak compressive JRF. Some muscle compensatory strategies appear to be present in the push-off phase; however the differences in muscles do not clearly explain the trend present in compressive JRF during the 2nd 50% of stance. Evidence from knee extension moment and muscle force contributions during the loading response phase indicates reduced muscle strength in the knee extensors of TKR patients. This result combined with greater flexor muscle force resulted in similar compressive JRF during loading response between groups.

Introduction

Total knee replacement (TKR) is commonly used to correct end stage knee osteoarthritis (OA) within the knee joint. The frequency with which TKRs are performed is expected to double in the US alone by 2015, reaching nearly 3.5 million by 2030 (Kurtz et al., 2007). The primary purposes of a TKR are to alleviate pain, restore normal range of motion (ROM), and restore the ability to perform activities of daily living. Several studies have reported reductions in the pain after a TKR (Bruyere et al., 2012; Hawker et al., 1998; Ko et al., 2011), however other studies reporting disappointment of patients due to post-surgery pain (Beswick et al., 2012; Dickstein et al., 1998) and difficulties with stair negotiation (Dickstein et al., 1998). Difficulty with stair climbing prolongs challenges of TKR patents which a TKR is intended to correct. Stair climbing is a common activity of daily living, with older adults utilizing stairs as frequently as younger adults (Startzell et al., 2000). Additionally, stair climbing is utilized in all clinical recovery assessments after a TKR including: the original knee society scoring system (Insall et al., 1989), the new knee society scoring system (Scuderi et al., 2012), and the oxford knee (Dawson et al., 1998). Therefore a holistic understanding of what occurs biomechanically during stair climbing is essential.

Traditional biomechanics has been utilized by several studies and a review to provide an understanding of how well a TKR actually restores a healthy gait in end-stage OA patients (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Mandeville et al., 2007; Standifird and Zhang, Accepted). The information provided by joint moments and joint reaction forces via inverse dynamics does not provide a true bone on bone loading at the knee joint. Loading at the joint results from the contraction of muscles, not just the GRF propagated up through the body. Several studies have provided a good understanding of these effects by utilizing an instrumented TKR while climbing stairs (Catani et al., 2009; D'Lima et al., 2007; D'Lima et al., 2005; D'Lima et al., 2006; Heinlein et al., 2009; Kim et al., 2009; Kutzner et al., 2010; Mundermann et al., 2008). While instrumented implants do provide additional information to the understanding of the kinetic requirements of stair ascent, there are limitations to consider such as the limited subject populations and the inability to examine the muscle contributions to knee joint loading.

Musculoskeletal modeling provides a means to accurately approximate joint loading and the corresponding muscle contributions during a movement. Kim et al. (2009) made comparisons between musculoskeletal simulation and instrumented implant data, showing “good” agreement between the two in over-ground walking. Musculoskeletal modeling has been utilized in applications for over-ground walking, running, crouch gait, and stair negotiation in both healthy and patient subject groups. However, only a limited number of studies have utilized musculoskeletal simulations to investigate stair negotiation (Ghafari et al., 2009; Piazza and Delp, 2001; Rouston, 2010; Taylor et al., 2004). Complete understanding of loading at the knee is of great interest in order to aid patient populations, implant manufacturers, rehabilitation, and future research. The outcome of a TKR is intended to reestablish normal motion and loading of TKR patients. Musculoskeletal simulation can provide an excellent means of determining true subject specific knee loading and aid in improving TKR design and restoring TKR patients to normal loading and movement patterns.

To date no studies have investigated the knee joint loading and the contributions of muscle forces in TKR patients during stair ascent via musculoskeletal modeling. Simulation studies have only utilized a single step-up task in their analysis failing to reflect the actual movement while climbing stairs. Contributions of knee joint muscles to the knee joint loading in stair ascent for TKR patients are not clear. Therefore, the purpose of the present study was to examine if the knee joint loading are recovered to the level of healthy individuals following a

TKR, and determine the contribution of the muscles to knee joint loading. It was hypothesized that knee joint compressive and shear loading and knee muscle forces would be different between TKR patients and their healthy counterparts.

Methods

Participants

All participants were recruited for a larger study currently underway in the UTK Biomechanics/Sports Medicine Lab. TKR patients were recruited through a local orthopedic clinic and all surgeries were performed by the same surgeon. All patients received a posterior stabilized TKR and were 14.6 ± 3.4 months post-surgery at the time of the data collection. All healthy participants had no knee pain in the past 6 months during daily activities and not been diagnosed of lower extremity joint OA. Additionally, healthy participants were excluded for any applicable criteria included for TKR patients (Table 1). All participants signed an informed consent approved by the University of Tennessee Institutional Review Board. Data from five healthy (57.8 ± 10.0 yrs, 1.8 ± 0.1 m, 89.0 ± 6.6 kg) and five TKR patients (63.6 ± 8.7 yrs, 1.7 ± 0.1 m, 87.0 ± 8.9 kg) were selected randomly from the total participant population for analysis.

Experimental Procedures

3-D kinematics was collected experimentally using a nine-camera motion analysis system (240 Hz, VICON Motion Analysis Inc., Oxford, UK). Reflective anatomical markers were place bi-laterally on the following anatomical landmarks: acromion processes, iliac crests, greater trochanters, medial and lateral femoral epicondyles, medial and lateral malleoli, 1^{st} and 5^{th} metatarsal heads, and toes (i.e. the most anterior aspect of the shoes). Reflective tracking markers in sets of four were connected to semi-rigid thermoplastic shells and secured to the trunk, pelvis, thighs, shanks, and on the posterior and lateral heel counter of a pair of standard lab shoes (Noveto, Adidas, USA). A three-step staircase (FP-Stairs, American Mechanical Technology Inc., Watertown, MA, USA) was securely bolted to two force platforms (1200 Hz, BP600600 and OR-6-7, American Mechanical Technology Inc., Watertown, MA, USA) in order to measure ground reaction forces (GRF) during stair negotiation (Figure 1). An additional two steps and a platform were also included (Paquette et al., Accepted; Paquette et al., 2014).

Before the reflective markers were placed on the participants walked 3-minute as a warm-up on a treadmill at a self-selected pace. All participants then performed functional assessments including: timed up and go (TUG), knee range of motion (ROM), stair ascent time, and stair descent time. Participants were asked to perform a minimum of three practice trials to determine their self-selected speeds. Participants then performed five successful trials of stair ascent at the self-selected speed (±5%) which was monitored by two photo cells and electronic timers (Lafayette Instrument Inc., IN, USA). Step 2 was the step of interest.

Data Analyses

Visual 3D (C-Motion, Inc., Germantown, MD, USA), a biomechanical analysis software suite was used to filter both kinematic and ground reaction force data at 8 Hz (Kristianslund et al., 2012), respectively, using a fourth-order Butterworth low-pass filter. An X-y-z Cardan rotational sequence was used in joint angle calculations and the right hand rule was used for determining the conventions for joint kinematics and kinetics. All joint moments were computed as internal moments.

Musculoskeletal Simulations

The processed individual trials were exported to OpenSim (3.0.1, SimTK, Stanford, CA, USA) to perform musculoskeletal simulations. A generic 12-segment, 19-degree of freedom, and 92 muscle OpenSim musculoskeletal model (Gait 2392 Model), originally developed by

Delp et al. (1990), was used and scaled to the height and weight of each individual participant to generate subject-specific models before simulations. In order to improve the accuracy of the simulation a residual reduction algorithm (RRA) was performed to minimize the amount of residual forces added to the model to account for dynamic inconsistency by making small adjustments to the motion trajectory and mass parameters (Delp et al., 2007). Kinematic changes were all kept below 5.5 cm of translation and 3.5 degrees of rotation. Peak residual forces and moments were each kept below 14% of body weight and 1.6 Nm/kg, respectively. Individual muscle forces and excitations were calculated via computed muscle control (CMC) to drive a simulation of the movement collected experimentally (Thelen and Anderson, 2006; Thelen et al., 2003). Joint reaction forces (JRF) were computed using the OpenSim tool.

The dependent variables included: peak vertical GRF, peak knee extension moment, peak knee abduction moment, peak knee compressive force, peak knee anterior shear forces, peak knee extensor and flexor muscle forces, velocity, and the functional assessments. In order to compare differences between TKR and healthy individuals, an independent samples t-test was run for each variable of interest (21.0, IBM SPSS, Chicago, IL) with an alpha level set at 0.05 a priori.

Results

No significant differences in age ($p = 0.358$), height ($p = 0.540$), and mass ($p = 0.688$) existed between TKR patients and healthy subjects. No significant differences were found in stair ascent velocity (actual velocity obtained during movement trials), knee ROM, TUG, and stair descent time between healthy controls and TKR patients (Table 6). However, TKR patients showed a trend of having slower stair ascent ($p = 0.055$) compared to healthy controls (Table 6).

Knee JRF curves for both controls and TKR patients are provided in Figure 3. Peak knee extensor moment ($p = 0.014$) was reduced in TKR patients compared to their healthy counterparts (Table 7). The second peak of the compressive knee force ($p = 0.051$) showed a trend of elevated compressive loading in the knees of TKR patients compared to healthy controls.

Knee extensor and flexor muscle force curves for both healthy individuals and TKR patients are provided in Figure 3 and 4, respectively. The 1st peak muscle force of the rectus femoris ($p = 0.005$), vastus lateralis ($p = 0.002$), and sum of knee extensor forces ($p = 0.001$) were reduced in TKR patients compared to healthy individuals (Table 8). The 1st peak muscle force of the bicep femoris short head ($p = 0.035$, Table 8), sartorius ($p = 0.009$), gracilis ($p = 0.045$), and lateral gastrocnemius ($p = 0.040$) were all greater in TKR compared to healthy controls.

The 2nd peak muscle force of the rectus femoris ($p = 0.026$) was reduced in TKR patients compared to their healthy counterparts (Table 9). The 2nd peak muscle force of the vastus medialis was greater in TKR patients compared to healthy controls ($p = 0.027$). The 2nd peak muscle force of the sartorius ($p = 0.030$) and lateral gastrocnemius ($p = 0.028$) were found to be greater in TKR patients, while the medial gastrocnemius ($p = 0.006$) was reduced compared to healthy controls.

Discussion

The primary purpose of this study was to determine the extent to which knee joint loading is recovered to the level of healthy individuals following a TKR, and determine the contribution of the muscles to knee joint loading. The hypothesis that knee joint compressive and shear loading and knee muscle forces would be different between TKR patients and their healthy counterparts was partially confirmed by the results of this study.

Reduced peak knee extensor moments in TKR patients compared to healthy individuals during stair climbing is commonly reported in the literature (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Mandeville et al., 2007). The findings of this study supported this conclusion (Table 7). Additionally, the results of the current study showed no differences in peak knee abduction moment which are supported in the literature (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003). However, it becomes important to note that implant design has been shown to play a role in the existence of differences in peak knee abduction moment. Some have found differences between controls and TKR patients in peak knee abduction moment when using a mobile bearing design (Catani et al., 2003; Fantozzi et al., 2003). All patients in this study received a posterior stabilized TKR which has been shown to result in no differences in peak abduction moment in the literature (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003).

The 1st peak compressive JRF was not different between groups. Increased knee muscle forces are likely to be present in order for the compressive loading to be the same between groups during the loading response (first 50%) of stance. Combined knee extensor muscle forces during the loading response phase were shown to be reduced in TKR patients compared to healthy controls (Table 8). The peak rectus femoris and vastus lateralis forces were both reduced in TKR patients resulting in a reduced sum of knee extensor muscles. Interestingly, the peak bicep femoris short head and lateral gastrocnemius forces were greater in TKR patients. Two accessory muscle forces, the sartorius and gracilis, also were larger in TKR patients. These findings complement the lack of differences in 1st peak compressive JRF and the previously mentioned reduction in knee extensor moment found here and in the literature. It is possible that the differences between TKR patients and healthy controls during the loading response phase lies primarily within the muscle differences. It might be assumed that differences in muscle force production would directly result in difference in the JRF, but clearly this is not the case.

The 2nd peak compressive JRF showed a trend toward being increased in TKR patients (Table 7). The values for compressive loading seen in this study were elevated slightly over those seen in the literature for stair ascent (D'Lima et al., 2007; D'Lima et al., 2005; D'Lima et al., 2006; Heinlein et al., 2009). TKR patients were found to have a 1st peak compressive loading of 2.73 ± 0.35 BW and 2nd peak of 4.15 ± 0.36 BW compared to instrumented implant literature which ranged from 2.5 to 3.06 BW (D'Lima et al., 2005; D'Lima et al., 2006; Heinlein et al., 2009). Differences in velocity could explain the discrepancy between the findings reported here and those seen in the literature. The present study found no differences in the velocity of TKR patients compared to healthy individuals. However, velocity has been shown in the literature to be reduced in TKR patients compared to healthy controls (Mandeville et al., 2007). Velocity was not reported in the instrumented implant research making it impossible to know what differences exist, if any. Instrumented implant studies are also limited in that they only have limited number of subjects and may not capture the true nature of the overall TKR patient population. It is possible that the trend towards greater 2nd peak compressive JRF in TKR patients is also a result of differences existing in the muscle forces. With a greater sample size, this difference could become significant.

During the push-off (second 50%) of stance, the results showed more variable differences in individual muscle force contributions than the loading response phase. While there were some increases and some reductions in muscle force for knee extensors and flexors neither summed group for flexors or extensors showed any differences during the push-off phase. While no differences existed in peak values it appears a different strategy is utilized by TKR patients to produce similar levels of muscle force contributions to healthy controls. It can be seen that the majority of force contribution during the push-off is from the rectus femoris in healthy individuals (Figure 7a). However, TKR patients utilized the vastus medialis more during the same part of stance than healthy controls (Figure 7b). Similarly, TKR patients utilized the medial and lateral gastrocnemius differently than healthy individuals. On the other hand, healthy controls employed the medial gastrocnemius more during the second half of stance while TKR patients primarily used the lateral gastrocnemius more (Figure 8a and 8b). It remains unclear the exact nature of these differences and the influence they have on JRF. TKR patients may be utilizing the muscles differently as a compensatory strategy for the reduced knee extensor strength that remains after rehabilitation. It is possible that gait compensation strategies seen in knee OA patients to relieve pain linger after the TKR rehabilitation is completed. Knee extensor muscle strength has been shown to be reduced in OA patients (Petterson et al., 2008). Based on the findings of this study and others (Berti et al., 2006; Catani et al., 2003; Fantozzi et al., 2003; Mandeville et al., 2007), the TKR or rehabilitation may have not addressed the reduced knee extensor strength. Rehabilitation focusing on re-establishing the rectus femoris and medial gastrocnemius as the primary force producers may help TKR patients reach similar levels of joint loading and muscle force production as seen in healthy individuals.

Peak shear JRF did not differ between healthy controls (2.61± 0.28 BW) and TKR patients (2.52 ± 0.57 BW) in the present study. However, shear loading in TKR patients was notably different from findings seen in the instrumented TKR literature which range between 0.26 and 0.36 BW in stair ascent (D'Lima et al., 2007; Heinlein et al., 2009; Kutzner et al., 2010). The findings of the present study suggest that TKR patients and healthy controls produce similar anterior shear loading and pattern during stair climbing (Table 7 and Figure 3). Differences between the present study and instrumented implant studies may be due to differences in speed and in implant design type. The present study only utilized a posterior stabilized design. The cruciate retaining (D'Lima et al., 2007) and mobile bearing (Kutzner et al., 2010) designs were commonly seen in the literature. The cruciate retaining and mobile bearing designs are more restrictive in the movement of the knee joint than the posterior stabilizing design. Also, the posterior stabilized design has been shown in literature to result in an anterior translation of the femur during a step up movement in 75% of a large cohort of TKR patients including over 40 knees (Banks and Hodge, 2004).

Conclusions

Evidence from knee extension moment and muscle force contributions during the loading response phase indicates reduced muscle strength in the knee extensors of TKR patients. This result combined with greater flexor muscle force resulted in similar compressive JRF during loading response between groups. TKR patients showed a trend of having higher 2nd peak compressive JRF than healthy individuals. Some muscle compensatory strategies appear to be present in the push-off phase; however the differences in muscles do not clearly explain the trend present in compressive JRF. Future research utilizing musculoskeletal modeling may investigate differences in muscle forces dependent on rehabilitation strategies. Also, different TKR design types always have a potential impact on joint loading and muscle contributions. Comparisons of pre- and post-surgery data would also provide a more clear insight into how well a TKR aids in correcting end-stage OA.

Table 5. Inclusion and Exclusion Criteria for the TKR Subjects.

Exclusion Criteria:	Inclusion Criteria:
- Any additional lower extremity joint replacement. - Any lower extremity joint arthroscopic surgery or intra-articular injection within the past month. - Systemic inflammatory arthritis (rheumatoid arthritis, psoriatic arthritis) (self-reported). - BMI greater than 35. - Inability to ascend/descend stairs without the use of a handrail. - Neurologic disease (e.g. Parkinson's Disease, stroke patients) (self-reported). - Any major lower extremity injuries/surgeries. - Inability to walk without a walking aid. - Any visual conditions affecting gait or balance. - Women who are pregnant or nursing. - Any cardiovascular disease or primary risk factor which precludes participation in aerobic exercise as indicated by the Physical Activity Readiness Survey.	- Men and women between the ages of 35 and 80. - Total knee replacement in one knee. - At least 6-months from TKR. - No more than 5-years from TKR

Table 6. Stair ascent velocity and functional assessments of healthy controls and TKR patients (Mean ± SD).

	Units	Healthy	TKR	P-value
Velocity	(m/s)	1.6 ± 0.2	2.1 ± 0.6	0.154
Knee ROM	(deg.)	121.4 ± 7.4	113.6 ± 7.3	0.133
TUG	(sec.)	7.4 ± 1.2	7.4 ± 0.5	0.991
Stair Ascent Time	(sec.)	6.2 ± 0.2	7.0 ± 0.7	0.055

Table 7. Peak GRF, knee moments, and knee JRF of healthy controls and TKR patients during stair climbing (Mean ± SD).

	Units	Healthy	TKR	P-value
1st Peak Vertical GRF	(N)	925.2 ± 117.7	850.1 ± 77.4	0.268
2nd Peak Vertical GRF	(N)	907.2 ± 95.7	985.8 ± 149.7	0.351
Peak Extension Moment	(Nm)	119.9 ± 25.9	77.1 ± 16.5	**0.014**
1st Peak Abduction Moment	(Nm)	-42.9 ± 13.1	-36.5 ± 18.5	0.546
2nd Peak Abduction Moment	(Nm)	-29.1 ± 12.6	-23.3 ± 11.1	0.489
Peak Anterior Shear JRF	(N)	2281.3 ± 294.4	2170.2 ± 624.5	0.732
1st Peak Compressive JRF	(N)	-2633.2 ± 208.7	-2332.5 ± 415.4	0.186
2nd Peak Compressive JRF	(N)	-2774.3 ± 456.5	-3560.6 ± 609.6	0.051

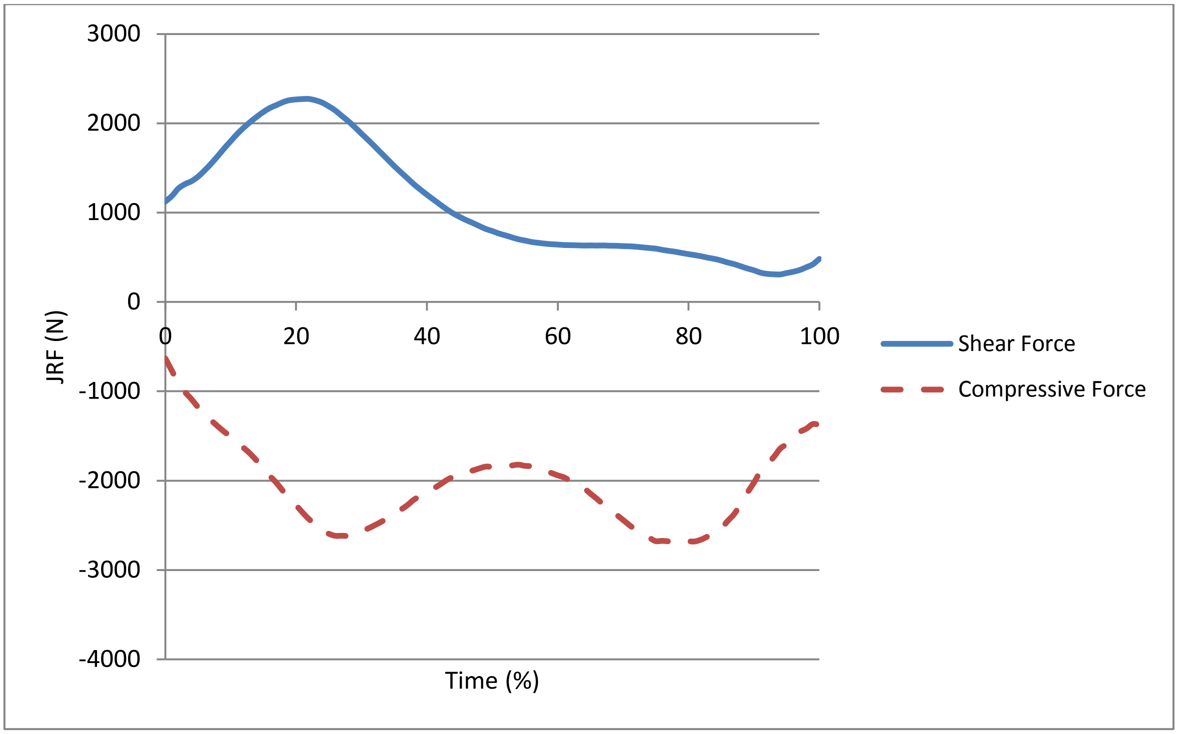
Table 8. 1st peak knee extensor and flexor muscle forces for healthy controls and TKR patients during stair climbing (Mean ± SD).

		Units	Healthy	TKR	P-value
Knee Extensors	1st Peak Rectus Femoris	(N)	516.6 ± 116.3	232.5 ± 120.2	**0.005**
	1st Peak Vastus Medialis	(N)	439.0 ± 59.1	649.1 ± 311.7	0.208
	1st Peak Vastus Intermedius	(N)	504.4 ± 68.0	412.3 ± 121.3	0.177
	1st Peak Vastus Lateralis	(N)	921.8 ± 124.0	527.3 ± 155.2	**0.002**
	1st Peak Sum	(N)	2124.2 ± 283.1	1340.8 ± 163.8	**0.001**
Knee Flexors	1st Peak Semimembranosus	(N)	419.1 ± 113.5	378.3 ± 106.8	0.575
	1st Peak Semitendinosus	(N)	61.6 ± 28.2	40.2 ± 13.8	0.167
	1st Peak Bicep Femoris Long Head	(N)	263.4 ± 35.0	255.2 ± 110.2	0.879
	1st Peak Bicep Femoris Short Head	(N)	281.1 ± 114.5	444.5 ± 86.8	**0.035**
	1st Peak Sartorius	(N)	27.9 ± 10.4	56.4 ± 15.6	**0.009**
	1st Peak Gracilis	(N)	9.8 ± 4.7	29.0 ± 17.5	**0.045**
	1st Peak Medial Gastrocnemius	(N)	537.2 ± 215.8	781.4 ± 293.4	0.172
	1st Peak Lateral Gastrocnemius	(N)	141.6 ± 80.6	321.6 ± 143.5	**0.040**
	1st Peak Sum	(N)	1343.1 ± 308.6	1511.0 ± 429.5	0.498

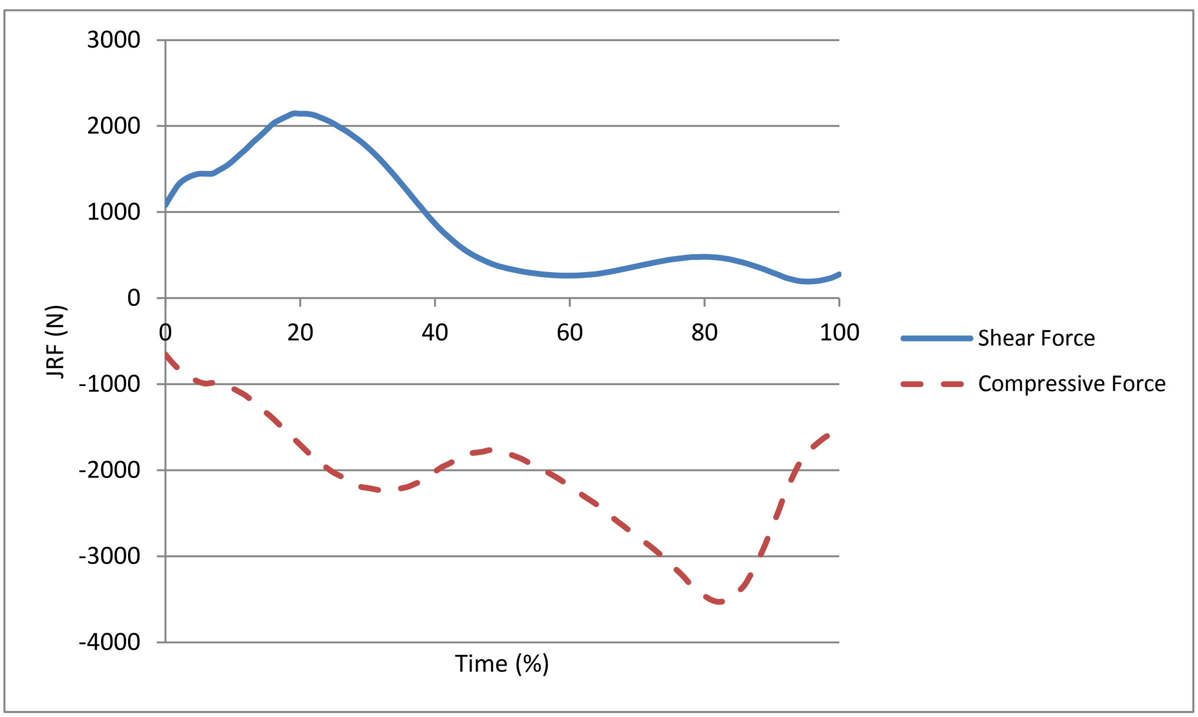
Table 9. 2nd peak knee extensor and flexor muscle forces for healthy controls and TKR patients during stair climbing (Mean ± SD).

		Units	Healthy	TKR	P-value
Knee Extensors	2nd Peak Rectus Femoris	(N)	730.7 ± 127.2	322.3 ± 310.4	**0.026**
	2nd Peak Vastus Medialis	(N)	92.6 ± 45.6	722.3 ± 415.5	**0.027**
	2nd Peak Vastus Intermedius	(N)	102.5 ± 53.5	63.1 ± 23.6	0.186
	2nd Peak Vastus Lateralis	(N)	184.8 ± 112.5	76.6 ± 29.3	0.098
	2nd Peak Sum	(N)	996.3 ± 227.2	1091.0 ± 271.2	0.566
Knee Flexors	2nd Peak Semimembranosus	(N)	359.6 ± 57.6	439.2 ± 116.3	0.207
	2nd Peak Semitendinosus	(N)	42.2 ± 18.5	43.4 ± 15.0	0.913
	2nd Peak Bicep Femoris Long Head	(N)	148.5 ± 77.1	129.7 ± 31.3	0.628
	2nd Peak Bicep Femoris Short Head	(N)	322.3 ± 45.8	398.4 ± 100.3	0.177
	2nd Peak Sartorius	(N)	38.6 ± 10.8	69.5 ± 23.9	**0.030**
	2nd Peak Gracilis	(N)	10.4 ± 3.2	10.9 ± 2.1	0.761
	2nd Peak Medial Gastrocnemius	(N)	847.5 ± 207.2	244.7 ± 296.7	**0.006**
	2nd Peak Lateral Gastrocnemius	(N)	242.7 ± 76.0	906.9 ± 446.8	**0.028**
	2nd Peak Sum	(N)	1367.3 ± 214.1	1587.5 ± 272.0	0.193

Figure 2. Complete setup of 5-step staircase for experimental data collections.

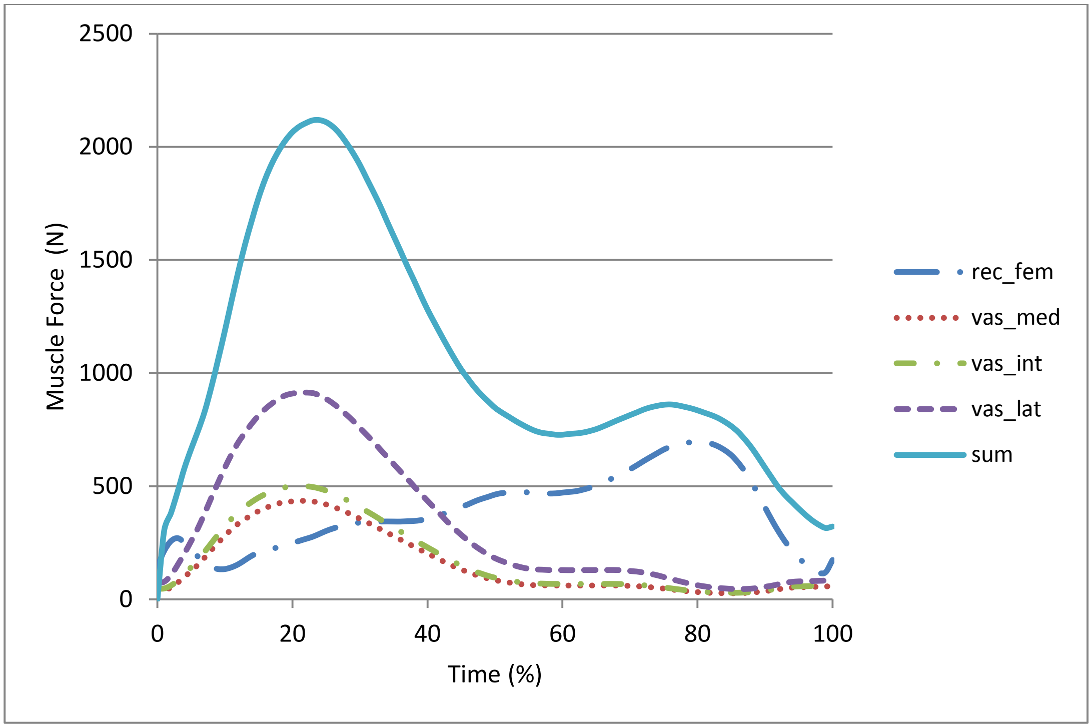


(a)

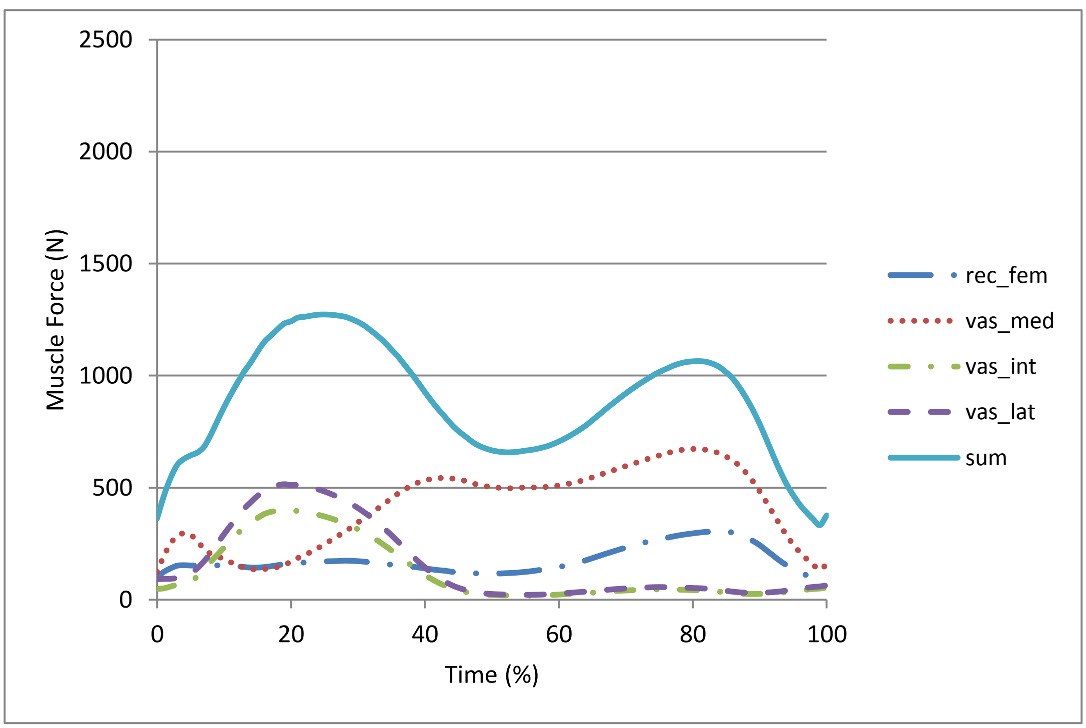


(b)

Figure 3. Knee joint reaction forces for healthy controls (a) and TKR patients (b).



(a)



(b)

Figure 4. Knee extensor muscle forces for healthy controls (a) and TKR patients (b). Note: rec_fem = rectus femoris, vas_med = vastus medialis, vas_int = vastus intermedius, vas_lat = vastus lateralis, sum = point by point summation of all knee extensors.

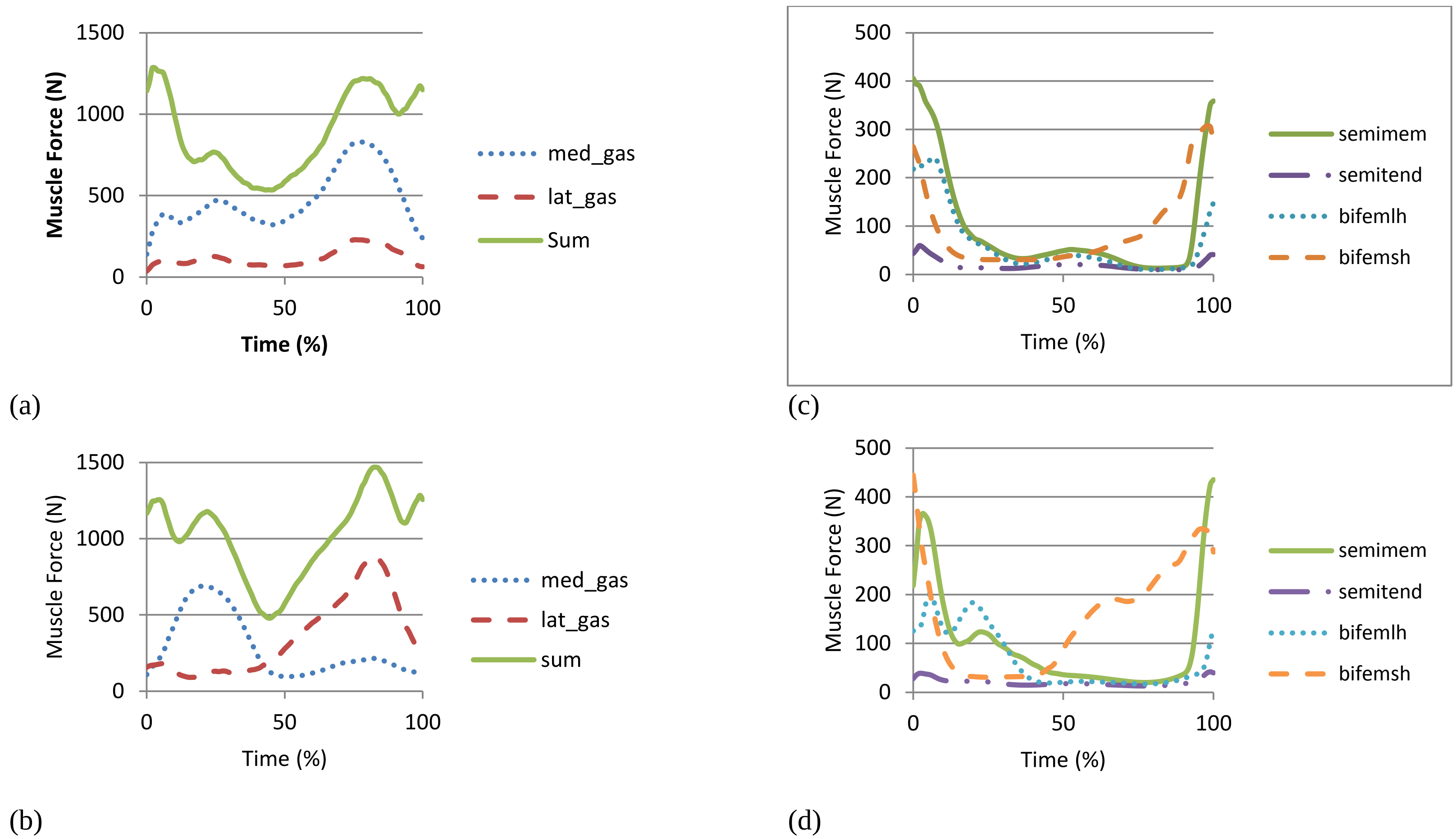


Figure 5. Knee flexor muscle forces of medial and lateral gastrocnemius and total knee flexor sum for healthy controls (a) and TKR patients (b); muscle forces of the semimembranosus, semitendinosus, biceps femoris long head, and biceps femoris short head for healthy controls (c) and TKR patients (d). Note: med_gas = medial gastrocnemius, lat gas = lateral gastrocnemius, sum = point by point summation of all knee flexors, semimem = semimembranosus, semitend = semitendinosus, bifemlh = biceps femoris long head, bifemsh = biceps femoris short head

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APPENDIX

APPENDIX A
Individual Subject Demographics

Table 10. Subject demographics.

		Age (yrs)	Mass (kg)	Height (m)	BMI
Healthy	1	68	79.80	1.76	25.76
	2	45	91.60	1.80	28.27
	3	55	96.60	1.75	31.54
	4	53	85.00	1.69	29.76
	5	68	92.19	1.89	25.81
	mean ± SD	57.8 ± 10.0	89.0 ± 6.6	1.8 ± 0.1	28.2 ± 2.5
TKR	1	63	95.70	1.88	27.22
	2	67	77.80	1.66	28.23
	3	62	77.00	1.69	27.12
	4	51	91.40	1.69	32.00
	5	75	92.98	1.80	28.70
	mean ± SD	63.6 ± 8.7	87.0 ± 8.9	1.7 ± 0.1	28.7 ± 2.0

APPENDIX B
Inclusion and Exclusion Criteria

Table 11. Inclusion and Exclusion Criteria for the Healthy Subjects.

Exclusion Criteria:	Inclusion Criteria:
- Knee pain for at least 6 months during daily activities. - Diagnosed with any type of lower extremity joint osteoarthritis (self-reported). - Any lower extremity joint replacement. - Any lower extremity joint arthroscopic surgery or intra-articular injection within past 3 months. - Systemic inflammatory arthritis (rheumatoid arthritis, psoriatic arthritis) (self-reported). - BMI greater than 35. - Inability to ascend/descend stairs without the use of a handrail. - Inability to walk without a walking aid. - Neurologic disease (e.g. Parkinson's Disease, stroke patients) (self-reported). - Any major lower extremity injuries/surgeries. - Any visual conditions affecting gait or balance. - Women who are pregnant or nursing. - Any cardiovascular disease or primary risk factor which precludes participation in aerobic exercise as indicated by the Physical Activity Readiness Survey.	- Men and women between the ages of 35 and 80.

Table 12. Inclusion and Exclusion Criteria for the TKR Subjects.

Exclusion Criteria:	Inclusion Criteria:
- Any additional lower extremity joint replacement. - Any lower extremity joint arthroscopic surgery or intra-articular injection within the past month. - Systemic inflammatory arthritis (rheumatoid arthritis, psoriatic arthritis) (self-reported). - BMI greater than 35. - Inability to ascend/descend stairs without the use of a handrail. - Neurologic disease (e.g. Parkinson’s Disease, stroke patients) (self-reported). - Any major lower extremity injuries/surgeries. - Inability to walk without a walking aid. - Any visual conditions affecting gait or balance. - Women who are pregnant or nursing. - Any cardiovascular disease or primary risk factor which precludes participation in aerobic exercise as indicated by the Physical Activity Readiness Survey.	- Men and women between the ages of 35 and 80. - Total knee replacement in one knee. - At least 6-months from TKR. - No more than 5-years from TKR

APPENDIX C
Individual Subject Data

Table 13. Peak GRF, knee moments, and knee JRF of healthy controls and TKR patients during stair climbing.

		Vertical GRF 1st Peak (N)	Vertical GRF 2nd Peak (N)	Knee Extension Moment (Nm)	Abduction 1st Peak (Nm)	Abduction 2nd Peak (Nm)	Anterior Shear (N)	Compressive 1st Peak (N)	Compressive 2nd Peak (N)
Healthy	1	775.86 ± 35.54	934.09 ± 21.36	108.55 ± 10.47	-21.88 ± 3.02	-11.54 ± 0.64	1967.46 ± 98.59	-2723.61 ± 283.71	-3019.85 ± 105.26
	2	879.82 ± 67.26	954.38 ± 74.40	98.31 ± 13.57	-22.26 ± 5.27	-14.60 ± 4.87	2306.65 ± 274.48	-3105.24 ± 660.34	-3224.18 ± 298.67
	3	1013.74 ± 6.03	966.31 ± 17.02	106.37 ± 3.11	Missing	-2.71 ± 0.72	2247.05 ± 156.93	-2491.35 ± 55.55	-3184.52 ± 63.84
	4	919.30 ± 26.58	1180.22 ± 40.33	87.88 ± 3.17	-16.35 ± 1.50	-20.75 ± 2.01	2807.23 ± 40.75	-2635.04 ± 194.55	-3983.85 ± 343.57
	5	923.65 ± 12.10	975.87 ± 46.55	72.69 ± 3.99	-54.03 ± 2.58	-32.75 ± 2.39	2973.29 ± 119.17	-2933.58 ± 256.16	-3562.48 ± 195.24
	mean ± SD	902.48 ± 29.50	1002.18 ± 39.93	94.76 ± 6.86	-28.63 ± 3.09	-16.47 ± 2.12	2460.34 ± 137.98	-2777.76 ± 290.06	-3394.98 ± 201.32
TKR	1	868.89 ± 54.67	1049.92 ± 46.98	56.45 ± 5.73	-24.77 ± 1.92	-25.21 ± 2.73	2037.96 ± 306.52	-2209.59 ± 335.36	-4419.73 ± 417.96
	2	785.08 ± 12.08	771.02 ± 20.75	58.99 ± 2.47	-48.96 ± 0.91	-35.52 ± 0.00	1519.40 ± 41.07	-2247.23 ± 86.35	-3014.24 ± 260.30
	3	747.93 ± 13.79	947.22 ± 31.27	77.85 ± 1.94	-24.10 ± 4.54	-14.97 ± 4.75	1751.77 ± 97.05	-1832.34 ± 94.28	-3048.24 ± 149.93
	4	856.39 ± 26.91	877.92 ± 28.95	108.26 ± 10.15	-1.09 ± 1.28	-12.43 ± 2.91	2388.01 ± 669.22	-2565.83 ± 184.95	-2596.18 ± 592.93
	5	923.65 ± 12.10	975.87 ± 46.55	72.69 ± 3.99	-54.03 ± 2.58	-32.75 ± 2.39	2973.29 ± 119.17	-2933.58 ± 256.16	-3562.48 ± 195.24
	mean ± SD	836.39 ± 23.91	924.39 ± 34.90	74.85 ± 4.86	-30.59 ± 2.25	-24.18 ± 2.56	2134.09 ± 246.61	-2357.71 ± 191.42	-3328.17 ± 323.27

Table 14. 1st peak knee extensor muscle forces for healthy controls and TKR patients during stair climbing

		Knee Extensors				
		1st Peak Rectus Femoris (N)	1st Peak Vastus Medialis (N)	1st Peak Vastus Intermedius (N)	1st Peak Vastus Lateralis (N)	1st Peak Sum (N)
Healthy	1	542.12 ± 121.50	372.42 ± 5.99	428.43 ± 6.19	781.02 ± 16.91	1770.23 ± 132.91
	2	437.63 ± 40.09	404.29 ± 51.40	464.00 ± 59.13	851.25 ± 106.85	1979.65 ± 297.22
	3	532.50 ± 37.59	452.25 ± 27.22	517.38 ± 31.60	951.22 ± 48.96	2251.30 ± 114.86
	4	723.07 ± 71.14	452.38 ± 48.76	521.98 ± 57.81	947.37 ± 96.72	2221.90 ± 262.83
	5	444.09 ± 84.22	530.23 ± 22.91	609.79 ± 26.18	1111.59 ± 49.78	2489.62 ± 143.78
	mean ± SD	535.88 ± 70.91	442.31 ± 31.26	508.32 ± 36.18	928.49 ± 63.84	2142.54 ± 190.32
TKR	1	264.04 ± 55.22	812.25 ± 159.32	324.86 ± 22.34	380.24 ± 23.03	1195.74 ± 177.84
	2	173.69 ± 24.42	872.57 ± 62.31	295.77 ± 9.40	350.02 ± 11.85	1230.81 ± 42.37
	3	453.77 ± 86.96	318.88 ± 17.28	368.58 ± 20.36	667.12 ± 34.71	1451.11 ± 70.87
	4	281.59 ± 28.84	313.43 ± 102.24	510.96 ± 18.09	604.98 ± 20.46	1420.43 ± 22.51
	5	104.76 ± 14.87	979.56 ± 35.24	593.11 ± 35.62	672.78 ± 38.48	1572.71 ± 76.68
	mean ± SD	255.57 ± 42.06	659.34 ± 75.27	418.66 ± 21.16	535.03 ± 25.71	1374.16 ± 78.05

Table 15. 1st peak knee muscle forces for healthy controls and TKR patients during stair climbing

		Knee Flexors								
		1st Peak Semimembranosus (N)	1st Peak Semitendinosus (N)	1st Peak Bicep Femoris Long Head (N)	1st Peak Bicep Femoris Short Head (N)	1st Peak Sartorius (N)	1st Peak Gracilis (N)	1st Peak Medial Gastrocnemius (N)	1st Peak Lateral Gastrocnemius (N)	1st Peak Sum (N)
Healthy	1	436.37 ± 43.50	111.71 ± 42.93	256.94 ± 31.46	425.77 ± 43.28	29.78 ± 4.49	19.31 ± 3.66	844.03 ± 152.51	289.90 ± 81.18	1847.41 ± 148.52
	2	617.32 ± 128.70	71.52 ± 23.04	336.50 ± 26.89	216.18 ± 91.94	31.73 ± 5.20	9.39 ± 2.51	873.47 ± 320.78	304.03 ± 131.22	1645.81 ± 196.99
	3	391.18 ± 101.69	51.94 ± 13.89	253.39 ± 41.39	182.81 ± 44.93	21.98 ± 3.98	7.02 ± 0.49	355.68 ± 85.06	70.68 ± 26.40	1106.37 ± 228.90
	4	441.26 ± 95.25	49.05 ± 9.30	291.48 ± 38.57	195.61 ± 60.82	21.57 ± 2.71	8.21 ± 1.00	405.17 ± 207.71	75.11 ± 40.44	1093.46 ± 100.82
	5	304.32 ± 89.19	43.19 ± 22.95	240.95 ± 46.25	394.29 ± 99.44	44.11 ± 13.83	8.25 ± 1.18	682.12 ± 163.65	200.83 ± 83.12	1386.49 ± 313.90
	mean ± SD	438.09 ± 91.67	65.48 ± 22.42	275.85 ± 36.91	282.93 ± 68.08	29.83 ± 6.04	10.44 ± 1.77	632.09 ± 185.94	188.11 ± 72.47	1415.91 ± 197.83
TKR	1	568.79 ± 134.67	63.85 ± 17.55	312.33 ± 90.88	436.33 ± 73.84	69.80 ± 13.51	22.53 ± 7.65	678.22 ± 39.61	485.44 ± 137.80	1798.74 ± 380.54
	2	286.63 ± 121.77	23.64 ± 7.83	88.27 ± 12.23	355.16 ± 10.75	38.20 ± 2.17	30.18 ± 2.73	618.68 ± 20.77	335.26 ± 97.37	909.42 ± 77.89
	3	298.71 ± 76.81	43.69 ± 3.15	240.76 ± 60.24	544.23 ± 27.49	74.86 ± 8.26	6.52 ± 0.10	525.05 ± 167.81	134.97 ± 52.17	1515.91 ± 311.56
	4	408.33 ± 57.34	38.64 ± 5.41	395.69 ± 114.48	370.75 ± 13.02	43.63 ± 2.16	35.63 ± 2.07	1073.47 ± 35.67	341.03 ± 98.22	2021.59 ± 142.05
	5	387.23 ± 114.99	46.28 ± 23.95	266.79 ± 35.31	526.41 ± 59.08	80.18 ± 16.35	53.13 ± 13.35	1141.36 ± 45.93	452.58 ± 99.56	1624.59 ± 197.95
	mean ± SD	389.94 ± 101.12	43.22 ± 11.58	260.77 ± 62.63	446.58 ± 36.84	61.33 ± 8.49	29.60 ± 5.18	807.36 ± 61.96	349.86 ± 97.03	1574.05 ± 222.00

Table 16. 2nd peak knee extensor muscle forces for healthy controls and TKR patients during stair climbing.

		Knee Extensors				
		2nd Peak Rectus Femoris (N)	2nd Peak Vastus Medialis (N)	2nd Peak Vastus Intermedius (N)	2nd Peak Vastus Lateralis (N)	2nd Peak Sum (N)
Healthy	1	834.30 ± 167.48	77.03 ± 21.50	85.37 ± 24.69	159.99 ± 53.08	1052.68 ± 199.04
	2	897.88 ± 165.65	65.08 ± 17.40	71.35 ± 20.51	118.87 ± 56.29	953.52 ± 178.48
	3	585.09 ± 41.22	59.60 ± 2.32	63.96 ± 3.73	97.83 ± 23.67	702.61 ± 75.81
	4	792.41 ± 154.22	152.48 ± 25.68	171.33 ± 28.80	331.85 ± 56.82	1331.88 ± 104.61
	5	682.93 ± 29.33	142.75 ± 31.22	161.45 ± 35.65	307.38 ± 66.73	1068.83 ± 171.18
	mean ± SD	758.52 ± 111.58	99.39 ± 19.62	110.69 ± 22.68	203.18 ± 51.32	1021.91 ± 145.83
TKR	1	238.61 ± 65.06	1061.64 ± 131.11	57.37 ± 1.94	65.07 ± 0.95	1343.66 ± 112.80
	2	194.61 ± 2.80	927.82 ± 136.60	48.41 ± 0.61	56.53 ± 0.65	1149.79 ± 172.29
	3	901.72 ± 91.73	48.71 ± 12.50	52.86 ± 13.91	74.26 ± 28.27	948.80 ± 97.90
	4	147.32 ± 9.84	1069.13 ± 31.41	110.63 ± 46.96	134.71 ± 57.41	1404.73 ± 106.90
	5	179.26 ± 9.44	623.78 ± 46.81	52.40 ± 4.31	60.41 ± 6.06	764.97 ± 84.80
	mean ± SD	332.30 ± 35.78	746.21 ± 71.69	64.33 ± 13.54	78.20 ± 18.67	1122.39 ± 114.94

Table 17. 2nd peak knee flexor muscle forces for healthy controls and TKR patients during stair climbing.

		Knee Flexors								
		2nd Peak Semimembranosus (N)	2nd Peak Semitendinosus (N)	2nd Peak Bicep Femoris Long Head (N)	2nd Peak Bicep Femoris Short Head (N)	2nd Peak Sartorius (N)	2nd Peak Gracilis (N)	2nd Peak Medial Gastrocnemius (N)	2nd Peak Lateral Gastrocnemius (N)	2nd Peak Sum (N)
Healthy	1	378.99 ± 43.65	68.63 ± 33.69	242.77 ± 98.23	328.08 ± 40.04	39.39 ± 11.32	15.03 ± 2.14	902.51 ± 79.59	342.73 ± 60.87	1486.00 ± 179.15
	2	394.73 ± 58.07	28.55 ± 9.86	107.98 ± 47.93	402.81 ± 30.02	46.65 ± 7.69	8.27 ± 1.92	995.10 ± 117.07	346.03 ± 23.42	1538.62 ± 138.47
	3	425.85 ± 23.16	53.71 ± 6.04	200.12 ± 29.02	287.07 ± 20.73	52.70 ± 1.80	11.92 ± 1.90	1076.41 ± 71.96	275.29 ± 46.96	1656.71 ± 82.33
	4	273.75 ± 71.60	23.88 ± 4.83	52.66 ± 14.47	297.82 ± 24.08	32.13 ± 3.34	7.49 ± 0.87	864.21 ± 45.70	217.28 ± 71.09	1178.03 ± 110.80
	5	343.63 ± 29.37	44.93 ± 5.24	158.07 ± 25.70	329.23 ± 32.09	25.23 ± 2.76	13.08 ± 2.04	616.16 ± 229.43	176.87 ± 97.38	1219.90 ± 145.71
	mean ± SD	363.39 ± 45.17	43.94 ± 11.93	152.32 ± 43.07	329.00 ± 29.39	39.22 ± 5.38	11.16 ± 1.77	890.88 ± 108.75	271.64 ± 59.94	1415.85 ± 131.29
TKR	1	468.96 ± 45.76	39.99 ± 7.72	130.33 ± 38.28	507.31 ± 126.77	111.12 ± 27.29	15.01 ± 3.84	82.44 ± 1.52	1258.27 ± 105.63	1918.78 ± 226.36
	2	256.45 ± 57.27	21.74 ± 2.48	79.17 ± 24.14	279.77 ± 38.13	58.14 ± 4.06	8.94 ± 0.18	69.72 ± 0.96	829.73 ± 58.05	1239.63 ± 75.46
	3	545.38 ± 93.63	62.24 ± 7.84	144.68 ± 46.79	363.48 ± 50.51	53.51 ± 5.10	10.23 ± 1.01	794.27 ± 37.09	199.01 ± 50.89	1491.66 ± 169.49
	4	539.32 ± 63.01	46.85 ± 14.39	137.11 ± 34.08	510.85 ± 80.83	61.44 ± 13.15	12.72 ± 2.79	243.87 ± 103.59	1058.84 ± 108.86	1657.02 ± 220.66
	5	431.16 ± 62.58	53.45 ± 13.15	164.29 ± 29.13	369.30 ± 59.05	70.64 ± 8.59	11.42 ± 2.26	78.11 ± 15.22	1309.11 ± 60.17	1798.34 ± 70.17
	mean ± SD	448.25 ± 64.45	44.85 ± 9.12	131.11 ± 34.48	406.14 ± 71.06	70.97 ± 11.64	11.66 ± 2.02	253.68 ± 31.68	930.99 ± 76.72	1621.09 ± 152.43

VITA

Robert J. Rasnick was born in Akron, OH on October 7th, 1988 to John and Janelle Rasnick. He grew up in northeast Ohio where he graduated from Cuyahoga Valley Christian Academy in 2007. He went on to pursue a Bachelor's degree in Mechanical Engineering from the University of Tennessee, Knoxville. In 2012, Robert began pursuing a Master's degree at the University of Tennessee, Knoxville. He graduated with a Master's of Science in Kinesiology with a concentration in Biomechanics in the year 2014