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Efficacy of a Cycling Intervention with Pedal Reaction Force Augmented Feedback on Reducing Inter-Limb Asymmetries in Patients with Unilateral Total Knee Arthroplasty

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I am submitting herewith a dissertation written by Erik T. Hummer entitled "Efficacy of a Cycling Intervention with Pedal Reaction Force Augmented Feedback on Reducing Inter-Limb Asymmetries in Patients with Unilateral Total Knee Arthroplasty." I have examined the final electronic copy of this dissertation for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Doctor of Philosophy, with a major in Kinesiology and Sport Studies.

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Efficacy of a Cycling Intervention with Pedal Reaction Force Augmented Feedback on Reducing Inter-Limb Asymmetries in Patients with Unilateral Total Knee Arthroplasty

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Dedication

This dissertation is dedicated to my wife Bethany and the rest of my family who were always there to support me along the way. My wife has helped in more ways than she will know, and always my rock in the good, and bad times. Without her support, I would not be where I am today, and I will always say that this is our PhD.

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Abstract

Fifteen patients with unilateral total knee arthroplasty (TKA) performed cycling at two workates (80 W and 100 W) and two walking conditions (preferred and fast speeds). Ten of these patients of TKA also participated in a short-term cycling intervention paired with visual augmented feedback of vertical pedal reaction forces for six sessions over two-three weeks. These ten patients of TKA participated in a 2nd post-training testing session. Study One compared the knee joint biomechanics for all fifteen participants during stationary cycling to ascertain if any biomechanical asymmetries may be present. The replaced limbs displayed significantly lower peak knee extension moment (KEM) and vertical pedal reaction (PRF) compared to non-replaced limbs during stationary cycling. Study Two examined the effect of the short-term cycling intervention on the knee joint biomechanics and biomechanical asymmetries during stationary cycling for the selected ten patients of TKA. The short-term cycling intervention had no significant effect for peak KEM or vertical PRF asymmetries during stationary cycling. Peak KEM asymmetries did decrease by 10% and 9.9% at 80 W and 100 W, respectively. Study Three examined the effect of the short-term cycling intervention on the knee joint biomechanics and biomechanical asymmetries during gait. Similarly, the short-term cycling intervention had no effect on peak KEM asymmetries and vertical ground reaction force (GRF) asymmetries during both walking condition. Study Four compared the estimated tibiofemoral joint forces during stationary cycling between the replaced and non-replaced limbs of the fifteen patients of TKA. The replaced limbs also had lower medical tibiofemoral contact force (MCF) compared to the non-replaced limbs during stationary cycling at 80 W. The non-replaced limb had greater peak MCF compared to the lateral tibiofemoral contact force (LCF). Unilateral TKA patients cycling with similar reductions of KEM in their replaced limbs. During cycling, there was no difference between MCF and LCF for the replaced limbs, potentially indicating a

successful operation to restore knee joint alignment. In summary, the use of a short-term cycling intervention with augmented feedback for six sessions were not significantly beneficial for addressing KEM asymmetries in both cycling and gait. However, the 10% reductions of peak KEM asymmetries may indicate some clinical benefits of this intervention. Future studies should examine similar interventions with an increased number of training sessions.

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CHAPTER I

INTRODUCTION

Background

One of the most common knee pathologies in an aging population is knee osteoarthritis (OA) (1-5). Knee OA is described as the progression of cartilage damage, along with the damage to underlying bone of the knee joint (1, 4-6). Knee OA follows a progression, with increasing amounts of damage to the point of necessary surgical interventions such as the total knee arthroplasty (TKA) (5). Studies have estimated a 673% increase in TKA procedures by the year 2030 (7). The goals of TKA procedures are to relieve pain and restore knee joint function (8-14). Knee joint functions after TKA are commonly quantified by an increase of joint range of motion (ROM) as well as restoration of gait biomechanics ranging from kinematics (joint angles) and kinetics (joint moments and ground reaction forces) (8, 9, 15-33).

Gait analysis has been used extensively in TKA patients to track and examine the effectiveness of the operation (9, 15, 16, 19, 34-36). During gait analysis, TKA patients are typically compared to (i) the replaced limb prior to surgery, (ii) the contralateral limb, (iii) and healthy controls with no clinical pathologies. The variables of interest during gait analysis are the vertical ground reaction force (GRF), peak knee flexion angles, knee flexion ROM, knee abduction angle, peak internal knee extension moment (KEM), and peak internal knee abduction moment (KAbM) (15, 20, 22, 24, 34, 36-42). The ways in which these critical variables differ prior to and following TKA gives insight into the capabilities during gait of TKA patients and potential underlying risks that may arise. TKA patients have been found to displayed decreased vertical GRF at both pre-surgery and post-surgery compared to their healthy limb (43-45). Additionally, the limb undergoing a TKA has significantly reduced vertical GRF when compared to healthy controls (46). However, some have demonstrated no differences in vertical GRF following TKA operation compared to contralateral limbs and healthy controls (47). Finally, there are even asymmetries between the replaced limb and non-replaced limb, with the replaced limb displaying significantly lower vertical GRFs (34, 48).

Other alterations during gait for TKA patients comes with a reduction of knee flexion ROM, typically termed “stiff knee gait” (20, 29, 30, 45, 49, 50). Patients demonstrated “stiff knee gait” walk with a more rigid leg that does not flex as much, which could lead to alterations in lower limb biomechanics. The alterations of sagittal plane knee kinematics have been found to persist in some TKA patients for as long as 12-months and potentially longer (20, 51). Peak knee adduction angles have been found to be reduced in the replaced limb following a TKA procedure (28, 51-53). Both peak knee flexion and adduction angles have been linked to loading of the tibiofemoral joint through joint moment measurements.

KEM is a good surrogate measure for the overall loading of the tibiofemoral joint during gait (54-57) while KAbM has been correlated directly to the loading of the medial tibiofemoral compartment (34, 36, 58, 59). While KAbM and KEM are used as surrogate measures for the medial compartment loading, they are popular measurements for patients with medial compartment knee OA and TKA patients. Several studies have found that people undergoing TKA have a lower peak KEM compared to both their contralateral limb and to those in healthy controls (45, 47, 60, 61). In addition to decreases in peak KEM, some studies have found significant decreases in peak KAbM following TKA operations, however others have found no differences (45, 47, 61, 62). Differences in knee joint loading variables such as KEM and KAbM present a concern and challenge for future surgical interventions. One major concern about loading asymmetries is the need for subsequent TKA to the healthy contralateral limb (35, 50, 63-65).

The rehabilitation following TKA procedures are crucial for the success and the recovery for these patients (66-75). As of now, there is no widely recognized set standards of rehabilitation for TKA patients. The primary consideration given is to participate in activities that result in lower knee joint loading and impact. The activities recommended range from swimming to one of the most common, cycling, as well as muscular strengthening (23, 76-81). Since cycling has been shown to result in a lower peak KEM and KAbM compared to gait, it has been recommended to be safe for TKA rehabilitation and is commonly prescribed (56, 82, 83). Cycling after TKA has been found to not be significantly better than rehabilitation not including stationary cycling (83). This negative result could be attributed to the stage of rehabilitation being very early post operation. Additionally, the study relied on self-reported measures without objective physical function assessments. While this previous study found no significant benefit in a TKA population, stationary cycling could still prove to be beneficial in many other ways. Stationary cycling is touted to be more beneficial by aiding in cardiovascular health, weight management, and some strengthening of the lower limb. However, there is no direct research done as of yet examining the efficacy of cycling as a means of rehabilitation in a TKA population and to examine the training effect of cycling on TKA patients. Additionally, when prescribing rehabilitation, there is a need for a progressive plan for increasing demand to allow for beneficial adaptations. During cycling, this is effectively done by increasing the intensity via workrate (56). Increases in workrate saw subsequent increases in KEM, which could prove to be beneficial post-TKA who demonstrate weakened quadriceps post-operation.

The use of KAbM and KEM are useful in estimating the amount of compressive force in the tibiofemoral joint, as this measure is extremely invasive to measure directly. Instrumented knee prosthesis have been used to directly measure tibiofemoral joint forces during cycling (82).

The main limitation of using net joint moments to estimate tibiofemoral loading is the absence of muscle forces acting upon the joint (84, 85). Musculoskeletal modeling and simulation have rose in prominence, as it utilizes biomechanical experimental data, and estimates joint contact forces and muscle forces. Musculoskeletal modeling for cycling has thus far been limited and has not extensively examined tibiofemoral joint loading or compressive forces (86-89). In a recent dissertation, static optimization along with joint reaction analysis has been used to estimate the loading of the tibiofemoral joint during cycling in a knee OA group (90). Further research is needed to examine the tibiofemoral joint loading during stationary cycling in rehabilitation settings, especially in the TKA population, which would fill a current gap in the literature.

Finally, in pathological populations such as TKA and knee OA, gait retraining and augmented feedback have been utilized to modify lower extremity biomechanics to elicit a more advantageous outcome (23, 32, 66, 91-95). The easiest way to classify most gait retraining avenues is a particular type of feedback called augmented feedback. Augmented feedback is an extrinsic feedback given to performers related to their performance to enhance feedback results (96-98). Examples of gait modifications that have been used include increasing step width, foot progression angle, vertical GRF, anterior GRF, and tibial accelerations (34, 38, 49, 99-102). Gait modifications have been beneficial in modulating internal KEM, leading to a reduction in knee joint loading, when given visual feedback to make their vertical GRF equal, and ± 10% in ACL reconstructed individuals (100). Another study used visual feedback to reduce the vertical GRF impulse by 5%, 10%, and 15% using augmented feedback of vertical GRF impulse during a countermovement jump in healthy individuals (103). The use of augmented feedback on kinetic variables has been found to be beneficial when addressing the loading variables in the biomechanics literature. To our knowledge, no studies have been done using these methods in a TKA population during stationary cycling.

Statement of Problem and Research Hypothesis

Biomechanical deficits and asymmetries following a TKA operation have been found persisting as long as 12 months and beyond during gait. KEM has been found to be reduced in the replaced limb compared to their healthy contralateral limb and to healthy controls (22, 34, 36, 47, 60). There have been some studies that have used gait interventions to address these asymmetries with little success. Furthermore, rehabilitation recommendations for TKA patients always include the use of stationary cycling, as it will result in reduced joint loading compared to other weight bearing activities (104-109). These recommendations appear to be made with little direct objective findings as there is only one study to date using cycling as an intervention for TKA patients (83). However, this study was conducted only on subjective function measures such as the WOMAC and self-reported function. These limitations to this cycling intervention study leave a wide gap in the literature on the efficacy of cycling interventions post TKA operation. There is no base of knowledge on the biomechanical efficacy of cycling as training in TKA patients. Additionally, there is no current research examining the biomechanical asymmetries in cycling in a TKA population. Some asymmetries have been found in patients with knee OA, which could persist or increase following surgical intervention similarly found in gait biomechanics. Finally, there have been no studies examining an intervention of stationary cycling combined with augmented feedback of vertical pedal reaction forces to address cycling and potential transfer to walking asymmetry reductions in TKA biomechanics.

Therefore, the purposes and hypotheses of the studies were as follows:

Study One: The primary purpose of this study was to examine the knee joint biomechanics of unilateral TKA patients during stationary cycling at two different workrates.

We hypothesized that:

1) Peak vertical PRF and KEM would be significantly lower in replaced limbs compared to non-replaced limbs.
2) Peak vertical PRF and KEM would be significantly greater at 100 W compared to 80 W.
3) There would be no significant interaction of limb and workrate on peak vertical PRF, peak KEM, peak hip extension moment, and peak ankle plantarflexion moment.

Study Two: The purpose of this study was to examine the effects of a short-term cycling training program with augmented feedback of vertical PRF on asymmetries of KEM and biomechanical inter-limb differences in patients of TKA.

We hypothesized that:

1) The inter-limb asymmetries for peak vertical PRF and KEM would decrease from pre- to post-training.

Study Three: The primary purpose of this study was to examine the transfer effects of a cycling training program with augmented feedback of vertical PRF on knee joint asymmetries and biomechanics in unilateral TKA patients in level walking at two different walking speeds.

We hypothesized that:

1) The asymmetries for peak vertical GRF and KEM would be reduced in gait at both preferred and fast speeds following the cycling intervention.

Study Four: The primary purpose of this study was to examine the tibiofemoral contact forces (total, medial compartment, and lateral compartment) and knee extensor and flexor muscle forces in TKA patients during stationary cycling.

We hypothesized that:

1) the replaced limb would have lower peak total tibiofemoral compressive force (TCF), tibiofemoral medial compartment compressive force (MCF), tibiofemoral lateral compartment compressive force (LCF), knee extensor force, and knee flexor force compared to the non-replaced limb.
2) peak MCF would be higher than peak LCF in both the replaced and non-replaced limbs .

Significance of the Study

Currently, most recommendations for rehabilitation post-operation of a TKA suggest using cycling as an exercise modality compared to jogging (76, 110). However, there is very little to no research in the literature on cycling biomechanics in the TKA population, or the use of cycling as a rehabilitation modality following TKA. There is a clear gap in the literature to give support to prescribing cycling as an effective and safe exercise for those undergoing TKA. Additionally, following TKA procedures patients display marked inter-limb asymmetries between their replaced and non-replaced limbs in gait. Information gathered from our studies would give insight into the cycling biomechanics of a TKA population as well as evidence for the use of cycling as a form of rehabilitation to aid in reducing inter-limb asymmetries.

Delimitations & Limitations

Delimitations

The inclusion criteria for TKA participant recruitment were as follows:

- Men and women between the ages of 50 and 80
- Total knee arthroplasty between 6 to 18 months ago

The exclusion criteria for TKA participant recruitment were as follows:

- Initial VAS pain scores greater than 5 in the replaced knee
- Diagnosed osteoarthritis of the ankles, hips, or contralateral knee that impacted walking
- Any other lower extremity joint replacement (other than single replaced knee)
- Systemic inflammatory arthritis (rheumatoid arthritis, psoriatic arthritis) as reported by the patient that impacts daily living
- BMI greater than 38 kg/m^2
- Neurologic disease (e.g. Parkinson's disease, stroke) as reported by the patient
- Any major lower extremity injuries/surgeries in the past 6 months
- 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 (PARQ)
- Cycling exercise of more than two times per week

Due to the intervention and augmented feedback being in the same group, it is unclear if any results were directly related to either the cycling intervention or based on the augmented feedback.

Limitations

The follow limitations were present in the current study:

- Testing data was collected in a laboratory setting, which could impact some results of the study.

- Kinematic data (and thusly inverse dynamics) are subject to error in the placement of reflective markers.
- Kinematic data may be subject to motion artifacts of the reflective markers during movement.
- Participants were required to self-report their activities in both groups to ensure they did not exercise in addition to the study requirements.
- An intervention period of 3 weeks may not be adequate for long-term adaptations.
- No control group was evaluated to compare the intervention group against.

CHAPTER II

LITERATURE REVIEW

Introduction

The purpose of the first study was to examine the knee joint biomechanical asymmetries between replaced and non-replaced limbs of unilateral TKA patients during stationary cycling at two workrates. The purpose of the second study was to examine the effects of a cycling training program with augmented feedback of pedal reaction force on knee joint biomechanical asymmetries in unilateral TKA patients during stationary cycling. The purpose of the third study was to examine the transfer effects of a cycling training program with augmented feedback on knee joint biomechanics and asymmetries in unilateral TKA patients in level walking. Finally, the purpose of the fourth study was to examine asymmetries of tibiofemoral joint compressive forces in unilateral TKA patients during stationary cycling at two workrates.

The purpose of this chapter is to summarize: 1) TKA pathology and epidemiology, 2) TKA rehabilitation guidelines, 3) TKA gait biomechanics, 4) healthy gait biomechanics, 5) bilateral asymmetries between TKA and healthy people, 6) cycling biomechanics, 7) augmented feedback, 8) how augmented feedback is used in biomechanics, 9) musculoskeletal modeling and simulation, and 10) cycling modeling and simulation.

Total Knee Arthroplasty (TKA)

Pathology

One of the most prevalent lower extremity diseases is knee osteoarthritis (OA). Knee OA is the progression of damage to the articular cartilage and underlying bone of the knee joint (1, 3-6, 111). The most common complaint from knee OA is joint pain caused from cartilage and bone damage (14, 112). There have even been differing pain patterns associated with risk factors for OA such as age, sex, BMI, and traumatic injury (14). Diffuse pain was most correlated to BMI while females were more likely to feel regional pain. Additionally, the role of genetic risk factors for knee OA have been found to have some impact on the risk of worsening knee pain (112). Degeneration of the articular cartilage of the knee typically is diagnoses in a range of 9 grades (5). The worst grade, 5b, is described as bone destruction that is equal to or greater than 5 mm. There is a positive correlation found between the cartilage degradation and the grade found on radiographs (5). Once knee OA has progressed and the damage is too great, a total knee arthroplasty (TKA) is done to restore function, relieve pain, and restore alignment to the replaced knee (8, 17, 113, 114)

Total knee arthroplasty is the surgical intervention to repair and reshape the knee joint due to damaged articular cartilage and its underlying bone (65). The distal end of the femur and proximal tibia are the surgical sites when performing a TKA procedure. Once the effected sections of bone have been removed, an artificial implant is then inserted onto these locations to create an artificial knee joint (115). These implants are designed with both a femoral and tibial component (65, 115). During the operation, the ligaments are either excised (discarded) or retained based on the TKA type and the state of these ligaments. The two major ligaments that are of interest during a TKA are the anterior cruciate ligament (ACL) and the posterior cruciate ligament (PCL) (116). A cruciate-retaining (CR) TKA procedure keeps the PCL intact, and thus its function of posterior stabilization of the joint is simulated (117-121). If the PCL is discarded, then a posterior-stabilizing TKA is performed, which includes a “cam” as a feature of the implant design to simulate the posterior stabilization of the PCL (39, 119, 122, 123). This posterior cam is a raised surface on the tibial tray of the implant, that will interact with the femoral component during flexion to act as the PCL would, preventing posterior translation of the tibia. If both the PCL and ACL are discarded, then a bi-cruciate stabilizing (BCS) procedure is done, which is similar to the PS, with an additional cam positioned on the anterior of the tibial implant to simulate the ACL (121, 124).

Epidemiology

Knee OA is one of the most prevalent forms of arthritis and is estimated to be rising in the coming years (7, 125). TKA procedures are predicted to be rising and are expected to grow by 673% by 2030 (7). TKA revisions are also expected to be rising by approximately 601% in the same time frame (7). The incidence has been steadily increasing for TKA procedures not only just in the United States. There has been an increase in the incidence of TKA procedures of 26.4 to 74.55 per 100,000 people between 1996 to 2010 in just Taiwan, with 154,533 total TKA procedures being done (126). From 2002 to 2005, 47,961 TKA procedures were performed in South Korea (127). This increase of incidence rates is hypothesized to be from the impact of an aging population, with older adults being more likely to undergo a TKA than their younger counterparts (126, 128). While age is one of the most related risk factors for TKA procedures, other factors such as genetics, traumatic injury to the knee, and weight are related to TKA (4, 112, 129-133).

A longitudinal study over 10 years examined the structural changes of the knee joint in people who had at least one parent with a TKA (130). When they were compared to people with no family history of TKA, those with family history had significantly higher cartilage deficit score (1.03 vs. 0.52), meniscal extrusion scores (0.28 vs. 0.10), and meniscal tear scores (0.40 vs. 0.10) (130). All these structural changes occurred in the medial tibiofemoral compartment and could put them at higher risk for knee OA and TKA. A similar study comparing the same groups found that those with family history of TKA were more likely to experience general knee pain using the WOMAC score (74 vs. 54%) (112).

Traumatic knee injuries can cause damage to the ligaments as well as the bone attached to them (134, 135). This damage leads to individuals being more predisposed to knee OA, and eventually TKA (131, 132, 136). This increased risk of OA will typically be accelerated, causing a TKA procedure at a much earlier age than typically found (131, 136). A retrospective study of military personal used 74 TKA procedures that all occurred under the age of 50 found the most common traumatic injury was an ACL rupture (n = 19) (132). The average age at time of injury was about 29.2 years with an average TKA procedure age of 44.3 years. Another study found that the average age for TKA following an ACL reconstruction (n = 122) was roughly 58 years old (137). It was found that these individuals will have a longer operation and some increased risks of TKA reoperation due to their previous injury and ACL reconstructions (131).

An individual's body weight is a risk factor for knee OA and TKA (64, 138-141). Increased body weight has been linked to a greater amount of vertical ground reaction force (GRF) that is related to loading of the lower extremity (99, 142, 143). While comparing step widths in obese compared to non-obese patients, obese people had on average 313.7 N greater vertical GRF during stair negotiations (99). Risk for knee OA is significantly greater when body mass index (BMI) is increased from <20 kg/m^2 to >36 kg/m^2 at a rate of 0.1 to 13.6 times more likely to have knee OA (139). This relative risk was found to be elevated if overweight or obese individuals have had previous knee injury (139). The risk factor for obesity also extends and impact younger populations. When analyzing a group of 18 to 59 year old participants, 52% of them have had a TKA with obesity being significantly linked when compared to a general population (141). Additionally, there was a greater likelihood of these participants who obtained a TKA (83%) to be obese compared to obtaining a total hip replacement (THR)(58%)(141). Patients who underwent a TKA also displayed a greater BMI and take fewer steps per day when compared to people with a THR as well as a general healthy population (144). Obesity and weight are a concern for managing risk factors for both knee OA and TKA, and is a key factor for prevention and rehabilitation for TKA (64, 138, 144).

Total Knee Arthroplasty Rehabilitation – Guidelines and Recommendations

Following a TKA procedures, the implementation of rehabilitation programs are an important step in healing (68, 76) Rehabilitation goals following a TKA will typically work to address the following concerns: return to function, increasing joint ROM close to pre-operative healthy ranges, pain management, increasing strength, and weight management. Rehabilitation protocols are designed to systematically address all of the previous concerns with the use of various modalities, training, and tests. The most common form of recommended rehabilitation comes in the form of regular exercise, first with direct supervision and then without (145). When choosing the exercise for TKA rehabilitation, it's critical to factor in the wear of the replacement as well as the joint loading being applied (145).

A major concern with performing exercises following a TKA is an increased wear of the implants over time. Damage and wear done to the polyethene components of the replacement are of particular concern (146). This damage is thought to be mainly controlled through modifiable risk factors such as exercise, exercise modalities, and dieting (145). Wear for joint replacements for their polyethylene components have been found to be a function of use (147). With wear being related more so to use than time, exercise recommendations aim to optimize the amount of load being applied to the replaced joint while maintaining an effective overall workload to be beneficial. Kuster et al. (148) examined the tibiofemoral compressive forces during both level and downhill walking. Level walking experienced a tibiofemoral compressive loading of 3.9 BWs whereas downhill walking reported a loading of 8 BWs (148). A concern for TKA designs is that based on this data and the replacement contact surface area (100 to 300 mm^2), the joint loading experienced during downhill walking will be too large and exceed yield points (148). Other studies have shown that greater peak external KEM were found in patients with knee OA and persisted 6 months post-surgery for a TKA (149). The increased loading during gait was thought to be linked to tibial loosening and increased wear on the implant, and therefore should be considered during rehabilitation (149).

Rehabilitation recommendations rely heavily on the joint loading based on the activities being performed (148). Variations in gait have been thoroughly explored with their relation to knee joint loading. Walking at different conditions such as level vs. downhill have previously been found to have differing knee joint loads (47, 148). Increasing the speed of gait (i.e. walking to jogging) have also been found to increase the knee joint loading (145). Following TKA, this increase in loading is the main consideration to reduce injury risk or potential wear on the implant. It is recommended to avoid high load and intensity exercise such as jogging or other activities that induce greater knee flexion (145). Tibiofemoral compressive loads during stationary cycling has been measured to be1.2 x BW, which is considerably lower than that of gait or other activities (108, 109). Recommendations for activities following a TKA are split into differing categories based on their relative joint loading in the following manner: recommended, allowed with experience (competitive or non-competitive sports), or not recommended (70). A consensus is that following a TKA, people should participate in low loading activities such as cycling, swimming, and low-impact aerobics. TKA patients should avoid high demand and high loading activities such as basketball, jogging, volleyball, or soccer (70, 145). Further comparison for endurance activities following TKA found cycling and power walking to be beneficial and below the yielding point for the implants (77). Most activities that involved going downhill or descent were not recommended (77). Surgeons as well as physical therapists should be involved with the rehabilitation planning and education to the patient.

Exercise is typically prescribed for patients to retain cardiorespiratory fitness, as well as weight management. Prior to requiring a TKA, obesity and weight play a role in the cartilage damage in knee OA (150). Inactivity leading to reduced cardiorespiratory fitness, increased weight gain, and muscle weakening are concerns for TKA patients (77, 151). Similar to the needs following a TKA, losing 5% BW in three months has been linked to a reduced risk of knee OA in older obese females (152). Huang et al. (71) also reported that weight reductions in knee OA rehabilitation are effective with reductions in pain scores using a visual analog scale (VAS) and should be considered for this rehabilitation (71). The main goal for these rehabilitation studies was to reduce the overall load being applied to the effected knee joint and prolong the damaged to the articular cartilage. Since wear is a function of use rather than time, if during exercise you reduce the amount of load being applied, the amount of wear will reduce (147). The compressive loading of the tibiofemoral joint is greatly impacted by the amount of one’s body weight. This is shown with the much higher compressive loading present in activities such as jogging, where one’s body weight is not supported and is accelerated, generating greater impact loads (77, 145, 148). The main goal of rehabilitation following TKA procedures focus mainly on limiting the amount of compressive loading of the impacted limb, primarily via modalities of exercise or through weight management (76, 80, 148).

One of the most commonly reported exercise modalities following TKA has been stationary cycling (77, 83, 148). The use of cycling has been selected due to the lower tibiofemoral loading when compared to activities such as walking or jogging (73, 77, 82, 109, 110). As previously mentioned, loads during cycling have been reported to be as low as 1.2 BW

(109). Accompanied with lower knee loading, with stationary cycling you are able to modify the bike to redistribute and modify lower extremity loading (i.e. saddle height or foot positions)(73). McLeod et al. (73) indicated the use of cycling as a pathway for quadriceps rehabilitation, with cycling propulsion being quadriceps dominant. Following TKA, there is a significant decrease in quadriceps strength indicated by quadriceps forces and maximal quadriceps torque measured during maximal voluntary contractions (MVICs) (26, 32, 45). Mizner et al. (32) reported that following TKA, patients had 64% lower maximal knee extension torque compared to healthy controls during a MVIC superimposed with a supramaximal electrical stimulation. Other studies have reported similar loses in quadriceps strength of about 60% post-operation (32). The use of cycling could be beneficial in terms of rehabilitation to strengthen the quadriceps muscles, regain muscle activity, while also generating lower loads at the knee and increasing cardiorespiratory fitness (77, 80, 153-155). One study has been done examining cycling rehabilitation starting at two weeks post TKA operation (83). The study found no significant benefit from a cycling rehabilitation protocol on TKA patients based on WOMAC scores. This study however did not include any objective testing of TKA patient function, such as gait analysis or functional testing like the timed up and go. More objective data is needed to examine the potential benefits of cycling following TKA.

Currently the recommendations for rehabilitation for TKA rely heavily on reducing knee joint loading through differing modalities (77, 145, 148). Some of the most recommended activities for rehabilitation include, walking, cycling, and other activities that are low in impact. Activities of higher impact and loading such as jogging, basketball, or volleyball should be avoided due to greater wear on the implants (148). Rehabilitation plans should focus on returning patients to function, maintaining or improving their cardiorespiratory fitness, increasing their quadriceps strength, as well as increasing balance control (25, 26, 32, 145). Very few studies have concluded on a recommended duration and frequency of exercise. One meta-analysis concluded that three-four times per week at 30-40 minutes of low impact aerobics would be beneficial following TKA (145).

Total Knee Arthroplasty Gait Biomechanics

A hallmark of TKA rehabilitation research is done examining the impact of the procedure on the patient's gait (8, 9, 15, 16, 18, 33, 42, 45, 51, 54, 60, 156, 157). Gait analysis using 3-dimensional motion capture and force platforms are typically used to quantify the biomechanical adaptations during gait. The biomechanical variables of interest during gait include the GRF, joint kinematics, and joint kinetics. Additionally, it is important to understand the bilateral asymmetries that are found during gait in TKA patients that could impact their rehabilitation (27, 61, 158).

Ground Reaction Force

Ground reaction force is an important measure for TKA as it relates to the amount of loading occurring during stance phase (44, 99, 148, 159). Ground reaction forces have been reported to decrease in the limb undergoing a TKA (43-45). Along with a decreased vertical GRF post TKA, the replaced limb displayed significantly lower GRF compared to the healthy contralateral limb. Participant two-years post-operation still displayed significantly lower vertical GRF on the replaced limb (1.06 BW) compared to their healthy contralateral limb (1.1 BW)(48). Similar results were reported in two difference groups, with and without lower back pain with a knee OA prior to surgery (43). The effected limb with OA had significantly decreased vertical GRFs regardless of back pain, with no significant group effects being present (43). Differences in GRFs have also been reported when comparing TKA patients to healthy controls. Females one-month post-operation had observable decreased vertical GRF compared to healthy controls (46). The vertical and posterior ground reaction forces for the TKA participants were significantly smaller compared to their control counterparts (46). The asymmetries found between the replaced limb and healthy contralateral limb have been reported not only in level walking, but during stair ascent (31). Push off vertical GRF for the replaced limb vs. non-replaced was 1.14 x BW and 1.19 x BW but did not reach statistical significance (31). Ground reaction forces have been reported to be decreased not only pre-operation, but also post-operation during gait (31, 46, 48). A primary concern appears to be that the operated limb experiences lower GRFs compared to the contralateral limb, increasing contralateral limb loading and potentially leading to a primary TKA on the contralateral limb (50, 160).

Kinematics

Sagittal Plane

Knee sagittal plane kinematics are a critical interest following a TKA operation as a sign of a knee joint function and is commonly used by clinicians (10, 161-163). Decreasing knee flexion ROM has previously been reported during passive and active ROM testing (164). Similar decreases have been previously reported during gait following TKA (28, 30). This stiff knee gait could potentially have impacts on the asymmetrical loading patterns between the replaced and contralateral limbs (20, 30, 50, 165).

Renaud et al. (166) examined two different TKA implants (Triathlon and Nexgen) compared to asymptomatic control knees during gait. The asymptomatic knees displayed greater total knee flexion ROM (55.4° vs. 47.1° and 48.2°, respectively) as well as knee flexion ROM during stance phase (17.4° vs. 15.3° and 13.6°, respectively) (166). The differences in ROM was mostly due to changes in peak knee flexion angles between the groups. The asymptomatic knees had greater peak knee flexion angles compared to both the Triathlon and Nexgen TKA groups, 57.6° vs. 50.4° and 52.8°, respectively (166). These deficits in knee flexion kinematics are present two months following a TKA procedure (61). Patients who underwent a TKA had their peak knee flexion angles reduced from 44.0° ± 10.8° to 35.0° ± 8.7° at two months (61). Their peak knee flexion angle was significantly different from the ones found in the healthy controls, 47.0° ± 5.0° (61). Not only were differences found in level walking, but also during stair ascent. The two months post operation knee flexion angle were significantly lower than at pre-operation and compared to healthy controls (54.0° ± 7.7° vs. 59.0° ± 6.4° and 62.0° ± 4.0°, respectively) (61).

Peak knee flexion angles have been shown to be asymmetric between the operated and non-operated limb not only at 3 months, but 12 months post-operation (45). Yoshida et al. (45) found that at both 3 and 12 months, patients had less peak knee flexion as well as resulting knee ROM. The operated limb however did not statically differ from matched healthy controls at 12 months (45). While most studies find that peak knee flexion increases following TKA, others have shown that peak knee flexion angles and dynamic ROM during gait don't change two months following TKA procedure (60). Levinger et al. (2013) found that 12 months post-operation values for both peak knee flexion and ROM did not improve from pre-operation knee OA values. It was also found that while pre-operation knee flexion ROM did differ from healthy controls, the post-operation value did not significantly differ (60). Joint ROM may not directly significantly improve from pre-operation values but could improve when compared to healthy controls.

Knee flexion ROM following TKA has been reported to improve compared to pre-operative levels ($p = 0.025$) but still was smaller when compared to healthy controls (167). The individuals who underwent TKA were also found to have a slower gait speed compared to their healthy controls. The alterations of the knee kinematics may have had an impact on the knee kinetics (167). The differences between TKA patients and healthy controls may persist for much longer post-operations (168). Ullrich et al. (168) compared females who underwent a TKA 10 years ago to healthy female controls. Females who had a TKA exhibited deficits compared to the healthy controls, with TKA participants exhibiting decreased peak knee flexion angles, indicating that some alterations to gait kinematics may persist long-term (168). A two-year longitudinal study followed patients post TKA procedure (20). The TKA patients displayed decreased knee flexion ROM compared to healthy controls at 6, 12, and 24 months post-operation: 48.9°, 49.7°, and 48.8° vs. 57.1°, respectively (20). Decreased knee flexion ROM was attributed to a decreased knee flexion compared to controls at the following gait parameters: loading response, toe off, and swing phase. Peak knee flexion angle at loading response was 10.4°, 12.1°, and 11.5° respectively compared to 16.7° for the healthy controls (20). Peak knee flexion at toe off was reported as 33.7°, 36.1°, and 33.9° compared to the healthy controls at 38.2° (20). Benedetti et al. (20) described the kinematic adaptations similar to that of a "stiff knee gait pattern", that is a concern often found following a TKA (50). Zeni et al. (50) found that individuals who exhibit a stiff knee gait pattern are at a higher risk of needing a contralateral TKA.

Overall, patients undergoing a TKA display a decrease in both peak knee flexion and knee sagittal plane ROM (29, 45, 50). These individuals display both a bilateral deficit in these measures, compared to their healthy contralateral limb, as well as to healthy controls used for comparison (45, 60). Decreases in knee motion in the sagittal plane has previously linked to increases in pain, co-contraction of muscles, or quadriceps avoidance gait. The adaptations following a TKA not only impact just the kinematics in the sagittal plane, but play a role in the loading of the joint, leading to a concern of further operations such as TKA revision or TKA of the contralateral limb (20, 50, 165). Excessive knee flexion has been reported linked to the overall compressive loading of the knee joint (109, 169). While there is clinical significance to static knee ROM following surgery, more attention is needed to address the long-term deficits found in dynamic knee ROM during gait.

Frontal Plane

Knee frontal plane kinematics gives some insight to the loading of the knee joint in the frontal plane (36). While during gait, most of the knee motion is directed in the sagittal plane, there is substantial motion in the frontal plane (15, 34, 52). During gait analysis comparing both Triathlon and Nexgen TKA groups to asymptomatic knees, peak stance frontal plane angle was significantly different (166). The asymptomatic knees had a peak knee adduction angle of 3.4° compared to 4.9° (Triathlon) and 6.6° (Nexgen) (166). There were no significant differences in either total or stance phase range of motions found in the frontal plane kinematics. Alnahdi et al. (15) compared the frontal plane knee angles of people who underwent a unilateral TKA compared to their contralateral limbs. The non-operated limb had a significantly greater peak knee adduction angles compared to the operated limb at both 6 and 12 months, -0.01° vs. 2.96° and 0.9° vs. 2.79°, respectively (15). Healthy controls were used for further comparison with both of their limbs displaying greater knee adduction angles during stance phase, 2.79° and 2.33° (15).

The deficits found in peak knee adduction angle comparing operated and non-operated limb could be due to a reduced joint angle following TKA (28). Naili et al. (28) used gait analysis to compare healthy controls to two groups of TKA patients, those who had a good or poor outcome, ranked using changes in knee-related Quality of Life surveys. The good outcome group (10.5° ± 6.1° to 5.7° ± 4.8°) and poor outcome group (8.5° ± 4.3° to 5.6° ± 2.4°) displayed significant decreases in their peak knee adduction angles following the TKA procedure (28). The control group displayed similar knee adduction angles compared to other healthy individuals that had knee adduction angles of 3.2° ± 3.3° (28). Significant decreases in peak knee adduction angles following TKA have been reported 6 months post-operation (51). Patients who underwent a PS TKA procedure displayed significantly lower peak knee adduction angles at 6 months (3.6° ± 5.8°) compare to their pre-operative values of 9.7° ± 6.5° (51). After 12 months, however, peak knee adduction angles increased and were not significantly different from their pre-operative levels, indicating a trend towards their pre-operative gait. Decreases in peak knee adduction angles are reported as early as 6 weeks post-operation (52). Debbi et al. (52) compared patients pre- and post-operation and found a relative decrease of 71% in the knee adduction angle. Peak knee adduction angle post-operation was 1.20° ± 4.94° compared to pre-operation values of 4.22° ± 8.50° (52).

Reductions in knee adduction angles were found previously when comparing two different TKA types, kinematically aligned (KA-TKA) and mechanically aligned (MA-TKA) (53). The MA-TKA is more commonly performed and gets its namesake from the femoral and tibial cuts being perpendicular to the mechanical axis of the femur and tibia, respectively (170). In general, the knee joint produces a knee abduction angle of approximately 4-5° and aims to distribute load evenly. The KA-TKA aims to restore an alignment that more closely resembles of pre-operation values. The femoral component has its distal and posterior joint line aligned in accordance with the femoral transverse axis (170). Niki et al. (53) reported reductions in peak knee adduction angles for the KA-TKA group of -5.9° comparing the pre- to post-operation values. The MA-TKA groups displayed a similar decrease in peak knee adduction angles of -6.7° following their operation. While some differences in knee adduction are clear, some studies also find no differences in peak knee adduction angles for TKA patients (171). Milner et al. (171) found no significant differences for operated, non-operated, and control limbs for peak knee adduction angles: 1.8° ± 3.6°, 4.3° ± 4.3°, and 2.4° ± 3.7°, respectively. It was inferred that peak knee adduction angle may not always be altered due to TKA that has been previously reported (15, 52, 171).

A main adaptation to gait biomechanics following a TKA procedure is a reduction in peak knee adduction angle in the operated limb (15, 28, 51, 53, 172). Knee adduction angles have been correlated to knee frontal plane kinetics, leading to an impact on the medial compartment loading of the knee joint (36, 52). There are however some studies that have concluded that no alterations in peak knee adduction angles occur following TKA procedures (171). Peak knee adductions angles should remain a critical variable to examine following TKA or rehabilitation due to the association to medial compartment loading.

Kinetics

Sagittal Plane

Knee extension moment (KEM) is a measure of the angular force in the sagittal plane that is commonly used to evaluate overall knee joint loading. It has been commonly used as an indication of overall joint loading during gait biomechanics (24, 54, 56, 57). This loading variable is of a high interest following TKA due to: increased loading to the implant causing wear, asymmetries, and quadriceps avoidance due to surgery and pain (20, 54, 148, 167). Adaptations during gait and locomotion following a TKA can be seen when examining modifications to the operated limb compared to pre-operative, contralateral limbs, or healthy controls (9, 15, 16, 20, 118).

Ouellet et al. (61) compared patients undergoing TKA at pre-operation and at 2 months to matched healthy controls. Knee extension moment was significantly reduced at the 2 months testing compared to the healthy controls, 0.13 Nm/kg to 0.44 Nm/kg , respectively ($p = 0.0003$) (61). Knee extension moments were also found to be significantly reduced following the procedure by.20 Nm/kg (61). Knee extensor moment asymmetries have been evident even 12 months and beyond following a TKA (45). Yoshida et al. (45) reported that the operated limb at 12 months had a KEM of 19.9 ± 15.5 Nm compared to the non-operated limb at 35.6 ± 18.4 Nm. The same patients however did not display this decreased KEM in the operated limb at 3 months post-operation (28.2 ± 14.8 Nm) compared to immediately post-operation (28.4 ± 10.4 Nm)(45). Lower peak KEM (quadriceps moments in the current study), have been found across all knee joint angles during gait when normalized to body mass, meaning there was lower peak KEM across the entire joint ROM (168). The individuals in this study were females who were 10 years post-operation from a TKA.

Internal KEM have been reported in TKA patients to be lower than those found in their contralateral limbs (22, 45). Internal KEM have also been found to be lower in TKA patients compared to healthy controls matched for age, mass, and sex (61). There is a wide range of time at which KEMs appear to be impacted from TKA, ranging from 2 weeks to 12 months and even 10 years post-operation.

Frontal Plane

Internal KAbM has been associated with the medial compartment loading of the tibiofemoral joint (34, 36, 99). This increased loading of the medial compartment is of interest for TKA, as to modulate forces and wear on either the implant or the healthy contralateral limb (15, 34, 36). Increases in KAbM has previously been found in patients with medial compartment knee OA, which in turns propagates having a TKA operation.

The first peak KAbM that occurs during loading-response has been reported to decrease in the operated limb following TKA compared to the healthy contralateral limb and to healthy controls (51). Orishimo et al. (51) reported a 15% decrease in the first peak KAbM 6 weeks following TKA. Peak KAbM was more similar to pre-operation levels when measured one-year post surgery (-5%). The second peak knee abduction moment during propulsion was 26% smaller at 6 months and 22% at 1 year (51). Similar trends in reduced peak KAbMs were seen spanning a 1-year period in 15 TKA patients (173). Shimada et al. (173) reported significant decreases in KAbM of the replaced limb at 3 weeks, 3 months, and 6 months following a TKA. Decreases in the peak KAbM were -0.24 Nm/kg, -0.21 Nm/kg, and -0.19 Nm/kg, respectively with respect to baseline pre-operation (173). Peak KAbM were not significantly different from baseline (0.80 ± 0.25 Nm/kg) when tested 1-year post-operation (0.67 ± 0.14 Nm/kg). (173).

Mandeville et al. (174) examined TKA patients at pre-operation and at 6 months following a TKA. Peak KAbM was significantly lower in level walking at 6 months (3.01 ± 0.30 Nm/BW*m) compared to pre-operation (4.07 ± 0.38 Nm/BW*m)(174). Additionally, decreases in peak KAbM has been found across multiple gait speeds, preferred and fast speed (41). McClelland et al. (41) reported that the TKA group had a peak knee KAbM of 2.91 ± 0.66 Nm/BW compared to a control of 3.59 ± 0.68 Nm/BW. Peak KAbM of 3.56 ± 0.98 Nm/BW were found once gait seed was increased.

While some studies have found significant differences in KAbM following TKA operations, there have been other research suggesting no significant differences (45, 47, 61).

Recently, Wen et al. (47) found no significant differences in loading-response peak KAbM when comparing TKA patients to healthy controls. Yoshida et al. (45) found that KAbM, along with other moments were similar between the operated and non-operated limbs at three and 12 months post-operation. Additionally, Milner et al. (171) found that the first peak KAbM during gait was similar between knees that underwent TKA and those found in heathy controls.

Peak KAbM is used as a surrogate measure of the medial compartment loading of the knee joint (36). One should also use peak KEM in addition to peak KAbM, as this increases the correlation and improves the estimate of loading to the medial tibiofemoral compartment (58). Coincidently, the most common form of knee OA is found in the medial compartment of the tibiofemoral joint. Following a TKA, peak KAbM has been reported to be significantly deceased at 3 weeks, 3 months, and even 6 months post-operation, indicating a modification in joint loading (45, 61, 174). There are some contrasting findings, with others suggesting no change in peak KAbM following TKA operations. Alterations to the peak KAbM following a TKA is a critical variable to relate knee joint loading in both the replaced and non-replaced limb.

Healthy Gait Biomechanics

Ground Reaction Force

Vertical GRF is a variable used to infer the loading of the entire body during gait (175). In a study examining different shoes during gait on healthy individuals, peak vertical ground reaction forces were reported as 1.08 ± 0.04 BW in normal running shoes during loading (176). Zhang et al. 2013 reported similar vertical ground reaction forces during push-off, 1.09 ± 0.05 BW. Another study examined the effect of short leg walking boots on ground reaction forces reported peak vertical GRFs ranged from 10.27 ± 0.72 N/kg to 10.77 ± 0.59 N/kg (177).

Toda et al. (2015) reported similar results comparing gait GRF based on participant sex and age. The elderly males and females displayed peak forces of 10.57 ± 0.94 N/kg and 10.71 ± 1.06 N/kg, respectively (159). There were no significant differences in the vertical peak force during loading response between the elderly and younger groups. There was a significant effect of age group in both males and females on walking speed. The older males walked on average 0.06 m/s slower while older females walked 0.14 m/s slower than their younger counterparts (159). Bennett et al. (34) reported peak vertical ground reaction forces ranging from 1.18 ± 0.09 to 1.20 ± 0.10 BW. Following an intervention of increased toe-in angle along with increased step width, vertical GRFs increased to a maximum of 1.27 ± 0.14 BW (34)

While there was no apparent difference in the previous study in peak vertical GRF with differing speeds, others have noted a significant effect of walking speed on the peak forces and loading rate during gait (178). Nilsson et al. (178) compared walking (1.0 – 3.0 m/s) and running (1.5 – 6.0 m/s) on the GRFs experienced by healthy males. Vertical GRF increased from 1.0 – 1.5 BW in walking to approximately 2.0 – 2.9 BW during running (178). Additionally, both anteroposterior and mediolateral GRFs increased with an increase of speed, twice as great and 2-4 times greater, respectively. Adjustments to step length were reported to have very little impact on the peak vertical GRFs during walking (179).

Kinematics

Sagittal Plane

Knee flexion, knee extension and knee range of motion (ROM) play a key role in the healthy function of a joint and to joint loading during gait (175). Average knee flexion ROM during stance phase has been found to be 46.7° ± 4.4° in a healthy younger population (176). Zhang et al. (176) reported that the knee extension angle at initial contact with the ground at -

5.2° ± 3.4°. Kerrigan et al. (180) compared the level walking kinematics of both elderly and young populations. Peak knee flexion during loading response was 19.2° ± 5.6° for the young population and 16.3 ° ± 6.0° for the elderly group, both at a comfortable pace. When walking cadence was increased, the elderly group displayed a peak knee flexion at loading of 21.3° ± 6.1° (180). Knee flexion ROM for the young group was 60.0° ± 4.5°, whereas the elderly group reported ROM of 57.9° ± 4.6° and 60.1° ± 4.7° at a comfortable and fast pace respectively (180). Knee kinematics in both elderly and younger people did not appear to be significantly different. Similar knee extension angles were found in older and younger individuals during level walking (181). Elderly participants had a peak knee extension angle of -8.3° ± 5.9° while younger people had knee extension angles of -4.4° ± 5.6, which were not significantly different. Maximum knee flexion during stance phase were -21.3° ± 5.5° and -26.3° ± 4.7°, respectively (181). Sagittal plane ROM and peak knee flexion during early stance phase were reproduced (182).

Knee sagittal plane ROM in healthy individuals typically lays between 40° to 60°, for both young and older people. Knee extension angle at initial contact has been indicated to be approximately -3° to -5°. Peak knee flexion angles during early stance (load response) range from 19° to 26°. The kinematics of the knee joint during gait appear to not be significantly different when comparing older to younger healthy populations.

Frontal Plane

The frontal plane knee angles during gait have been previously linked to the frontal plane moments and joint loading (175). Yu et al. (175) examined level walking for frontal plane kinematics. Peak knee adduction angles that were associated with the peak abduction moments were 3.6° ± 2.1°, 6.7° ± 5.2°, and 6.5° ± 4.2 (175). A regression analysis showed that the corresponding joint angle to joint moment were significantly increased in the stair ascent compared to level walking. Individuals with a neutral alignment displayed a peak knee adduction angle of 2.4° ± 2.7° and peak abduction angle of -2.1° ± 2.5° (34). Bennett et al. (34) reported knee frontal plane ROM in healthy and neutrally aligned individuals of -4.5° ± 1.8°.

Similar knee frontal plane angles during stance phase were reported in a gait comparison of healthy controls compare to TKA patients (28). Naili et al. (28) found peak varus angles during stance phase of gait of 3.2° ± 3.3°. Peak knee abduction angles of the knee averaging 2° - 4° appears to be very common in a healthy population. Alnahdi et al. (15) reported similar frontal plane knee angles in healthy populations during gait. Peak knee adduction angles of 2.4° ± 3.7° were reported in another gait analysis that utilized heathy controls (171).

Knee frontal plane kinematics appear to be very small in magnitude during the stance phase of gait in healthy populations (34, 171). Peak knee adduction angles have been reported to range from 2.4° to roughly 6.7°, with the maximum of the range occurring during stair ascent (175). Frontal plane ROM of the knee joint appears to be relatively small, with most studies agreeing on a range from 4° to about 5° (34).

Kinetics

Sagittal Plane

Knee extension moment (KEM) during load response has been measured to be approximately 0.53 ± 0.13 Nm/kg for healthy individuals (176) The peak knee extension moment during late stance near push off was reported as 0.40 ± 0.006 Nm/kg (176). Another study reported smaller increases in KEM during early stance in younger males, 0.71 ± 0.20 Nm/kg, and younger females, 0.76 ± 0.27 Nm/kg compared to older males (0.58± 0.29 Nm/kg) and females (0.74 ± 0.26 Nm/kg) aged 65 years or older (159). Toda et al. (159) also examined the peak KEM in elderly males and females, which were 0.58 ± 0.29 Nm/kg and 0.74 ± 0.26

Nm/kg, respectively. However, no significant effects of gender or age were found in the peak knee extensor moment during early stance.

Another study reported peak KEMs for both young and older populations (180). The younger group had peak KEMs of 0.41 ± 0.13 Nm/kgm (180). Kerrigan et al. (180) found that elderly individuals had knee extensor moments of 0.27 ± 0.11 Nm/kgm at their comfortable walking speed and 0.46 ± 0.18 Nm/kgm when their walking speed was increased. There were significant differences between both groups at their own comfortable pace as well as when comparing the elderly group between their comfortable walking speed to a faster speed. There was a significant decrease in knee extension moment in the older population, with the KEM significantly increasing with gait speed.

The first peak KEM, found during early stance phase, is associated with the loading response of the lower extremity. During gait in a healthy population, the 1st peak KEM has been found to range from 0.40 to 0.75 Nm/kg, with some variation on the age of the individual. Increasing walking speed has been found to increase the KEM. Increasing walking speed increases the joint moments to accommodate the increase demand. This also could be linked to increases in GRFs also found with increases in speed.

Frontal Plane

Zhang et al. (176) reported loading-response peak internal knee abduction moments (KAbMs) in healthy individuals on average of -0.41 ± 0.11 Nm/kg during gait. Alnahdi et al. (15) previously reported similar KAbMs in healthy controls during gait. While examining both limbs, peak frontal plane moments were -0.41 ± 0.13 and -0.38 ± 0.13, which were reported as external knee adduction moments (15). Bennett et el., (34) reported peak KAbM in healthy individuals of 0.48 ± 0.12 Nm/kg during early stance. The second KAbM was lower at 0.37 ±

0.11 Nm/kg (34). Naili et al. (28)found slightly higher peak frontal plane moments during gait in healthy controls averaging 0.60 ± 0.10 Nm/kg.

Peak KAbM has been compared during level walking(175). The mean peak KAbM reported for level walking, stair ascent, and stair descent were 3.27 ± 0.73, 3.12 ± 1.10, and 2.81 ± 1.00 Nm/BW*m respectively. Mandeville et al. (174) reported peak KAbM of 2.70 ± 0.35 Nm/BW*m and 3.07 ± 0.30 Nm/BW*m at two different testing points for healthy controls.

Peak KAbM appear to be in a range of 0.37 Nm/kg to 0.60 Nm/kg (34, 177). The reported ranges when in terms of body weight multiplied by body height (m) appears to be within a range of 2.70 to 3.27 BW*BH (174). Peak KAbM is observed to be lower than the sagittal plane knee moments during stance phase of gait. However, with the link between frontal plane moments and the medial compartment loading of the knee joint, small alterations in this joint moment could be of significance (36). Additionally, adding the peak knee moment to this correlation increases its strength, lending support that both peak KAbM and peak KEM should be used to estimate medial tibiofemoral compartment loading (58).

Bilateral Asymmetries (TKA vs. Healthy)

Gait asymmetries and deficits in TKA patients could potentially explain how function of the knee joint is impaired following the operation. The deficits also potentially put the patients at increased risk for increased wear on their implant or a TKA on their non-replaced healthy limb (20, 50). There are reported asymmetries in vertical GRFs, knee flexion angles, knee flexion ROM, knee extensor moment, and peak KAbM (15, 28, 36, 52, 157, 174). In the case of the TKA replaced limb; all the reported variables have been found to be decreased compared to healthy controls. A common theory for the deficits found compared to healthy individuals is a reduction in gait speed due to quadriceps avoidance, weakness, or pain levels (20, 45, 61, 167).

The main deficit appears when comparing the replaced limb to the healthy contralateral limb, where asymmetrical patterns could last as long as 12 months following TKA. Healthy individuals have been shown to not have a noticeable asymmetry in their gait kinematics or kinetics.

Cycling Biomechanics

Stationary cycling has quickly become a mode of exercise for rehabilitation for various procedures and illness. Cycling is a preferred mode of exercise due to the pedal reaction forces (PRF) and joint moments experience by the body during movement when compared to walking or running (105, 183-185). During cycling, most of the patient's body weight is supported via the saddle (seat) and the handlebars. The reduced impact force while having their body supported is also split between two pedals, while one leg is in power phase, the other enters recovery phase. The cyclic repetitive motion allows for cardiovascular exercise while remaining quadriceps dominant, allowing for some strength adaptations to potentially address quadriceps weakness post-operation (107, 109, 169, 183, 186). The reduced loading experienced during cycling has made this mode of exercise preferred in rehabilitation following TKA (79, 187).

Kinematics

Sagittal Plane

Cycling is a task that is primarily a knee driven movement. When compared to both over ground walking and treadmill walking, cycling had greater amounts of knee extension and flexion and thusly ROM (104). The most common joint impacted by modifications during cycling is the knee joint, especially in terms of knee kinematics (188-192). Changes in the sagittal plane kinematics have been linked to modifications to the bike as well as body position (106, 189, 193-195). One major constraint to the lower extremities during cycling comes with the pelvis being placed on the saddle with very little movement. Additionally, the feet of the individual is usual in constant contact with the pedal, being fixed to the pedal via clips or a toe cage, joint kinematics are heavily impacted in the sagittal plane with changes in positions. This can be shown when saddle height increases, the knee ROM is increased due to having to reach for the pedal, through an increase knee extension angle (189).

The sagittal plane ROM during cycling is typically much larger when compared to gait or other activities. Nordeen et al. (188) found that knee ROM can range between 69° to 82.9°, when cycling with a seat height between 95% and 105% trochanter heights (188). Other studies have reported average knee ROM of 65.3° to 70° with saddle heights between 96% or 100% of the persons trochanteric height (TH) (196). The respective maximum knee angle in both conditions were 108.4° and 103.7° (196). Ericson et al. (106) described the knee joint having an average ROM of 66° (112° to 46°) (106). In the same study, an increase in workrate had a significant impact on knee joint extension (49° to 42°) but no impact on either peak knee flexion or knee flexion ROM (106). Additionally, when using three saddle heights (100% TH,100% TH + 3 cm, and 100% TH – 3 cm), knee joint ROM was significantly decreased when saddle height was decreased (194). In the same study, both workload and cadence had no effect on knee joint kinematics. The decreased ROM and increased knee flexion angle that is associated with decreasing saddle height has been indicated as a concern about knee loading (169). Decreasing saddle height is linked to an increased in knee flexion angle (and decreased knee flexion ROM), that could be linked to a greater compressive force (169).

While saddle height plays a role in knee joint kinematics, moving forward or backward on the saddle has also been found to elicit changes in knee joint kinematics (191). Cyclist were instructed to use their preferred saddle position, and then simulate a more forward or backwards position on the saddle in a sprinting position. Knee flexion angles were measured at the 3°clock (90°) and 6°clock (180°) crank positions. There were significant increases in the knee flexion angle at both times when comparing the more forwards (22%) to backwards saddle (36%) positions (191). However, these differences in knee angle were not found to be impactful (5-6°) on the knee joint loading. Bini et al. (195) again repeated a similar design on comparing the forward/backwards position in both cyclists and triathletes. Both groups saw an increase in their mean knee angle at the more forwards position (6% and 5%, respectively) and a decreased knee ROM of about 8% in the triathletes group (195). Moving forward on the saddle appears to increase the knee flexion angle while decreasing the joint ROM during cycling.

The use of lateral wedges during cycling was proposed to address rehabilitation needs in reducing joint loading (197). The addition of adding either a 5° and 10° lateral wedge had a significant effect on knee flexion angles during cycling in people with medial compartment knee OA (197). When compared to the neutral (-44.9°)condition, both the 5° lateral wedge (-47.2°) and 10° lateral wedge (-48.8°) displayed greater knee flexion angles. Modifications made proximally at the saddle and distal at the pedal have effects on the knee flexion angles of individuals during cycling.

Frontal Plane

Even with the foot being constrained by contacting the pedal and the pelvis in contact with the saddle, knee joint kinematics during cycling involved frontal plane motions (abduction/adduction) (56, 197). The use of 3D kinematics has become more popular to quantify the joint kinematics in the frontal plane, rather than just sagittal plane found in 2-D motion analysis. Gardner et al. (2016) compared healthy to knee OA individuals using lateral wedges and increasing toe-in angles. The knee OA group had a peak adduction angle in the power phase of 4.4° ± 5.6° while healthy individuals had an angle of 2.2° ± 5.3°(197). The two groups displayed no significant difference and did not significantly differ with the addition of the lateral wedges. Another study examined the impact of workrates on knee joint biomechanics ranging from 0.5 kg to 3.0 kg in 0.5 kg increments. Older healthy adults have been shown to have a peak adduction angle during cycling of approximately 6.56° ± 5.88° cycling at a middle workrate of 1.5 kg (56). When the workload was modified, their peak adduction angle changed slightly, but not statistically significant. Similarly, the older healthy population did not display a significant effect of cadence on their peak knee adduction angle at a 1kg workload (56). Peak knee adduction angles during cycling appear to be small, ranging from 2.2° to 6.56° depending on the health state of the individuals (56, 197). Modifications of workload/workrate, cadence, and the addition of lateral wedges do not appear to significantly change the peak knee adduction angles.

Kinetics

Sagittal Plane

Internal knee extensor moments reflect on the amount of overall loading at the knee joint (56, 107, 109, 154, 190, 197, 198). Compressive loading has also been reported at the knee joint as a kinetic variable during cycling (108, 109, 169). Due to its cyclic motion and high amount of knee flexion, knee joint loading is a concern for lower limb injuries, especially overuse injuries (169). Peak KEM and joint loading appear typically in mid-propulsive phase (0-180°) (198).

Ericson et al. (1986) examined the effect of workload, pedal rate, saddle height, and foot position on the knee joint moment and tibiofemoral compressive load (154). On average, the peak tibiofemoral compressive load was found to be lower than gait while external knee extension moment was decreased with increased saddle height (154). The peak knee joint moments observed were significantly lower than the ones found in gait (154). Lower saddle heights (100% TH + 3 cm) are linked to an increased work contribution of the knee compared to the hip and ankle (192). While examining other participants cycling at 120W, 60 RPM, and mid saddle height, tibiofemoral compressive forces were 812N (1.2x BW) found using the joint reaction forces from an inverse dynamics approach (109). Patellofemoral compressive forces while cycling at similar conditions were on average 905N (1.3x BW), and were increased with work load or a decrease in saddle height (108). Peak knee extensor moment at the same condition (120W and 60 RPM) was roughly 28.8 Nm (107). Workrate modifications were found to be the most impactful on modulating the knee extensor moment during cycling (56, 107). Recently, another study found that decreases in saddle height (20° to 40° knee flexion angle) significantly increases peak knee extensor moments (199).

When workload was increased from 0.5 kg to 2.5 kg while cycling at 60 RPM, knee extensor moment increased from 11.61 to 37.16 Nm with accompanied increases to peak vertical PRF (56). Bini et al. (2010) measured knee extensor moment with trained cyclists who took part in a cycling to exhaustion protocol, based on maximal power output. Knee extension moment increased by 39% between cycling done at 75% power output compared to 100% power output (190). Accompanied with increases in the joint moment, the contribution from the knee to total net moment increased with increased workload (5-8%) (190). Bini et al. (2010) confirmed again that with an increase of workload (0N – 10N), knee joint moments and contribution to total network increased, to meet demands of a greater workload (192).

Cycling biomechanics have been found to be altered following a lower extremity injury to the knee such as ACL injuries or knee pathologies such as OA(153, 200). Compared to healthy individuals, people with deficient ACL displayed a decreased knee extensor moment as well as decreased quadriceps electromyography (EMG) activation (200). These adaptations during cycling were proposed to be an attenuation process to aid in protecting the impacted limb, in attempt to reduce knee joint loads. Another concern with rehabilitation is the asymmetrical deficits found in some diseases like knee OA. During submaximal cycling, participants with knee OA displayed a larger asymmetry index (%) at two workrates (75 W and 100W) and two different cadences (60 and 90 RPM) (153). For the OA individuals, the asymmetry index was based on the amount of crank power of their less affected and more effected limb (Equation 1) while the healthy controls were based on leg dominance (Equation 2) (153). In this study, if the index was greater than or equal to 10%, it was considered to be meaningful. These asymmetry indexes ranged from -9.8% to -13.1% in the knee OA group compared to a range of 1.0% to 4.5% for the healthy controls (153).

$$Index = \frac{Less\ affected\ leg\ power - More\ affected\ leg\ power}{Less\ Affected\ Leg\ Power} * 100 \qquad (1)$$

$$Index = \frac{Dominant\ leg\ power - Nondominant\ leg\ power}{Dominant\ Leg\ Power} * 100 \qquad (2)$$

Sagittal plane knee kinetics during cycling can be impacted by a variety of factors (107, 109, 154, 190, 191, 194). It appears that the most sensitive measure to modify and modulate knee extension moments is workrate (56, 107). Cadence did not have an effect on the knee joint loading during cycling (107{Fang, 2016 #62{Fang, 2016 #62)}. Increasing saddle heights have been shown to result in decreases of KEM, and vice versa (107, 154, 194). Asymmetries have been found during cycling for a variety of injuries that could play a role in the use of cycling rehabilitation (200). Knee extension moment is a key variable to modulate when using cycling as a form of exercise.

Frontal Plane

Few studies have been done examining the frontal plane kinetics during cycling (56, 154, 197). The peak KAbM has previously been linked to greater loading of the medial compartment of the tibiofemoral joint during gait (36). An earlier study done found that the frontal plane joint moment during cycling was similar to that of walking (154). Ericson et al. (1986) found that modifications made to workrate were the most impactful on joint moments, including the frontal plane knee moments compared to pedaling rate, saddle height, or foot position. Increases in workrate found increases in joint moments, whereas decreases found decreases in joint moments. This was found again in a study examining the effect of both workload and cadence on the frontal plane biomechanics of cycling (56). While increasing workload and maintaining constant cadence, the knee abduction moment displayed an increase from 5.82 to 14.36 Nm (56). This increase in frontal plane moment was accompanied by an increase in both medial and vertical peak PRF (56). It was again concluded that there was no significant effect of workload on the knee frontal plane moment during cycling.

Attempts to modify cycling to reduce KAbM in cycling have also been done to aid in reducing medial compartment loading to the tibiofemoral joint (197). Gardner et al. (197) examined the effect of adding lateral wedges to the pedal surface to reduce KAbM, using neutral (no wedge), and a 5° and 10° lateral wedges. During cycling at 60 RPM and 80W, and found that there was a significant decrease in KAbM (-22%) when using a 10° lateral wedge (197). Interestingly, there was an increase in both the vertical and medial PRF at the same lateral wedge condition, which could be problematic when attempting to reduce KAbM. Gregersen et al. (201) had a similar study examining the effect of foot inversion/eversion angles on knee frontal plane joint moments during cycling. Participants cycled at foot angles corresponding to neutral, and either 5° and 10° of either inversion or eversion (201). The peak frontal plane moment was significant reduced by about 55% when cycling was performed at 10° eversion (201). The lateral wedges in Gardner et al. (197) would react similar to adjusting the foot angle into eversion found in Gregersen et al. (201). Finally, frontal plane moments have been previously measured at 15.3 Nm during a steady state cycling of 90 RPM and 225 W workrate (56, 202). Peak KAbM during cycling have been shown to range from 5.82 Nm to 15.3 Nm, depending on the workrate being used being the most impactful modification (56, 202).

Lateral wedges have been shown to reduce the KAbM when implemented at 10°. Recently, two studies have been done to investigate the effect of saddle height and inter-pedal distance (Q-factor) on the frontal plane knee moments during cycling (199{Thorsen, 2019 #532{Thorsen, 2019 #532)}. Modifying saddle height within a range of 20° and 40° knee flexion angle did not have a significant effect on KAbM during submaximal cycling (199). Increasing Q-Factor did however significantly increase KAbM (203). More research is needed to fully comprehend the knee frontal plane kinetics during cycling, and how to modulate loading during rehabilitation.

Augmented Feedback

Background & Introduction

While performing any task or activity, we are given multiple forms of feedback to become more proficient, reduce injury risk, or learn a new skill. Feedback s typically used during such tasks include intrinsic (sensory) or extrinsic (augmented) feedback. Sensory feedback is latent feedback that is provided from the body and its sensory organs (204). Augmented feedback is a board category of task-oriented feedback to an individual. Augmented feedback is delivered from an external source and provides feedback to the performer, with the hopes of enhancing their latent feedback and performance (204-206).

Augmented feedback is typically separated into two different types, knowledge of results (KR) or knowledge of performance (KP). Both forms of augmented feedback are given post performance, with the content and context of feedback changing While the content is of great concern, the amount of feedback and its frequency plays a crucial role for enhancing performance. The use of concurrent (real-time) compared to terminal (delayed) feedback has been debated for which is the most effective timing schedule to improve performance

Augmented feedback has been provided in a wide range of methods in biomechanics. Biomechanists aim to use augmented feedback to modify a particular variable, like muscle activity or joint angle, to elicit a better outcome after training interventions (207-213). Others may use a force platform to measure GRF following an anterior cruciate ligament (ACL) injury (91, 100). All of these methods can be performed to obtain an objective value and provide feedback with instruction to "train" the participant. It is important to identify the optimal approach to using augmented feedback in biomechanics to attempt to elicit the best outcomes.

The most effective way to compare augmented feedback to that of sensory feedback is to view them in terms of extrinsic and intrinsic feedback (214). Augmented feedback is given typically after someone performs a skill or desired action and provides some input on how the skill was performed, thus is extrinsic in nature (96-98). Augmented feedback can be given in a variety of methods, ranging from verbal to visual information (211, 215, 216). Sensory feedback comes during the movement or skill and is given based on the performer and feedback their body provides them and is an example of intrinsic feedback. Sensory feedback can also come in the form of a performer's own observation of the movement, or how they perceive their performance. Sensory feedback allows for individuals to gain some insight into how they performed their task and adjust based on only information they gathered on their own. Augmented feedback allows for external sources of information during or following the activity, and for a lack of better term, "augment" their performance. In biomechanics, there is an aim to modify certain skills to either increase performance, reduce injury risk, or rehabilitate following surgery (34, 37, 66, 92, 99, 103, 217). Augmented feedback has quickly become popular to allow researchers to provide external information to make these changes.

Types of Augmented Feedback

The content of provided information is a critical component of augmented feedback during skill modification and acquisition. The two major categories for feedback content are knowledge of results (KR) and knowledge of performance (KP) (218). KR content is focused on presenting the performer with an outcome assessment of their performance, or how well they performed a certain task (97, 98, 219). KR feedback can also be focused more so on whether or not the skill was performed to the standard or optimal goal (218). KP content, however, is focused on a direct aspect of the skill being performed that could impact performance, but not a direct measure of the performance itself (218, 220). An example in biomechanics research that could be used to identify the differences between KR and KP is the countermovement jump (212, 221-223). If a researcher gave feedback on how high the subject jumped during a countermovement jump, that would qualify as KR. The end result or goal of a countermovement jump would be the height of the jump, which would count as the objective outcome. Since KR is based on the final result of an action, this performance criterion would qualify as KR. Some other examples could be the distance of a shot-put throw, long jump, or the height of a pole vault. If instead the researcher gave feedback on their hip movement or GRF with the hopes of improving that aspect of the movement, that would qualify as KP. Where KR gives feedback on the end result of the jump (i.e. jump height), KP gives information on an aspect of the activity that could impact performance. Giving this information will still hopefully lead to an increased performance, but not through direction feedback on the end result.

`When comparing the two types of augmented feedback, the main concerns regard the skill acquisition during practice or skill retention following intervention (212, 218, 224-227). For example, Sharma et al. (218) compared KR and KP using a ball throw as their skill. The participants were split into either an KR group or a KP group, and performed ball throwing for 4 weeks, 6 days per week, and 40 trials per day The KR group was given feedback on the furthest distance they were able to throw after every 10 trials. The KP group was given feedback via verbal queues and taped videos of their own performance. There was a significant increase in the ball throw distance in both groups pre- to post-testing, with the KP group showing a greater increase compared to KR. The current study however did not test retention of the skill following a wash out period, which would have helped compared both forms of feedback in terms of their skill acquisition effectiveness. In the short term, KP appeared to outperform the use of the KR feedback paradigm.

When comparing KR and KP feedback, it is important to note that both have been found to be beneficial with skill acquisition and reducing overall error (97, 98, 226). Sidaway et al. (98) found that immediate and summary forms of the KR improved performance during the training testing conditions during a timing task. Another study have also examined the impact of KR feedback during retention following a similar timing task (227). Knowledge of results has also been examined for an impact of frequency and complexity of a motor skill (225). Groups of young children (age 11-13) were split into eight corresponding groups based on frequency of feedback and task complexity (low vs. high). Interestingly, there were no significant effects of task complexity on the outcomes measured, meaning that KR effectiveness does not appear to be impacted by task complexity (225).

Both forms of feedback have an important role and place in skill acquisition (204). Knowledge of results and performance both have been found to be a proper pathway for giving augmented feedback during a task (96, 97, 225). A consensus has been that the type of task, performer, and situation are all factors to consider when formulating the type of augmented feedback you wish to provide (204, 228, 229). There is no one size fits all model to providing feedback. In biomechanics research, there is a wide range of studies using both forms of feedback (66, 91, 230, 231). The type of feedback given in biomechanics research follows with this assertion that both knowledge of results and performance have a place. However, KP typically will fit into biomechanics as it will more likely than not consist of a kinematic, kinetic, or muscle activities (204). Since the content will vary depending on situations, the timing of feedback (immediate vs. delayed) and the frequency of said feedback is the next critical part of augmented feedback.

Timing & Frequency of Augmented Feedback

Timing of Augmented Feedback: Concurrent vs. Terminal

When providing augmented feedback during practice, the timing of providing feedback can impact the effectiveness of skill acquisition or retention (98). First, giving feedback concurrently, or in real-time, has been used previously for many different tasks (210-213, 232). The other timing paradigm used is terminal feedback scheduling (97, 98, 219, 233, 234). Terminal feedback is best described as delaying feedback until after the trial has been completed. Skill acquisition, performance during practice, and skill retention remains the focus on prescribing a specific timing and frequency schedule. Skill retention has become a greater concern, as to have a modification have longer lasting effects compared to a short-term impact from learning a skill (98, 214, 225, 235, 236). Following an intervention, the ideal outcome would be to have any modifications be retained, with individuals no longer requiring feedback.

Among studies comparing both concurrent and terminal feedback, a trend occurs that shows that each form of feedback is beneficial in in their own rights. Schmidt et al. (97) compared concurrent feedback compared to terminal feedback every 5, 10, and 15 trials during a ballistic task. The ballistic task used by Schmidt et al. (97) was to grasp an apparatus, move it 30 cm to the left, reverse direction 15 cm, and then to move backward again to complete the task. The goal time to perform the action was 550-ms, with no measure of accuracy being recorded. Feedback was provided at each appropriate interval per group and was displayed as error with respect to time to finish the task (KR). When no feedback was given, the time was filled as "empty time", where the same amount of time was given with no present feedback being given. When examined for skill acquisition, the immediate concurrent group displayed a greater performance compared to terminal feedback groups (97). This was however the opposite when examining a retention test following the intervention. The terminal feedback groups displayed a greater outcome during the retention test, with the longer the interval (15) displaying the most optimal retention performance (97, 219).

During the majority of studies that compare the use of concurrent versus terminal feedback, the differences become clear when comparing results based on training and retention tests (98, 204, 214, 227, 237, 238). A common comparison is made between the two frequencies, that concurrent feedback is very beneficial and most effective during acquisition (96, 98, 204). Terminal feedback has been found to be less effective during training but provides a greater increase in performance during a retention test. When designing an augmented feedback schedule, it then becomes critical of the outcome you desire. Concurrent feedback appears to be a better choice for immediate results whereas terminal feedback is more beneficial for long-term outcomes, when the feedback is taken away. Concurrent feedback could be beneficial first to validate that specific feedback can indeed lead to a significant change. Additionally, concurrent feedback could be beneficial to those with neurological deficits that could impair learning of a modified task in the long-term time scale. One more factor to determine using terminal feedback is how often you provide augmented feedback, typically based on a time or trial interval. Some individuals may need increased frequency to modify a task whereas others may be able to with less frequent feedback.

Frequency of Augmented Feedback

With terminal feedback displaying a greater long-term effect on skill retention, the next factor to consider is how frequent feedback should be given. In the previous section, there was a comparison between concurrent feedback and terminal feedback given following a specific set number of trials. It was reported that even within terminal feedback designs, there can be feedback schedules that are better or worse compared to others (97, 219). Summary feedback schedules will provide feedback after a predetermined amount of time, typically scheduled based on trials being performed. Schmidt et al. (97) reported that when comparing different frequencies of terminal feedback (every 5, 10 or 15 trials) that the longer interval provided that greatest benefit in skill retention testing. Schmidt et al. (97) was then supported through a study examining giving summary feedback given during 15, 7, 3, or 1 trials of a 15 trial study (98). It was found that immediate feedback given every trial did have the least amount of error in training, but the long-term retention test showed better performance with less error during retention (15 trial). The frequency of feedback was the underpinning of the results originally displayed by Schmidt et al. (97, 98).

An additional study replicated those findings suggesting the use of longer intervals for providing augmented feedback (239). Weeks et al. (239) compared two groups during a soccer throw-in task. One group received KP feedback at a 100% frequency, while the other received KP feedback 33% frequency. The group that received only 33% relative frequency of feedback had better scores in acquisition, retention, and transfer tests (239). Providing feedback less often was thought to make the performers less reliant on feedback, which allows for better retention of a skill once the KP feedback was then taken away. Butki et al. (214) used a similar design in which they compared 100%, 50%, and 0% relative frequency KR augmented feedback during golf putting. In line with previous studies, those receiving 100% continuous feedback performed better during the acquisition phase of the study, but 50% relative feedback performed greater during retention tests following the study (214). When formulating a feedback-based program, manipulating the frequency of feedback can be beneficial based on the wanted outcome or situation. Providing feedback with a higher frequency, such as after each trial, could be more beneficial for skill acquisition. Lower frequency found in summary feedback schedules could prove to be more beneficial in long-term learning and skill retention.

Augmented Feedback in Biomechanics

Augmented feedback has been implemented in biomechanics as a mean to provide additional information to participants for training studies (213, 229, 236, 240-243). Most commonly these training studies are used to change some facets of the movement to either increase performance, reduce injury risk, or optimize the movement (207, 210, 211, 217, 244). In the realm of biomechanics, studies use augmented feedback in gait retraining studies, counter movement jumps, cycling mechanics, and even with clinical populations to improve their daily life function (66, 91, 206, 212, 245-247). The main difference between most of these studies deal with the method that augmented feedback is given. Feedback is typically given using some form of biomechanics testing equipment with additional verbal instructions and guidance. The most common methods for providing augmented feedback in biomechanics include EMG, force plate data, joint angles found via motion capture, and visual cues such as mirror training (206, 208, 213, 224, 247-252).

Kinetic Based Augmented Feedback

A common tool in biomechanics research are force platforms. This instrument allows for the collection of GRF three-dimensionally (medio-lateral, anterior-posterior, and vertical). The force measurements are a key variable when examining and trying to influence loading on the body. Along with GRF, which acts as an external force applied to the body, other critical kinetic variables are joint moments and compressive loading. A joint moment is a measure of angular force acting upon a specific joint, giving insight to loading at a specific joint. These kinetic variables have been used to monitor and give feedback during biomechanics (94, 100).

Luc-Harkey et al. (100) recently used an instrumented treadmill to provide real-time feedback on vertical GRF for people following anterior cruciate ligament (ACL) reconstruction during walking. Researchers collected real-time data from their vertical forces and displayed them directly in front of participants. Participants were instructed to modify their walking patterns to either make their left and right GRF to be equal or increase/decrease it by 5% (100). However, the participants in this study were no compared to a control group that was not given any form of feedback. The goal was to reduce knee extension moment, which is a key variable for knee joint loading. Anterior GRF has also been previously used to modify gait during the propulsive phase of the gait cycle (102). Healthy individuals were presented with real-time feedback of their anterior GRF at the time of propulsion and given guidance to increase that force by 20-30%. Schenck et al. (102) wanted to use an increased anterior GRF training modalities for clinical implications for populations such as post-stroke patients to modify gait parameters to a healthier level (101, 103). During an 11-minute gait retraining session, participants were encouraged to increase their peak anterior GRF by 20-30% and was compared to baseline measurements. The use of feedback was able to display an immediate effect on increasing peak anterior GRF (102). Another variable of interest for reducing lower limb loading during gait is vertical GRF impulse which incorporates vertical GRF as a function of time. Effectively it would be the area under the vertical GRF curve during a movement (103). Golyski et al. (103) had healthy uninjured young people walk at four walking speeds with three conditions for reducing vertical GRF impulse of 5%, 10%, and 15% given real time feedback. Reducing the amount of vertical GRF impulse would lead to a decreased loading of the body, which could be beneficial for clinical populations. The "control" used for this study was considered to be baseline testing, and not a control group with no feedback. During tasks like walking, using GRFs or relevant loading variables have been used to provide feedback to elicit more beneficial outcome, such as reduced loading (100, 253).

Kinetic augmented feedback has been used in other tasks instead of just gait. Another popular movement that has been examined using kinetic based augmented feedback is jumping and landing (222, 247). Ericksen et al. (222) performed a systematic review that examined expert feedback, self-analysis, or a combination of both. The results indicated that a combination of both self-analysis feedback combined with external expert feedback were able to reduce peak vertical GRF during jump-landing tasks (222). Just as in walking, jumping or landing tasks involve a great amount of loading to the lower extremity. Onate et al. (247) wanted to investigate whether providing feedback back could help individuals land "softer". Similar to previous studies, their participants were landing on a force platform and given both visual feedback of their vertical GRF as well as verbal ques to help them modify their landing. The ability to land "quieter" would help reduce loading to the lower extremity and reduce injury risk in sports where jumping is done. Participants who were given feedback significantly reduced their vertical GRF compared to the control group that was not given any form of feedback (247)

While GRF information can be used to estimate the loading of joints, other forms of kinetic based feedback include measuring compressive loading in real-time or measuring segment accelerations (217, 245, 254-256). Pizzolato et al. (217) used motion capture in conjunction with force plate data to drive a model to estimate the medial tibiofemoral compartment loading during gait. Participants were given immediate feedback and were instructed to modify the medial tibiofemoral loading either by increasing or decreasing the value. This method was unique in that this estimate may be the most accurate means to estimate the medial tibiofemoral joint loading (217). One other means to estimate the loading of the tibiofemoral joint is through the use of accelerometers placed on the tibia, which would measure acceleration of the tibia. Creaby et al. (245) used this method of accelerometry data on a gait training program to reduce the tibiofemoral contact loads. Groups were given real-time feedback on vertical tibial peak accelerations during gait and were compared to a control group only given clinician guided feedback (245). Peak tibial acceleration was significantly reduced from baseline testing following 10 minutes of feedback (-19%) and after an additional 10 minutes without feedback (-29%). However, there was no significant difference between baseline testing and re-testing at a one-week follow-up (245). Wood et al. (256) used the same variable of peak positive acceleration of the tibia during gait, however instead of using a visual feedback paradigm, they implemented an auditory feedback program. Data was run into a custom program with a set threshold, with the pitch of sounds corresponding to the degree accelerations exceeding the set threshold (256). In conjunction, participants were instructed to run at two difference conditions: without any beeps or with the lowest pitch beeps as possible, in an effort to reduce peak tibial accelerations. Participants were able to successfully reduce their tibial peak positive accelerations when given audio feedback, and could potentially be an avenue for further feedback interventions (256).

Kinetic based augmented feedback in biomechanics presents a direct way to measure and modify a task parameter based on the desired loading outcome. While other forms of feedback modify a muscle activity or kinematic variable to reduce loading, kinetic feedback gives a direct load measurement to modify. GRFs, joint moments, tibiofemoral joint loads, or even tibial accelerations have been used as a means to modulate the loading of the lower extremity. Situational considerations should be factored in which type of augmented feedback is done in biomechanics. Tibial acceleration can be measured very easily and implemented in the real world, not in the laboratory setting and could prove to be more clinically relevant. Measuring tibiofemoral joint loading may be a more accurate means to estimate and modify knee loading, however this method takes longer to process causing delays in feedback. Ground reaction forces give a relatively good estimate of loading and can be measured both easily and rapidly in the laboratory setting. The most relevant feedback methodology may depend on the nature of the study and the desired outcomes.

Typical Timing & Frequency

The timing and frequency of augmented feedback can dictate the effectiveness of acquiring or retaining a task (97, 98). The same is true for augmented feedback studies conducted in biomechanics. The purpose of retraining studies in biomechanics is to modify a movement pattern to reduce risk or optimize performance. Therefore, when implementing augmented feedback, one must find the best paradigm to give feedback. Biomechanists have typically used two timing factors for feedback: real-time (concurrent) or following a trial or set of trials (terminal). The most common timing and frequency of augmented feedback is concurrent based feedback given at 100% frequency (100, 103, 210-213, 217, 232).

The timing and frequency used in kinetic based studies followed a similar trend to using concurrent 100% frequency feedback as the schedule of choice (100-103, 217, 221). Kinetic based studies give a visual representation of feedback on GRFs (100) or even estimate compressive loading of the tibiofemoral joint (217). Concurrent feedback on the anterior GRF was used to assist patients with increasing their propulsion on push-off (253). While not a direct kinetic measure, tibial accelerations have been used in real-time to reduce the acceleration of the tibia and reduce loading of the knee joint (245, 255, 256). Accelerometer based studies set a specific threshold that the acceleration cannot exceed, and will set off a warning to the participant (256). Feedback is given in real-time to the participant on whether or not each step they take is either below or above (how far above) the threshold

In the biomechanics field, it is apparent that the main schedule for providing augmented feedback is concurrent feedback. Forms of concurrent feedback range from muscle activity, trunk lean, ground reaction force, or tibial acceleration during a movement. Very few studies aim to use a terminal feedback paradigm when providing augmented feedback. Going forward, more studies are needed to fully describe the benefits of augmented feedback using different timing as well as frequency, especially when examining long-term benefits for training programs.

Musculoskeletal Modeling and Simulation

Estimating Muscle Forces

A common pitfall in traditional biomechanics research using 3-dimensional motion capture, force platforms, and electromyography (EMG) is that these methods are limited to the use of inverse dynamics. Inverse dynamics utilizes kinematic data from motion capture, external forces from force platforms, and subject specific anthropometric data to estimate the net moments about joints. These net joint moments calculated by inverse dynamics are the generalized forces that do not consider muscle forces and ligamentous forces that produce motion (257). The use of musculoskeletal modeling allows for an estimation of muscle forces to be used to calculate joint loading . The most common open source musculoskeletal modeling software is OpenSim (258). OpenSim allows for simulations to be run to estimate joint loading with the inclusion of muscle forces. OpenSim has three commons tools to estimate muscle forces: static optimization (SO), forward dynamics (FD), and computed muscle control (CMC) (257, 259-262).

Static Optimization

Static optimization first and foremost is one of the least taxing calculations wise, meaning it will not take as much computational power as its counterparts. SO is the tool for estimating muscle forces that will utilize kinematic and external kinetic data based inverse dynamics (263-265). While the name may suggest it, SO is not entirely a “static problem”. SO works to optimize the muscle force at each time point without respect to the previous time, which will be subject to a predestined objective function (257, 266). To run SO in OpenSim, typically a four-step process is run in the following fashion: scale the model, inverse kinematics (IK), inverse dynamics, and

SO (266-268). The final output given will be the optimized muscle forces to achieve the motion of the model (258, 266, 269, 270).

The first step in the process is to scale the model in OpenSim to subject specific parameters. The generic model used in OpenSim will have a generic set of parameters that may not be accurate between subjects. Scaling the model will allow for differences in subjects height, mass, and muscle moment arms between subjects (258, 271). The model is scaled based on subject specific data to match the model's segment lengths and widths as well as muscle attachments. These properties that differ between subjects play a crucial role in solving for joint torques and muscle forces. The model is further scaled using the experimental marker coordinate data collected during motion capture. OpenSim takes experimental marker data and uses the coordinates to adjust the model virtual markers. Experimental marker data gives OpenSim specific distances between markers, that can be used to adjust the model markers to match experimental marker data. Therefore, the scaling of the model ensures that the marker data on the model will match experimental data collected to make sure simulated results will use accurate data collected in the laboratory setting.

Next SO uses IK, which solves for the generalized joint angles and translations that best represent the experimental marker data (266, 271, 272). IK will solve to minimize the amount of squared error between the experimental marker locations ($x_i^{subject}$) and angles ($\theta_j^{subject}$) to those associated with the model (x_i^{model} θ_j^{model}) (Equation 3) (258).

$$Squared\ Error = \sum_{i=1}^{markers} \omega_i \left(x_i^{subject} - x_i^{model}\right)^2 + \sum_{j=1}^{angles} \omega_j \left(\theta_j^{subject} - \theta_j^{model}\right)^2$$

(3)

The third step in the process is standard inverse dynamics (265, 266, 273). Inverse dynamics will use kinematics obtain in the IK step along with external kinetic data and subject anthropometric data to solve for net joint moments.

Finally, the last step will be to apply the SO procedure to find the "optimal" muscle forces to solve for the net joint moments found previously. This set of optimal muscle forces is related to optimizing the movement to minimize metabolic cost, which has been debated to be the best method for movements such as gait (274, 275). The optimal muscle forces are found by minimizing a cost function (J) (Equation 4).

$$MIN\ J(F_{MT}) \qquad (4)$$

Where F_{MT} is the sum of the musculotendon forces produced by the muscles in the model. The optimization algorithm will generate solutions that will solve for the torques produced with a set of muscle forces. If the function is not minimized, the new set of muscle forces will be fed back into the beginning of the process, to then obtain the next set of F_{MT}. The optimization algorithm will continue to run until the set of muscle activations squared produced are the lowest possible that accomplish the motion. Since the F_{MT} is a function of muscle activation, the minimization criteria could also be written to directly optimize muscle activation of the i^{th} muscle, written as $a^2{}_i$ (Equation 5) (276).

$$\min \sum_{i=1}^{N} c_i a_i^2 \qquad (5)$$

Additional muscle specific constraints (c_i) are applied to ensure proper activation of each set of muscle activations. In OpenSim, the optimization criteria will minimize the muscle activation (a_t) squared and use those muscle activations to run the muscle contraction dynamics, to estimate muscle forces (277). Muscle contraction dynamics uses information from a_t profiles, musculotendon length (l_{MT}), and contraction velocity (v_{MT}) to produce tendon forces (F_T) which will be assumed equal to muscle force (277).

There has been debate on the best objective or criteria used during SO that would best represent muscle forces (274). Prior to the optimization criteria we use now, many people set optimization parameters that did not have any basis in muscle physiology during human movement. The main physiological concept that was proposed for the optimization criteria was that humans will walk in the most "efficient" manner and most efficient muscle forces. Crowninshield et al. (1981a) examined this and the concept of the force-endurance relationship. Additionally, it was assumed that the force a muscle produces will be linearly related to the physiological cross sectional area (PSCA) of the muscle (274, 277). Crowninshield et al. (274) found that minimizing muscle stress to the third power will work to maximize endurance, which is the primary physiological parameter. This criterion has also been found to have a great deal of agreeance with EMG results. It should be noted that minimizing muscle activity to maximize endurance may not be ideal for every type of activity (274). Dynamic activities like jumping, sprinting, or cutting maneuvers may not be suitable activities for using this specific criterion since the body will not optimize for endurance in dynamic tasks.

SO usually will be constrained by the intrinsic properties of muscles, the force-length and force-velocity properties (Equation 6) (266).

$$\sum_{m=1}^{n}[a_m f(F_m^o, l_m, v_m)]r_{m,j} = t_j \qquad (6)$$

Where a_m is muscle activation, $F^o_{m\ i}$ is the optimal force of the muscle, l_m is the muscle length, and v_m is the muscle velocity. This constraint is utilized to ensure that the muscle forces generated in SO does not violate the properties of muscle, such as the force-velocity property. Additional equality and inequality constraints (g,h) can be placed on the model during SO depending on the joint of interest during simulation (266). One example given by Erdemir et al. (2007) were constraints placed on the joint reaction forces of the glenohumeral joint to ensure there was no dislocation of the joint during motion.

SO uses experimental data consisting of marker trajectories, external forces (via force plate or instrumented pedals), and anthropometric information to find the optimal set of muscle forces to achieve the motion (265, 266, 273). When using SO in OpenSim, there are three inputs needed to use the tool. The results of inverse kinematics, the file containing external forces, and the scaled model generated from the scaling tool. SO will have three outputs generated: the optimized muscle activations, a storage file of the muscle activations, and finally the muscle forces over time. These results can then be used in the joint reaction analysis tool to compute the joint contact forces and joint contact moments during motion.

While SO may not be the best means to estimate muscle force in every situation, is has been used in musculoskeletal modeling of various tasks accurately. SO has found agreeable results when conducted during gait studies (264, 275, 278-281). SO has however not been used heavily in running studies where the speed of the movement may require a dynamic optimization methodology. Others have used SO heavily in upper limb movements of the elbow (267, 282-284) and shoulder (285). Recently, inverse dynamic SO has been used to estimate the muscle forces during submaximal stationary cycling (90).

The final step, which is indicative of all methodologies, is to validate the model and ensure the simulation is similar to experimental data. The main method for validating the model and simulation is through the use of EMG data (264, 266, 274, 275). Since EMG data is based on muscle action potential, there is no direct relationship to muscle activation. Instead, EMG data is compared for how close the waveform match and are in synchronization with muscle excitation during a movement. This comparison ensures that the simulated muscle excitations will be in good agreeance with experimentally collected muscle action potential (266). Another common validation method is comparing model outputs of forces to previous studies that used direct measurement methods, such as instrumented prosthesis (286). Validating the model gives more confidence that the model was simulating the movement more accurately. Without providing validation to the model, there is no way to assure the accuracy of your results.

Static optimization has successfully been used to estimate muscle forces at each individual time point to fit a specific objective function. SO uses objective functions to solve an optimization problem to minimize (optimize) a set of muscle activations to achieve the model's positions, velocities, and accelerations. SO is relatively straight forward and simple to use and it accompanied with a fast computation speed. However, SO does have some limitations that could be problematic for some types of movements. SO solves for the objective function at each time point without concern to the time before or after, which may not follow how the human body works. This issue proves to be problematic for more dynamic activities that may require a different technique to accurately estimate muscle forces. Overall, SO is an appropriate method for estimating muscle forces depending on the task and the research question at hand.

Joint Reaction Analysis

One analysis tool in OpenSim is the joint reaction analysis (JRA) tool (271, 287-290). JRA is utilized to estimate various loading parameters that are generated between two bodies (i.e. the femur to the tibia). The loading parameters, which will also comprise the outputs, are three joint compressive forces along with three joint moments (289, 291). The three contact forces that are estimated include: anterior-posterior shear, medio-lateral shear, and compressive loading. The three joint moments that are estimated using JRA are flexion-extension, internal-external rotation, and adduction-abduction moments (90, 291). To utilize JRA in OpenSim, there are several necessary inputs. JRA will use joint kinematics, external forces (ground reaction forces), and the muscle force estimations derived from one of the aforementioned procedures.

Joint Contact Forces

Steele et al. (291) used joint reaction analysis to estimate the tibiofemoral forces during crouch gait, which provides a useful example on using JRA (Equation 7).

$$R_{Knee} = [M]_{tibia} a_{tibia} (R_{ankle} + \sum F_{Muscles} + F_{Gravity}) \qquad (7)$$

where R_{Knee} is designated as the forces from the femur being placed upon the tibia, $[M]_{tibia}$ represents the inertial properties of the tibia, a_{tibia} as the accelerations (linear and angular) of the tibia, R_{ankle} are the forces placed on the tibia from the foot, $F_{Muscles}$ and $F_{Gravity}$ represent all muscle forces and force due to gravity, respectively, acting on the tibia (291).

In this example, the estimation of knee joint forces utilizes the inertial forces, forces applied from gravity and the forces applied by the ankle. A major difference in the joint reaction force algorithm is the inclusion of all the muscle forces that impact the knee joint in the form of $\sum F_{muscle}$ (291). R_{Knee} includes the three joint compressive forces along with the three joint moments as described above. Following the estimation of R_{Knee}, then the loading parameter of choice is derived depending on the orientation. Steele et al. (291) used the measurement of tibiofemoral compressive force which was the component of R_{knee} described as the component parallel to the longitudinal axis of the tibia. Therefore, the anterior-posterior shear and medio-lateral forces are those orthogonal to the longitudinal axis (290).

Others have used similar mathematical procedures to JRA to estimate joint contact loads to those measured directly with implants (289, 292). Since JRA and similar protocols produce a joint contact load (joint contact force, then the estimate will be similar to those found in implants that will directly measure the contact loading between the two bodies (271, 292). Lerner et al. (2016) used joint reaction analysis to compare hip contact forces to those found in instrumented hips. The hip contact loads produced where the resultant forces again due to muscle forces, external loads, and the inertial loads applied to the joint (289). Similar to that of Steele et al. (291), they were interested in the contact forces measured in the sagittal, frontal and transverse plane. Hip joint contact loads were not exactly similar to those measured *in-vivo* but were did not have a large degree of error.

Joint Contact Moments

The other output given through joint reaction analysis that is not as widely used are the joint contact moments (JCM). The JCM given by joint reaction analysis is the resultant moment between the two bodies that factor in the same variables in inverse dynamics, with the addition of the muscle forces previously estimated by one of the earlier techniques discussed (291, 293). Steele et al. (291) describes that both the joint contact forces and JCMs found in joint reaction analysis are the resultant forces and moments that are required to balance the loads and motions of the body in question (Equation 8).

$$R_o = \frac{t_o}{F_0} = M_i(q)u_i + F_{Contraint} - (\sum F_{External} + \sum F_{Muscle} + R_{i+1}) \qquad (8)$$

Where t_o will be the vector of JCMss (torques) and F_0 are the vector of joint contact forces. M_i is the mass matrix of the segment and q and u_i are generalized position and velocities of a given segment. Additionally, R_{i+1} represents joint contact forces and torques of the joint distal to the once in question. $F_{constraint}$ are the constraint forces applied to the body, when needed. $\sum F_{muscle}$ are the sum of all muscle forces and moments acting upon the joint of interest where as $\sum F_{external}$ is the sum of all external forces being applied.

To our knowledge only two dissertation studies have examined the use of JCMs that are generated using joint reaction analysis in OpenSim (90, 294). The most commonly reported JCM is the knee contact moment in the frontal plane, also called the varus-valgus contact moment (VVCM) (294). The main concept behind JCMs are that external and muscle forces are unbalanced during motion, and due to this imbalance, an internal moment is produced by joint contact to balance the forces and maintain the motion. In the example of a VVCM, the imbalance of loading between the two compartments, medial and lateral, would cause the JCM to compensate and maintain the joint position. When the medial compartmental loading is greater than the lateral, this would cause a counter-clockwise moment about the anterior-posterior axis of the knee joint center, leading to a positive VVCM.

It should be noted that there should be no contact moments reported in directions that the joint will not be able to make contact or resist motion. It was found that during cycling, there was no flexion-extension JCM for the knee joint (90). This could be changed theoretically in extreme alignment situations that would then cause different contact situations in the sagittal plane. This is compounded when you examine the hip contact moments. The hip is modeled as a ball and socket joint, which leads to the joint not being able to resist rotations or generate an internal torque (https://simtk-confluence.stanford.edu:8443/display/OpenSim/Joint+Reactions+Analysis).

Cycling Modeling and Simulation and Joint Contact Loading

While knee joint loading during cycling has been examined using inverse dynamics, to accurately measure this direct loading *in-vivo* is quite invasive (82, 107-109, 154). Kutzner et al. (82) used instrumented prosthesis as a means to measure tibiofemoral loading *in vivo.* Since it is difficult to perform these studies, musculoskeletal modeling and simulation has been used to accurately estimate the loading *in vivo (84, 88, 258, 295, 296).*

Musculoskeletal modeling has been extensively used during activities such as walking, running, and jumping (258, 281, 288, 295, 297, 298). Some previous works have examined muscle forces and synergies during stationary cycling (87-89, 299, 300). Early work of mathematical modeling on cycling used bivariate optimization to find the optimal cadence and crank arm length for specific subjects (301). Another study used simulations to run a forward dynamics problem to solve for neuromuscular quantities such as muscle activation and timing of activation during cycling, and to optimize these quantities based on cadence (300). They found the neuromuscular fatigues was minimized at a cadence of 90 RPM compared to both 75 and 105 RPM while at a workrate of 265 W. Additional work has been done looking at muscle synergies during forward and backward pedaling (89). The biomechanical functions of muscles appeared to not change when comparing the direction of cycling.

While there is limited research using musculoskeletal modeling, others have used inverse dynamics to estimate loading or directly measuring knee contact forces using an instrumented knee (107-109, 185). D'Lima et al. (185) utilized a custom tibial component of a TKA prothesis to measure tibial forces during various activities following a TKA. Peak tibial forces during stationary cycling peaked around 1.03 BWs and were not significantly impacted by increases in cadence ranging from 60 to 90 rpm with no direct measure of workrate give. Kutzner et al. (82) used a similar methods and report peak resultant tibial forces of 119% BW, while shear forces were approximately 5-7% BW. In the absence of said instruments, inverse dynamics does allow for computation of forces found at the tibiofemoral joint (109, 302). Peak tibiofemoral compressive force using this approach has been estimated to be 1.2 BW while cycling at 120 W, 60 rpm, and at a middle saddle height. It was found that peak knee extension moments of 28.8 Nm while cycling at 120 W and 60 rpm (107). Ruby et al. (202) found peak knee varus moment during stationary cycling at 225 W and 90 rpm to be approximately 15.3 Nm and peak knee valgus moments of 11.2 Nm. However, since inverse dynamics does not account for muscle forces or co-contraction of muscles crossing the knee joint, their estimations may be reduced to actual loading.

Simulation tibiofemoral joint loading during cycling has not been as thoroughly examined in the prior literature (88, 90). A recent dissertation at The University of Tennessee by Thompson et al. (90) conducted musculoskeletal modeling during stationary cycling in knee OA patients. Patients with knee OA and healthy participants performed cycling at various conditions consisting of neutral, 5° lateral wedge or toe-in, and 10° lateral wedge or toe-in. Musculoskeletal models were generated in OpenSim using a modified gait2392 model with added patella and increased knee flexion. Static optimization was run to estimate muscle forces that would generate the experimental positions, velocities, and accelerations. Further research is needed using musculoskeletal modeling to estimate knee joint loading during stationary cycling, allowing for inclusion of muscle forces. There is especially a gap in literature for using musculoskeletal modeling to estimate knee compressive forces during stationary cycling in a TKA patient population. Estimating joint loading through this method in TKA patients will greatly improve the clinical significance of measurements compared to those found in inverse dynamics, that do not account for muscle forces.

CHAPTER III

MATERIALS AND METHODS

Participants

Individuals who have undergone a TKA within the past 6-18 months were recruited from the Tennessee Orthopedic Clinic (TOC). Potential participants were identified by the TOC and were mailed letters giving study information. Participants that met both the inclusion and exclusion criteria (Table 1) were invited to participate in this study. Participants recruited into this study were randomized into either the intervention or control group. However, due to recruitment difficulties and issues stemming from COVID-19, the control group was not included for analysis in studies 2 and 3.

Table 1. Inclusions and Exclusion Criteria for TKA participants

Exclusion Criteria • Initial VNS pain score greater than 5 in the replaced knee • Osteoarthritis of ankles, contralateral knee, or hips that impacted walking • Any other lower extremity joint replacement other than the single knee • Systemic inflammatory arthritis (rheumatoid arthritis, psoriatic arthritis) as reported by the patient that impacts daily living • BMI greater than 38 • Neurologic disease (e.g. Parkinson’s disease, stroke) as reported by the patient. • Any major lower extremity injuries/surgeries in the past 6 months • 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.
Inclusion Criteria • Men and Women between the ages of 50 and 80 • Total knee replacement between 6 to 18 months prior

Interested participants were given study information by the TOC or contacted via mailed letters by the TOC. All of the TKA operations were conducted by the same surgeon. Participants were required to meet the remaining inclusion/exclusion criteria. Prior to participation, participants were contacted and passed additional screening questions (Appendix A).

Sample size for this study was determined using GPower (Ver. 3.1, Heinrich Heine Universistat Dusseldorf). Sample size for studies 1 and 4 were determined using *a priori* power analysis that used asymmetry indices of power output that displayed a need for 12 participants with an alpha of 0.05 and beta level of 0.80 (153). This study was approved by the Institutional Review Board at the University of Tennessee (UTK IRB-19-05110-XP). Prior to participation, all participants read, signed, and gave the informed consent (Appendix B).

Instrumentation

A twelve-camera motion capture system (240 Hz, Vicon Motion Analysis Inc., UK) was used to obtain the three-dimensional (3D) kinematics during the testing. Reflective anatomical markers were placed on the acromion processes, iliac crests, greater trochanters, medial and lateral epicondyles, medial and lateral malleoli, the head of 1st and 5th metatarsals, and tip of the second toe. A cluster of four reflective markers affixed to a semi-rigid thermoplastic shell were placed on the trunk, thighs, and legs respectively to track segment motions during testing. Four additional discrete reflective markers were placed on the heel of each shoe to track the motion of the foot.

Two force platforms (1200 Hz, BP600600 and OR-6-7, American Mechanical Technology Inc., Watertown, MA, USA) were used to measure the ground reaction forces (GRF) and the moments of forces during the walking trials, using the Vicon Nexus Software (Ver 2.8, Vicon Motion Analysis Inc., UK). Simultaneous collection of the 3D kinematics and ground reaction forces were conducted during the gait testing trials using Vicon Nexus (2.9, Vicon Motion Analysis Inc., UK).

A 16-channel wireless EMG system (1200 Hz, Delsys Trigno Wireless EMG., Delsys, Natick, MA, USA) was used to record muscle activity during gait and cycling trials. The sensors were placed on the following muscle bilaterally: vastus medialis (VM), vastus lateralis (VL), biceps femoris (BF), semitendinosus (ST), and medial gastrocnemius (MG). Participant's skin was shaved to remove any hair and cleaned with alcohol swabs prior to electrode placement. Electrode placement on each muscle followed the guidelines by the SENIAM (303, 304). Sensors were attached with double sided adhesive tape and anchored with athletic pre-wrap.

An isokinetic dynamometer (System 4, Biodex Medical System, Shirley, New York, USA) was used to test the patient's quadriceps and hamstring strength. Participants performed two trials of submaximal and three trials of maximal contraction concentric/concentric isokinetic testing at 80°/sec. Participants were positioned to maintain a 90° angle between their trunk and thigh.

An electromagnetically braked stationary ergometer (Excalibur, Lode B.V., Groningen, Netherlands) was used for the stationary cycling testing and the subsequent training program. Two customized bike pedals instrumented with two 3D force sensors (1200 Hz, Type 9027C, Kistler, Switzerland) coupled with two industrial charge amplifiers (Type 5073A and 5072A, Kistler, Switzerland) for each pedal were used to measure 3D forces and moments. The charge amplifiers can convert the charges measured by the force sensors to voltage values used by Vicon Nexus. The kinetic data from the instrumented pedal was recorded by the Vicon Nexus software suite simultaneously with the 3D kinematic and EMG data during the cycling testing trials.

Experimental Protocol

Experimental Data Collection Protocol (Studies 1-4)

TKA participants were recruited into two groups: an intervention group and control group. Each participant participated in two testing sessions, a pre- and post-training session, separated by two-three weeks of either the training or a control period. During the first test session, participants completed an informed consent (Appendix B), the Knee Injury and Osteoarthritis Outcome Score (KOOS) questionnaire (Appendix C), and a physical readiness questionnaire (PARQ) (Appendix E). Participants were only included in the study if they answer "no" to every question on the PARQ, to ensure they did not need physician approval prior to exercise as a means to minimize risk. The pre- and post-training testing days were identical to one another.

Participants completed a three-minute treadmill warm-up at a self-selected pace. Next participants completed a series of functional tests commonly used to test general functionality. Participants first performed a timed up-and-go (TUG) test, in which participants were tasked to rise from a chair without using their arms to push off and walk 10 feet around a cone and return to their seat at a regular pace. The next functional test conducted was the chair rise test. Participants started in a seated position, with their knees flexed at a 90° angle. Upon starting, participants were told to stand up, and sit back down into the seat a total of 10 times at a comfortable pace. During these trials, participants were instructed to cross their arms across their chest and rest their hands on their shoulders. Time to completion was recorded for both functional tests. For the TUG, timing was recorded from when the participant began to move and until they came to rest sitting after completing the trial. For the chair rise, the timer started once participants began to move on initiation and once, they have returned to a resting seated position, following their 10^{th} repetition.

Participants then completed isokinetic testing of their quadriceps and hamstrings strength for both their replaced and non-replaced limbs in a randomized order. Participants performed two submaximal trials proceeded three trials of maximal effort muscle testing of their quadriceps and hamstrings at 90°/sec on an isokinetic dynamometer. Prior to testing, participants were set into the dynamometer to elicit a 90° hip angle and with the axis of rotation in line with the lateral femoral epicondyle. Participants started with their knee at a 90°. Their limb was fastened to the dynamometer with a padded Velcro strap just superior to the malleoli. During testing, participants were told to flex and extend their knee as forcefully as possible within their pain tolerances and were given verbal instruction during testing. Following the maximal isokinetic testing, participants were given at least 5 minutes prior to moving to the next protocol.

Participants were then instrumented with the wireless surface EMG bilaterally (303, 304). Following EMG placement, participants completed a series of "functional" tests that were used to normalize EMG data. Three trials of each movement were completed. A chair was placed within reach of participants for each test as a safety mechanism if participants required additional support for balance. Functional tests were completed in the following order: body weight quarter squat, standing unilateral hamstring curl, and calf raise (ankle plantarflexion, going onto toes). Participants were then instrumented with reflective markers.

Participants performed two cycling conditions (80 and 100 Watts) on a stationary ergometer at a constant cadence of 80 revolutions per minute (RPM). The ergometer was set up to adjust handlebars to elicit a 90° hip angle. The saddle height was modified to elicit to 30° knee flexion while the pedal was at the bottom crank measured by a handheld goniometer. The ergometer saddle fore aft position was adjusted to place the knee above the pedal spindle at the 3o'clock position of crank arm using a plumb bar. Participants were given a three-minute warm up at their first workrate. Next, participants cycled at each workrate for one-minute with data being collected in the final 10 seconds. Participants were unaware of when data was being collected during every condition. A total of five crank cycles were truncated into five separate trials for further analyses. Three-dimensional kinematic data was captured using a 12-camera motion capture system. Pedal reaction forces were collected using two customized instrumented pedals. Kinematics and kinetics were recorded synchronously using Vicon Nexus software.

Participants then perform two over ground gait conditions at the participants preferred speed (±10% speed) and a fast speed (preferred speed + 0.4 m/s). A total of five successful trials were collected for each speed for each limb. Prior to collecting, participants completed at least three practice trials to become accustomed to the walkway. During these practice trials, the participant's preferred walking speed was monitored using two sets of photocells (63501 IR, Lafayette Instrument Inc., IN, USA) and Universal Timer and software (Model 35930, Lafayette Instrument Inc., IN, USA), which were placed three meters apart. A total of three practice trials were recorded and averaged to determine participants' preferred walking speed. Once the preferred speed range were determined, a total of five recorded trials were then completed for each walking condition. Conditions were randomized first by speed and then by limb. Kinetic data from inground force platforms were recorded in conjunction with kinematic data using Vicon Nexus software. Participants were asked to rate their pain level using an enlarged visual numeric scale (VNS) (Appendix F) and rating of perceived exertion (RPE) (Appendix G) before and after each testing condition and throughout testing.

Following the pre-training session, participants completed either the training intervention or the control period of the study for two-three weeks After the three weeks, participants returned to complete a post-training test session, which followed the same protocol as pre-training.

Training Intervention

The training program consisted of 6 training sessions: two sessions per week for three weeks, which included cycling training enhanced with augmented feedback of PRFs. The initial training session started with a duration of 10 minutes and progressed to a final session lasting 20 minutes. Breaks were given in 5-minute intervals to ensure proper rest. The cadence was kept constant at 80 RPM throughout the training sessions. Participants started the training program at a workrate of 60 W and progresses by 20 W increment based on the RPE, VNS pain levels, and asymmetry index (Appendix H). Following each training session, participants performed a cool down for five minutes followed by light static stretching including: seated hamstring stretch, assisted quadriceps stretch, and wall calf stretch. All participants were asked to maintain their exercise levels to what they were doing, but not increase or decrease their extracurricular activities. Logs kept during the training period tracked the number of bouts completed for each participant, their workrate for each bout, and their RPE, VNS, and asymmetry index.

Augmented feedback was provided on a consistent interval and displayed as visual feedback to the intervention group. Augmented feedback was given to each participant at this schedule during each exercise bout that lasted five minutes in length: minute one, minute two, minute three, minute four, and minute five. Vertical PRFs using the same instrumented pedals were collected in Vicon Nexus (1200 Hz) for 30-seconds (seconds 20-50) with feedback being processed and provided immediate using a custom MATLAB (2019a, MathWorks Inc., Natick, MA, USA) as a post-processing pipeline in Vicon Nexus. Feedback was displayed on the screen for a total of 10 seconds. PRFs were filtered using a fourth-order zero-lag Butterworth filter at a cut-off frequency of 6 Hz (38, 56, 197). Peak PRFs of cycles were then identified and averaged over the 30-seconds period. Visual representation of the averaged peak data was presented as a bar graph on a screen in front of the participant (Figure 1). The replaced and non-replaced limbs were in line with the participant's view (i.e. if the right limb is replaced, the right bar will be data for the right limb). Each bar was displayed as the average data for the peaks for each corresponding limb with an upper and a lower threshold boundary. The thresholds were calculated based off of the non-replaced limb. These thresholds were equal to ± 10% of the non-replaced limb average force (Equation 9) and were presented on the graph as horizontal bars.

$$Threshold = NRForce \pm NRForce * 0.10 \qquad (9)$$

Where NRForce is defined as the average peak vertical PRF on the non-replaced limb. This is in accordance to asymmetries of greater than 10% is considered to be clinically relevant (305). Participants were instructed to keep both bars of the figure within the threshold bars. Prior to starting the intervention, participants were instructed on how to read the graphs during feedback, and what would need to be accomplished in any of the three situations. During the intervention, no direct verbal feedback was given to not overcomplicate the intervention for participants (96-98).

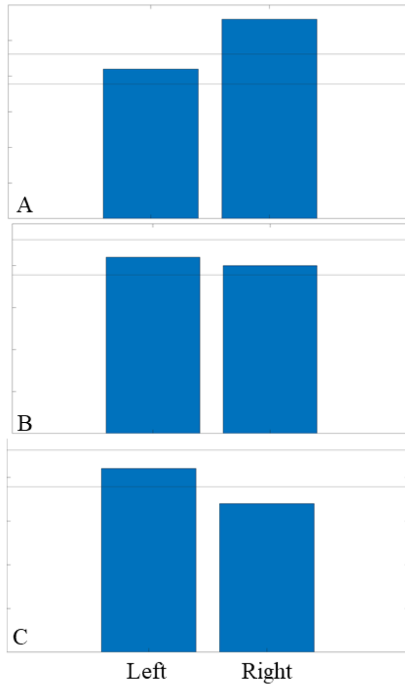


Figure 1. Graphical representation of pedal reaction force based augmented feedback for a right limb replacement. Horizontal black lines correspond with ±10% of the pedal reaction force of the non-replaced limb (left in this example). A) Displays a greater asymmetry towards the replaced limb. B) Displays asymmetries within the threshold of ±10%. C) Displays a lower asymmetry for the replaced limbs

Data Analysis

Study One/Two/Three

Three-dimensional kinematics, kinetics, ground reaction forces, pedal reaction forces, moments, and center of pressure were computed in Visual3D (6.01, C-Motion Inc., Germantown, MD, USA). Kinematics and PRFs from cycling collections were filtered using a fourth-order zero-lag Butterworth filter at a cutoff frequency of 6 Hz (197). Ground reaction forces were filtered using a fourth-order zero-lag Butterworth filter at a cutoff frequency of 50 Hz. Three-dimensional kinematics were calculated using a Cardan sequence of X-Y-Z and the joint coordinate system (306). Joint moments were calculated via an inverse dynamics method in Visual3D. Kinematics and kinetics were both reported in the joint coordinate system following the right-hand rule. Joint moments and PRFs were not normalized to body mass as the majority of the participants body weight were supported on the saddle and handlebars (56, 197). Joint moments were normalized to body mass (Nm/kg) and GRFs were normalized to body weight (BW) for study three. Critical peak events were identified and organized using a custom computer program (VB_V3D and VB_Tables, MS Visual Basic 6.0, USA). Events were selected for five sequential cycles for each interested variable and the averages of the five trials were used in statistical analyses.

Musculoskeletal Modeling and Simulation (Study 4)

Musculoskeletal modeling was performed using the open source software OpenSim (258). Experimental data was exported from Visual3D for use in OpenSim, that included the computation of scaling factors based on experimental marker data and inverse kinematics. A generic musculoskeletal model with 23 degrees-of-freedom and 92 musculotendon actuators was used for the musculoskeletal modeling and simulation (307). The hip joint is modeled as a ball-and-socket while the ankle and subtalar joints were modeled as revolute joints. The subtalar and metatarsophalangeal joints were locked to model the foot as a single rigid segment. The knee joint has been modified to include two revolute joints to estimate forces in the medial and lateral tibiofemoral joint compartments and remains a single degree-of-freedom joint allowing for flexion and extension movement (307).

The process for analyzing the cycling trials used an inverse dynamics static optimization approach that resulted in estimated muscle activations (308, 309). Each subject specific model was scaled based on the participants height, mass, and segment lengths based on experimental marker data. Inverse dynamics was run based on the kinematics, external pedal reaction forces recorded at each pedal. Muscle activations and forces were then estimated using static optimization to solve to minimize the sum of the squared muscle activations (Equation 10) (274, 297).

$$Min \sum_{i=1}^{N} C_i \, a_i^2 \qquad (10)$$

Finally, joint reaction analysis was used to solve for TCF, MCF and LCF expressed in the tibia reference frame (291).

Electromyography

Experimental EMG data was filtered using a 4th order zero-lag Butterworth bandpass filter with a high-pass cutoff frequency of 10 Hz and a low-pass cutoff frequency of 450 Hz. The filtered EMG data was then full wave rectified. Finally, a moving root-mean-squared (RMS) filter was conducted with a 91 ms moving window size. The RMS EMG data was normalized to the peak value of each muscle during functional tests. The quarter weight squat was used for both the VM and VL. The standing unilateral hamstring curl was used for the ipsilateral BF and ST. The ankle plantarflexion functional test was used to normalize the MG. The normalized waveforms for EMG were used for validation for the musculoskeletal modeling.

The TCF, MCF, and LCF generated by joint reaction analysis were exported and peak values during the power phase of the crank cycle (crank angle between 0 – 180°) were determined interactively using customized codes in Matlab. The summed muscle force for the knee extensor and flexor groups were selected in addition to the tibiofemoral contact forces. To validate the musculoskeletal model used in this current study, experimental EMG muscle activations were qualitatively compared to the computed muscle activations.

Statistical Analysis

Study One

Primary variables of interest included peak KEM and vertical PRF. Several secondary supporting variables were included for discussion including knee kinematics as well as the sagittal plane moments of the ankle and hip joints. A 2 x 2 (limb x workrate) repeated measure analysis of variance (ANOVA) was conducted to examine effects of limb, workrate and their interactions on the primary and secondary variables with an alpha set at 0.05 *a priori* (IBM SPSS Statistics 25, Chicago, IL, USA). The assumptions of normality and sphericity were assessed using a Shapiro-Wilk test and Greenhouse-Geisser test, respectively. Paired sample t-tests were run for planned post-hoc comparisons in the presence of significant interactions of limb and workrate with an adjusted alpha level of 0.0125. Effect sizes for ANOVAs are reported as partial eta squared (η^2_p) (310) while effect sizes for main effects of limb and workrate were computed as Cohen's d (311).

Study Two

A one-way repeated measures analysis of variance (ANOVA) was run on AI of peak vertical PRF, posterior PRF, and KEM comparing pre- and post-training measurements at 80 and 100 W separately. A 2 x 2 (limb x time) repeated measure ANOVAs were run on the mean data for: vertical PRF, posterior PRF, KEM, knee extension ROM, knee abduction ROM, hip extension moment, and ankle plantar flexion moment at 80 and 100 W separately. Alpha levels were set *a priori* of 0.05. Effect sizes were reported as partial eta squared (η^2_p) and were interpreted as small ($\eta^2_p < 0.06$), medium ($0.06 \leq \eta^2_p < 0.15$), and large ($\eta^2_p \geq 0.15$) (310).

Study Three

Separate one-way repeated measures analysis of variance (ANOVA) were run on AI of load-response KEM, load-response vertical GRF, push-off KEM, and push-off vertical GRF comparing pre- and post-training measurements for both walking speeds. Individual 2 x 2 (limb x time) repeated measure ANOVAs were run on the selected dependent variables at each walking speeds separately. Paired t-tests were run on gait velocities and VNS pain outcomes comparing pre- and post-training. An alpha level was set 0.05 *a priori*. Effect sizes were reported as partial eta squared (η^2_p) and were interpreted as small ($\eta^2_p < 0.06$), medium ($0.06 \leq \eta^2_p < 0.15$), and large ($\eta^2_p \geq 0.15$) (310).

Study Four

A 2 x 2 (limb x workrate) repeated measure analysis of variance (ANOVA) was run on peak TCF, MCF, LCF, and summed knee extensor and flexor muscle forces (IBM SPSS Statistics 25, Chicago, IL, USA). A separate 2 x 2 (compartment x limb) repeated measure ANOVA was run on peak MCF and LCF. An alpha level was set at 0.05 *a priori*. Normality and sphericity were assessed via a Shapiro-Wilk test and Greenhouse-Geisser, respectively. Paired sample t-tests were conducted for planned post-hoc analysis when an interaction was present with Bonferroni adjustment using an adjusted alpha level of 0.0125. Effect sizes were reported as partial eta squared (η^2_p) and interpreted as large ($\eta^2_p \geq 0.14$), medium ($0.06 \leq \eta^2_p < 0.14$) and small ($\eta^2_p < 0.06$) (310). Effect sizes for main effects and post-hoc pairwise comparisons were reported as Cohen's D (307, 311).

CHAPTER IV

Knee Joint Biomechanics of Patients with Unilateral Total Knee Arthroplasty During Stationary Cycling

Abstract

Stationary cycling is typically recommended following total knee arthroplasty (TKA) operations. However, knee joint biomechanics during cycling remains mostly unknown for TKA patients. Biomechanical differences between the replaced and non-replaced limb may inform applications of cycling in TKA rehabilitation. The purpose of this study was to examine the knee joint biomechanics of TKA patients during stationary cycling. Fifteen TKA participants cycled at 80 revolutions per minute and workrates of 80 Watts and 100 Watts while kinematics (240 Hz) and pedal reaction forces using a pair of instrumented pedals (1200 Hz) were collected. A 2x2 (limb x workrate) repeated measures ANOVA was run with an alpha of 0.05. There was a main effect of limb on peak knee extension moment (KEM) ($p = 0.034$) and vertical pedal reaction force ($p = 0.038$). Both peak KEM and vertical pedal reaction were significantly lower in the replaced limb compared to the non-replaced limb. Peak KEM did not change for TKA patients with the increased workrate ($p = 0.750$). However, both peak hip extension moment ($p = 0.009$) and ankle plantarflexion moment ($p = 0.017$) increased due to increased workrate. Patients following TKA showed similar decreases in peak KEM and vertical pedal reaction force as previously seen in gait. Future research should examine tibiofemoral joint contact forces via musculoskeletal modeling, as well as training implications using stationary cycling following TKA.

Keywords: total knee replacement, ergonomic cycling, knee extension moment, hip and ankle extension moment, bilateral deficits

Introduction

Knee osteoarthritis (OA) is a degenerative disease of articular cartilage and subchondral bone of the knee joint and is one of the most common knee pathologies in older adults (1, 4, 5). End-stage knee OA brings with it a considerable amount of pain, that can decrease joint function and impair activities of daily living (5). The primary surgical intervention to end-stage knee OA is a total knee arthroplasty (TKA) (16, 17, 76). While TKA operations can improve function and reduce pain levels, there are still concerns following the operation. One important facet of undergoing TKA is the rehabilitation that follows, generally, with the aim of regaining as much knee joint function as possible. Additionally, deficits between replaced and non-replaced limbs could predispose patients to a TKA revision or TKA of the contralateral limb (47, 76).

While there are more complete guidelines addressing exercise rehabilitation and activity for knee OA patients (150, 312-314), no comprehensive and universally accepted rehabilitation guidelines following TKA have been adopted (79, 315, 316). Current suggestions include increasing knee range of motion (ROM), quadriceps strengthening, activities for cardiovascular health and weight management, and decreased knee joint loading during activity (17, 76, 77, 80, 145, 148, 185). One preferred exercise modality is stationary cycling, an activity with lower tibiofemoral joint loading compared to weight-bearing exercises, which also promote cardiovascular health and muscle strengthening (82, 148). However, there is a lack of evidence in the literature that supports stationary cycling as a rehabilitation modality following TKA.

It is not clear how TKA patients would respond to changes in workrates in stationary cycling, which may provide an evidence-based recommendation when prescribing exercises post TKA. Although cycling biomechanics data of TKA patients are scarce in the literature, cycling biomechanics of healthy participants in relationship to rehabilitation applications has been studied extensively (38, 56, 106, 154, 197, 302). Increasing workrate has been found to increase both internal knee extension moment (KEM) and internal knee abduction moment (KAbM) (56), both of which are directly related to the amount of tibiofemoral joint loading (58). While increases in workrate lead to increased KEM and KAbM, increasing cadence during stationary cycling does not increase either (56). However, the impact of workrate changes remain unknown on the knee joint kinetics for TKA patients.

While literature on cycling biomechanical deficits of TKA patients are limited, TKA patients show clear deficits in other common daily activities. Following TKA, bilateral deficits are present in key biomechanical variables such as knee flexion angles, vertical ground reaction forces (GRF), and KEM during activities such as walking and stair negotiation (20, 29, 30). TKA patients walk with decreased knee flexion ROM (stiff knee gait), which is proposed to be due to a quadriceps weakness and avoidance (317). TKA patients also demonstrate decreased vertical GRF and KEM in their replaced knee compared to their non-replaced contralateral limb and healthy matched controls (31, 47, 318). Decreases in peak KEM in the replaced limb compared to their non-replaced ranged from 12.2% to 20% during stair negotiation (31, 318) and were about 5.7% in level walking (47). These deficits in key loading variable raise concerns of disproportionate loading between limbs, potentially increasing risk for TKA on the contralateral limb, or a revision of the current replacement (124, 319).

Therefore, the purpose of this study was to examine the knee joint biomechanics of unilateral TKA patients during stationary cycling at two different workrates. Our primary hypothesis was that peak KEM and vertical pedal reaction force (PRF) would be significantly lower in the replaced limb compared to the non-replaced limb. Our secondary hypothesis was that there would be significantly greater peak KEM and vertical PRF at 100 W compared to 80

W. Finally, our tertiary hypothesis was that there would be no limb by workrate interaction for peak KEM and vertical PRF.

Methods

Participants

Fifteen unilateral TKA patients (10 males and 5 females, 64.3 ± 8.2 yrs, 94.1 ± 20.4 kg, 1.74 ± 0.1 m) were recruited from a local orthopaedic clinic. All patients were 6 to 18 months post unilateral TKA, completed by the same surgeon, and were between 50-80 years old. Potential participants were excluded from the study if they had any other forms of debilitating lower limb joint OA that impacted the way they walk, other joint arthroplasties, BMI greater than 38 kg/m^2, any neurological diseases that would impact gait or balance, arthroscopic surgeries within three months, or required aid during gait (walkers or cane) or during stationary cycling. All study procedures and protocols were approved by the university Institutional Review Board. Prior to participation, all participants read and signed an approved informed consent.

Instrumentation

Participants wore tight fitting spandex shorts as well as standard laboratory shoes during testing (Zoom Pegasus 34, Nike, Portland, OR, USA). Three-dimensional kinematics were collected using a twelve-camera motion capture system (240 Hz, Vicon Motion Capture Inc., Oxford, UK). Reflective markers were placed bilaterally on the following anatomical landmarks for the static calibration prior to collection: acromion processes, iliac crests, greater trochanters, lateral and medial femoral epicondyles, lateral and medial malleolus, 1st and 5th metatarsal heads, and the distal phalanx of the second tarsal. Segment motion was tracked using cluster sets of four reflective markers mounted on thermoplastic shells using Velcro attached to neoprene straps on the trunk, pelvis, and both thighs and shanks. Additionally, four individual reflective makers were placed on the posterior-lateral heels to track the foot segments' motion. To track the motion of the pedals, three markers were rigidly attached to the lateral side of each pedal, with a fourth being placed on the anterior surface of the pedal. Two reflective makers were placed on the crank arm axes and one additional marker on the front of the bike.

An electromechanically braked stationary cycle ergometer (Excalibur Sport, Lode B.V., Groningen, Netherlands) was used during all cycling test conditions. Workrate was controlled via a control unit placed in front of the ergometer with a display of workrate and the current cadence. Three-dimensional kinetics were collected with two customized instrumented pedals, with each equipped with two tri-axial force sensors (1200 Hz, Type 9027C, Kistler, Switzerland), in conjunction with two amplifiers (Type 5073A, Kistler, Switzerland). To ensure the pedal coordinate system of the ergometer was aligned with the global coordinate system, the stationary ergometer was secured via a metal jig that affixed it to the ground. Three-dimensional kinetic data was recorded in conjunction with 3D kinematic data using Vicon Nexus data collection software (version 2.8.2, Vicon Motion Capture Inc., Oxford, UK).

Experimental Procedures

Upon arrival, participants completed the informed consent, the Physical Readiness Questionnaire (PARQ). Next, participants completed a three-minute warm up at a self-selected pace on the treadmill. Participants were then instrumented with the reflective makers prior to commencement of data collection.

The stationary ergometer was adjusted to fit for each participant. The saddle height was set to elicit a 30° knee flexion angle with the pedal at the dead bottom position (180°) measured by a handheld goniometer (192, 320, 321). The saddle fore-aft position was set as to have the knee set directly above the pedal spindle while the pedal was at the 3'oclock position measured with a plum bob. The handlebars were adjusted to elicit a 90° angle between the thigh and trunk while the pedal was at the 3'oclock position measured.

The two workrate conditions (80 and 100 Watts) were randomized prior to participant arrival. These workrates have been previously employed in examining knee OA patients during stationary cycling (38, 197). Participants were given three-minutes to warm up on the stationary ergometer at their first workrate. Cadence was kept at 80 revolutions per minute (RPM) with a range of ± 2 RPM (78 – 82 RPM), which was displayed visually in front of the participant. Participants then cycled for one minute and data were collected for ten seconds at the end of the first minute. Following the first condition, participants were given a minimum of one-minute rest before completing the second condition.

Data Analyses

The ten second data of marker trajectories and PRF were truncated into five individual trials consisting of a complete crank cycle for both limbs. A crank cycle was defined as the crank arm beginning at top dead center (0°) and finished once one complete revolution returned the crank arm to the top dead center (360°). PRF, COP, joint kinematics, and joint moments were calculated using Visual3D (Version 6.01, C-Motion Inc., Germantown, MD, USA). Kinematic and kinetic data were filtered using a fourth order zero-lag Butterworth lowpass filter with a cutoff frequency of 6 Hz (38, 56, 197). Joint angular kinematics were calculated using the joint coordinate system with a Cardan rotational sequence (X-Y-Z). Joint moments were calculated using inverse dynamics expressed in the proximal segment of the joint (e.g. knee moments were expressed in the thigh). Segment masses were equated from established regression equations using body mass (322). Joint kinematics and moments were expressed following the right-hand rule convention. Peak values were identified and organized in custom programs (VisualBasic

6.0, Microsoft, Redmond, WA, USA). PRF and joint moments were not normalized to body mass, since participants had the majority of their body mass supported via the saddle and handlebars (38, 56, 197).

Statistical Analyses

Primary variables of interest included peak KEM and vertical PRF. Several secondary supporting variables were included for discussion including knee kinematics as well as the sagittal plane moments of the ankle and hip joints. A 2 x 2 (limb x workrate) repeated measure analysis of variance (ANOVA) was conducted to examine effects of limb, workrate and their interactions on the primary and secondary variables with an alpha set at 0.05 *a priori* (IBM SPSS Statistics 25, Chicago, IL, USA). The assumptions of normality and sphericity were assessed using a Shapiro-Wilk test and Greenhouse-Geisser test, respectively. Paired sample t-tests were run for planned post-hoc comparisons in the presence of significant interactions of limb and workrate with an adjusted alpha level of 0.0125. Effect sizes for ANOVAs are reported as partial eta squared (η^2_p) (310) while effect sizes for main effects of limb and workrate were computed as Cohen’s d (311).

Results

Primary Outcome Variables

No significant interaction (p = 0.375) nor workrate main effect (p = 0.750) were observed for peak KEM. There was a significant effect of limb for peak KEM ($F[1,12] = 5.49$, p = 0.034, η^2_p = 0.32, d = 0.87, Table 1) with greater peak knee KEM found in the non-replaced limb. There was no significant interaction for peak vertical PRF (p = 0.14, Table 2). Significant effects of limb ($F[1,12] = 5.42$, p = 0.038, η^2_p = 0.31, d = 0.30) and workrate ($F[1,12] = 31.615$, $p < 0.001$, η^2_p = 0.73, d = 0.45) were found for peak vertical PRF. Peak vertical PRF was greater in the non-replaced limb, regardless of workrate. Peak vertical PRF was also greater at 100 W, regardless of limb (Table 2). Ensemble curves of knee, hip and ankle sagittal-plane moments in the 100 W condition are presented in Figure 2.

Secondary Supporting Variables

Peak posterior PRF displayed a significant effect of limb (F[1,12] = 7.50, $p = 0.018$, $\eta^2_p = 0.39$, $d = 0.61$) but no effect of workrate ($p = 0.855$) or interaction ($p = 0.677$). The non-replaced limb had a greater peak posterior PRF compared to the replaced limb. Peak medial PRF exhibited no significant interaction ($p = 0.811$) or effect of limb ($p = 0.564$). There was a main effect of workrate for peak medial PRF (F[1,12] = 2.44, $p = 0.026$, $\eta^2_p = 0.35$, $d = 0.38$) with greater peak medial PRFs found at 100 W compared to 80 W.

Knee extension ROM did not display significant interaction ($p = 0.748$) or effect of workrate ($p = 0.688$, Table 2). However, there was a main effect of limb on knee extension ROM (F[1,12] = 13.84, $p = 0.003$, $\eta^2_p = 0.54$, $d = 0.50$) with greater knee extension ROM found in the non-replaced limb (Table 3). Knee abduction ROM displayed a significant interaction of limb and workrate (F[1,12] = 9.264, $p = 0.010$, $\eta^2_p = 0.44$). Post hoc t-tests found that the non-replaced limb at 80 W differed from the non-replaced at 100 W ($p = 0.012$, $d = 0.29$), and the replaced limb at 80 W ($p = 0.008$, $d = 1.29$). The non-replaced limb at 100 W was additionally different from the replaced limb at 100 W ($p = 0.006$, $d = 0.83$). No significant interaction ($p = 0.204$) or effects of limb ($p = 0.376$) or workrate ($p = 0.146$) were found for peak KAbM.

Peak plantarflexion moment did not display a significant interaction ($p = 0.945$) or effect of limb ($p = 0.196$, Table 3). There was a significant effect of workrate (F[1,12] = 7.74, $p = 0.017$, $\eta^2_p = 0.39$, $d = 0.22$) on peak plantarflexion moment. Peak plantarflexion moment was greater at 100 W compared to 80 W. Similarly, peak hip extension moment displayed a significant effect of workrate ($F[1,12] = 9.702$, $p = 0.009$, $\eta^2_p = 0.45$, $d = 0.55$) but no interaction ($p = 0.658$) or effect of limb ($p = 0.465$). Peak hip extension moment was greater at 100 W compared to 80 W.

Discussion

The purpose of this study was to examine the knee joint biomechanics of unilateral TKA patients during stationary cycling at two different workrates. Our primary hypothesis was that the replaced limb would exhibit decreased peak KEM and vertical PRF. This hypothesis was supported, in that both KEM and peak vertical PRF were significantly lower in the replaced limb compared to the non-replaced limb (Table 1).

Peak KEM and vertical PRF were on average 21.3% and 5.3% lower on the replaced limb across both workrates, respectively. These results indicate a large deficit in KEM which suggest a quadriceps avoidance strategy for the TKA patients in their replaced limb. Decreases in peak vertical PRF indicate a decrease of lower extremity loading for the replaced limb and are consistent with the observed decreased KEM (22, 45). The large decreases in KEM may be further explained by a significant decrease in posterior PRF (16.3%) and decreased knee extension range of motion (3.8%). A combination of both the vertical and posterior PRFs contributes to the magnitude of KEM and, the decreased posterior PRF is likely a driving factor for lower KEM found in the replaced limb. Therefore, it is essential to also examine the anterior/posterior PRF during cycling, as the differences between limbs are greater compared to those for the vertical PRF. Our results are similar to those found in TKA gait literature examining KEM (31, 47, 318). The replaced limb has shown 5.7% reduction of peak KEM compared to the non-replaced limb during level walking (47), 12.2% - 20.0% for stair ascent (31, 318), and 25.0% during walking on a 10° incline (47). The 21.3% deficit of peak KEM observed for replaced limb during stationary cycling is larger than those during level walking, and similar to stair ascent and ramp walking. The disproportionate loading between the replaced and non-replaced limb should be considered when prescribing stationary cycling for TKA patients.

Our secondary hypothesis was that there would be greater peak KEM and vertical PRF at 100 W compared to 80 W. Our hypothesis was partially supported, with 8.0% greater peak vertical PRFs at 100 W compared to 80 W (Table 1). Previous cycling work depicting the impact of workrate on vertical PRF have found similar results in healthy individuals (56, 323). When workrate was increased from 60 W to 90 W, peak vertical PRF increased by 15.6% in healthy college aged individuals (324). Similarly, increases of 40 W (80 W to 120 W), yielded a 16.1% increase in vertical PRF (323). While previous work of healthy participants had greater workrate changes, TKA patients experienced increased peak vertical PRF even at the smaller workrate increase of 20 W. However, peak KEM did not change significantly (1.4%) with an increase of workrate from 80 W to 100 W (Table 1). This is contrary to the results of increasing workrate in a healthy population, which found increases of 22.3% in peak KEM due to increases of workrate from 60 W to 90 W (56). While vertical PRF in our study did increase due to workrate, there was no significant change in peak posterior PRF. This may partially explain why no change in KEM was found for our TKA patients. It may also be that these TKA patients may still be attempting to avoid use of their quadriceps by using other joints to accommodate the increased demand of the 100 W condition. The peak hip extension and ankle plantar flexion moments increased due to increased workrate (Table 2). The peak hip extension moment is typically achieved early in the crank cycle, around 15° (Figure 2a), which seems to make up the KEM deficit during early power phase. Conversely, the peak ankle plantarflexion moment occurs at the transition from the power phase to the recovery phase, about 180° (Figure 2c). The increased peak plantarflexion moment seems to help lower limb complete the power phase and transition into the recovery phase with the presence of KEM deficit at the higher workrate. These results suggest that the TKA patients may rely on the hip extensors and ankle plantar flexors to compensate for weak knee extensors at a workrate higher than 80 W in stationary cycling.

Our tertiary hypothesis was that there would be no significant interactions of limb and workrate on peak KEM or vertical PRF. This hypothesis was supported for both variables (Table 1). These findings suggest that any limb differences for key biomechanical variables for TKA patients do not exacerbate further due to increases in workrate. When workrate was increased from 80 W to 100 W, responses in peak KEM and vertical PRF were similar for both the replaced and non-replaced limb. However, it is unknown if greater increases in workrate would elicit similar responses.

This study is not without its limitations. First, the non-replaced limb for these TKA patients used for comparisons were not equivalent to healthy limbs. Thirteen of the participants had been diagnosed with knee OA in their non-replaced limbs. However, all patients did not report any issues walking or pain in their non-replaced knees. This limitation was unavoidable, as many elderly TKA patients will have some degree of knee OA in their contralateral limbs. Second, we only examined the lower extremity biomechanics during short bouts of cycling. Third, this study only examines the acute difference of lower extremity biomechanics between limbs and workrates. There is no direct indication of whether any limb deficits lead to negative consequences and if symmetrical patterns are desirable for long-term health. More research is needed to ascertain the long-term biomechanical effects of cycling and its implications for TKA patients. Finally, we only examined KEM as it can be correlated to tibiofemoral joint contact forces. Future research should use musculoskeletal modeling approach to examine tibiofemoral contact forces and related muscle forces in stationary cycling for unilateral TKA population. Comprehending the contact forces for the tibiofemoral joint and limb deficits can begin to inform rehabilitation protocols for TKA patients using stationary cycling.

Conclusions

TKA patients exhibit decreased peak KEM and vertical PRF in their replaced limb compared to their non-replaced limb during stationary cycling. However, they do not increase their knee joint moments when workrate is increased from 80 W to 100 W. To accommodate the increased demand, TKA patients may relay more greatly on their hip and ankle extension moments. Finally, increasing workrate did not exacerbate the inter-limb differences for peak KEM or vertical PRF. These limb deficits during stationary cycling are similar to those found during gait.

Table 2. Peak pedal reaction forces (N) and knee joint moments (N•m) for the replaced and non-replaced limbs at 80 and 100 W.

	80 W		100 W		*P* values		
	Replaced Limb	Non-Replaced Limb	Replaced Limb	Non-Replaced Limb	Int	Limb	Workrate
Vertical PRF*%	210.2±42.2	227.3±43.7	233.9±39.3	241.7±43.2	0.144	**0.038**	**<0.001**
Posterior PRF*	-54.4±15.9	-65.7±22.1	-55.5±15.7	-65.6±17.8	0.677	**0.018**	0.855
Medial PRF%	-14.4±24.6	-16.5±27.9	-27.8±26.3	-23.4±29.9	0.811	0.564	**0.026**
KEM*	18.7±5.2	24.7±6.9	19.8±5.1	24.2±7.0	0.375	**0.034**	0.750
KAbM	-7.4±4.0	-13.1±7.5	-12.3±9.0	-12.8±6.2	0.204	0.376	0.146

Notes: * - main effect of limb, % - main effect of workrate, Bolded *p* values are significant at 0.05 level, Int: interaction, PRF: pedal reaction force, KEM = knee extension moment, KAbM = knee abduction moment.

Table 3. Knee joint range of motions (°) and peak ankle and hip joint sagittal plane moments (N•m) for the replaced and non-replaced limbs at 80 and 100 W.

	80 W		100 W		*P* values		
	Replaced Limb	Non-Replaced Limb	Replaced Limb	Non-Replaced Limb	Int	Limb	Workrate
Extension ROM *	70.8±5.7	73.4±5.3	70.0±5.9	72.9±5.0	0.748	**0.003**	0.688
Abduction ROM #*	-13.3±3.8	-8.4±3.8[a,c]	-13.0±3.8[a,c]	-9.6±4.4	**0.010**	**0.001**	0.058
Plantarflexion moment %	-15.5±5.3	-16.8±7.5	-18.2±5.4	-19.6±8.9	0.945	0.196	**0.017**
Hip Extension moment %	-26.5±9.2	-26.9±6.6	-30.9±11.2	-32.0±7.9	0.658	0.465	**0.009**

Notes: # - interaction between limb and workrate, * - main effect of limb, % - main effect of workrate, [a] Significantly different from Replaced at 80 W, [c]Significantly different from Non-Replaced at 100 W. Int: interaction. Bolded values are significant at 0.05 level, ROM : range of motion.

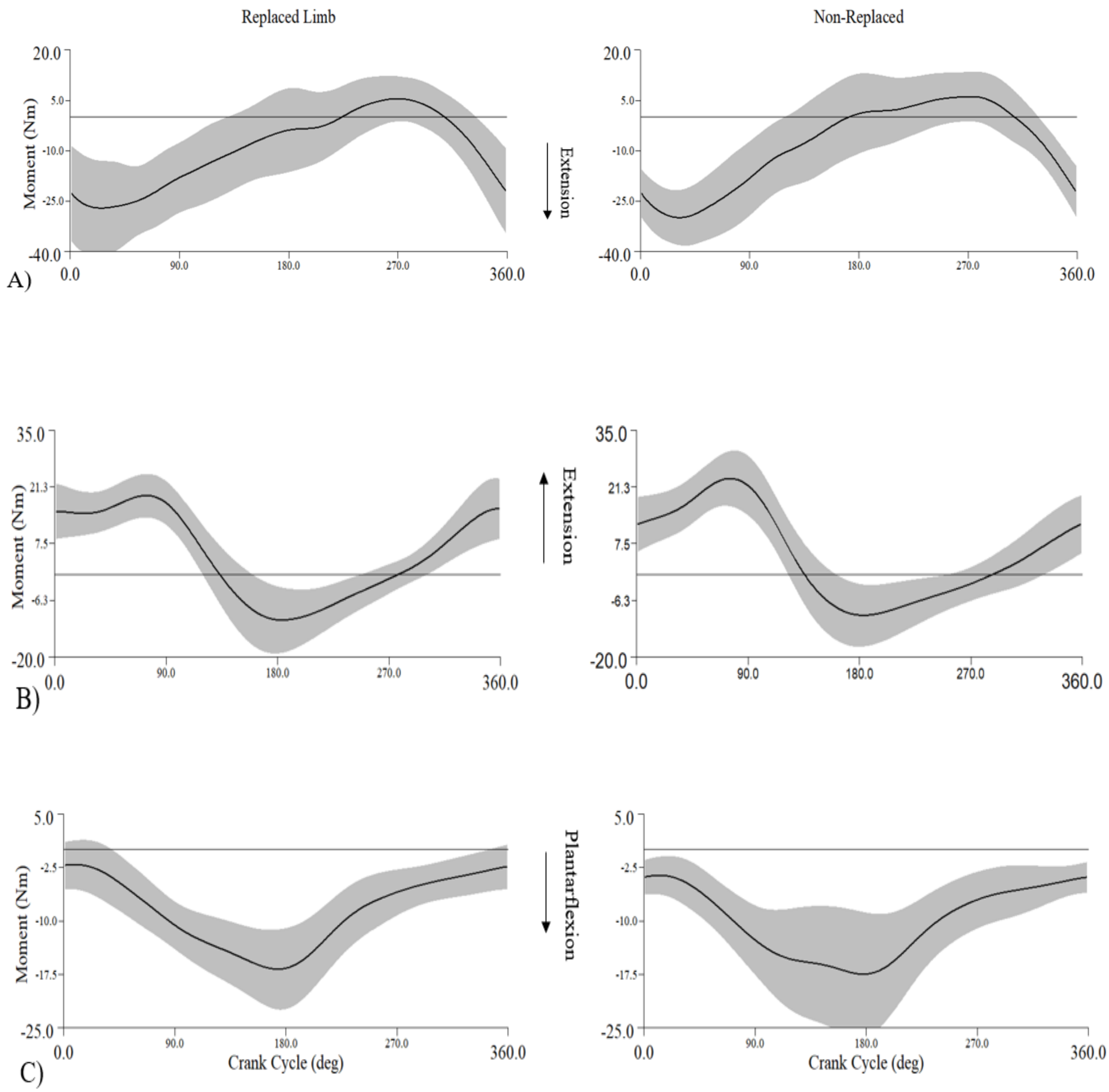


Figure 2. Ensemble curves of a) hip extension moment, b) knee extension moment, and c) ankle plantarflexion moment for both the replaced and non-replaced limbs across an entire crank cycle (deg) at the 100 W condition.

CHAPTER V

Effects of a Short-Term Cycling Intervention on Knee Joint Biomechanics of Patients with Unilateral Total Knee Arthroplasty during Stationary Cycling

Abstract

Following total knee arthroplasty (TKA), there are limb deficit has been found for both vertical ground reaction force and internal knee extension moment (KEM). One key factor that is could influence limb deficits post-operation is the rehabilitation and physical activity. Currently, it is unknown how stationary cycling impacts the lower limb biomechanics for patients following TKA during stationary cycling. The purpose of this study was to examine the effect of a short-term cycling intervention paired with augmented feedback on lower extremity biomechanics for patients following TKA. We hypothesized that the inter-limb asymmetries for peak KEM and vertical pedal reaction force (PRF) would decrease following the intervention. Ten unilateral TKA patients participated in two cycling conditions (80 and 100 Watts) at 80 RPM before and after the intervention. Kinematics (240 Hz) and kinetics (120 Hz) were collected to calculate net joint moments. The intervention included six sessions of stationary cycling across two weeks. During cycling, participants were given augmented feedback on their vertical PRF and instructed to reduce their asymmetry to be less than 10%. Following the intervention, peak KEM AI decreased from 25.7% to 15.7% at 80 W (p = 0.499, $\eta^2_p = 0.052$) and 23.6% to 13.7% at 100 at 100 W (p = 0.395, $\eta^2_p = 0.092$). Peak vertical PRF AI decreased from 5.4% to -3.0% at 80 W (p = 0.256, $\eta^2_p = 0.141$) and 1.4% to -3.9% at 100 W (p = 0.479, $\eta^2_p = 0.064$). While not statistically significant, reductions in AI of 10% or greater could indicate clinically relevant changes. The cycling intervention paired with augmented feedback may have some clinical relevance for shifting the loading for replaced and non-replaced limbs to be more symmetrical during stationary cycling.

Keywords: knee extension moment, cycling, asymmetry index, visual feedback

Introduction

The rates of total knee arthroplasty (TKA) are increasing as an intervention for late-stage knee osteoarthritis (OA) (325). Like most surgical interventions, the post-operation rehabilitation and modes of rehabilitation exercise are critically important. Currently, no widely recognized rehabilitation guidelines exist for patients of TKA, like those presented for knee OA (150). However, a general consensus is that activities with lower tibiofemoral joint loading should be prioritized and those with high forces should be avoided (70, 76, 77).

One such lower impact activity is stationary cycling. Peak knee extension moment (KEM) for healthy individuals has shown increased values with greater workloads ranging from 11.6 Nm (0.15 Nm/kg) to 37.2 Nm (0.47 Nm/kg) (56). Additionally, patients with medial compartment knee OA displayed peak KEM of 27.9 Nm (0.35 Nm/kg) at a workrate of 80 Watts (197). The peak KEM during gait is greater, typically ranging from 0.33-0.35 Nm/kg for patients of TKA and 0.49-0.57 Nm/kg for healthy participants (47). These differences have been further supported by the results with *in vivo* measurements of knee contact forces. Tibiofemoral compressive forces during cycling ranged from 1.03 – 1.19 body weight (BW) (82, 185) compared to 2.05 – 2.60 BW for walking (185). In addition, there have been deficits for patients of TKA for peak vertical ground reaction force (GRF) that follows peak KEM, with decreased vertical GRF in the replaced limb (31, 45, 48). Decreased peak vertical GRF indicates a reduced overall loading for the replaced limb and highly corresponds with peak KEM, which coincides typically with patient's pain levels or quadriceps weakness (8, 32).

Although these suggestions appear to be merited, there is still very little research evidence on the efficacy of stationary cycling post TKA. Liebs et al. (2010) found no difference in a standard care rehabilitation program compared to a stationary cycling centric rehabilitation program following TKA for scores on Knee Injury and Osteoarthritis Outcome Score (KOOS) indicating no benefit from cycling based on functional KOOS scores. However, Liebs et al. (2010) performed their cycling intervention very early into rehabilitation with no progression of intensity (cadence and resistance), and effects of cycling intervention might have been reduced by pain and quadriceps avoidance. To our knowledge, there has been no studies examining the impact of cycling rehabilitation following TKA on biomechanical objective outcomes, such as knee joint loading, as well as the impact of cycling during the later stages of TKA rehabilitation. Understanding the impact of cycling rehabilitation on knee joint biomechanics may provide evidence-based recommendations for the use of cycling in patients of TKA.

The current literature on TKA biomechanics during stationary cycling are sparse. Our initial work with patients of TKA during stationary cycling have found significant inter-limb differences of knee joint kinetics (326). The replaced limbs of unilateral patients of TKA displayed lower peak KEM compared to their non-replaced limbs in ergonomic cycling. These findings of reduced peak KEM in stationary cycling are in agreement with similar results found in gait (31, 47, 318). Reductions of KEM may be also accompanied by a stiff-knee gait which is attributed to quadriceps avoidance. This avoidance could be due to quadriceps weakness or knee joint pain following TKA (20, 50). With the replaced limb deficit in KEM persisting longer than one-year post-operation, the focus on rehabilitation is key. Since cycling has lower tibiofemoral joint loading (185), are a quadriceps dominant exercise (327, 328), and enhance cardiovascular health (329), it may be a safe and effective rehabilitation post-operation for patients of TKA.

Therefore, the purpose of this study was to examine the effects of a short-term cycling training program with augmented feedback of vertical PRF on asymmetries of KEM and inter-limb differences in patients of TKA. Our hypothesis was that the inter-limb asymmetries for peak vertical PRF and KEM would decrease from pre- to post-training.

Methods

Participants

A total of ten participants were recruited and participated in the current study (Table 4). Participants took part in two testing sessions (pre- and post-training), with the post-training session occurring after three weeks following their training intervention. Participants were recruited from a local orthopaedic clinic after undergoing a unilateral TKA by the same surgeon. Participants were enrolled in the study if they met the following criteria: between 6- and 18-months post TKA and were within the age range of 50-80 years of age. Potential participants were excluded from the current study if they had: debilitating OA of other lower extremity joints that impacted how they walk, required walking aids (e.g. canes, walkers, etc..), arthroplasty of other lower extremity joints, body mass index greater than 38 kg/m^2, neurological disease that impacted walking or balance, systemic inflammatory arthritis, arthroscopic surgery within six months, lower extremity injury within past six months, and had a pain level greater than five in their replaced knee. All participants passed a Physical Readiness Questionnaire (PARQ) to ensure safety of exercise. All procedures for the current study was approved by the University Institutional Review Board. All participants reviewed and signed an informed consent prior to participation.

Instrumentation

Three-dimensional kinematics was collected using a twelve-camera motion capture system (240 Hz, Vicon Motion Capture Inc., Oxford, UK). Anatomical reflective markers were placed on the following landmarks bilaterally: acromion processes, iliac crests, greater trochanter, medial and lateral femoral epicondyles, medal and lateral malleoli, 1st and 5th metatarsal heads, and the distal phalanx of the 2nd toe. Reflective tracking markers affixed to semi-rigid plastic were attached to the trunk, thighs, and shanks bilaterally to track segment motion. An additional four reflective markers were placed on the anterior lateral aspects of each foot, three placed on the lateral side of each pedal, and finally a single marker on the anterior aspect of each pedal.

In conjunction with kinematics, three-dimensional pedal reaction force data were collected, using two customized instrumented pedals. Each pedal was outfitted with two tri-axial force sensors (1200 Hz, Type 9027C, Kistler, Switzerland) along with two amplifiers (Type 5073A, Kistler, Switzerland). Both kinematics and kinetics were collected simultaneously using Vicon Nexus (version 2.8.2, Vicon Motion Capture Inc., Oxford, UK).

Testing Protocol

On each testing day, participants completed stationary cycling at two conditions based on workrates of 80 W and 100 W. Conditions were randomized prior to their arrival each session. Participants first completed a three-minute warm up at a self-selected speed on a treadmill. Following the warm-up, participants were instrumented with the reflective markers and changed into a pair of standard laboratory shoes (Pegasus 32, Nike Inc., Portland, OR, USA). Next, the stationary ergometer was fitted to each participant. The saddle height was set to 30° knee flexion angle while the pedal was at the bottom dead center measured via handheld goniometer (56, 197, 323). The saddle fore-aft position was adjusted to ensure the anterior aspect of the knee was directly above the pedal spindle while at the three o'clock position (197).

Once the stationary ergometer was fitted, the workrate was set to the first condition (80 W or 100 W). The two conditions were randomized prior to each participant arrival. Participants warmed up for three minutes at their first workrate. For each condition, participants cycled for one minute, with a ten second data collection occurring in the final ten seconds. A minimum of one-minute rest was given between conditions.

Training Protocol

Training sessions took place across two-three weeks. Each session consisted of multiple five-minute bouts of cycling exercise. The first session consisted of two bouts, the second session consisted of three bouts, and the remaining sessions had four bouts. In between each bout, participants were allowed to rest a minimum of one minute. Cycling for each bout was done at a constant cadence of 80 RPM. Workrates were controlled between each bout of exercise, with participants starting at 60 W. Workrates were moderated between bouts based on three factors: i) RPE, ii) VAS pain, and iii) asymmetry index. Decisions on maintaining, increasing, or decreasing workrate between workrates are described in Appendix H. The goal of these criteria was to minimize pain levels, maintain a moderate intensity of exercise, and to minimize asymmetries as much as possible.

In addition to the cycling exercise, participants were given visual augmented feedback of their vertical PRF data for both their replaced and non-replaced limbs (100). Within each bout, feedback was presented on a computer monitor directly in front of participants. Vertical PRF data were collected using Vicon Nexus and immediately processed using Matlab. The vertical PRF data was first lowpass filtered using a fourth order zero lag Butterworth filter with a cutoff frequency of 6 Hz (38). Next, each individual peak vertical PRF was found corresponding to each individual crank cycle. Finally, an average was taken across all peak vertical PRF for both the replaced and non-replaced limb.

Feedback was given in a bar graph with two bars, corresponding right or left limbs of participants. Participants were given a target range of ±10% of their non-replaced limbs data. Participants were instructed to keep both bars within this target range, to reduce the asymmetry between the two limbs to be less than 10% (153). Participants were all instructed at the first session on how to interpret the feedback and the feedback figure (Figure 1). Throughout the rest of training, participants were only reminded of which bar corresponded to which leg.

Data Analyses

Five individual trials consisting of a full crank cycle for each limb were truncated from the ten seconds of marker trajectory and PRF data. A full crank cycle consists of starting at the top center position (0° crank angle) and completes at the return of the same position (360° crank angle). Kinematics and kinetics were calculated in Visual3D (Version 6.01, C-Motion Inc., Germantown, MD, USA). Marker trajectories and PRF data were filtered using a fourth-order zero lag Butterworth lowpass filter at a cutoff frequency of 6 Hz (56, 197). Angular joint kinematics were calculated using the joint coordinate system and a Cardan rotational sequence (X-Y-Z) (306). Net joint moments were calculated via inverse dynamics. Conventions of kinematic and joint moments were expressed using the right-hand rule. Peak variables were identified using a custom program (VisualBasic 6.0, Microsoft, Redmond, WA, USA). Net joint moments and PRF data were not normalized to body mass, as the majority of the participant's mass was supported by the saddle and handlebars (56, 197).

Asymmetry index (AI) for each peak variable: vertical PRF, KEM, and KAbM were computed based on the replaced and non-replaced limbs (Equation 11) and reported as percentages (%).

$$AI = \frac{X_{non-replaced} - X_{replaced}}{X_{non-replaced}} * 100 \quad (11)$$

Where $X_{non\text{-}replaced}$ is the peak variable for the non-replaced limb and $X_{replaced}$ is the variable for the replaced limb. A negative AI value indicates a replaced limb asymmetry, whereas a positive value is indicative of an asymmetry of the non-replaced limb. An AI of zero indicates a complete symmetry, whereas increasing AI values away from zero indicates larger asymmetries. An AI was computed for each variable were exported and used for statistical analysis. The following variables were also included in the analyses: vertical PRF, posterior PRF, KEM, knee extension range of motion (ROM), knee abduction ROM, hip extension moment, and ankle plantar flexion moment.

Statistical Analyses

A one-way repeated measures analysis of variance (ANOVA) was run on AI of peak vertical PRF, posterior PRF, and KEM comparing pre- and post-training measurements at 80 and 100 W separately. A 2 x 2 (limb x time) repeated measure ANOVAs were run on the mean data for: vertical PRF, posterior PRF, KEM, knee extension ROM, knee abduction ROM, hip extension moment, and ankle plantar flexion moment at 80 and 100 W separately. Alpha levels were set *a priori* of 0.05. Effect sizes were reported as partial eta squared (η^2_p) and were interpreted as small ($\eta^2_p < 0.06$), medium ($0.06 \leq \eta^2_p < 0.15$), and large ($\eta^2_p \geq 0.15$) (310).

Results

For the AI, there was no significant difference for peak vertical PRF AI between pre- and post-training at 80 W ($F_{(1,9)} = 1.472$, $p = 0.256$, $\eta^2_p = 0.141$) or 100 W ($F_{(1,8)} = 0.550$, $\eta^2_p = 0.479$, $d = 0.064$ Table 5). No significant difference was found for peak posterior PRF AI at both 80 W ($F_{(1,9)} = 0.043$, $p = 0.840$, $\eta^2_p = 0.005$) and 100 W ($F_{(1,8)} = 1.199$, $p = 0.305$, $\eta^2_p = 0.130$). Finally, no significant difference based on time was found for peak KEM AI for 80 W ($F_{(1,9)} = 0.496$, $p = 0.499$, $\eta^2_p = 0.052$) and 100 W ($F_{(1,16)} = 1.908$, $p = 0.395$, $\eta^2_p = 0.092$).

The ANOVA results for discrete variables showed a significant effect of time for peak posterior PRF at 100 W ($F_{(1,8)} = 10.588$, $p = 0.012$, $\eta^2_p = 0.570$), which was increased following the intervention compared to pre-training values (Table 6). Additionally, there was significantly greater posterior PRF in the non-replaced limb (Table 6 and 7) at both 80 W ($F_{(1,9)} = 29.269$, $p < 0.001$, $\eta^2_p = 0.765$) and 100 W ($F_{(1,8)} = 17.866$, $p = 0.003$, $\eta^2_p = 0.691$). There were also significant main effects of limb for peak KEM at both 80 W ($F_{(1,9)} = 24.804$, $p = 0.001$, $\eta^2_p = 0.734$) and 100 W ($F_{(1,8)} = 21.006$, $p = 0.002$, $\eta^2_p = 0.724$) with lower peak KEM for the replaced limb. A significant effect of limb was present for knee extension ROM at 80 W ($F_{(1,9)} = 24.006$, $p = 0.001$, $\eta^2_p = 0.727$) and 100 W ($F_{(1,8)} = 13.043$, $p = 0.007$, $\eta^2_p = 0.620$) with the replaced limbs displaying decreased ROM compared to the non-replaced limbs (Tables 6 and 7. Finally, knee abduction ROM showed a significant main effect of limb at 80 W ($F_{(1,9)} = 30.560$, $p < 0.001$, $\eta^2_p = 0.772$) and 100 W ($F_{(1,8)} = 7.148$, $p = 0.028$, $\eta^2_p = 0.472$). The replaced limbs displayed a greater amount of knee abduction ROM compared to the non-replaced limbs.

Discussion

The purpose of this study was to examine the effects of a short-term cycling training program with augmented feedback of vertical PRF asymmetry on asymmetries of KEM and inter-limb differences in patients with TKA. Our hypothesis was that the inter-limb AI for peak vertical PRF and peak KEM would be reduced at post-training compared to the pre-training values.

Our hypothesis was rejected, with no significant differences found for the AI for peak vertical PRF, posterior PRF, and KEM after training. While no significant effect was found, there was a medium effect size ($\eta^2_p = 0.141$) for peak vertical PRF AI at 80 W. Following the training program, the range of AI changes was 8.4% (from 5.4% to -3.0%), leading to a shift of asymmetry towards the replaced limbs, but still within the acceptable 10% AI range. The effect was similar at 100 W ($\eta^2_p = 0.064$) but not quite as profound, also with an AI shift towards the replaced limbs. While the augmented feedback provided information on the vertical PRF, the posterior PRF AI showed similar shifts towards the replaced limbs at both 80 and 100 W due to cycling training. This AI shift for posterior PRF was more profound at 100 W compared to 80 W. Both of these changes in vertical and posterior PRF AI could have impacts on peak KEM AI, as both PRF components are directly linked to KEM. These results indicate slightly increased vertical loading to the replaced limb following the short-term intervention. The analysis of related discrete sagittal-plane loading variables supported these findings, showing no significant interaction for either peak vertical or posterior PRF at both workrates. There were also main effects of limb for both variables, indicating that patients with TKA displayed a decreased sagittal-plane PRF (combination of vertical and posterior PRF) in their replaced limbs.

The current study did not find a significant change due to time for peak KEM AI (Table 3). The peak KEM AIs at both 80 W and 100 W indicated greater loading of the non-replaced limbs and a deficit in the replaced limbs, which is commonly observed in gait for patients following TKA (31, 318). Interestingly, the patients with TKA in this current study were relatively high functioning and very little reported pain before training, and still had mean peak KEM AI ranging from 23.6 – 25.7%. Along with the KEM AI, the replaced limbs at pre- and post-training displayed decreased knee extension ROM during the power phase in cycling. The goal of this training program with augmented feedback was to reduce any present AI of vertical PRF to be smaller than 10%, as previous work indicates this may be clinically relevant (153). While the goal of this intervention was to reduce vertical PRF AI, the KEM AI was indirectly impacted. Even though the peak KEM AI was not reduced to less than 10%, we did see approximate 10% reductions of AI. While these reductions did not reach statistical significance, they did show medium effect sizes. These reductions of KEM asymmetry may be a result of combined changes observed in peak vertical PRF and posterior PRF AIs. Both vertical and posterior PRF AIs changed non-significantly towards the replaced limb following training, suggesting more knee extension muscle efforts by the replaced limbs during power phase. However, our patients with TKA did not display an AI for peak vertical PRF greater than 10%, whereas both KEM (23.6 – 25.7%) and posterior PRF (17.5 – 20.6 %) AIs were larger than the proposed 10% clinically significant threshold (153). Utilizing a more sensitive variable such as KEM may be more effective in giving feedback during cycling training compared to vertical PRF, as it is a directly related to knee joint loading and asymmetry in patients with TKA. The cycling training could be beneficial for rehabilitation for individuals following TKA who may still have quadriceps deficits, avoiding loading their replaced limbs (32, 45).

The use of cycling training with augmented feedback appeared to not have any significant impact of the sagittal plane joint moments for both the hip and ankle. While patients of TKA displayed deficits for their knee joint of replaced limbs, it did not appear that these individuals compensated significantly via hip or ankle of their replaced limbs during the power phase of cycling. In our limb comparisons, the peak hip extension and ankle plantarflexion moments were similar between replaced and non-replaced limbs. The cycling intervention used in the current study focused on providing augmented feedback of vertical PRF during training, but also increased in intensity (workrate) over time. Previous work has found that increases in workrate saw increases in net joint moments of the lower limb during stationary cycling (56). The progressive workrate increases in the current intervention may have also contributed to the training effect and reduction of KEM and PRF asymmetries.

The current study is not without its limitations. First and foremost, due to difficulties with participant recruitment and testing due to COVID-19, we were unable to fully collect the rest of recruited participants and finish the training of two participants to reach the desired statistical power. The smaller sample size may have partially to the lack of significant changes in KEM AI after cycling training. In this, we were unable to include a full control group to compare the intervention group against. The originally proposed control group would have participated in two testing sessions, separated by a control period similar in duration to the intervention. This comparison would aid in examining if changes seen in the intervention group were due to the intervention, or potentially due to time and recover. Second, the intervention used in the current study was only conducted over six sessions. The long-term impacts and implications of such training programs remain unclear for biomechanical asymmetries. Third, we used peak vertical PRF data as our feedback variable, primarily to elicit changes in KEM. Our participants did not show significant vertical GRF asymmetry in their replaced limbs prior to training, but they did show approximately 25% asymmetry in their knee extension moment. Further work should use KEM as control and feedback variable, as this may be a more sensitive measurement of knee joint loading and related asymmetry.

In summary, the short-term cycling intervention paired with visual augmented feedback did not significantly impact the AI for key peak vertical and posterior PRFs. Peak KEM asymmetry did show an approximate 10% reduction following the intervention. These findings may indicate a clinically relevant decrease in asymmetry of knee extension moment. Further work is needed to fully explore benefits of the cycling training program including more participants and for longer intervention durations.

Table 4. Participant characteristics for the intervention and control groups (Mean ± STD)

Age (yr)	64.8 ± 7.7
Mass (kg)	89.2 ± 21.3
Height (m)	1.7 ± 0.1
Post-Op (mo)	8.6 ± 2.4

Table 5. Asymmetry index (AI) for peak vertical PRF, posterior PRF, and KEM (%) at pre- and post-training for 80 and 100 Watts. Mean ± STD.

	80 Watts			100 Watts		
	Pre-Training	Post-Training	P(η^2_p)	Pre-Training	Post-Training	P(η^2_p)
Vertical PRF	5.4±8.2	-3.0±20.7	0.256(0.141)	1.4±11.7	-3.9±12.6	0.479(0.064)
Posterior PRF	17.5±17.5	15.7±15.1	0.840(0.005)	20.6±19.0	12.7±13.9	0.305(0.130)
KEM	25.7±23.9	15.7±26.7	0.499(0.052)	23.6±14.2	13.7±16.2	0.395(0.092)

PRF: pedal reaction force, KEM: knee extension moment, η^2_p: partial eta squared effect size

Table 6. Mean peak data for secondary variables for pre- and post-training at 80 W (mean ± STD)

	Pre-Training		**Post-Training**		**P(η^2_p)**		
	Non-Replaced	**Replaced**	**Non-Replaced**	**Replaced**	**Inter**	**Limb**	**Time**
Vertical PRF (N)	233.2±43.9	209.8±43.8	221.0±39.2	223.6±40.9	0.210(0.188)	0.534(0.050)	0.622(0.032)
Posterior PRF (N)	-66.1±21.7	-52.3±16.9	-67.6±19.1	-56.9±18.0	0.627(0.027)	**<0.001**(0.765)	0.066(0.328)
KEM (Nm)	25.5±6.6	17.9±5.2	24.5±6.9	20.1±7.3	0.405(0.078)	**0.001**(0.734)	0.681(0.020)
Knee Ext ROM (°)	72.9±5.7	70.2±6.8	71.5±5.7	67.8±4.7	0.292(0.122)	**0.001**(0.727)	0.211(0.168)
Knee Abd ROM (°)	-8.2±4.2	-13.5±3.8	-8.0±4.0	-11.2±3.2	0.198(0.177)	**<0.001**(0.772)	0.267(0.134)
Hip Ext Moment (Nm)	-24.7±4.9	-25.2±8.5	-26.7±5.7	-29.1±5.2	0.433(0.070)	0.253(0.142)	0.176(0.193)
Ankle PF Moment (Nm)	-16.4±9.0	-15.0±6.2	-15.5±6.7	-15.1±5.5	0.407(0.078)	0.352(0.097)	0.764(0.010)

Bolded values indicate a significant effect ($p < 0.05$). PRF: pedal reaction force, KEM: knee extension moment, Ext: extension, Abd: abduction, PF: plantar flexion, η^2_p: partial eta squared effect size, Inter: Interaction

Table 7. Mean peak data for secondary variables for pre- and post-training at 100 W (mean ± STD)

	Pre-Training		**Post-Training**		**P(η^2_p)**		
	Non-Replaced	**Replaced**	**Non-Replaced**	**Replaced**	**Inter**	**Limb**	**Time**
Vertical PRF (N)	249.3±47.9	209.8±43.8	212.9±83.7	218.0±81.4	0.473(0.066)	0.922(0.001)	0.300(0.133)
Posterior PRF (N)	-71.0±18.7	-55.4±18.4	-75.2±19.7	-64.0±16.6	0.410(0.086)	**0.003**(0.691)	**0.012**(0.570)
KEM (Nm)	27.0±6.0	19.9±5.3	28.1±7.2	23.5±6.8	0.383(0.096)	**0.002**(0.724)	0.053(0.391)
Knee Ext ROM (°)	72.5±5.7	69.4±7.0	71.5±6.2	67.5±5.3	0.455(0.072)	**0.007**(0.620)	0.393(0.455)
Knee Abd ROM (°)	-9.5±5.0	-13.6±3.7	-9.4±4.7	-11.5±3.3	0.2360.170)	**0.028**(0.472)	0.355(0.107)
Hip Ext Moment (Nm)	-30.7±6.4	-31.6±9.4	-28.0±9.2	-32.0±6.6	0.329(0.119)	0.225(0.178)	0.695(0.020)
Ankle PF Moment (Nm)	-19.0±9.7	-17.6±6.2	-18.1±8.2	-17.5±5.8	0.570(0.042)	0.497(0.060)	0.751(0.013)

PRF: pedal reaction force, KEM: knee extension moment, Ext: extension, Abd: abduction, PF: plantar flexion, η^2_p: partial eta squared effect size, Inter: interaction

CHAPTER VI

Transfer Effects of a Short-Term Cycling Intervention on Asymmetries and Knee Joint Biomechanics in Gait for Patients following Unilateral TKA

Abstract

Current rehabilitation recommendations for total knee arthroplasty patients includes stationary cycling. However, it is currently unknown if cycling rehabilitation has any impact on the inter-limb loading deficits of vertical ground reaction force (GRF) and knee extension moment (KEM), which are prevalent in this patient population during gait. The purpose of this study was to examine the effect of a short-term cycling intervention paired with augmented feedback on the asymmetries of vertical GRF and KEM, and other knee joint biomechanics in patients of TKA during gait. We hypothesized that vertical GRF and KEM asymmetries would be reduced post-training. Ten unilateral patients of TKA participated in two testing sessions separated by six training sessions over 2-3 weeks. Three-dimensional kinematics (240 Hz) and GRFs (1200 Hz) were collected at preferred and fast speed (preferred + 0.4 m/s). Six training sessions included five-minute bouts of stationary cycling with progressive workrate (intensity) increases and visual feedback of vertical pedal reaction forces. No differences were found for both the loading-response and push-off vertical GRF asymmetry at preferred ($p = 0.641$, $p = 0.229$) and fast ($p = 0.600$, $p = 0.303$) speed after intervention. The intervention also had no effect on the loading-response or push-off KEM asymmetry at preferred ($p = 0.363$, $p = 0.225$) or fast ($p = 0.267$, $p = 0.144$) speed. The intervention may not have included enough training days to have caused beneficial changes in knee asymmetries during gait. The use of direct measures of knee joint loading, such as KEM, may prove to be more applicable with patients of TKA.

Keywords: asymmetry, knee extension moment, ground reaction force, cycling

Introduction

The rate of total knee arthroplasty (TKA) are estimated to be on a rise with an increased incidence of knee osteoarthritis (OA) (330). While the goal of TKAs are to alleviate pain and restore knee joint function, there are concerns post-operation that the operation may not fully restore function (17). One significant concern post TKA is the loading deficit of the replaced limbs typically associated with decreased knee flexion angles during gait (9, 20). This decrease of knee flexion is proposed to be due to pain and is commonly referred to as "stiff knee gait". The decreased knee flexion is commonly accompanied with peak internal knee extension moment (KEM) and quadriceps weakness (31, 47, 318). These inter-limb deficits of loading have been a cause of concern for other joint arthroplasties in the contralateral limb or revision of the primary TKA (50).

Currently, there are no widely recognized rehabilitation guidelines post TKA (331), as seen for knee OA (150). The common goals for TKA rehabilitation exercise are to increase knee joint range of motion (ROM), moderate pain levels , regain function for activities of daily living, increase strength, and manage weight (19, 76). One recommendation is to participate in low loading activities (e.g. stationary cycling) to avoid loosening of and wear on the TKA implant (70, 76, 83, 185). Stationary cycling appears to be beneficial for patients, meeting the aforementioned ideal criteria for exercise post TKA. Stationary cycling is widely used for rehabilitation post-TKA. However, to our knowledge there has only been one study examining the impact of a cycling focused rehabilitation program for patients of TKA (83). The TKA patients participated in cycling rehabilitation starting two weeks post-operation and did not show any improvements in physical function or pain up to twenty-four months post-operation (83).

One aspect of rehabilitation exercise is the benefits of exercise transferring to other activities. Currently, there is a lack of literature on the transfer effects from stationary cycling-based training to gait. If patients following TKA are able to adequately exercise using stationary cycling, this may aid in addressing loading deficits found in gait through lower limb muscle strengthening. Another aspect used in some training programs is the use of augmented feedback to optimize or alter human movement to provide additional benefits (100, 253). Augmented feedback (e.g. visual feedback) has been used previously to address the inter-limb deficits for peak KEM in patients following anterior cruciate ligament repair (ACLR) (100). Individuals were able to manipulate their GRF while given feedback that directly impacted their peak KEM (100). Peak KEM was greater during the high loading condition (±5% vertical GRF) compared to the low loading condition (-5% vertical GRF). A combination of cycling training with augmented feedback on vertical PRF may be beneficial in reducing the KEM inter-limb deficits found following TKA not only in cycling but may also be reflected in gait. If the inter-limb deficits can be address during training, then there could be a transfer to the inter-limb deficits found during gait.

Therefore, the purpose of this study was to examine the impact of a short-term cycling intervention with augmented feedback had on vertical GRF and KEM asymmetry index (AI) as well as the knee joint biomechanics in gait for patients of TKA. We hypothesized AI would be reduced for peak vertical GRF and KEM in gait at both preferred and fast speeds following the cycling intervention.

Methods

Participants

A total of ten participants were recruited from a local orthopaedic clinic after undergoing a unilateral TKA by the same surgeon and participated in the current study (age: 64.8±7.7 yrs, mass: 89.2 ± 21.3 kg, height: 1.7 ± 0.1 m, months post operation: 8.6 ± 2.4). One participant was unable to complete the fast walking condition at post-training. Participants were enrolled in the study if they met the following criteria: between 6- and 18- months post TKA and were within the age range of 50-80 years of age. Potential participants were excluded from the current study if they had: debilitating OA of other lower extremity joints that impacted how they walk, required walking aids (e.g. canes, walkers, etc.), arthroplasty of other lower extremity joints, body mass index greater than 38 kg/m^2, neurological disease that impacted walking or balance, systemic inflammatory arthritis, arthroscopic surgery within six months, lower extremity injury within past six months, and had a pain level greater than five in their replaced knee. All participants filled out and passed a Physical Readiness Questionnaire (PARQ) to ensure safety of exercise. All procedures for the current study was approved by the University Institutional Review Board. All participants reviewed and signed an informed consent prior to participation.

Instrumentation

Three-dimensional kinematics was collected using a twelve-camera motion capture system (240 Hz, Vicon Motion Capture Inc., Oxford, UK). Anatomical reflective markers were placed bilaterally on acromion processes, iliac crests, greater trochanter, medial and lateral femoral epicondyles, medal and lateral malleoli, 1st and 5th metatarsal heads, and the distal phalanx of the 2nd toe. Tracking marker clusters of four reflective markers were attached to the trunk, thighs, shanks and feet.

In conjunction with kinematics, three-dimensional ground reaction force (GRF) data were recorded with an imbedded force platform (1200 Hz, Advanced Mechanical Technology Inc., Watertown, MA, USA). Both kinematic and kinetic data were recorded synchronously using Vicon Nexus (version 2.9, Vicon Motion Capture Inc., Oxford, UK).

Testing Protocol

Participants took part in two testing sessions (pre- and post-training), with the post-testing session occurring after three weeks following their training intervention. Participants first completed a three-minute warm up at a self-selected speed on a treadmill. Following the warm-up, participants were instrumented with the reflective markers and changed into a pair of standard laboratory shoes (Pegasus 32, Nike Inc., Portland, OR, USA). On each testing day, participants completed five level walking trials in four test conditions of two walking speeds (preferred and fast) for both limbs (replaced and non-replaced), respectively. The preferred speed condition was at a speed each participant would typically walk during the respective test day. Preferred speeds were determined as the average speed of three practice trials. The fast speed equals the preferred speed + 0.4 m/s. Successful trials included trials when the participant's speed was within ±10% of the desired speed and without targeting of force platform. The order of the test conditions was first randomized by speed and then by limb. Gait speed was recorded using two photocells (63501 IR, Lafayette Instrument Inc., IN, USA), which were placed three meters apart, and Universal Timer and software (Model 35930, Lafayette Instrument Inc., IN, USA) (Table 8). Participants were asked to rate their pain levels of their replaced limb using an enlarged Visual Numeric Scale (VNS) on a scale of 1-10 (Table 8). Pain levels were recorded upon arrival for testing during the pre- and post-training sessions (initial), as well as during the preferred and fast speeds of walking.

Training Protocol

Six training sessions took place across a period of 2 - 3 weeks. Each session consisted of multiple five-minute bouts of cycling exercise. The first session consisted of two bouts, the second session consisted of three bouts, and the remaining sessions had four bouts. In between bouts, participants were allowed to rest a minimum of one minute with no participants taking more than five minutes. Cycling for each bout was done at a cadence of 80 RPM. Workrates were controlled and progressed between each bout of exercise, with participants starting at 60 W and were increased as much as they could. Workrates were moderated between bouts based on three factors: i) RPE, ii) VNS pain, and iii) asymmetry index. Decisions on maintaining, increasing, or decreasing workrate between workrates are described in Table 9. The goal of these criteria was to minimize the vertical PRF AI, increase intensity (workrate), and maintain or reduce pain levels.

In conjunction with the cycling exercise, participants were given visual augmented feedback of the vertical PRF data for the replaced and non-replaced limbs. Feedback was presented on a computer monitor placed directly in front of participants. Thirty seconds of vertical PRF data were collected using Vicon Nexus and immediately processed using a Matlab program. Data was first collected at 20 seconds and then every minute thereafter (e.g. 20 seconds, 1 minute 20 seconds, 2 minute 20 seconds). Peak vertical PRF was found for every crank cycle, and the average was computed for each of the replaced and non-replaced limb, respectively. Feedback was given on a figure with two bars, corresponding to the average peak vertical PRF for right and left limbs of participants (Figure 1). Participants were given a target range of ±10% of the vertical PRF of their non-replaced limbs and were instructed to keep both bars within this target range. The goal during training was to reduce and maintain the asymmetry so the bars of both limbs would fall within ±10% (153). Participants were all instructed at the first session on how to interpret the feedback and the feedback figure. Throughout the rest of training, participants were only reminded of which bar corresponded to which leg.

Data Analyses

Five successful trials were collected for each speed condition for both the replaced and non-replaced limbs. Kinematics and kinetics were calculated in Visual3D (Version 6.01, C-Motion Inc., Germantown, MD, USA). Marker trajectories were filtered using a fourth-order zero lag Butterworth lowpass filter at a cutoff frequency of 6 Hz. Raw GRF data was filtered using a fourth-order zero lab Butterworth lowpass filter with a cutoff frequency of 50 Hz (47). Angular joint kinematics were calculated using the joint coordinate system (306) and a Cardan rotational sequence (X-Y-Z). Net joint moments were calculated via an inverse dynamics approach. Conventions of kinematic and joint moments were expressed using the right-hand rule. Peak variables were identified using a custom program (VisualBasic 6.0, Microsoft, Redmond, WA, USA). Net joint moments were normalized to body mass (Nm/kg) where as GRF data were normalized to body weight (BW). The vertical PRF data used during cycling training was first lowpass filtered using a fourth order zero lag Butterworth filter with a cutoff frequency of 6 Hz (56, 197) .

Asymmetry index (AI) for each peak variable: vertical GRF and KEM were computed based on the replaced and non-replaced limbs (Equation 12) and reported as percentages (%).

$$AI = \frac{X_{non-replaced} - X_{replaced}}{X_{non-replaced}} * 100 \quad (12)$$

Where $X_{non\text{-}replaced}$ is the peak variable for the non-replaced limb and $X_{replaced}$ is the variable for the replaced limb. A negative AI value indicates a replaced limb asymmetry, whereas a positive value is indicative of an asymmetry of the non-replaced limb. An AI of zero indicates a complete symmetry, whereas increasing AI values away from zero indicates larger asymmetries. An AI was computed for each variable were exported and used for statistical analysis.

Statistical Analyses

Separate one-way repeated measures analysis of variance (ANOVA) were run on AI of load-response KEM, load-response vertical GRF, push-off KEM, and push-off vertical GRF comparing pre- and post-training measurements for both walking speeds. Individual 2 x 2 (limb x time) repeated measure ANOVAs were run on the selected dependent variables at each walking speeds separately. Paired t-tests were run on gait velocities and VNS pain outcomes comparing pre- and post-training. An alpha level was set 0.05 *a priori*. Effect sizes were reported as partial eta squared (η^2_p) and were interpreted as small ($\eta^2_p < 0.06$), medium ($0.06 \leq \eta^2_p < 0.15$), and large ($\eta^2_p \geq 0.15$) (310).

Results

Gait Velocities and VNS Pain

There were no significant differences found for the gait velocities for preferred (p = 0.278) or fast (p = 0.086) speeds between pre- and post-training (Table 8). The velocity for the preferred speed was significantly lower compared to the fast speed ($p < 0.001$). No significant difference between pre- and post-training was found for VNS pain at the beginning of pre- and post-training (p = 0.343), and VNS pain following preferred speed walking (p = 1.00) or fast speed walking (p = 0.758).

Asymmetry Indices

Following the intervention, there was no effect of time on the load-response KEM AI at both preferred ($F_{1,9} = 0.929$, $p = 0.363$, $\eta^2_p = 0.104$) or fast ($F_{1,8} = 0.963$, $p = 0.267$, $\eta^2_p = 0.151$) speeds (Table 10). Similarly, there was no significant differences found for push-off- KEM AI for either the preferred ($F_{1,9}$ = 1.700, *p* = 0.225, η^2_p = 0.159) or fast ($F_{1,8}$ = 2.620, *p* = 0.144, η^2_p = 0.247**)** walking conditions. The load-response vertical GRF AI showed no differences based on time for preferred ($F_{1,9}$ = 0.233, *p* = 0.641, η^2_p = 0.025) or fast walking ($F_{1,8}$ = 0.298, *p* = 0.600, η^2_p = 0.036). Finally, push-off vertical GRF AI displayed no effect of time for preferred ($F_{1,9}$ = 1.668, *p* = 0.229, η^2_p = 0.156) or fast walking ($F_{1,8}$ = 1.214, *p* = 0.303, η^2_p = 0.132).

Supporting Variables

The overall findings for the supporting variables for the preferred walking speed condition can be found in Table 11. Peak push-off KAbM showed a main effect of limb ($F_{1,9}$ = 6.009, *p* = 0.037, η^2_p = 0.400). The non-replaced limbs had a greater KAbM compared to the replaced limbs (Table 11). The knee contact angle displayed an effect of time ($F_{1,9}$ = 10.595, *p* = 0.010, η^2_p = 0.541) with post-training displaying a more flexed initial contact angle. The knee adduction ROM also displayed a significant effect of time ($F_{1,9}$ = 11.840, *p* = 0.007, η^2_p = 0.568). There were smaller knee adduction ROM at post-training compared to pre-training values.

Overall results for the supporting variables during the fast walking speed condition can be found in Table 12. KAbM at push-off had a significant main effect of limb ($F_{1,8}$ = 6.018, *p* = 0.004, η^2_p = 0.429) with a greater KAbM for the non-replaced limbs (Table 12). Knee contact angle had a significant main effect of time ($F_{1,8}$ = 26.291, *p* = 0.001, η^2_p = 0.767). Following the training intervention, these individuals had a more flexed knee at initial contact. Knee adduction ROM also displayed a significant main effect of time ($F_{1,8}$ = 10.868, *p* = 0.011, η^2_p = 0.576). There were smaller knee adduction ROM found at post-training compared to pre-training.

Discussion

The purpose of this study was to examine the effects of a short-term cycling intervention paired with augmented feedback of vertical PRF on the KEM AI and knee joint biomechanics during gait for patients of TKA.

We hypothesized that the peak vertical GRF AI and KEM AI would decrease following the intervention period, indicating a more symmetrical loading pattern. Our hypothesis was rejected, with no differences found between the pre- and post-training vertical GRF AI and KEM AI for either preferred or fast walking speeds. During loading-response, the peak vertical GRF AI changed minimally from pre- to post-training at the preferred speed (1.61% vs. 1.24%) and the fast walking speed (1.34% vs. 0.06%). Similar results were found for vertical GRF AI during push-off, with no change in vertical GRF AI following the training intervention. However, there was a medium effect size of limb for the push-off vertical GRF at the fast speed, indicating there could be a lower magnitude of push-off vertical GRF in the replaced limbs. One reason to explain the lack of significant decrease in vertical GRF asymmetries is that our participants did not show a large AI prior to training. The asymmetries found for vertical GRF appear to be similar to those found during stationary cycling of the same group of TKA patients (326). Vertical PRF asymmetries ranged from 5.4 – 1.4% at pre-training at testing workrates of 80 W and 100 W, respectively. Since these patients started with relatively symmetrical vertical PRF, then perhaps that played an integral role in no changes found post-training. Patients of TKA who have greater asymmetries of vertical PRF or GRF may benefit more from the training compared to those with less asymmetries. Comparisons between subgroups of asymmetry magnitudes may prove to be beneficial in identifying individuals who may find benefits of this type of training intervention.

Our results also showed no changes in KEM asymmetry as a result of the cycling training, during both load-response and push-off at either walking speed. Despite the lack of statistical significance, there were non-significant reduction of KEM AI at preferred (16.3%) and at the fast speed (17.97%) during loading-response. Additionally, there were large decreases in push-off KEM AI for preferred and fast speeds (11.9% and 15.4%, respectively). When examining AI values, some have used ±10% to indicate some clinical relevance for knee OA patients (153). These decreases in KEM AI display a shift of loading from the non-replaced to replaced limbs, with the push-off KEM AI actually shifted to a greater loading of the replaced limbs at both speeds. One consideration for the lack of significant findings is the very large standard deviations for KEM AIs, especially during the load-response. A closer examination showed that two of the participants had high asymmetry for their non-replaced limbs compared to asymmetry for the replaced limbs for the other participants. Further analysis found that when both outliers were removed, the load-response KEM AI did not significantly differ before and after training: 38. ±31.0% vs. 24.2±28.0 % (p = 0.394, η^2_p = 0.123). Their removal did not alter the statistical results, but reduced the large standard deviations previously found. While all participants reported that OA in the other joints did not impact their walking, these two participants may have contralateral knee OA that has precipitated some form of compensation mechanism. Further research could aim to group participants based on their response to examine the training effect more accurately for AI. Interestingly, the magnitudes for vertical PRF asymmetries appear to be much lower than those found for KEM and symmetrical. The large magnitude of KEM asymmetries may not be directly and mainly linked to peak vertical GRFs. The large KEM asymmetries may be a consequence of both the vertical and posterior GRFs combined, along with deficits for knee flexion ROM during initial loading response. At the fast pace, there was a large effect of limb for knee flexion ROM, with a larger ROM found for non-replaced limbs compared to replaced. Decreased knee flexion ROM may indicate a stiffer knee joint, leading to a decreased KEM during load response. This is consistent with others who found that patients of TKA may walk with a stiff knee joint (9, 20, 22, 318). However, the patients of the current study did not display a significant difference between limbs for peak KEM which has previously been shown in other studies. One cause for this could be the small sample size in the current study. During load-response, there was a large effect size of limb that did not reach statistical significance. Additionally, participants in the current study were mostly pain free for their replaced limb (VNS: 0.6 ± 1.3). Others have also found that unilateral TKA patients between 6 to 60 months post-operation do not have significantly different peak KEM during level walking (47). These patients may not have adopted a compensation mechanism to reduce their replaced limbs KEM (9, 22).

This study is not without its limitations. First, due to recruitment difficulties and cessation of in-person human research activities due to the COVID-19 pandemic, we were unable to enroll and train enough participants to reach the *a priori* statistical power. The smaller sample size in this study may have partially had an impact on the lack of significant findings for KEM asymmetries. In addition, due to this we were unable to include a control group for comparison against the short-term cycling intervention. Inclusion of a control group would allow for a comparison to ensure any effect of time was due to the training intervention, and rather natural healing over time. Second, this study only used six training sessions over three weeks for the intervention. It appears that the cycling training program may require more sessions in a longer time span to be effective. Thus, the impact of this training intervention remains unknown and merit further research. Third, we used vertical PRF data to provide augmented feedback during training sessions to elicit changes in KEM. Our participants did not show an asymmetry of vertical PRF but a rather large asymmetry in KEM. Using KEM for feedback would provide more meaningful feedback and have greater effects on asymmetry of patients with TKA. Fourth, the participants in the current study had completed their post-TKA rehabilitation, high functioning and relatively pain free (VNS: 0.6 ± 1.3) at the time of the study. Implementing a training program with stationary cycling may be more beneficial in an earlier stage of rehabilitation for this population.

In conclusion, the short-term cycling training intervention used in the current study did not alter the AI for vertical PRF or KEM during gait. While no statistical significance was found for either AI variables, in the reduction of KEM AI exceeded 10%, which may be clinically relevant. Future work should aim to implement this type of intervention with the inclusion of a control group, fully powered intervention group, at earlier stages of rehabilitation, for longer durations with more training sessions, and more direct measures of knee joint loading used for augmented feedback.

Table 8. Participant velocities and VAS pain outcomes for both walking speeds at pre- and post-training

	Pre-Training	Post-Training	P
Preferred Gait Velocity (m/s)	1.25±0.33	1.30±0.18	0.278
Fast Gait Velocity (m/s)	1.55±0.38	1.64±0.18	0.086
Initial VNS Pain	0.60±1.34	0.95±1.45	0.343
Preferred Speed VNS Pain	0.60±1.58	0.60±1.26	1.000
Fast Speed VNS Pain	0.65±1.56	0.60±1.26	0.758

VNS: Visual Numeric Scale,

Table 9. Criteria for regulating workrate during the training intervention based on rating of perceived exertion (RPE), knee pain (VNS), and asymmetry index (AI).

	RPE	VNS Pain	Index (%)
Increase Workrate	< 15	< +2 of Previous Bout	< 20%
Maintain Workrate	15	< +2 of Previous Bout	> 20%
Decrease Workrate	>15	≥ +2 of Previous Bout	N/A

VNS: Visual Numeric Scale

Table 10. Peak vertical GRF and KEM AI (%) for the preferred and fast walking for pre- and post-training (mean ± STD)

	Preferred			Fast Pace		
	Pre-Training	Post-Training	P(η^2_p)	Pre-Training	Post-Training	P(η^2_p)
LR KEM	25.97±42.97	9.67±69.31	0.363(0.104)	19.84±36.40	1.87±47.52	0.267(151)
PO KEM	1.96±22.59	-9.89±24.98	0.225(0.159)	11.24±26.92	-4.14±26.62	0.144(0.247)
LR vertical GRF	1.61±3.49	1.24±4.18	0.641(0.025)	1.34±10.31	0.06±6.52	0.600(0.036)
PO vertical GRF	0.81±3.98	1.50±4.69	0.229(0.156)	2.85±7.58	0.80±6.03	0.303(0.132)

LR: load response, KEM: knee extension moment, PO: push off, GRF: ground reaction force, AI: asymmetry index, η^2_p: partial eta squared

Table 11. Mean data for peak GRF, knee joint kinematic and moments variables, hip kinetics, and ankle kinetics for replaced and non-replaced limbs during self-selected speed walking at pre- and post-training (mean ± STD)

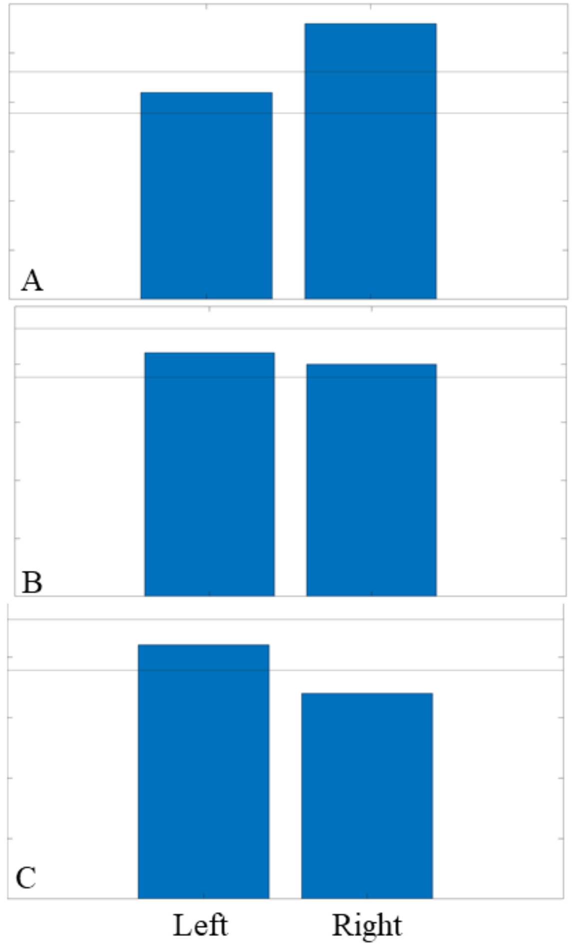
	Pre-Training		Post-Training		P(η^2_p)		
	Replaced	Non-Replaced	Replaced	Non-Replaced	Inter	Limb	Time
LR vertical GRF (BW)	1.08±0.07	1.09±0.08	1.08±0.05	1.10±0.09	0.450(0.065)	0.633(0.026)	0.674(0.021)
PO vertical GRF (BW)	1.06±0.05	1.07±0.07	1.05±0.05	1.07±0.07	0.200(0.175)	0.343(0.100)	0.913(0.001)
LR posterior GRF (BW)	-0.19±0.05	-0.20±0.04	-0.20±0.04	-0.20±0.04	0.964(0.001)	0.823(0.006)	0.820(0.006)
LR KEM (Nm/kg)	0.24±0.12	0.32±0.20	0.25±0.16	0.28±0.18	0.372(0.089)	0.193(0.180)	0.774(0.010)
PO KEM (Nm/kg)	0.15±0.05	0.15±0.05	0.15±0.04	0.14±0.05	0.463(0.061)	0.732(0.014)	0.969(0.000)
LR KAbM (Nm/kg)	-0.49±0.16	-0.53±0.21	-0.46±0.10	-0.53±0.19	0.421(0.073)	0.323(0.108)	0.542(0.043)
PO KAbM (Nm/kg)	-0.26±0.06	-0.35±0.15	-0.28±0.05	-0.34±0.13	0.205(0.172)	**0.037**(0.400)	0.727(0.014)
Knee Contact Angle (°)	-0.85±4.43	0.17±4.81	1.78±3.91	1.90±6.72	0.601(0.032)	0.706(0.017)	**0.010**(0.541)
Knee Flexion ROM (°)	-15.67±3.66	-16.98±4.18	-16.35±3.15	-16.52±5.32	0.212(0.167)	0.526(0.046)	0.858(0.004)
Knee Adduction ROM (°)	4.93±2.14	5.21±2.22	4.00±1.46	4.41±1.57	0.904(0.002)	0.555(0.040)	**0.007**(0.568)
Hip Extension Moment (Nm/kg)	-0.97±0.21	-0.93±0.14	-0.96±0.27	-0.92±0.19	0.931(0.001)	0.229(0.156)	0.857(0.004)
Ankle Plantar Flexion Moment (Nm/kg)	-1.42±0.14	-1.44±0.25	-1.42±0.16	-1.42±0.19	0.452(0.064)	0.820(0.006)	0.649(0.024)

Bolded values indicate a significant effect ($p < 0.05$). LR: load response, PO: push off, GRF: ground reaction force, KEM: knee extension moment, KAbM: knee abduction moment, Inter: interaction, η^2_p: partial eta squared.

Table 12. Mean data for peak GRF, knee joint kinematics and kinetics, hip kinetics, and ankle kinetics for replaced and non-replaced limbs during fast walking at pre- and post-training (mean ± STD)

	Pre-Training		Post-Training		P(η^2_p)		
	Replaced	Non-Replaced	Replaced	Non-Replaced	Inter	Limb	Time
LR vertical GRF (BW)	1.20±0.08	1.22±0.11	1.23±0.06	1.24±0.11	0.518(0.054)	0.634(0.030)	0.728(0.016)
PO vertical GRF (BW)	1.07±0.05	1.10±0.09	1.07±0.06	1.09±0.10	0.255(0.158)	0.305(0.130)	0.570(0.042)
LR posterior GRF (BW)	-0.25±0.04	-0.25±0.04	-0.25±0.04	-0.25±0.06	0.767(0.012)	0.797(0.009)	0.693(0.021)
LR KEM (Nm/kg)	0.37±0.15	0.52±0.30	0.43±0.21	0.51±0.31	0.254(0.159)	0.099(0.303)	0.839(0.005)
PO KEM (Nm/kg)	0.16±0.05	0.18±0.05	0.20±0.05	0.21±0.08	0.624(0.031)	0.365(0.104)	0.120(0.274)
LR KAbM (Nm/kg)	-0.58±0.18	-0.64±0.24	-0.59±0.12	-0.64±0.22	0.935(0.001)	0.517(0.054)	0.314(0.126)
PO KAbM (Nm/kg)	-0.25±0.06	-0.34±0.16	-0.28±0.07	-0.33±0.12	0.159(0.232)	**0.040**(0.429)	0.991(0.000)
Knee Contact Angle (°)	-1.75±3.81	-0.80±3.90	0.93±3.47	2.40±4.07	0.904(0.002)	0.350(0.110)	**0.001**(0.767)
Knee Flexion ROM (°)	17.15±3.32	19.08±4.52	18.08±3.99	19.68±4.36	0.385(0.096)	0.115(0.282)	0.839(0.005)
Knee Adduction ROM (°)	5.47±2.31	5.95±2.50	4.57±2.36	5.01±2.38	0.998(0.000)	0.550(0.046)	**0.011**(0.576)
Hip Extension Moment (Nm/kg)	-1.23±0.19	-1.20±0.15	-1.24±0.31	-1.25±0.20	0.376(0.099)	0.919(0.001)	0.763(0.012)
Ankle Plantar Flexion Moment (Nm/kg)	-1.48±0.14	-1.52±0.29	-1.49±0.19	-1.50±0.22	0.192(0.202)	0.354(0.108)	0.386(0.095)

Bolded values indicate a significant effect (*p* < 0.05). LR: load response, PO: push off, GRF: ground reaction force, KEM: knee extension moment, KAbM: knee abduction moment, Inter: interaction, η^2_p: partial eta squared.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CHAPTER VII

Knee Compressive and Associated Muscle Forces in Total Knee Arthroplasty Patients during Stationary Cycling

Abstract

Unilateral total knee arthroplasty (TKA) patients have been shown to cycle with a lower peak knee extension moment in their replaced compared to non-replaced knees. The purpose of this study was to use musculoskeletal modeling to estimate total (TCF), medial (MCF), and lateral (LCF) tibiofemoral compressive contact forces for TKA patients during stationary cycling. Fifteen unilateral TKA patients were recruited from the same surgeon. Each participated in two cycling condition, 80 Watts and 100 Watts, while at a constant cadence of 80 RPM. A knee model (OpenSim 3.2) was used to estimate TCF, MCF, and LCF for the replaced and non-replaced limbs. A 2×2 (limb×workrate) ANOVA and a 2×2 (compartment×limb) ANOVA were run on the selected variables. No difference was found for peak TCF. Peak MCF was 32.3% lower in the replaced limb compared to non-replaced limbs at 80 W ($p = 0.003$). Our compartment by limb ANOVA found that peak MCF was 52.2% greater than peak LCF in the non-replaced limb at 80 W ($p < 0.001$). At 100 W, there was a 37.4% greater peak MCF compared to LCF. Following TKA, patients appear to have greater medial compartment loading on their non-replaced compared to their replaced limb. These findings may suggest that the TKA may be successful in correcting varus alignments in the replaced limb, but may still be present in the non-replaced limbs. Future research should examine the long-term impacts of the differences found, especially with clinical and functional testing.

Keywords: knee loading, medial tibiofemoral compartment, TKA, total knee replacement, ergometer cycling

Introduction

One of the most prevalent lower limb diseases is knee osteoarthritis (OA), which in the end-stage leads to total knee arthroplasty (TKA). Following TKA, patients avoid high tibiofemoral joint loading to recover from the invasive surgical operation and to ensure survivability of the implant (70). While there are no widely recognized set of rehabilitation guidelines for TKA, a common recommendation for TKA patients is to partake in activities that have lower tibiofemoral contact forces (70, 80). Stationary cycling is a popular form of exercise for rehabilitation from TKA due to the decreased tibiofemoral compressive forces (70, 76, 79, 155). Stationary cycling has a wide range of benefits, as it is effective for cardiovascular health, weight management, and lower extremity muscular strengthening (80). Currently, there is sparse information on the tibiofemoral contact forces in TKA patients during stationary cycling (82, 185).

In vivo tibiofemoral contact forces measured with an implanted knee replacement during stationary cycling are considerably lower than other common forms of exercises (82, 185). While cycling at a moderate intensity, 60 watts (W) and 40 revolutions per minute (RPM), peak resultant tibiofemoral forces were approximately 1.19 body weights (BW) in individuals following TKA (82). Increases in workrate increased tibiofemoral forces, whereas increases in cadence caused decreases in tibiofemoral forces. Others have found peak tibiofemoral compressive forces of 1.03 BWs during stationary cycling following TKA (185). Treadmill walking, over ground walking, and jogging elicited peak tibial forces of 2.05 BWs, 2.6 BWs, and 3-4 BWs, respectively (185). However, TKA with *in vivo* measurement capacity is expensive and not practical in normal clinical practice for large scale studies examining tibiofemoral compartment forces. Additionally, *in vivo* measurements can give insight only into implanted knee, not the contralateral non-operated knee.

Musculoskeletal modeling and simulation allows for an estimation of tibiofemoral contact forces and related muscle forces without *in vivo* measurements (258, 291, 307-309). Thus far, there has been limited musculoskeletal modeling research on tibiofemoral contact loads in clinical populations during stationary cycling (308). Previous work with medial compartment knee OA patients used musculoskeletal modeling to compare the effects of lateral toe wedges and foot angles during cycling on tibiofemoral contact force and moments (308). To our knowledge, there have been no studies examining the tibiofemoral contact forces in unilateral TKA patients comparing operated and non-operated limbs during stationary cycling. Current literature has observed a noticeable decrease in the internal knee extension moment (KEM) for the replaced limb compared to the non-replaced contralateral limb during gait and stair negotiation (31, 47, 318).

In addition to total tibiofemoral compressive force, the forces acting upon each individual tibiofemoral compartment is of great importance for TKA and knee OA patients (307). Musculoskeletal simulation data of TKA instrumented knee data using a knee model has shown a 14.6% greater force in the medial compared to the lateral tibiofemoral compartment compressive forces (835 N vs. 713 N) (307). Increased medial compartment loading in knee OA patients is typically linked with an increased varus alignment and reduced medial joint space (34, 307). Increased loading of the medial tibiofemoral compartment has been linked to progression and severity of knee OA and have further implications following TKA (58, 173, 332). The excessive varus alignment seen in knee OA patients is attempted to be corrected during TKA. Since medial tibiofemoral compartment loading is difficult to determine, internal knee abduction moment

(KAbM) using inverse dynamics has been commonly used as a surrogate measure (58, 333, 334). Currently, this is no consensus on whether or not there is a clear inter-limb difference for KAbM in TKA patients during gait (47, 51, 318). These findings may be indicative of success of TKA procedures correcting the malalignment of knee OA patients. Understanding the differences in the medial tibiofemoral contact forces in both replaced and non-replaced limbs could provide insight into potential risks for contralateral knee OA, and TKA revision. Additionally, the increased medial compartment contact load may indicate if the malalignment of OA limbs is corrected by TKA surgery and if there still exists a discrepancy between the replaced and non-replaced limbs. Therefore, musculoskeletal modeling of stationary cycling with unilateral TKA patients can provide insight into the kinetic deficits that have been previously documented for TKA patients during gait. This information may provide scientific evidence for prescribing exercises during rehabilitation for unilateral TKA patients.

The purpose of this study was to examine the tibiofemoral contact forces (total, medial compartment, and lateral compartment) and knee extensor and flexor muscle forces in TKA patients during stationary cycling. Our primary hypothesis was that the replaced limb would have lower peak total tibiofemoral compressive force (TCF), tibiofemoral medial compartment compressive force (MCF), tibiofemoral lateral compartment compressive force (LCF), knee extensor force, and knee flexor force compared to the non-replaced limb. Our secondary hypothesis was that peak MCF would be higher than peak LCF in both the replaced and non-replaced limbs.

Methods

Participants

Fifteen unilateral TKA patients participated in the data collection session (10 males and 5 females, 64.3 ± 8.2 yrs, 94.1 ± 20.4 kg, 1.74 ± 0.1 m). All participants were recruited from a local orthopedic clinic, at which they underwent a unilateral TKA operation performed by the same surgeon. Participants were required to be between 6- and 18-months post TKA operation, and between the ages of 50-80. Potential participants were excluded from the study if they had any of the following: debilitating OA of their other lower limb joints that impacted locomotion, other joint arthroplasties, BMI greater than 38 kg/m^2, neurological disease that would impact gait or balance, systemic inflammatory arthritis, lower extremity injuries within the past six months, or arthroscopic surgeries within that past three months. All testing procedures were approved by the Institutional Review Board. Participants read and signed an informed consent form, prior to participation in the current study. Additionally, all participants passed a Physical Readiness Questionnaire (PARQ), to ensure they did not require physician approval to exercise.

Instrumentation

Three-dimensional kinematics were collected using a twelve-camera motion capture system (240 Hz, Vicon Motion Capture Inc., Oxford, UK). Anatomical and tracking reflective markers were placed on the following anatomical landmarks bilaterally to track motion of the trunk, pelvis, thighs, shanks, and feet (323). Pedal motion was tracked using three markers on the lateral aspect and a fourth located on the anterior side of the pedal. To track the crank arm during each crank cycle, markers were placed on each crank arm axes.

Three-dimensional kinetic data was collected using custom-made instrumented pedals outfitted with two tri-axial force sensors (1200 Hz, Type 9027C, Kistler, Switzerland) accompanied with two amplifiers (Type 5073A, Kistler, Switzerland) for each pedal (56, 323). Additionally, a sixteen-channel wireless surface EMG system (1200 Hz, Delsys Trigno, Delsys Inc., Natick, MD, USA) was used to collect EMG data of the following muscles bilaterally: vastus medialis (VM), vastus lateralis (VL), semitendinosus (ST), biceps femoris (BF), and medial gastrocnemius (MG). Surface electrode placements for the examined muscles followed established guidelines using anatomical landmark and muscle palpations (303). Kinetic and EMG data was collected simultaneously with kinematic data using Vicon Nexus (version 2.8.2, Vicon Motion Capture Inc., Oxford, UK).

Experimental Protocol

Participants completed a three-minute self-selected warm up on a treadmill. Participants were then instrumented with the EMG sensors bilaterally on their lower limb in accordance with provided guidelines for sensor placement (303). Next, each participant completed functional tests for the purpose of normalizing EMG in leu of maximal voluntary isometric contractions. The functional tests included a body weight quarter squat for VL and VM, standing unilateral hamstring curl for BF and ST, and bilateral standing calf raise for MG. Each movement speed was performed at a frequency of 60 beats/min set via a digital metronome.

Following the functional tests, participants and the stationary ergometer were instrumented with the reflective markers. The stationary ergometer was then fitted for each participant based on saddle height, saddle fore-aft position, and handlebar distance (38, 197). The saddle height was modified to elicit a knee flexion angle of 30° with the pedal at the bottom dead center position (169, 197). Participants cycled for two minutes to become accustomed to the stationary ergometer at the first workrate. Cadence during the cycling warm up and trials were kept at 80 ± 2 RPM displayed on a control monitor placed in front of the ergometer. Participants cycled at each workrate for one minute with ten seconds of data being collected at the end of the trial (50 seconds – 60 seconds). Following the first workrate condition, participants were allotted a minimum of one-minute rest to avoid fatigue. Kinematic, kinetic, and EMG data were recorded simultaneously (Vicon Nexus Version 2.9, Vicon Motion Systems, UK) and exported for further analysis.

Experimental data was exported into Visual3D (Version 6.01, C-Motion Inc., Germantown, MD, USA). Three-dimensional marker trajectories were identified and filtered using a fourth-order zero lab Butterworth Lowpass filter with a cutoff frequency of 6 Hz (38, 56, 197). Pedal kinetic and COP data were computed for each pedal and transformed into the laboratory coordinate system for inverse dynamics analysis. Raw EMG signals were processed using a linear envelope including: a Butterworth bandpass filter with cutoff frequencies of 10-450 Hz, full-wave rectification, and a moving root mean square with a window of 91 ms and normalized to the peak value from the maximum of functional test trials for each appropriate muscle.

Musculoskeletal Simulation

Musculoskeletal modeling was performed using the open source software OpenSim (258). Experimental data was exported from Visual3D for use in OpenSim, that included the computation of scaling factors based on experimental marker data and inverse kinematics. A generic musculoskeletal model with 23 degrees-of-freedom and 92 musculotendon actuators was used for the musculoskeletal modeling and simulation (307). The hip joint is modeled as a ball-and-socket while the ankle and subtalar joints were modeled as revolute joints. The subtalar and metatarsophalangeal joints were locked to model the foot as a single rigid segment. The knee joint has been modified to include two revolute joints to estimate forces in the medial and lateral tibiofemoral joint compartments and remains a single degree-of-freedom joint allowing for flexion and extension movement (307).

The process for analyzing the cycling trials used an inverse dynamics static optimization approach that resulted in estimated muscle activations (308, 309). Each subject specific model was scaled based on the participants' height, mass, and segment lengths based on experimental marker data. Inverse dynamics was run based on the kinematics, and external pedal reaction forces recorded at each pedal. Muscle activations and forces were then estimated using static optimization to solve to minimize the sum of the squared muscle activations (Equation 13) (274, 297).

$$Min \sum_{i=1}^{N} a_i^2 \tag{13}$$

Finally, joint reaction analysis was used to solve for TCF, MCF and LCF expressed in the tibia reference frame (291).

Data Analyses

The TCF, MCF, and LCF generated by joint reaction analysis were exported and peak values during the power phase of the crank cycle (crank angle between 0 – 180°) were determined interactively using customized codes in Matlab. The summed muscle force for the knee extensor and flexor groups were selected in addition to the tibiofemoral contact forces. To validate the musculoskeletal model used in this current study, experimental EMG muscle activations were qualitatively compared to the computed muscle activations.

A 2 x 2 (limb x workrate) repeated measure analysis of variance (ANOVA) was run on peak TCF, MCF, LCF, and summed knee extensor and flexor muscle forces (IBM SPSS Statistics 25, Chicago, IL, USA). A separate 2 x 2 (compartment x limb) repeated measure ANOVA was run on peak MCF and LCF. An alpha level was set at 0.05 *a priori*. Normality and sphericity were assessed via a Shapiro-Wilk test and Greenhouse-Geisser, respectively. Paired sample t-tests were conducted for planned post-hoc analysis when an interaction was present with Bonferroni adjustment using an adjusted alpha level of 0.0125. Effect sizes were reported as partial eta squared (η^2_p) and interpreted as large ($\eta^2_p \geq 0.14$), medium ($0.06 \leq \eta^2_p < 0.14$) and small ($\eta^2_p < 0.06$) (310). Effect sizes for main effects and post-hoc pairwise comparisons were reported as Cohen's D (307, 311).

Results

We validated our model by comparing some of the muscle activations from static optimization with the EMG muscle activity collected experimentally. Out of the five muscles we collected EMG data, activation results of four muscles from static optimization: VM, VL, MG, and BF appear to agree with our EMG activity profiles (Figure 3). The ST muscle activation from simulation was minuscule during power phase compared to that of the ST EMG activation.

Peak TCF did not display any interaction ($p = 0.556$), effect of limb ($p = 0.181$) or effect of workrate ($p = 0.577$) (Figure 2). Peak MCF displayed no significant interaction ($p = 0.219$) or effect of workrate ($p = 0.233$). However, there was a significant main effect of limb for peak MCF ($F_{1,11} = 6.441$, $p = 0.028$, $\eta^2_p = 0.369$) with greater peak MCF found in the non-replaced compared to replaced limb (Table 13). Peak power phase LCF displayed no significant interaction ($p = 0.179$), effect of limb ($p = 0.255$) or effect of workrate ($p = 0.994$).

Peak knee extensor muscle force displayed no significant interaction ($p = 0.224$), effect of limb ($p = 0.846$) or effect of workrate ($p = 0.875$). Similarly, peak knee flexor muscle force showed no significant interaction ($p = 0.997$), effect of limb ($p = 0.633$) or effect of workrate ($p = 0.159$).

At 80 W, there was a significant interaction ($F_{1,13} = 19.706$, $p = 0.001$, $\eta^2_p = 0.603$) and effect of compartment ($F_{1,13} = 14.218$, $p = 0.002$, $\eta^2_p = 0.522$) but no effect of limb ($p = 0.087$). Post-hoc analysis found that MCF was lower in the replaced limb compared to the non-replaced ($p = 0.003$, $d = 1.29$, Figure 5a), and was greater than LCF in the non-replaced limb ($p < 0.001$, $d = 1.82$). LCF was greater in the replaced limb compared to non-replaced ($p = 0.004$, $d = 0.88$). At a workrate of 100W, there was no significant interaction ($p = 0.518$) or effect of limb ($p = 0.276$). There was however a main effect of compartment ($F_{1,11} = 7.81$, $p = 0.017$, $\eta^2_p = 4.15$) with greater forces experienced in the medial compartment (Figure 5B).

Discussion

The purpose of this study was to examine the tibiofemoral contact forces and the knee extensor and flexor muscle forces in TKA patients during stationary cycling. Our primary hypothesis was that the replaced limb would display a decreased peak TCF, MCF, LCF, knee extensor muscle force, and knee flexor muscle force compared to the non-replaced limb.

Our primary hypothesis was partially rejected, as peak TCF displayed no significant differences between the replaced and non-replaced limbs. While we did not find a significant difference, our peak power phase TCF was approximately 10.1% lower in the replaced limb compared to the non-replaced (Figure 2, Table 1). Our TCF results in the replaced limb at 80 W are in agreement with open-source data set of *in vivo* tibiofemoral contact forces during cycling (ortholoads.com). The *in vivo* data for cycling at 75 W and 60 RPM, the closest available condition, had a peak TCF of 903.6 N compared to 918.6 N in the current study. While not a direct measurement, KEM has been correlated and is a common surrogate measure for TCF (58). Even though our difference of TCF did not reach a statistically significant level, it could be clinically meaningful. Others examining knee OA patients during cycling considered a deficit of

10% or greater could be clinically significant (*d* = 0.44), and relevant to prescribing exercise (153). These findings were contrary to the results found using inverse dynamics in the same population (326). We found the peak KEM of replaced limbs was 21.3% lower than non-replaced limbs of TKA patients in cycling, indicating an unloading of the replaced knee joint. The inclusion of muscle forces appeared to have impacted knee joint loading, potentially providing a more accurate estimation of overall tibiofemoral loading during stationary cycling for TKA patients. Current literature of gait has also found a significant decrease in KEM in the replaced limb compared to the non-replaced in TKA patients (31, 47, 318). The replaced limb has shown decreased KEM of 5.7% in level walking (47) and 12.2% to 20.0% for stair negotiation (31, 318). The non-significant changes in peak TCF coincide with no significant differences in both knee extensor and flexor muscle forces between the replaced and non-replaced limbs (Table 1), as muscle forces directly impact tibiofemoral loading (335). Another reason for the lack of significant difference is that the level of tibiofemoral joint loading for cycling is much lower than weight bearing activities such as walking (185), which might have not been sufficient enough to impact on the bilateral difference for TCF.

While peak TCF and LCF did not differ, peak MCF was lower in the replaced limb compared to the non-replaced limb. Post hoc analysis showed that MCF was 32.3% lower with a large effect size (*d* = 1.24), in the replaced limb only at the 80 W condition. Interestingly, our results also indicated that increasing workrate to 100 W saw a non-significant 14.6% reduction of peak MCF in replaced limbs compared to the non-replaced limbs (*d* = 0.39, Table 1). MCF is often evaluated via a surrogate measure of internal KAbM in conjunction with KEM in the gait and cycling literature (36, 47, 56, 58, 197). Our results of the same TKA participants showed no significant changes in KAbM between the replaced and non-replaced limbs (326). However, peak KEM was 21.3% greater in the non-replaced limb compared to the replaced. With no difference in KAbM, the difference found in MCF in the current study may be the result of the differences in KEM and the knee joint alignment of the non-replaced limb. Knee OA patients tends to display a more excessive varus alignment, decreased medial knee joint space, and increased MCF. In healthy populations, increases in workrate found subsequent increases in both KAbM and KEM in cycling, which would indicate increased MCF (56). Our results showed that this may not be necessarily true for the replaced limbs of TKA population (Table 13). Currently, there is conflicting information on KAbM following TKA during gait, with some describing decreases following operation (51, 173), and others showing no difference (47). Our current study showed that in cycling, TKA patients loaded their replaced limb medial knee compartment less, which is similar to previous results shown in gait (173, 318). This could be due to the correction of excessive knee varus alignment during TKA procedure. Further research is required to examine the impact of greater workrate increases on peak MCF in replaced and non-replaced limbs.

Our secondary hypothesis was that peak MCF would be higher than peak LCF in both replaced and non-replaced limbs. This hypothesis was partially refuted, with no significant differences found between peak MCF and LCF in the replaced limb (Figure 3). One goal of a TKA is to restore knee joint alignment, from an excessive varus alignment found in knee OA pre-operation. The current study, to our knowledge, is the first to examine the loading conditions of the tibiofemoral medial and lateral compartments for TKA patients during stationary cycling. It appears that following TKA, these patients cycled with a relatively balanced mediolateral compartment loading in their replaced limbs. However, peak MCF was 52.2% greater at 80 W (d = 1.82) and 42.9% at 100 W (d = 1.41) compared to peak LCF on the non-replaced limb.

Increased MCF, has been previously linked to the progression of medial compartment knee OA severity (36). These findings suggest that the TKA may be successful in fixing the excessive varus alignment in replaced limbs. Non-replaced limbs may have a natural varus alignment similarly to that found in healthy individuals. This could explain the increased MCF for non-replaced limbs. Previous simulation work with the same knee model using TKA *in vivo* data, showed an estimated 835 N vs. 713 N for the loading-response peak MCF and LCF during gait, respectively, which showed that MCF was 14.6% higher compared to LCF (307). When using a uniformed model (with natural lower limb alignments), the peak MCF was shown to be 43.8% greater than the peak LCF (307). The current results appear to agree with the uninformed model when examining cycling post TKA. Future work examining mediolateral knee joint load in TKA patients should employ models incorporating subject specific knee alignment along with subject specific medial and lateral compartment contact points to improve contact force accuracy.

This study is not without its limitations. First, the muscle activations computed for this study were found using static optimization, which are time independent and are solved without respect to the previous time frame. However, they appear to agree with our experimentally measured EMG muscle activations. Second, the knee model used in the current study has been validated in walking using instrumented knee data, but not in cycling. The magnitudes of TCF in the current study were similar to those found during cycling studies using *in vivo* instrumented knee data (orthoload.com), but no *in vivo* data is available to confirm MCF or LCF. Third, we only examined a short bout of cycling in the current study. We cannot discern the long-term impact or adaptations to cycling in TKA patients. Fourth, the model used in the current study utilized generic contact locations for the medial and lateral compartments of the tibiofemoral joint. Generating subject specific contact points via computerized tomography may provide more accurate results for each individual subject. Lastly, a larger sample size may be necessary to provide enough statistical power to discern small changes in TCF that would be both clinically relevant and statistically significant.

This study used musculoskeletal modeling to estimate TCF, MCF, and LCF for TKA patients during stationary cycling in both the replaced and non-replaced knee. We found a non-significant decrease of 10.05% in peak TCF in the replaced compared to non-replaced limb. In addition, peak MCF was lower in the replaced limb at 80 W which is consistent with some current literature indicating a decreased medial compartment load following TKA. Our data also indicated that there was no difference in peak MCF and LCF for the replaced limbs. However, the non-replaced limb displayed an increased MCF compared to LCF at both 80 and 100 W. Future studies should aim to examine the long-term potential effects of cycling as a rehabilitation modality and its impact on inter-limb deficits found in the current study.

Table 13. Peak TCF, MCF, LCF, knee extensor and knee flexor muscle forces (N) for the replaced and non-replaced limbs at 80 and 100 W (mean ± std)

	80 Watts		100 Watts			p (η^2_p)	
	Replaced	Non-Replaced	Replaced	Non-Replaced	Inter	Limb	Workrate
TCF	-918.6±253.4	-1021.4±212.7	-952.5±208.6	-1024.7±276.2	0.556 (0.03)	0.181 (0.16)	0.577 (0.16)
MCF	-477.4±182.7	-705.2±183.2	-597.2±296.7	-699.0±213.9	0.219 (0.13)	**0.028** (0.37)	0.233 (0.13)
LCF	-472.2±133.7	-339.8±154.5	-411.9±238.9	-399.3±212.5	0.179 (0.16)	0.255 (0.12)	0.994 (0.00)
Extensor	876.1±263.7	910.0±214.4	901.6±234.9	894.4±271.3	0.224 (0.13)	0.846 (0.00)	0.875 (0.00)
Flexor	230.5±101.4	242.3±80.2	255.9±141.1	267.8±117.1	0.997 (0.00)	0.633 (0.02)	0.159 (0.17)

TCF: total contact force, MCF: medial contact force, LCF: lateral contact force, η^2_p: partial eta squared value, Inter: interaction.

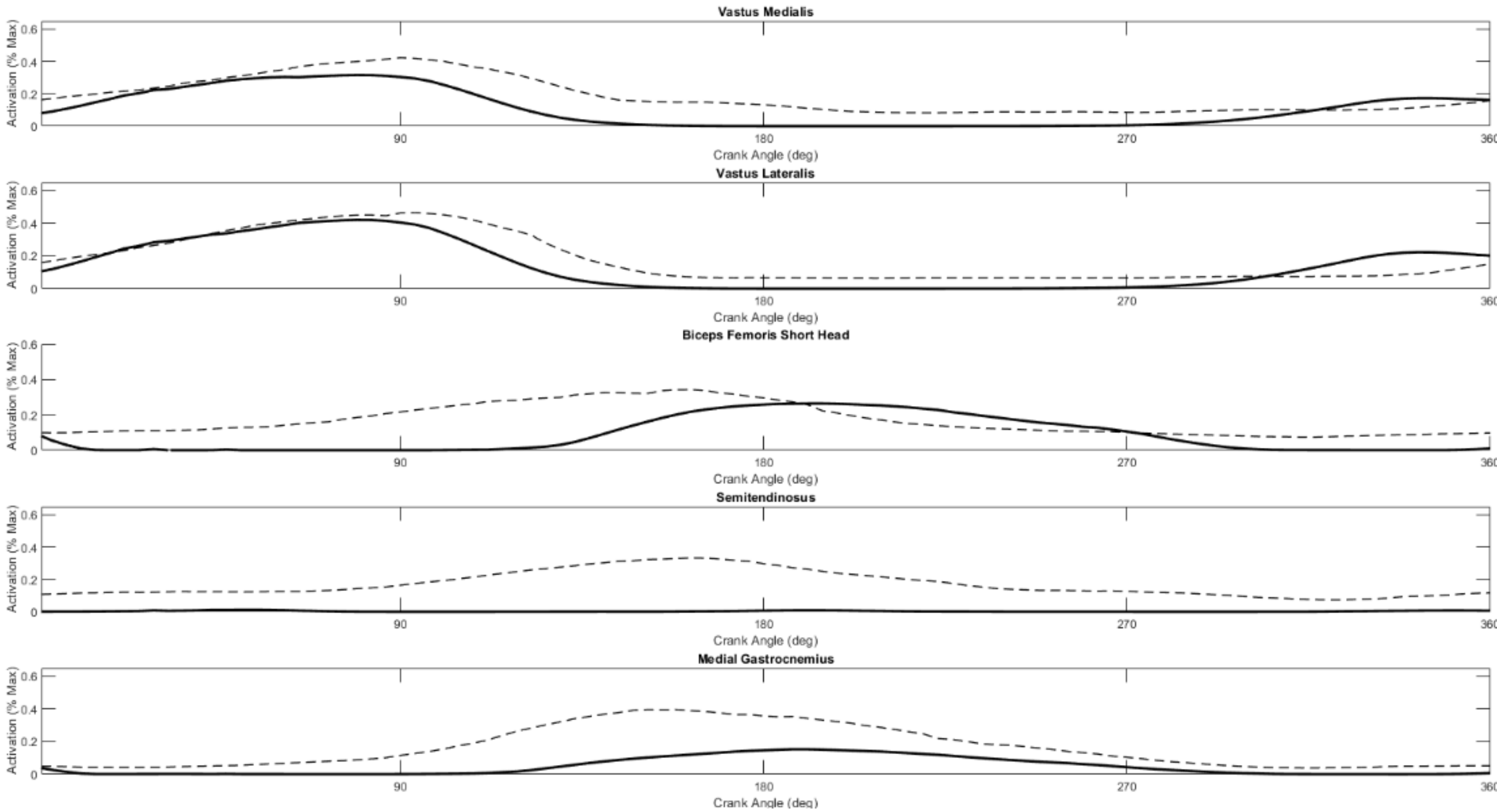


Figure 3a. Ensemble curves for activations for selected knee joint muscles (solid) computed from the static optimization and the experimental EMG muscle activities (dashed) for the replaced limb at 80 W

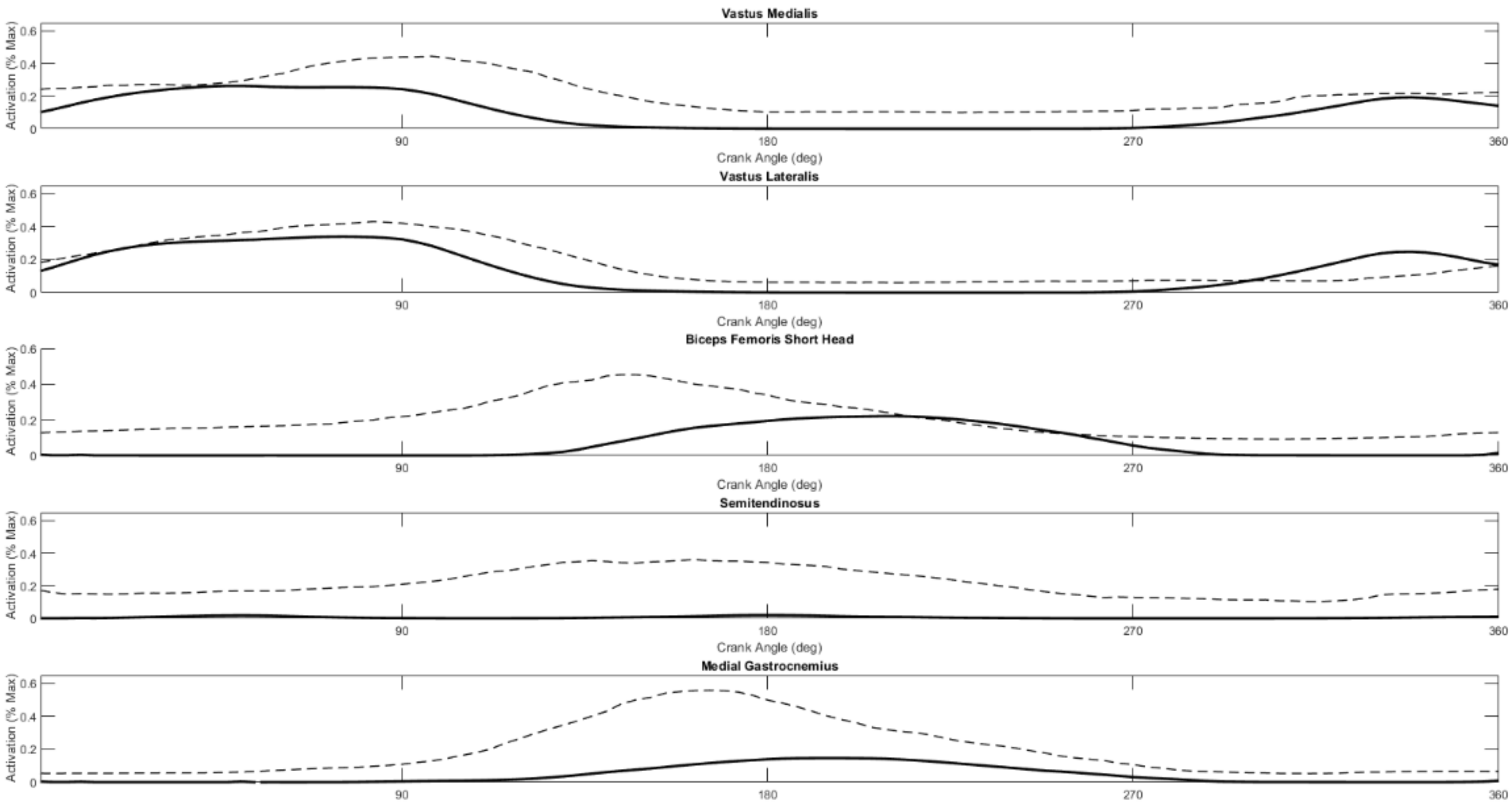


Figure 3b. Ensemble curves for activations for selected knee joint muscles (solid) computed from the static optimization and the experimental EMG muscle activities (dashed) for the replaced limb at 80 W

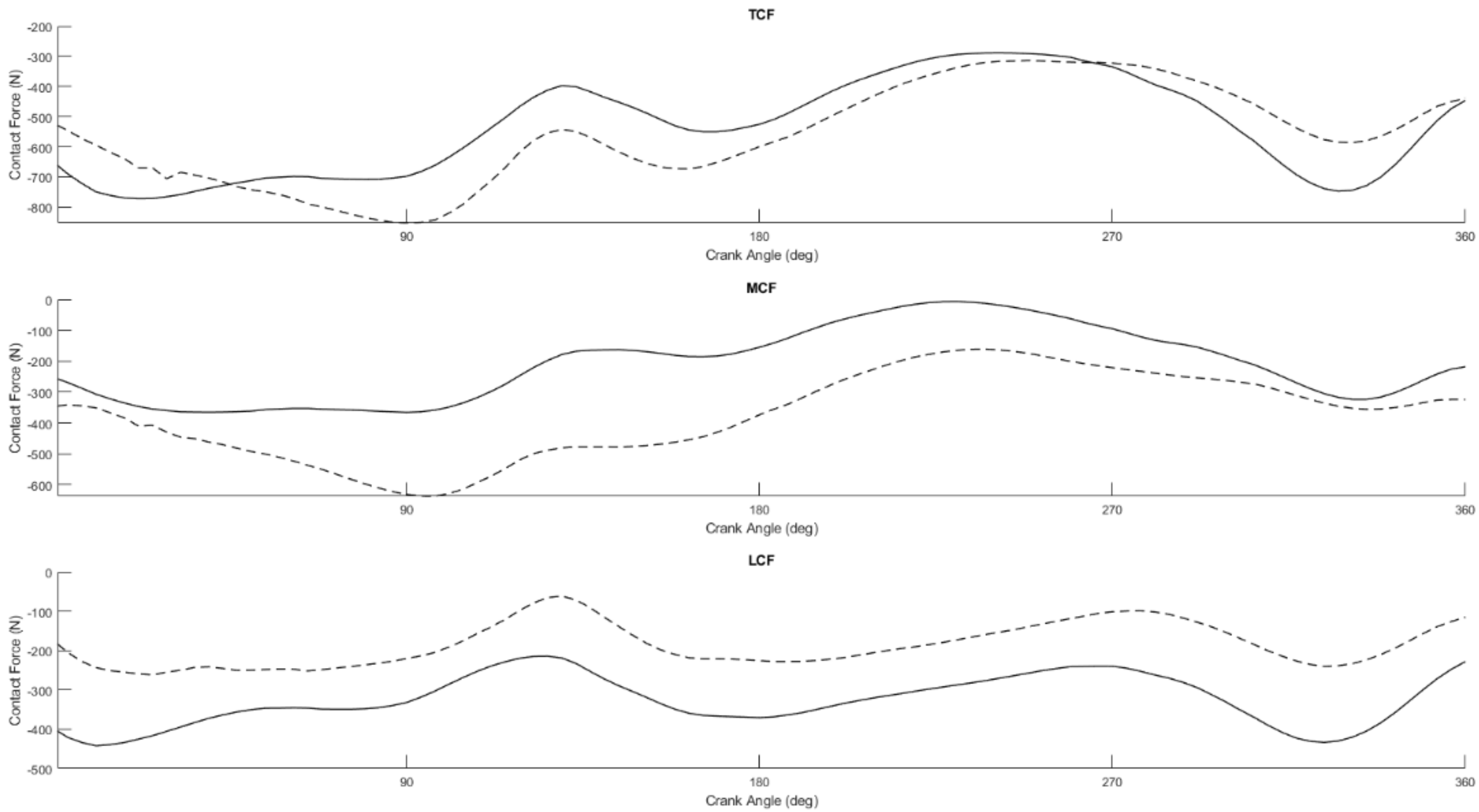
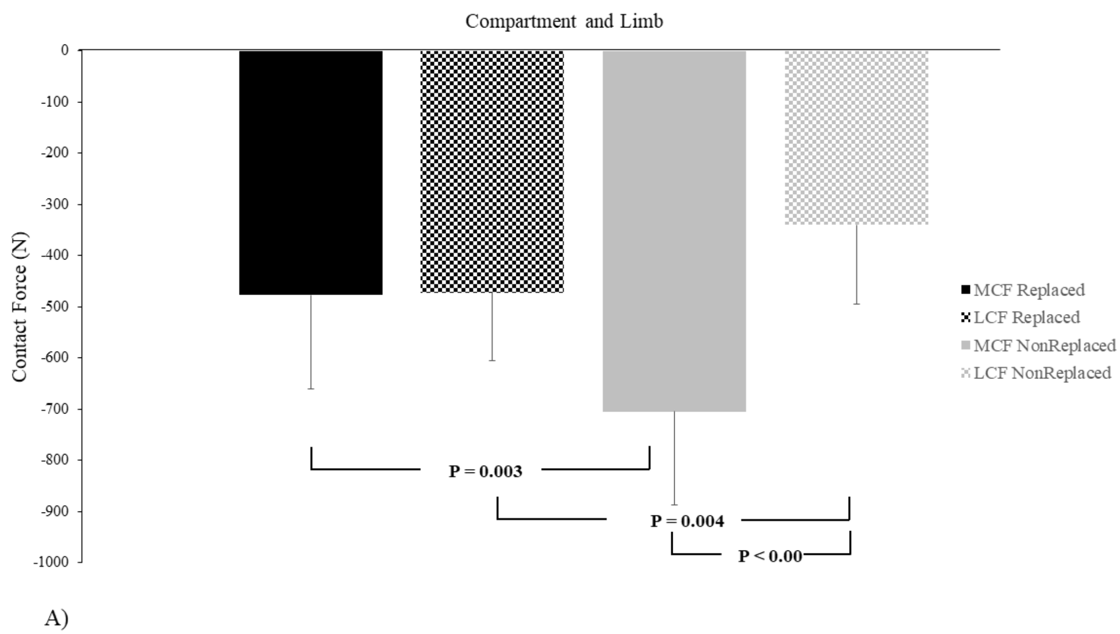


Figure 4. Ensemble curves for total tibiofemoral compressive force (TCF), medial compartment compressive force (MCF), and lateral compressive force (LCF) for the replaced limbs (solid) and the non-replaced limbs (dashed).



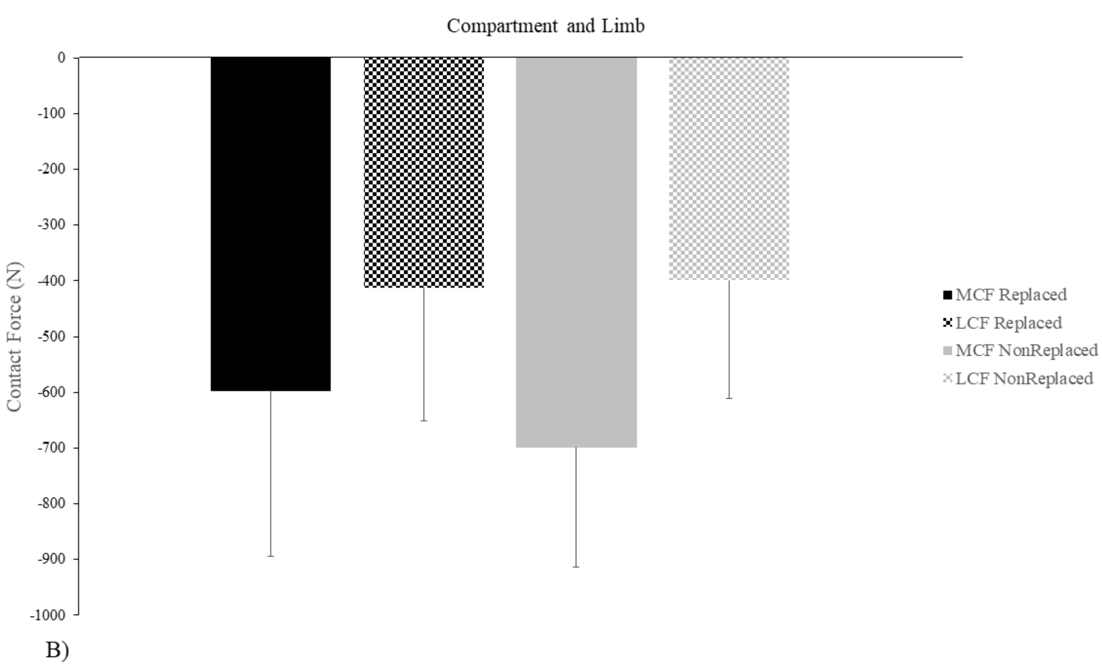


Figure 5. Comparison of MCF and LCF (Mean ± STD) for both the replaced and non-replaced limb at 80 W (A): with an interaction ($p = 0.001$) and effect of compartment ($p = 0.002$); at 100 W (B): with an effect of compartment ($p = 0.017$). Significant differences of post hoc comparisons between compartments and limbs are represented by the p values.

CHAPTER VIII

CONCLUSIONS

The purpose of this dissertation was to identify biomechanical inter-limb deficits for patients of unilateral TKA during stationary cycling using experimental biomechanics and musculoskeletal modeling, and to examine the effect of a short-term cycling intervention with augmented feedback on both cycling and gait knee joint biomechanics. In study one, analysis of stationary cycling at 80 and 100 W found that the replaced limbs had lower peak KEM and vertical PRF compared to non-replaced limbs. In study two, analysis of the pre- and post-training found that the short-term cycling intervention had no significant impact on the peak KEM or vertical PRF asymmetries during stationary cycling. KEM asymmetries did decrease with a moderate effect size by 9.9-10%. In study three, analysis examining the effect of the short-term cycling intervention found there was no effect on peak KEM or vertical GRF asymmetries during level walking. In study four, musculoskeletal modeling analysis found that the replaced limbs had lower peak MCF compared to non-replaced limbs. Additionally, the non-replaced limbs had a significantly greater peak MCF compared to peak LCF.

This dissertation has provided novel information on the knee joint biomechanics for patients following unilateral TKA. Interestingly, the patients in the current study showed no limb difference in peak KEM but had a significant inter-limb deficit during stationary cycling. The replaced limbs saw no differences in peak MCF compared to LCF, which may be indicative of a successful TKA to restore knee joint alignment. In addition, this dissertation has provided some novel data on the impact of a cycling intervention paired with augmented feedback on knee joint biomechanics both during stationary cycling and walking. While no significant benefits were found for this intervention, there reductions in peak KEM asymmetries of 10%. There may be some clinical benefits and relevance for using a cycling intervention with augmented feedback to reduce peak KEM asymmetries during cycling. The lack of significant findings could be linked to a smaller sample size than desired, due to COVID-19, as well as the intervention no including enough training days/dose. Therefore, future work should aim to include a larger sample size, increased number of training sessions, as well as during an earlier stage of rehabilitation.

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APPENDICES

Appendix A. Participant Phone Call Screening Questions

Phone Interview

PI – Hello, this is Erik Hummer from the University of Tennessee. I am calling because you were identified by the Tennessee Orthopedic Clinics as a potential participant in our study on knee replacements and had expressed interest in participating in our research once you were in formed. I just have a few brief questions to cover to ensure that you qualify for our study.

PI – The following questions are in regards of your eligibility to be included in the research:

1. Have you been diagnosed with osteoarthritis at the ankle or hip joint that impacts your walking and other activities?
2. Besides the one knee replacement, have you had any additional lower extremity joint replacement?
3. Within the past three months, have you had any lower extremity joint arthroscopic surgery or intra-articular injection?
4. Have you been diagnosed with systemic inflammatory arthritis (rheumatoid arthritis, psoriatic arthritis)?
5. Have you been diagnosed with a neurologic disease (e.g. Parkinson's disease, stroke patients)?
6. Have you had any additional major lower extremity injuries/surgeries except for the replaced knees?
7. Are you able to walk without a walking aid?
8. Are you able to cycle without aid on a stationary bike?
9. Do you have any visual conditions affecting gait or balance?
10. Are you currently pregnant or nursing?

If the participant passes all the exclusion criteria, he or she may then be allowed to participate in the study.

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Appendix B. Informed Consent Form

Consent for Research Participation

Research Study Title: *Efficacy of Cycling Training with Augmented Visual Feedback on Reducing Bilateral Asymmetries of Pedal Reaction Force of Patients with Total Knee Arthroplasty*

Researcher(s): Erik Hummer, University of Tennessee, Knoxville
Dr. Songning Zhang, University of Tennessee, Knoxville

Why am I being asked to be in this research study?

We are asking you to be in this research study because you have met all the inclusion and exclusion criteria and we believe that you would be a good candidate for the study.

What is this research study about?

The purpose of the research study is to see the effect of short-term cycling training with feedback on biomechanical functions of patients with total knee replacements. We want to see if cycling training with visual feedback can help with your rehabilitation, and to make your legs more symmetrical (the same).

Who is conducting this research study?

The research team at the University of Tennessee, Knoxville is receiving funding through the International Society of Biomechanics, College of Education, Health and Human Science, and the Department of Kinesiology, Recreation and Sport Studies for this study.

How long will I be in the research study?

If you agree to participate in this study, you will be involved with 8 study visits. Two of the study visits will last approximately 2 hours while the other 6 will last approximately 1/2 hour.

What will happen if I say "Yes, I want to be in this research study"?

If you agree to be in this study, we will ask you to do the following:

- Participate in 8 visits to our school. Two visits will include testing while the other 6 will be cycling training.
- All participation will take place in our lab in the HPER building
- Complete a physical readiness questionnaire to ensure your safety for exercise
- On testing days, you will be asked to the do the following:
  - Change into appropriate attire either owned by you or provided by the lab
  - Complete a brief 3 minute warm up of walking
  - Complete muscle strength testing on our dynamometer on your leg muscles
  - Be fitted with retroreflective markers and electromyography (EMG) sensors. The sensors are small boxes that will be taped to your skin and pose no risk.

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    - Complete 5 successful walking trials at your own walking pace, as well as at an increased speed.
    - Complete two trials of cycling at 80 RPM while at 80 and 100 Watts
    - Complete questionnaires on how hard the activities were and how much pain you had
- On training days, you will be asked to do the following:
    - Complete a training workout on a stationary bike
    - Training duration will be no longer than 20 minutes
    - You will be given rest at 5-minute intervals
    - You will be given feedback (verbal and visual) on how to cycle

What happens if I say "No, I do not want to be in this research study"?

Being in this study is up to you. You can say no now or leave the study later. If at any time you decide to leave the study, your relationship with UT and the researchers will not be impacted.

What happens if I say "Yes" but change my mind later?

Even if you decide to be in the study now, you can change your mind and stop at any time.

If you decide to stop before the study is completed, no negative consequences will occur to you. Any information we have collected so far will be destroyed and not used in the study.

Are there any possible risks to me?

There is minimal risk in this study. The duration and intensity of exercise are kept at a mild to moderate level. We will ensure you pass the PARQ form, which will help ensure you are safe to exercise. Both walking and cycling pose minimal risk to you during exercise. The researchers in this study are certified in first aid, CPR, and AED. Other people could potentially see you or find out of your participation in the study. We have procedures in place to minimize this risk such as locking all personal information in a locked cabinet. All information will be stored with no identifying information and will be kept using a subject code. All documents will be kept in a locked cabinet and only the investigator will have access to them.

Are there any benefits to being in this research study?

There is a possibility that you may benefit from being in the study, but there is no guarantee that will happen. Possible benefits include physical benefits that come from training. You may also have a reduced asymmetry between your operated and non-operated leg, which could be beneficial long-term. Being in a training protocol may lead to increased overall health as well. Even if you don't benefit from being in the study, your participation may help us to learn more about how cycling with feedback can be beneficial for people following a total knee replacement. We hope the knowledge gained from this study will benefit others in the future. You will also be compensated for your time during this study, which will be detailed in "Will I be paid for being in this research study". Parking at the University of Tennessee will be provided at no cost to you.

Who can see or use the information collected for this research study?

We will protect the confidentiality of your information by deidentifying the data such that only a subject number will be assigned to it. If information from this study is published or presented at scientific meetings, your name and other personal information will not be used.

We will make every effort to prevent anyone who is not on the research team from knowing that you gave us information or what information came from you. Although it is unlikely, there are times when others may need to see the information, we collect about you. These include:

- Government agencies (such as the Office for Human Research Protections in the U.S. Department of Health and Human Services), and others responsible for watching over the safety, effectiveness, and conduct of the research.
- If a law or court requires us to share the information, we would have to follow that law or final court ruling.
- International Society of Biomechanics who is the study sponsor providing funding for this research.
- The Tennessee Orthopedic Clinic (TOC) who will be our site for recruitment.

What will happen to my information after this study is over?

We will not keep your information to use for future research, presentations, and/or publications. Your name and other information that can directly identify you will be deleted from your research data collected as part of the study.

We may share your research data with other researchers without asking for your consent again, but it will not contain information that could directly identify you

Will I be paid for being in this research study?

Upon completing your participation, you will be finically compensated for your time. You will be provided $150.00 for your time given as a Target gift card after you complete all test and training sessions.

Will it cost me anything to be in this research study?

It will not cost you anything to participate in this study. All costs will be taken care of by the researchers.

What else do I need to know?

We use procedures to lower the possibility of these risks happening. Even so, you may still experience problems or injury, even when we are careful to avoid them. Please tell the researcher in charge, Erik Hummer at (865) 974-2091, about any injuries or other problems that you have during this study.

The University of Tennessee does not automatically pay for medical claims or give other compensation for injuries or other problems.

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Who can answer my questions about this research study?

If you have questions or concerns about this study, or have experienced a research related problem or injury, contact the researchers, Erik Hummer at **ehummer@vols.utk.edu** or over the phone at (865) 974-2091. You may also contact my faculty advisor, Dr. Songning Zhang via email at **szhang@utk.edu**

For questions or concerns about your rights or to speak with someone other than the research team about the study, please contact:

Institutional Review Board
The University of Tennessee, Knoxville
1534 White Avenue
Blount Hall, Room 408
Knoxville, TN 37996-1529
Phone: 865-974-7697
Email: utkirb@utk.edu

STATEMENT OF CONSENT

I have read this form and the research study has been explained to me. I have been given the chance to ask questions and my questions have been answered. If I have more questions, I have been told who to contact. By signing this document, I am agreeing to be in this study. I will receive a copy of this document after I sign it.

Name of Adult Participant	Signature of Adult Participant	Date

Researcher Signature (to be completed at time of informed consent)

I have explained the study to the participant and answered all of his/her questions. I believe that he/she understands the information described in this consent form and freely consents to be in the study.

Name of Research Team Member	Signature of Research Team Member	Date

Appendix C. Knee Injury and Osteoarthritis Outcome Score (KOOS)

KOOS KNEE SURVEY

Today's date: _____/______/______ Date of birth: _____/______/______

Name: ____

INSTRUCTIONS: This survey asks for your view about your knee. This information will help us keep track of how you feel about your knee and how well you are able to perform your usual activities.
Answer every question by ticking the appropriate box, only <u>one</u> box for each question. If you are unsure about how to answer a question, please give the best answer you can.

Symptoms
These questions should be answered thinking of your knee symptoms during the **last week**.

S1. Do you have swelling in your knee?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S2. Do you feel grinding, hear clicking or any other type of noise when your knee moves?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S3. Does your knee catch or hang up when moving?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S4. Can you straighten your knee fully?

Always	Often	Sometimes	Rarely	Never
☐	☐	☐	☐	☐

S5. Can you bend your knee fully?

Always	Often	Sometimes	Rarely	Never
☐	☐	☐	☐	☐

Stiffness
The following questions concern the amount of joint stiffness you have experienced during the **last week** in your knee. Stiffness is a sensation of restriction or slowness in the ease with which you move your knee joint.

S6. How severe is your knee joint stiffness after first wakening in the morning?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

S7. How severe is your knee stiffness after sitting, lying or resting **later in the day**?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

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Pain

P1. How often do you experience knee pain?

Never	Monthly	Weekly	Daily	Always
☐	☐	☐	☐	☐

What amount of knee pain have you experienced the **last week** during the following activities?

P2. Twisting/pivoting on your knee

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P3. Straightening knee fully

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P4. Bending knee fully

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P5. Walking on flat surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P6. Going up or down stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P7. At night while in bed

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P8. Sitting or lying

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P9. Standing upright

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, daily living

The following questions concern your physical function. By this we mean your ability to move around and to look after yourself. For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A1. Descending stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A2. Ascending stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

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For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A3. Rising from sitting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A4. Standing

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A5. Bending to floor/pick up an object

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A6. Walking on flat surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A7. Getting in/out of car

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A8. Going shopping

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A9. Putting on socks/stockings

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A10. Rising from bed

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A11. Taking off socks/stockings

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A12. Lying in bed (turning over, maintaining knee position)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A13. Getting in/out of bath

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A14. Sitting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A15. Getting on/off toilet

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

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For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A16. Heavy domestic duties (moving heavy boxes, scrubbing floors, etc)

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

A17. Light domestic duties (cooking, dusting, etc)

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

Function, sports and recreational activities

The following questions concern your physical function when being active on a higher level. The questions should be answered thinking of what degree of difficulty you have experienced during the **last week** due to your knee.

SP1. Squatting

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

SP2. Running

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

SP3. Jumping

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

SP4. Twisting/pivoting on your injured knee

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

SP5. Kneeling

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

Quality of Life

Q1. How often are you aware of your knee problem?

Never	Monthly	Weekly	Daily	Constantly
□	□	□	□	□

Q2. Have you modified your life style to avoid potentially damaging activities to your knee?

Not at all	Mildly	Moderately	Severely	Totally
□	□	□	□	□

Q3. How much are you troubled with lack of confidence in your knee?

Not at all	Mildly	Moderately	Severely	Extremely
□	□	□	□	□

Q4. In general, how much difficulty do you have with your knee?

None	Mild	Moderate	Severe	Extreme
□	□	□	□	□

Thank you very much for completing all the questions in this questionnaire.

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Appendix D. 2012 KOOS Scoring Form

KOOS Scoring 2012

A change in how to manage missing items was introduced in 2012. Previously, 2 missing items were allowed in each subscale. From 2012, at least 50% of the items should be responded to. The differences in subscale level are outlined in the following table:

	Number of items needed for calculation of subscale score (2012 rule for missing items)	Number of items needed for calculation of subscale score (1998 rule for missing items)
Pain	5	7
Symptoms	4	5
ADL	9	15
Sport/Rec	3	3
QOL	2	2

KOOS Scoring instructions

Assign the following scores to the boxes:

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐
0	1	2	3	4

Each subscale score is calculated independently. Calculate the mean score of the individual items of each subscale and divide by 4 (the highest possible score for a single answer option). Traditionally in orthopedics, 100 indicates no problems and 0 indicates extreme problems. The normalized score is transformed to meet this standard.

Missing data: If a mark is placed outside a box, the closest box is chosen. If two boxes are marked, that which indicates the more severe problem is chosen.
As long as at least 50% of the subscale items are answered for each subscale, a mean score can be calculated. If more than 50% of the subscale items are omitted, the response is considered invalid and no subscale score should be calculated. For the subscale Pain, this means that 5 items must be answered; for Symptoms, 4 items; for ADL, 9 items; for Sport/Rec, 3 items; and for QOL, 2 items must be answered in order to calculate a subscale score. Subscale scores are independent and can be reported for any number of the individual subscales, i.e. if a particular subscale is not considered valid (for example, the subscale Sport/Rec 2 weeks after total knee replacement), the results from the other subscale can be reported at this time-point.

KOOS Excel scoring files

Excel spreadsheets with formulae to calculate the five subscale scores are available from www.koos.nu. If, for any reason, you prefer to use your own spreadsheets, the Excel formulae are given below.

Excel formulation: When the raw data have been entered in the order the items occur in the KOOS questionnaires available from koos.nu, these Excel formulations can be copied and pasted directly into an English version of an Excel spreadsheet to automatically calculate the five subscore scales. Please note that it has been assumed that the items in the subscale symptoms appear first in the questionnaire.

KOOS Pain:	=IF(COUNT(B2:H2)>=(COLUMNS(B2:H2)/2),100-(AVERAGE(B2:H2))/4*100,"")
KOOS Symptoms:	=IF(COUNT(I2:Q2)>=(COLUMNS(I2:Q2)/2),100-(AVERAGE(I2:Q2))/4*100,"")
KOOS ADL:	=IF(COUNT(R2:AH2)>=(COLUMNS(R2:AH2)/2),100-(AVERAGE(R2:AH2))/4*100,"")
KOOS Sport/Rec:	=IF(COUNT(AI2:AM2)>=(COLUMNS(AI2:AM2)/2),100-(AVERAGE(AI2:AM2))/4*100,"")
KOOS QOL:	=IF(COUNT(AN2:AQ2)>=(COLUMNS(AN2:AQ2)/2),100-(AVERAGE(AN2:AQ2))/4*100,"")

KOOS Manual Score calculation

The slightly updated version of the formulae (presented above and used from August 2012 in the spreadsheets available from www.koos.nu) does not need any manual imputation: Apply the mean of the observed items within the subscale (e.g. KOOS Pain), divide by 4, and multiply by 100; when this number is then subtracted from 100, you have the KOOS subscale estimate for that particular cross-sectional assessment of the individual patient. For manual calculations, please use the formulae provided below for each subscale:

1. PAIN $$100 - \frac{\text{Mean Score (P1–P9)} \times 100}{4} = KOOS\ Pain$$

2. SYMPTOMS $$100 - \frac{\text{Mean Score (S1–S7)} \times 100}{4} = KOOS\ Symptoms$$

3. ADL $$100 - \frac{\text{Mean Score (A1–A17)} \times 100}{4} = KOOS\ ADL$$

4. SPORT/REC $$100 - \frac{\text{Mean Score (SP1–SP5)} \times 100}{4} = KOOS\ Sport/Rec$$

5. QOL $$100 - \frac{\text{Mean Score (Q1–Q4)} \times 100}{4} = KOOS\ QOL$$

WOMAC - How to score from the KOOS

Assign scores from 0 to 4 to the boxes as shown above. To get original WOMAC Scores, sum the item scores for each subscale. If you prefer percentage scores in accordance with the KOOS, use the formula provided below to convert the original WOMAC scores.

$$\text{Transformed scale} = 100 - \frac{\text{actual raw score} \times 100}{\text{maximum score}}$$

WOMAC subscores	Original score = sum of the following items	Maximum score
Pain	P5-P9	20
Stiffness	S6-S7	8
Function	A1-A17	68

KOOS Profile

To visualize differences in the five different KOOS subscores and change between different administrations of the KOOS (e.g. pre-treatment to post-treatment), KOOS Profiles can be plotted. The example from Nilsdotter et al. [17] shows KOOS profiles prior to and at three time points following total knee replacement (TKR).

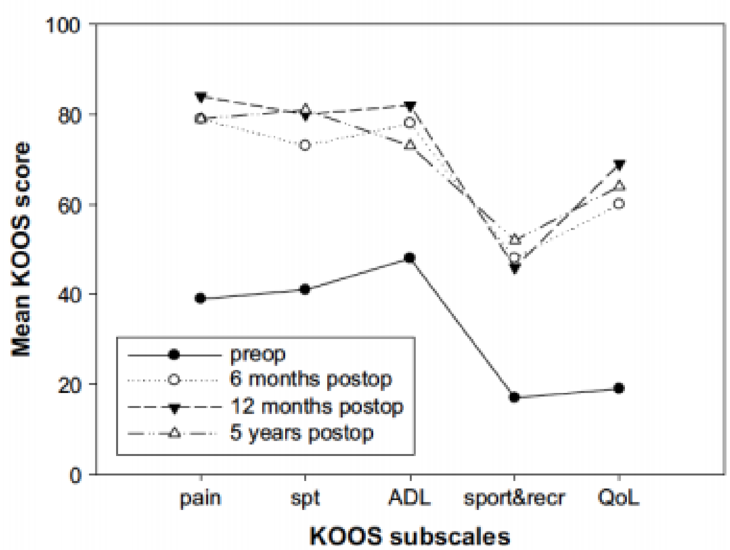

Fig. 1. KOOS profiles prior to and up to 5 years after TKR. Mean KOOS scores ($n = 80$) at the preoperative, 6 months, 12 months and 5 year assessments after TKR.

Appendix E. Physical Readiness Questionnaire (PARQ)

Physical Activity Readiness Questionnaire - PAR-Q (revised 2002)

PAR-Q & YOU

(A Questionnaire for People Aged 15 to 69)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age, and you are not used to being very active, check with your doctor.

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: check YES or NO.

YES	NO		
☐	☐	1.	**Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?**
☐	☐	2.	**Do you feel pain in your chest when you do physical activity?**
☐	☐	3.	**In the past month, have you had chest pain when you were not doing physical activity?**
☐	☐	4.	**Do you lose your balance because of dizziness or do you ever lose consciousness?**
☐	☐	5.	**Do you have a bone or joint problem (for example, back, knee or hip) that could be made worse by a change in your physical activity?**
☐	☐	6.	**Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?**
☐	☐	7.	**Do you know of any other reason why you should not do physical activity?**

If you answered

YES to one or more questions

Talk with your doctor by phone or in person BEFORE you start becoming much more physically active or BEFORE you have a fitness appraisal. Tell your doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want — as long as you start slowly and build up gradually. Or, you may need to restrict your activities to those which are safe for you. Talk with your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

NO to all questions

If you answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:

- start becoming much more physically active – begin slowly and build up gradually. This is the safest and easiest way to go.
- take part in a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively. It is also highly recommended that you have your blood pressure evaluated. If your reading is over 144/94, talk with your doctor before you start becoming much more physically active.

DELAY BECOMING MUCH MORE ACTIVE:

- if you are not feeling well because of a temporary illness such as a cold or a fever – wait until you feel better; or
- if you are or may be pregnant – talk to your doctor before you start becoming more active.

PLEASE NOTE: If your health changes so that you then answer YES to any of the above questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

Informed Use of the PAR-Q: The Canadian Society for Exercise Physiology, Health Canada, and their agents assume no liability for persons who undertake physical activity, and if in doubt after completing this questionnaire, consult your doctor prior to physical activity.

No changes permitted. You are encouraged to photocopy the PAR-Q but only if you use the entire form.

NOTE: If the PAR-Q is being given to a person before he or she participates in a physical activity program or a fitness appraisal, this section may be used for legal or administrative purposes.

"I have read, understood and completed this questionnaire. Any questions I had were answered to my full satisfaction."

NAME ____________________

SIGNATURE ____________________ DATE ____________________

SIGNATURE OF PARENT or GUARDIAN (for participants under the age of majority) ____________________ WITNESS ____________________

Note: This physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if your condition changes so that you would answer YES to any of the seven questions.

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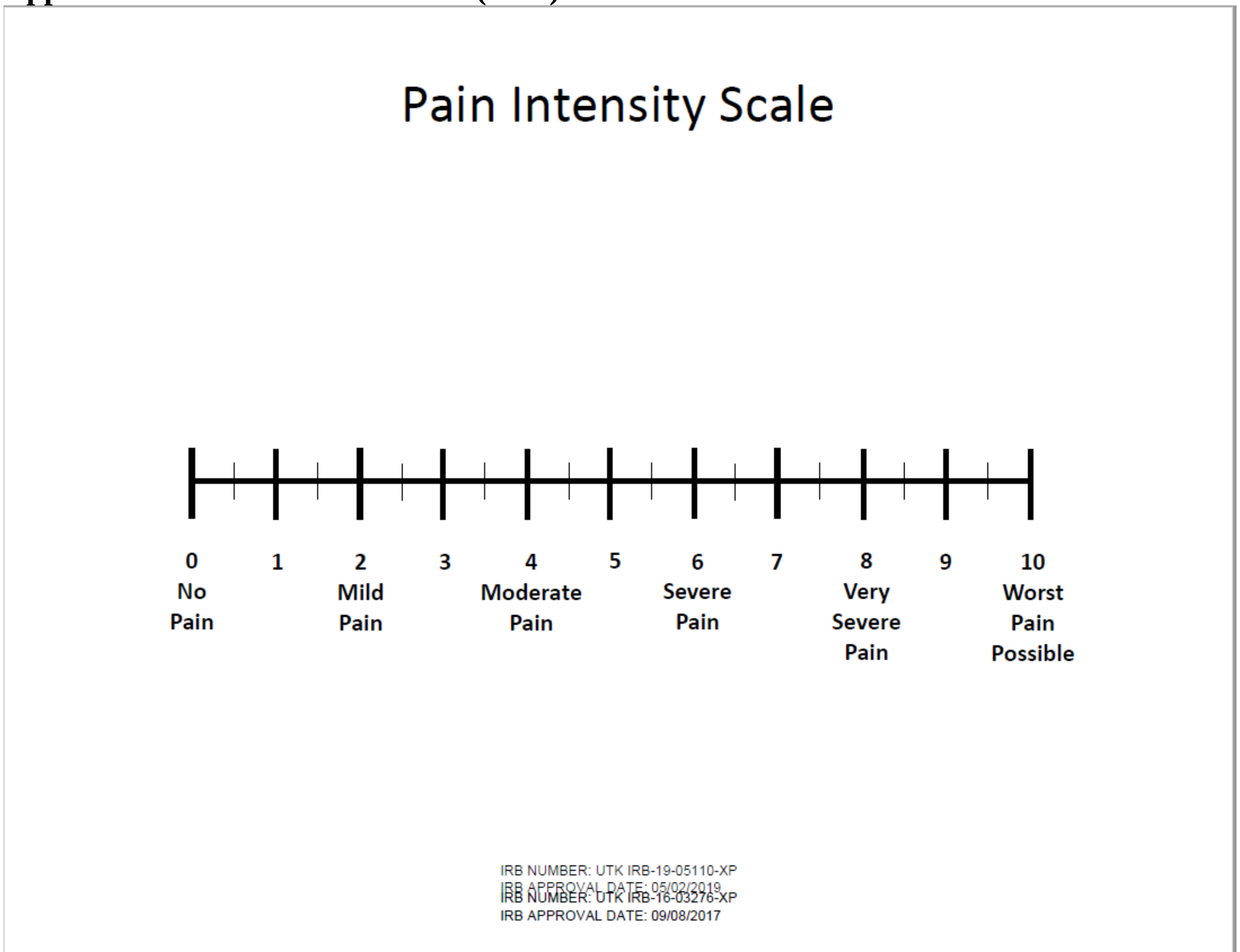
continued on other side...

Appendix F. Visual Numeric Scale (VNS) for Pain



Appendix G. Rating of Perceived Exertion (RPE)

6	No exertion at all
7	Extremely light
8	
9	Very light
10	
11	Light
12	
13	Somewhat hard
14	
15	Hard (heavy)
16	
17	Very hard
18	
19	Extremely hard
20	Maximal exertion

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Appendix H. Intervention Progression Based on RPE, VNS, and Asymmetry Index (AI)

Criteria for regulating workrate during the training intervention based on rating of perceived exertion (RPE), knee pain (VNS), and asymmetry index (AI).

	RPE	VNS Pain	Index (%)
Increase Workrate	< 15	< +2 of Previous Bout	< 20%
Maintain Workrate	15	< +2 of Previous Bout	> 20%
Decrease Workrate	>15	≥ +2 of Previous Bout	N/A

VNS: Visual Numeric Scale

Appendix I. Raw Data Tables for Variables of Interest

Table 14. Chapter 4 Subject Data for Peak Vertical PRF (N)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	225.36	213.88	177.89	217.33
2	170.83		170.93	
3	239.83	222.09	225.06	212.78
4	177.91	186.03	158.26	201.81
5	287.77	326.06	293.32	328.15
6	228.95	247.93	191.36	216.80
7	286.06	311.48	237.57	264.33
8	167.39	177.65	169.79	194.57
9	246.98	258.50	244.13	257.08
10	221.50	244.01	214.99	226.20
11	240.89	248.23	242.20	286.65
12	209.65	259.31	188.96	204.41
13	175.43	204.19	158.06	204.95
14	222.97	242.21	203.35	226.29
15	308.52		276.62	

Table 15. Chapter 4 Subject Data for Peak Posterior PRF (N)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	-69.42	-70.22	-55.93	-71.01
2	-23.89		-25.77	
3	-54.04	-57.22	-66.33	-55.99
4	-52.40	-55.08	-49.88	-68.24
5	-91.28	-86.93	-79.69	-76.38
6	-40.61	-42.50	-47.13	-50.60
7	-84.74	-91.57	-49.74	-54.44
8	-49.10	-46.26	-33.17	-31.30
9	-76.05	-94.96	-73.91	-84.45
10	-66.16	-65.57	-59.50	-52.70
11	-83.48	-75.69	-58.88	-49.71
12	-76.37	-66.25	-45.20	-34.25
13	-40.64	-41.76	-41.28	-44.15
14	-71.11	-60.08	-48.95	-47.98

Table 16. Chapter 4 Subject Data for Peak Medial PRF (N)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	2.80	-19.49	-17.30	-25.68
2	-.94		-11.65	
3	20.59	16.81	-26.84	-27.62
4	-21.28	-23.73	-41.97	-57.59
5	-30.53	-35.36	-9.49	-14.91
6	-7.97	-6.11	-36.60	-41.11
7	4.96	2.11	-30.26	-36.69
8	-3.42	-4.55	-13.57	-15.34
9	-34.61	-45.36	-41.56	-8.10
10	-35.01	-96.58	-27.59	-28.81
11	-32.29	-26.79	5.79	-83.96
12	11.03	-26.57	-10.33	-11.25
13	9.19	10.35	-26.25	-36.67
14	-42.48	-49.02	26.99	26.18
15	-87.11		45.31	

Table 17. Chapter 4 Subject Data for Peak KEM (Nm)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	28.83	28.11	17.12	25.04
2	17.01		15.75	
3	24.27	22.99	22.70	18.40
4	18.89	21.40	28.39	27.29
5	27.51	29.81	24.77	25.49
6	16.57	13.75	18.60	18.18
7	29.55	35.57	13.70	17.88
8	20.25	21.70	10.88	13.03
9	25.39	30.87	23.30	27.85
10	24.25	24.21	21.20	18.55
11	36.83	30.61	23.89	20.36
12	30.26	26.83	11.32	12.97
13	15.21	15.23	16.84	18.21
14	19.08	13.29	14.90	14.22
15	36.29		17.61	

Table 18. Chapter 4 Subject Data for Peak KAbM (Nm)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	-8.01	-2.72	-8.83	-13.17
2	-8.10		-4.62	
3	-25.27	-17.55	-12.65	-11.47
4	-8.20	-7.23	-12.55	-19.14
5	-20.53	-25.15	-14.26	-16.12
6	-6.16	-10.30	-6.73	-7.68
7	-18.62	-18.52	-10.78	-13.34
8	-7.78	-8.32	-4.75	-5.68
9	-12.04	-15.29	-6.74	-5.93
10	-7.75		-5.06	-6.14
11	-13.57	-10.00	-2.19	-36.68
12	-5.45		-2.41	-3.88
13	-11.22	-14.94	-5.03	-7.86
14	-12.50	-10.59		
15	-30.63			

Table 19. Chapter 4 Subject Data for Knee Extension ROM (°)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	76.52	76.71	74.26	76.09
2	75.55		76.75	
3	77.84	77.45	73.32	73.25
4	69.02	69.19	69.25	72.33
5	70.01	69.27	70.62	69.68
6	71.64	74.23	71.51	71.83
7	70.33	69.67	65.87	65.89
8	86.15	84.18	82.55	80.87
9	70.13	75.74	66.27	67.43
10	71.59	71.29	68.02	67.69
11	67.16	67.35	63.98	65.49
12	66.61	65.25	59.91	56.82
13	74.54	73.31	74.25	74.40
14	72.32	73.63	69.29	68.38
15	80.86		76.49	

Table 20. Chapter 4 Subject Data for Knee Abduction ROM (°)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	-1.48	-1.72	-13.47	-13.69
2	-5.46		-10.07	
3	-8.46	-10.81	-16.97	-17.04
4	-5.28	-4.80	-6.89	-7.56
5	-16.06	-17.76	-22.53	-21.66
6	-7.82	-10.78	-10.62	-9.19
7	-5.43	-4.12	-10.86	-10.14
8	-9.34	-10.47	-13.62	-14.06
9	-11.37	-10.95	-16.41	-15.65
10	-10.16	-12.99	-12.08	-10.99
11	-4.04	-5.97	-9.16	-9.38
12	-8.09	-8.14	-14.01	-12.87
13	-10.69	-13.35	-13.21	-13.91
14	-11.20	-12.60	-12.41	-13.10
15	-26.94		-16.73	

Table 21. Chapter 4 Subject Data for Peak Hip Extension Moment (Nm)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	-25.48	-29.72	-20.93	-23.87
2	-15.04		-13.21	
3	-39.93	-26.60	-30.01	-29.91
4	-22.12	-26.00	-17.06	-15.78
5	-31.24	-39.67	-34.75	-42.18
6	-35.29	-37.90	-19.48	-20.01
7	-23.54	-24.67	-25.25	-28.37
8	-23.26	-23.47	-24.16	-27.82
9	-28.36	-39.29	-33.43	-38.07
10	-18.96	-25.28	-13.75	-18.48
11	-28.74	-33.10	-35.21	-46.76
12	-28.77	-35.38	-32.40	-34.91
13	-23.24	-25.76	-18.50	-23.98
14	-34.90	-49.30	-36.36	-52.14
15	-24.07		-43.26	

Table 22. Chapter 4 Subject Data for Peak Ankle Plantar Flexion ROM (°)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	-17.97	-17.50	-17.43	-18.23
2	-7.02		-6.94	
3	-15.37	-11.09	-16.23	-15.37
4	-18.83	-21.63	-18.08	-22.54
5	-25.04	-23.59	-24.52	-25.33
6	-15.88	-20.49	-12.55	-17.91
7	-31.04	-31.61	-22.30	-22.89
8	-12.12	-12.22	-11.87	-14.84
9	-28.67	-35.57	-22.01	-25.69
10	-13.51	-15.61	-14.32	-13.25
11	-13.79	-18.49	-12.27	-18.20
12	-7.79	-9.93	-8.85	-7.43
13	-6.68	-6.80	-9.53	-12.48
14	-24.04	-30.62	-20.82	-21.80
15	-14.71		-15.51	

Table 23. Chapter 5 Subject Data for Peak Vertical PRF AI (%)

	Pre-Test		Post-Test	
Subject	80 W	100 W	80 W	100 W
1	20.4825	-2.2671	26.4332	-5.3071
2	-0.3939		9.10663	
5	-2.2048	-0.6931	11.1737	10.6749
7	16.7574	14.8876	-11.583	-6.9307
8	-1.992	-9.5143	-9.778	-2.594
9	1.06459	0.09966	-1.5381	3.83464
10	2.7456	6.77423	0.80575	5.46609
11	-1.183	-17.528	-4.1384	-3.0196
12	8.97837	21.0741	-52.811	-33.763
13	9.91465	-0.4169	2.10949	-3.508

Table 24. Chapter 5 Subject Data for Peak Posterior PRF AI (%)

	Pre-Test		Post-Test	
Subject	80 W	100 W	80 W	100 W
1	18.8284	-1.9087	37.885	8.79365
2	-8.7804		38.6412	
5	12.312	11.9729	10.0482	19.1317
7	40.6049	40.4645	-0.1562	11.601
8	32.0779	32.3968	7.8038	3.12579
9	2.38415	7.81792	7.36341	8.29633
10	9.84298	19.1477	34.8545	40.9881
11	29.0992	33.1459	8.49525	24.3421
12	40.2377	48.0915	5.29632	4.83267
13	-1.6125	-5.7818	6.22866	-6.6501

Table 25. Chapter 5 Subject Data for Peak KEM AI (%)

	Pre-Test		Post-Test	
Subject	80 W	100 W	80 W	100 W
1	39.8546	8.4404	52.4584	12.8084
2	7.4897		55.4943	
5	8.80566	14.024	10.245	16.9755
7	51.7686	33.4374	-27.807	1.66024
8	45.9866	45.4903	9.57194	10.0736
9	8.02406	12.1059	5.11582	-1.652
10	11.1428	20.8856	41.0703	47.6879
11	34.2177	22.322	-7.4896	28.1922
12	61.5119	44.2937	-8.4244	-4.6643
13	-11.644	11.4368	26.9202	11.921

Table 26. Chapter 5 Subject Data for Mean Peak Vertical PRF (N) at 80 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	225.360	177.893	206.252	148.496
2	170.829	170.933	210.776	191.553
5	287.771	293.319	294.419	261.661
7	286.058	237.569	242.472	269.883
8	167.391	169.793	183.237	200.794
9	246.976	244.133	253.705	257.611
10	221.496	214.993	247.410	244.640
11	240.887	242.201	222.548	231.482
12	209.645	188.963	165.309	250.312
13	175.425	158.064	183.715	179.615

Table 27. Chapter 5 Subject Data for Mean Peak Vertical PRF (N) at 100 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	213.879	217.332	191.140	200.495
2				
5	326.058	328.147	305.939	273.065
7	311.482	264.326	262.935	279.257
8	177.650	194.568	207.651	212.512
9	258.495	257.077	260.173	249.862
10	244.005	226.200	252.703	238.226
11	248.226	286.653	253.570	261.430
12	259.311	204.413	193.253	257.502
13	204.194	204.951	201.317	207.643

Table 28. Chapter 5 Subject Data for Mean Peak Posterior PRF (N) at 80 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	-69.415	-55.928	-84.053	-51.686
2	-23.891	-25.765	-30.887	-18.945
5	-91.282	-79.686	-87.838	-78.956
7	-84.743	-49.739	-62.824	-62.819
8	-49.102	-33.170	-54.212	-49.988
9	-76.046	-73.909	-83.539	-76.806
10	-66.159	-59.498	-81.300	-52.524
11	-83.483	-58.881	-76.109	-68.868
12	-76.370	-45.196	-70.972	-66.597
13	-40.637	-41.282	-44.597	-41.849

Table 29. Chapter 5 Subject Data for Mean Peak Posterior PRF (N) at 100 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	-70.218	-71.007	-75.211	-68.409
2				
5	-86.925	-76.381	-91.582	-73.789
7	-91.573	-54.436	-83.413	-71.875
8	-46.264	-31.301	-49.384	-47.391
9	-94.955	-84.452	-104.518	-95.626
10	-65.565	-52.696	-74.271	-43.163
11	-75.692	-49.708	-88.024	-66.504
12	-66.251	-34.248	-66.560	-62.719
13	-41.763	-44.146	-43.768	-46.671

Table 30. Chapter 5 Subject Data for Mean Peak KEM (Nm) at 80 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	28.828	17.124	32.170	14.934
2	17.008	15.746	16.205	7.229
5	27.514	24.770	36.129	32.463
7	29.549	13.698	18.697	23.867
8	20.246	10.878	24.492	22.135
9	25.386	23.299	26.805	25.029
10	24.247	21.202	31.233	18.078
11	36.827	23.893	23.009	24.277
12	30.262	11.317	20.096	21.110
13	15.208	16.839	16.488	12.041

Table 31. Chapter 5 Subject Data for Mean Peak KEM (Nm) at 100 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	28.107	25.038	29.269	25.349
2				
5	29.812	25.487	41.187	34.003
7	35.565	17.884	28.481	25.870
8	21.703	13.028	21.400	18.963
9	30.871	27.853	32.859	33.170
10	24.211	18.549	29.871	15.378
11	30.612	20.357	30.939	22.201
12	26.825	12.966	20.296	20.652
13	15.233	18.211	18.189	16.022

Table 32. Chapter 5 Subject Data for Knee Extension ROM (°) at 80 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	76.517	74.262	80.862	74.280
2	75.552	76.752	77.114	73.761
5	70.010	70.624	62.789	60.909
7	70.328	65.869	71.074	62.224
8	86.145	82.549	75.513	72.387
9	70.133	66.273	70.702	67.193
10	71.591	68.018	71.522	67.569
11	67.159	63.976	63.079	64.627
12	66.612	59.912	69.556	65.307
13	74.541	74.248	72.864	69.454

Table 33. Chapter 5 Subject Data for Knee Extension ROM (°) at 100 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	76.711	76.086	79.664	76.393
2				
5	69.269	69.681	61.657	59.456
7	69.673	65.890	72.056	63.112
8	84.177	80.866	75.546	72.963
9	75.736	67.426	79.422	70.498
10	71.289	67.688	70.628	65.560
11	67.354	65.492	63.945	65.444
12	65.252	56.822	68.639	64.694
13	73.305	74.396	71.766	69.263

Table 34. Chapter 5 Subject Data for Knee Abduction ROM (°) at 80 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	-1.484	-13.470	-4.628	-8.605
2	-5.457	-10.068	-4.588	-9.151
5	-16.055	-22.533	-12.206	-9.004
7	-5.434	-10.855	-4.217	-12.825
8	-9.336	-13.624	-10.132	-11.750
9	-11.369	-16.410	-10.283	-10.499
10	-10.164	-12.076	-14.976	-13.328
11	-4.040	-9.161	-4.139	-8.121
12	-8.094	-14.005	-4.370	-10.321
13	-10.690	-13.207	-10.343	-18.703

Table 35. Chapter 5 Subject Data for Knee Abduction ROM (°) at 100 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	-1.717	-13.686	-4.645	-9.261
2				
5	-17.759	-21.660	-12.924	-10.186
7	-4.120	-10.144	-6.034	-13.145
8	-10.469	-14.060	-14.721	-12.495
9	-10.949	-15.648	-11.725	-10.798
10	-12.988	-10.985	-15.177	-11.110
11	-5.967	-9.376	-3.502	-7.203
12	-8.142	-12.869	-4.353	-10.096
13	-13.346	-13.910	-11.680	-19.113

Table 36. Chapter 5 Raw Data for peak hip extension moment (Nm) at 80 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	-25.476	-20.931	-25.309	-27.149
2	-15.043	-13.212	-31.209	-30.581
5	-31.239	-34.745	-20.637	-31.821
7	-23.541	-25.246	-25.689	-25.393
8	-23.258	-24.155	-21.245	-19.382
9	-28.357	-33.433	-36.987	-28.082
10	-18.963	-13.748	-18.909	-26.163
11	-28.744	-35.211	-26.700	-38.297
12	-28.771	-32.399	-32.678	-34.353
13	-23.242	-18.501	-27.204	-29.839

Table 37. Chapter 5 Raw Data for peak hip extension moment (Nm) at 100 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	-29.715	-23.871	-20.281	-24.101
2				
5	-39.665	-42.175	-16.208	-30.370
7	-24.670	-28.368	-17.273	-21.930
8	-23.471	-27.815	-32.876	-29.654
9	-39.294	-38.073	-42.565	-31.263
10	-25.279	-18.482	-25.986	-35.789
11	-33.095	-46.758	-30.176	-42.486
12	-35.380	-34.911	-38.811	-38.960
13	-25.763	-23.980	-28.164	-33.657

Table 38. Chapter 5 Raw data for ankle plantar flexion moment (Nm) at 80 W

	Pre-Test		Post-Test	
Subject	Non-Replaced	Replaced	Non-Replaced	Replaced
1	-17.971	-17.429	-13.729	-14.217
2	-7.018	-6.938	-6.782	-7.291
5	-25.035	-24.519	-18.683	-16.213
7	-31.037	-22.303	-22.594	-20.890
8	-12.123	-11.873	-10.212	-10.697
9	-28.671	-22.008	-29.580	-24.574
10	-13.508	-14.315	-15.203	-21.236
11	-13.791	-12.268	-15.742	-11.853
12	-7.791	-8.846	-12.646	-11.332
13	-6.675	-9.532	-10.272	-12.241

Table 39. Chapter 6 Subject Data for Push-Off KEM AI (%)

	Pre-Test		Post-Test	
Subject	80 W	100 W	80 W	100 W
1	9.605602	11.96468	-49.1204	-41.1159
2	-4.16783	-19.0218	2.404501	8.13379
5	5.725599	24.87837	-44.2465	-24.1837
7	8.738971	21.22217	-2.30845	22.90141
8	21.18088	13.87099	-2.25129	17.82537
9	20.29881	0.122356	-5.79118	-35.5936
10	13.35032	47.35676	22.21179	26.14547
11	-25.2845	38.63201	9.746118	11.97973
12	-48.7539	-37.8499	-38.7464	-23.327
13	18.95097	4.241009	9.218692	

Table 40. Chapter 6 Subject Data for Load Response vertical GRF AI (%)

	Pre-Test		Post-Test	
Subject	80 W	100 W	80 W	100 W
1	0.641689	3.225837	-1.23278	-1.09596
2	-3.80593	-25.3121	-5.42253	-11.5602
5	0.490718	4.474431	2.267813	0.923664
7	5.680017	9.669222	6.059685	3.294235
8	6.243574	7.324344	7.497231	11.32292
9	-0.80655	3.160786	1.398793	1.677905
10	5.115863	5.195555	-0.77755	0.575473
11	2.176866	1.317325	4.034864	2.76924
12	-2.78658	3.012261	-4.16497	-7.36063
13	3.186018	-0.84609	2.713785	

Table 41. Chapter 6 Subject Data for Push Off vertical GRF AI (%)

	Pre-Test		Post-Test	
Subject	80 W	100 W	80 W	100 W
1	3.185762	7.666999	4.153272	3.210008
2	-8.40585	-14.5855	-9.25753	-4.66702
5	4.678474	6.625811	5.40519	6.309989
7	1.014064	1.164151	3.457727	-8.18857
8	4.078044	9.04241	5.733519	11.03832
9	-1.40965	-1.35084	0.271511	-2.91249
10	1.307031	3.181088	-0.68246	-0.54605
11	4.04715	10.51068	6.500115	4.838803
12	1.897279	3.384865	-0.01198	-1.85205
13	-2.31283	-7.72659	-0.61415	

Table 42. Chapter 6 Subject Data for Mean Loading Response Vertical GRF (N)

	Pre-Test				Post-Test			
	Preferred Speed		Fast Speed		Preferred Speed		Fast Speed	
Subject	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced
1	1.089	1.098	1.173	1.213	1.099	1.086	1.157	1.144
2	1.200	0.984	1.340	1.071	1.007	0.956	1.212	1.087
5	1.024	1.029	1.106	1.158	1.020	1.044	1.138	1.149
7	1.139	1.207	1.217	1.349	1.168	1.243	1.303	1.341
8	1.140	1.217	1.273	1.374	1.105	1.194	1.225	1.383
9	1.071	1.062	1.172	1.211	1.085	1.101	1.181	1.202
10	1.044	1.100	1.239	1.307	1.120	1.112	1.332	1.340
11	1.126	1.151	1.244	1.262	1.135	1.183	1.257	1.295
12	1.036	1.009	1.172	1.208	1.044	1.002	1.255	1.171
13	0.964	0.997	1.063	1.040	1.046	1.077		

Table 43. Chapter 6 Subject Data for Mean Push-Off Vertical GRF (N)

	Pre-Test				Post-Test			
	Preferred Speed		Fast Speed		Preferred Speed		Fast Speed	
Subject	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced
1	1.125	1.162	1.163	1.261	1.095	1.143	1.164	1.203
2	1.085	1.000	1.108	0.968	1.066	0.976	0.992	0.949
5	0.968	1.016	0.997	1.068	0.961	1.016	1.009	1.078
7	1.092	1.103	1.081	1.093	1.034	1.072	1.048	0.964
8	1.066	1.112	1.101	1.211	1.050	1.113	1.082	1.217
9	0.977	0.964	1.010	0.997	0.979	0.981	1.017	0.989
10	1.029	1.043	1.104	1.141	1.046	1.039	1.124	1.119
11	1.119	1.166	1.006	1.124	1.106	1.183	1.057	1.111
12	1.030	1.050	1.062	1.099	1.053	1.053	1.164	1.143
13	1.058	1.035	1.104	1.033	1.110	1.104		

Table 44. Chapter 6 Subject Data for Mean Load Response Posterior GRF (N)

	Pre-Test				Post-Test			
	Preferred Speed		Fast Speed		Preferred Speed		Fast Speed	
Subject	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced
1	-0.199	-0.217	-0.245	-0.264	-0.212	-0.197	-0.234	-0.251
2	-0.281	-0.193	-0.315	-0.257	-0.203	-0.161	-0.243	-0.175
5	-0.127	-0.155	-0.181	-0.200	-0.147	-0.154	-0.192	-0.199
7	-0.187	-0.207	-0.209	-0.272	-0.202	-0.218	-0.232	-0.268
8	-0.227	-0.212	-0.277	-0.264	-0.214	-0.208	-0.268	-0.291
9	-0.152	-0.153	-0.235	-0.223	-0.164	-0.162	-0.221	-0.211
10	-0.241	-0.265	-0.299	-0.326	-0.257	-0.272	-0.341	-0.355
11	-0.239	-0.247	-0.262	-0.269	-0.226	-0.261	-0.238	-0.285
12	-0.156	-0.152	-0.234	-0.221	-0.155	-0.144	-0.262	-0.211
13	-0.120	-0.145	-0.206	-0.168	-0.165	-0.190		

Table 45. Chapter 6 Subject Data for Knee Flexion Range of Motion (°)

	Pre-Test				Post-Test			
	Preferred Speed		Fast Speed		Preferred Speed		Fast Speed	
Subject	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced
1	-19.317	-17.151	-20.485	-23.443	-20.523	-16.432	-23.351	-19.645
2	-16.438	-18.687	-18.892	-17.969	-14.763	-13.990	-20.292	-14.430
5	-14.185	-17.260	-16.025	-19.765	-14.621	-15.564	-16.058	-19.382
7	-17.704	-18.631	-19.531	-22.190	-16.978	-17.500	-17.650	-20.736
8	-18.231	-19.205	-19.457	-18.724	-18.157	-19.437	-20.835	-21.516
9	-16.952	-11.967	-17.333	-15.776	-17.659	-11.565	-17.788	-16.652
10	-18.504	-23.355	-21.159	-26.632	-21.030	-27.790	-22.039	-29.497
11	-16.870	-18.875	-12.557	-18.099	-15.690	-21.002	-11.555	-19.412
12	-8.934	-16.481	-12.011	-18.196	-11.211	-12.334	-13.143	-15.851
13	-9.584	-8.222	-14.084	-9.950	-12.875	-9.555		

Table 46. Chapter 6 Subject Data for Knee Extension Angle at Initial Contact (°)

	Pre-Test				Post-Test			
	Preferred Speed		Fast Speed		Preferred Speed		Fast Speed	
Subject	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced
1	5.698	-2.261	5.831	-1.303	7.242	1.549	7.406	2.705
2	-3.623	-3.222	-5.380	-4.450	3.438	-4.835	0.027	-2.889
5	-3.040	0.393	-4.641	0.290	3.276	-0.773	1.095	-0.178
7	-2.858	-0.174	-3.128	-1.027	-3.228	0.813	-4.387	1.464
8	4.152	3.002	2.886	0.949	2.562	2.674	3.801	1.546
9	-4.999	-5.239	-2.900	-3.781	-3.700	-3.684	-3.083	-1.931
10	0.097	2.848	-2.675	-1.429	4.631	11.398	1.156	4.082
11	5.669	6.114	1.491	3.625	4.498	11.866	0.505	9.559
12	-4.776	7.581	-3.914	6.279	2.830	7.904	1.848	7.222
13	-4.856	-7.302	-5.098	-7.115	-3.724	-7.880		

Table 47. Chapter 6 Subject Data for Knee Adduction Range of Motion (°)

	Pre-Test				Post-Test			
	Preferred Speed		Fast Speed		Preferred Speed		Fast Speed	
Subject	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced
1	6.228	0.105	6.178	0.231	2.489	3.455	2.346	4.173
2	8.934	5.852	9.294	7.288	5.460	5.649	8.882	6.215
5	5.172	7.790	7.479	9.522	4.028	7.243	5.469	8.968
7	6.825	5.493	7.037	6.010	7.251	5.102	7.909	5.820
8	3.911	3.994	5.400	4.403	2.712	3.276	3.556	3.790
9	3.004	5.559	4.137	6.964	3.542	5.309	3.132	7.156
10	4.528	5.944	5.231	7.861	3.352	3.371	3.968	5.363
11	3.516	7.624	2.254	5.612	4.256	3.225	2.620	1.700
12	1.429	3.629	1.717	4.835	2.697	2.024	3.264	1.931
13	5.766	6.085	5.999	6.800	4.214	5.418		

Table 48. Chapter 6 Subject Data for Mean Load Response KEM (Nm/kg)

	Pre-Test				Post-Test			
	Preferred Speed		Fast Speed		Preferred Speed		Fast Speed	
Subject	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced
1	0.355	0.683	0.454	1.021	0.359	0.348	0.483	0.510
2	0.296	0.405	0.516	0.424	0.066	0.269	0.264	0.247
5	0.019	0.191	0.117	0.264	0.093	0.103	0.219	0.268
7	0.386	0.498	0.479	0.834	0.378	0.523	0.491	0.823
8	0.292	0.354	0.411	0.577	0.334	0.388	0.477	0.668
9	0.276	0.169	0.398	0.322	0.380	0.147	0.485	0.228
10	0.278	0.472	0.523	0.896	0.515	0.586	0.888	1.133
11	0.238	0.224	0.364	0.440	0.130	0.176	0.230	0.397
12	0.199	0.045	0.374	0.261	0.203	0.044	0.360	0.300
13	0.047	0.108	0.078	0.145	0.042	0.249		

Table 49. Chapter 6 Subject Data for Mean Push-Off KEM (Nm/kg)

	Pre-Test				Post-Test			
	Preferred Speed		Fast Speed		Preferred Speed		Fast Speed	
Subject	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced
1	0.191	0.212	0.220	0.258	0.186	0.118	0.205	0.146
2	0.216	0.207	0.226	0.144	0.162	0.166	0.223	0.244
5	0.062	0.068	0.080	0.107	0.085	0.054	0.123	0.100
7	0.181	0.199	0.183	0.233	0.200	0.214	0.285	0.369
8	0.136	0.174	0.172	0.200	0.199	0.197	0.209	0.256
9	0.078	0.098	0.166	0.167	0.108	0.103	0.182	0.135
10	0.082	0.108	0.105	0.201	0.114	0.147	0.147	0.230
11	0.185	0.149	0.116	0.204	0.143	0.159	0.192	0.219
12	0.174	0.118	0.213	0.154	0.155	0.102	0.218	0.179
13	0.150	0.143	0.149	0.151	0.155	0.172		

Table 50. Chapter 6 Subject Data for Mean Load Response KAbM (Nm/kg)

	Pre-Test				Post-Test			
	Preferred Speed		Fast Speed		Preferred Speed		Fast Speed	
Subject	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced
1	-0.569	-0.472	-0.709	-0.495	-0.537	-0.537	-0.593	-0.651
2	-0.789	-0.510	-0.913	-0.689	-0.532	-0.453	-0.791	-0.573
5	-0.291	-0.573	-0.395	-0.746	-0.330	-0.578	-0.430	-0.766
7	-0.596	-0.912	-0.647	-1.089	-0.614	-0.917	-0.710	-1.072
8	-0.596	-0.576	-0.718	-0.610	-0.572	-0.579	-0.601	-0.612
9	-0.372	-0.447	-0.505	-0.603	-0.372	-0.490	-0.473	-0.613
10	-0.367	-0.180	-0.446	-0.227	-0.374	-0.203	-0.488	-0.249
11	-0.566	-0.833	-0.644	-0.888	-0.524	-0.742	-0.607	-0.771
12	-0.429	-0.458	-0.527	-0.607	-0.437	-0.410	-0.621	-0.491
13	-0.292	-0.377	-0.301	-0.391	-0.351	-0.414		

Table 51. Chapter 6 Subject Data for Mean Push-Off KAbM (Nm/kg)

	Pre-Test				Post-Test			
	Preferred Speed		Fast Speed		Preferred Speed		Fast Speed	
Subject	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced
1	-0.211	-0.415	-0.237	-0.424	-0.326	-0.295	-0.305	-0.307
2	-0.295	-0.303	-0.246	-0.372	-0.286	-0.280	-0.391	-0.305
5	-0.213	-0.427	-0.232	-0.455	-0.252	-0.438	-0.294	-0.448
7	-0.358	-0.624	-0.373	-0.619	-0.328	-0.563	-0.352	-0.440
8	-0.285	-0.381	-0.275	-0.411	-0.336	-0.427	-0.287	-0.488
9	-0.268	-0.295	-0.232	-0.255	-0.268	-0.252	-0.255	-0.216
10	-0.148	-0.076	-0.163	-0.053	-0.178	-0.116	-0.130	-0.130
11	-0.280	-0.471	-0.196	-0.380	-0.300	-0.431	-0.233	-0.325
12	-0.329	-0.287	-0.315	-0.263	-0.279	-0.318	-0.292	-0.300
13	-0.195	-0.256	-0.249	-0.208	-0.236	-0.276		

Table 52. Chapter 6 Subject Data for Mean Hip Extension Moment (Nm/kg)

	Pre-Test				Post-Test			
	Preferred Speed		Fast Speed		Preferred Speed		Fast Speed	
Subject	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced
1	-0.712	-0.771	-0.960	-1.044	-0.779	-0.761	-0.966	-1.009
2	-1.303	-0.905	-1.550	-1.304	-0.924	-0.891	-1.411	-1.324
5	-0.857	-0.725	-1.096	-0.977	-0.665	-0.819	-0.891	-1.019
7	-1.251	-1.118	-1.485	-1.295	-1.335	-1.182	-1.900	-1.572
8	-1.112	-0.985	-1.319	-1.165	-1.190	-1.124	-1.346	-1.354
9	-1.091	-1.121	-1.269	-1.445	-1.117	-1.065	-1.239	-1.375
10	-0.935	-0.904	-1.301	-1.356	-0.788	-0.728	-1.118	-1.332
11	-0.846	-1.047	-1.034	-1.184	-1.045	-1.083	-1.268	-1.285
12	-0.806	-0.864	-1.156	-1.157	-0.517	-0.596	-0.982	-1.007
13	-0.793	-0.813	-1.159	-1.030	-1.220	-0.934		

Table 53. Chapter 6 Subject Data for Mean Ankle Plantar Flexion Moment (Nm/kg)

	Pre-Test				Post-Test			
	Preferred Speed		Fast Speed		Preferred Speed		Fast Speed	
Subject	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced	Replaced	Non-Replaced
1	-1.645	-1.784	-1.794	-2.052	-1.711	-1.688	-1.873	-1.896
2	-1.326	-1.183	-1.400	-1.129	-1.333	-1.173	-1.204	-1.076
5	-1.246	-1.275	-1.416	-1.434	-1.265	-1.377	-1.466	-1.520
7	-1.614	-1.817	-1.665	-1.869	-1.606	-1.662	-1.622	-1.586
8	-1.286	-1.547	-1.370	-1.688	-1.276	-1.519	-1.333	-1.640
9	-1.407	-1.203	-1.471	-1.285	-1.354	-1.238	-1.456	-1.348
10	-1.321	-1.339	-1.439	-1.505	-1.315	-1.333	-1.441	-1.421
11	-1.574	-1.662	-1.371	-1.602	-1.557	-1.618	-1.493	-1.505
12	-1.324	-1.423	-1.386	-1.501	-1.293	-1.381	-1.502	-1.546
13	-1.448	-1.176	-1.503	-1.166	-1.516	-1.252		

Table 54. Chapter 7 Subject Data for Peak TCF (N)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	-769.32	-761.54	-917.74	-994.81
2	-711.34		-692.26	
3	-1087.27	-908.26	-1039.31	-946.60
4	-986.45	-861.89	-1355.50	-1136.56
5	-1352.24	-1566.24	-1149.40	-1279.50
6	-688.00	-754.30	-663.65	-712.89
7	-1176.93	-1346.52	-1302.30	-1301.75
8	-862.80	-942.52	-587.12	-723.60
9	-1117.59	-1327.15	-931.26	-1042.33
11	-1320.25	-1140.73	-915.61	-932.88
12	-1085.96	-1143.96	-750.67	-696.31
13	-784.99	-824.94	-761.84	-851.61
14	-1024.77	-718.04	-649.32	-811.60
15	-1719.11			

Table 55. Chapter 7 Subject Data for Peak MCF (N)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	-575.56	-388.60	-437.27	-509.91
2	-483.04		-288.36	
3	-1081.04	-762.31	-607.56	-534.92
4	-537.90	-511.87	-834.77	-983.78
5	-865.31	-1053.30	-554.96	-666.14
6	-590.75	-719.68	-461.53	-509.16
7	-870.05	-973.42	-764.54	-742.27
8	-426.31	-561.84	-308.14	-393.76
9	-756.47	-922.73	-396.11	-487.04
11	-838.84	-722.09	-360.81	-1302.30
12	-694.03	-391.45	-293.75	-300.48
13	-584.76	-726.91	-453.64	-490.85
14	-640.91	-654.25	-256.05	-245.83
15	-1326.14	-388.60	-378.51	-509.91

Table 56. Chapter 7 Subject Data for Peak LCF (N)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	-305.69	-431.94	-482.62	-541.49
2	-207.87		-405.99	
3	-29.35	-245.21	-448.47	-427.27
4	-476.99	-369.41	-565.82	-342.48
5	-490.43	-526.24	-599.42	-622.71
6	-175.47	-202.34	-207.49	-233.10
7	-262.09	-381.01	-552.07	-591.58
8	-530.14	-555.10	-305.50	-387.35
9	-395.50	-435.41	-590.46	-605.47
11	-484.13	-445.95	-626.55	230.49
12	-443.82	-903.75	-480.42	-435.65
13	-234.62	-205.59	-304.85	-364.40
14	-249.02	-89.77	-502.41	-621.21
15	-624.48		-779.92	

Table 57. Chapter 7 Subject Data for Peak Knee Extensor Muscle Group Force (N)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	698.06	1014.77	929.23	1014.77
2	567.88		693.40	
3	917.64	956.66	946.89	956.66
4	1027.96	1129.77	1355.49	1129.77
5	1249.76	1191.82	1023.07	1191.82
6	584.76	694.70	643.14	694.70
7	960.42	1353.98	1326.14	1353.98
8	807.05	650.75	523.99	650.75
9	1105.98	942.39	897.30	942.39
11	1210.79	724.67	790.13	724.67
12	919.11	655.13	737.80	655.13
13	791.86	808.29	750.39	808.29
14	646.83	696.71	589.29	696.71
15	1621.02		784.60	

Table 58. Chapter 7 Subject Data for Peak Knee Flexor Muscle Group Force (N)

	Non-Replaced Limb		Replaced Limb	
Subject	80 W	100 W	80 W	100 W
1	268.57	346.75	186.09	200.32
2	61.65		54.52	
3	247.72	152.55	176.76	170.05
4	176.87	201.43	121.81	124.54
5	186.96	224.87	253.53	285.93
6	357.81	288.65	145.70	153.80
7	208.85	205.06	193.95	213.38
8	173.18	158.33	185.65	156.64
9	174.79	246.49	282.08	276.08
11	204.89	249.49	292.83	378.28
12	306.84	552.34	438.50	542.20
13	184.99	180.27	111.64	98.61
14	415.74	406.85	376.97	470.52
15			293.41	

VITA

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