LOWER LIMB BIOMECHANICS AFTER TOTAL KNEE ARTHROPLASTY

by

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Abstract

SHANGCHENG WANG. Lower limb biomechanics after total knee arthroplasty. (Under the direction of DR. NIGEL ZHENG)

Knee osteoarthritis is a painful and disabling disease that is prevalent among elder adults. Total knee arthroplasty (TKA) can effectively alleviate knee arthritic pain and improve patient quality of life. Unfortunately, there are evidences showing that (1) patients after TKA still exhibit function limitations in ascending/descending stairs; (2) ankle, knee and hip biomechanics are not restored to normal after TKA; (3) about 20% patients are not satisfied with their current TKA. The objective of this study is to investigate ankle, knee and hip joint biomechanics of TKA patients during activities of daily living (i.e. level walking, stair ascent/descent, sit-to-stand and pivoting), and how three different factors affect their joint mechanics.

The first factor is the existence and progression of contralateral knee osteoarthritis (CKOA). After unilateral TKA, a high percentage of patients have CKOA. The existence of CKOA has been associated high knee adduction moment on the contralateral knee and at least 35% incidence of future contralateral knee replacement in 10 years. However, few studies have examined the effect of CKOA progression on the mechanics of other joints. Thirteen moderate and 13 severe CKOA patients were tested during level walking (LW), stair ascent (SA), stair descent (SD), and sit-to-stand (S2S). The severity of CKOA were classified by clinical decision of contralateral knee replacement and radiographical scoring. As we expected, the results supported that both contralateral ankle and hip biomechanics were altered. Contralateral ankle reduced dorsiflexion and moment/power, while contralateral hip increased hip internal rotation, hip moment and contribution to dynamic support of the body. The progression of CKOA also impacted task performance (reduced speed and increased time) and decreased loading on contralateral leg. Unexpectedly, operated knee axial rotation at heel strike during level walking and knee moment of pulling up body at stair ascent was increased, and hip of the operated side increased abduction angle and moment

during stair ascent/descent. These findings suggested that the progression of CKOA not only changes biomechanics of the affected knee but also impacts the operated knee and hip joints on both sides.

The second factor is the replacement of both knees. Bilateral TKA (BTKA) replaces both knees so it is hypothesized that they would have different biomechanical outcomes and asymmetry from unilateral TKA (UTKA). The biomechanics of ten staged BTKA patients and thirteen UTKA patients during level walking, stair ascent/descent and sit-to-stand were compared. BTKA was associated symmetrical biomechanics, despite that the latest TKA side had a lower peak hip adduction moment during level walking and stair ascent than the first TKA side. UTKA was associated asymmetrical biomechanics. Knee flexion angle and ankle dorsiflexion angle, hip extension moment/power, knee flexion moment and power, and ankle power were lower on the operated side than on the non-operated side. The operated side also reduced knee contribution but increased hip contribution to total support moment during sit-to-stand. Compared to UTKA operated side, BTKA had higher flexion moment and total support moment at weight acceptance of stair ascent, knee power generation at sit-to-stand. BTKA patients also had less extended hip and more anteriorly tilted pelvic during level walking and standing than UTKA patients.

The third factor is the bearing mobility in TKA. Mobile-bearing TKA may promote natural knee rotation and reduce rotation torque at proximal tibia. Twenty MB knees and 17 FB knees were tested during LW, SA, SD, step turn (outside turn) and spin turn (inside turn). Knee rotation angle and knee rotation moment were compared. The results showed that bearing mobility did not significantly change transverse plane biomechanics of the operated knee.

In summary, this dissertation provided biomechanical targets for physical therapists to improve outcomes for different subgroups of TKA patients and evidences for the transverse plane comparison between MB and FB TKA.

Dedication

This dissertation is dedicated to two great women in my life. One is my mother who got healed from bone fracture fifteen years ago. The other one is my choir director who got heavenly heal from cancer six months ago.

Shangcheng Wang

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Table of Contents

LIST OF TABLES

LIST OF FIGURES

LIST OF ABBREVIATIONS

TKA	total knee arthroplasty
UTKA	unilateral TKA
BTKA	bilateral TKA
OA	osteoarthritis
CKOA	contralateral knee osteoarthritis
MB	mobile-bearing
FB	fixed-bearing
OP	operated side
NOP	non-operated side
ASI	absolute symmetry index
SA	stair ascent
SD	stair descent
LW	level walking
S2S	sit-to-stand
PKAM	peak knee adduction moment
PTSM	peak total support moment

Chapter 1 Introduction

Knee Osteoarthritis

Osteoarthritis (OA) is the most common chronic condition of the joints, and the knee joint is one of the most commonly affected area. A study published in 2008 showed that the average lifetime risk of developing symptomatic knee OA was 44.7% (Murphy et al., 2008). Knee OA can result in a painful, stiffness and swelling knee. Although it is not a deadly disease, it affects the quality of life and causes disability for the affected person. It produces abnormal knee joint motion and stress concentration around the knee joint. In clinic, knee OA is indicated by a loss of knee joint cartilage, the appearance of bone spurs and the narrowing of knee joint space.

Knee OA is chronic. It becomes severe as the patients age, and this process cannot be reversed. Currently, there is no cure for knee OA. Some non-surgical treatments to relieve the related pain and disability includes lifestyle modifications, physical therapy, assistive devices, medications and other alternative therapies. When non-surgical treatments are not working to relieve the pain and patients suffer disability, surgical replacement of knee joint surfaces is recommended.

Total Knee Arthroplasty

Total Knee Arthroplasty (TKA) is the gold standard to treat end-stage knee OA. It replaces the diseased joint with an artificial joint: knee implant. The procedure removes the damaged cartilage and bone, and then fix new metal or plastic joint surfaces to restore the knee function. It is generally effective in alleviating knee pain, correcting limb deformity if existed and restoring patient mobility. Different from TKA, partial knee replacement treats knee diseases which affects only partial of the joint such as medial knee OA.

It was first performed in 1968. Since then, a lot of improvements have been made and now it is one of the most successful procedures in all medicine. Between 1991 and 2010, there were more than 3.2 million patients who received primary (first time) TKA identified in Medicare Part A data file and annual primary TKA volume increased 161.5% (Cram et al., 2012). On average, knee replacement costs more than 20k U.S. dollars in 2009 (Ruiz et al., 2013). Although it costs, about 85% of TKA patients are satisfied with TKA and reported a significant pain reduction (Schulze & Scharf, 2013).

Issues with TKA

Wear and Loosening of TKA

TKA fails around 15-20 years after surgery in more than 95% of patients. A failed TKA needs a revision surgery to remove the failed implants and install new ones. There are both mechanical and biological reasons for TKA failure. Just like moving mechanical pairs such as roller bearing, the articulation wears away the implant surface and creates wear debris. Biological reaction to the produced wear particles can also induce osteolysis (bone resorption) which would cause bone loss. The bone loss can increase the risk of bone fracture and loosening of the implants (Gundry, Hopkins, & Knapp, 2017). Impact force during activities in TKA can be higher than intact knee. This may also induce micro-fracture of cancellous bone and produce loosening (Hoshino & Wallace, 1987).

Functional Deficits

It has been reported that at least 20% of the patients are not satisfied with their TKA outcomes (Schulze & Scharf, 2013). Knee instability (poor knee kinematics/kinetics), loss of range of knee motion and function limitations accounts for the dissatisfactions or even failures after TKA. Although many studies that followed TKA patients at both pre- and post-surgery supported a significant improvement in pain and functional status measures, functional deficits and abnormal joint function persists even years after surgery (C. E. Milner, 2009). Many patients with TKA take significant longer time and have a slower velocity to walk, rise from a chair and climb stairs than age-matched healthy people. Improvements are still needed in terms of restoring the ability to normally perform daily living activities (Kim, Bamne, Song, Kang, & Kim, 2015).

Impacts on Other Lower Extremity Joints

The kinematic and kinetic alteration of lower extremity joints may result in poor function performance, the progression of degenerative arthritis in other joints and shortened longevity of knee implants. After unilateral TKA, about 37.2% of patients after TKA also replace their contralateral knee joint and the risk increases with knee arthritis severity (McMahon & Block, 2003). One recent study reported that the chances of a surgical procedure (joint replacement) in contralateral knee, contralateral hip and ipsilateral hip were 45%, 3% and 2% at 20 years, and older age was a significant predictor of ipsilateral or contralateral total hip arthroplasty. So, the risk of hip replacement after TKA in 20 years was 5% in total (Sanders, Maradit Kremers, Schleck, Larson, & Berry, 2017).

Outcome Assessment after Knee Replacement

Clinically-used survey forms to assess patient outcomes after TKA are widely used yet mostly subjective. Simple timed function tests such as 6-minute walking test and 30-second stair climb test are more objective than questionnaires but still provide limited mechanical information regarding TKA. Gait analysis is a powerful tool to understand the mechanics of musculoskeletal system after knee replacement.

Knee joint motion and loading after TKA have been shown to affect the wear of knee prosthesis and have been used as input to simulate implant wear. Biomechanical outcomes also reflect a good or poor mechanics of the knee joint to climb stairs and stand up. The current knowledge in the literature supports the idea that a TKA has good mechanics and long life if it replicates normal knee motion and loading during both static, dynamic, passive and active situations.

Gait analysis using motion capture system is one of the widely used methods in the lab to assess in-vivo biomechanical outcomes after TKA. Gait is the manner or style of walking. The classical gait analysis is studying the pattern of movement of the limbs of animals, including humans, during locomotion. Simple gait analysis, which is just solely based on the observations of human eyes and brain, can qualitatively detect obvious gait abnormality. Modern gait analysis is augmented by instrumentation for measuring not only the movement (kinematics) but also the mechanics of the movement (kinetics) including force, torque, power and work. The detailed theory and techniques will be described in chapter 5.

Assessing biomechanical outcomes after TKA through gait analysis can provide information regarding the range of motion, joint motion, and loading pattern during tested tasks. It advances our understanding on if and how patients following TKA ambulate with different mechanics from healthy counterparts during daily activities. Based on biomechanical findings, physical therapist can more effectively devise certain modifiable target (biomechanical variable) in their physical training program to improve patient function. It can also be used to test the influence of different implant designs on the joint motion and loading pattern after TKA.

Objectives of this Study

The objective of this study is to investigate the factors that may be linked to the listed TKA issues, including the following: (1) patient characteristics such as the progression of contralateral knee arthritis and the replacement of both knees and (2) implant design that may be linked to natural knee motion and longevity of the implants. Previous studies on biomechanical outcomes after TKA neglected these factors and most studies investigated only biomechanics during level walking. Effects of these factors on biomechanical alterations and compensations in individuals following TKA are investigated. Compensations are indicated the differences between the involved limb and the contralateral limb (opposite to the involved limb). Biomechanical compensation reflects inter-limb asymmetry and compensatory strategies.

Arthritis of the Contralateral Knee

Arthritis of the contralateral knee (CKOA) is prevalent among patients after unilateral TKA. Up to 87% of patients awaiting unilateral TKA are affected by bilateral knee osteoarthritis. Previous studies have shown that among unilateral TKA patients with asymptomatic contralateral knee, about 21% developed symptomatic CKOA at 7 years and 37% to 46.0% required a contralateral TKA at 10 years (McMahon & Block, 2003). The contralateral knee joint is the most common joint second to undergo replacement among all remaining joints. The biomechanics and gait asymmetry of the non-operated knee after unilateral TKA have been investigated to explain the high prevalence. The importance of managing CKOA before and after surgery is increasingly recognized.

The progression of CKOA is irreversible and hurts the contralateral knee. It may also bring adverse changes to other joints. To our knowledge, few investigations have examined the effect of increasing levels of CKOA severity on biomechanics of all the lower extremity joints. This thesis filled this gap and shed light on the biomechanical changes of lower extremity joints as the CKOA progresses from moderate status to severe status. The findings help us to see the potential benefits of treating and delaying the progression of CKOA on other joints as well as the contralateral knee joint.

Bilateral versus Unilateral TKA

Eighty percent of elder people (age over 64) are affected by osteoarthritis in the knee and about one third of them have symptoms in both knees. Patients who need two knees replaced can choose to stage the two surgical procedures within an interval of months. Staging two knee replacement procedures increase the cost because patients need two hospitalization and two rehabilitation experiences. But staged bilateral TKA places less burden to the patients during recovery than simultaneous bilateral TKA (replacing two knees in one hospitalization), because patients have at least one good knee to stand on during recovery. According to a report published in April 2016 by Canadian Institute for Health Information, about 9.0 % of all patients receiving TKA received staged bilateral TKA. Bilateral TKA replaces both knees and should have improved gait symmetry and biomechanical outcomes.

Unilateral TKA is more common than bilateral TKA. It generally alleviates the symptoms in the most affected knee. The knee contralateral to the operated knee may still suffer from non-severe arthritis (if severe, both knees should be replaced). Inter-limb biomechanical asymmetries for patients after unilateral TKA have been reported in the literature.

This thesis examined whether staged bilateral TKA result in better functional outcomes and gait adaptations/symmetry than unilateral TKA. No studies have been performed on this question. What is more, a comparison between unilateral and bilateral TKA patients is needed to investigate the effect of recruiting mixed unilateral and bilateral TKA patients as subjects on biomechanical outcomes. McClelland et al. 2007 reviewed gait studies about TKA. They found most studies included unilateral TKA patients and were not consistent in excluding or including bilateral TKA patients.

Mobile- versus Fixed-Bearing TKA

The nature knee rotates about the vertical axis of the tibia during dynamic activities. During passive knee flexion-extension, the healthy knee relies on a passive system of joint contact surfaces, ligaments and menisci to provide internal control of knee motion. TKA results in a knee joint without menisci and typically lacking cruciate ligaments. Yet axial rotation is still allowed in TKA. The function of the lost structures in regulating axial rotation is replaced by the shapes of articulating surfaces, ligament substituting mechanism and the mobility of the tibial bearing. Fixed-bearing TKA is the conventional type and allows limited degree of axial rotation. However, as the knee axially rotates, the congruity of the articulating surfaces decreases and contact stresses increase. The concepts of meniscal-bearing and rotation-platform was introduced into the TKA community around late 1970s to optimize contact area and reduce wear. Mobile-bearing TKA adopts these design concepts and is believed to be able to facilitate knee axial rotation. It may be able to promote load sharing and dissipating between knee implant and surrounding soft tissues. The mobile bearing is also believed to reduce rotation torque transmitted to the proximal tibia and thus improves the rate of implant loosening.

This study investigated the effect of the implant design on the transverse plane biomechanics during multiple daily activities. To provide a comprehensive comparison between mobile- and fixed-bearing TKA, this thesis investigated biomechanical outcomes not only during level walking, but also during four other walking-related tasks: stair ascent, stair descent, step turn after level walking and spin turn after level walking. Step turn is making turn to the swing side and spin turn is making turn to the standing side. These two turning activities challenged the TKA and might show the advantage of the mobile-bearing design.

Chapter 2 Knee Anatomy and Total Knee Arthroplasty

Anatomy of Intact Knee

Two Articulations and Three Compartments

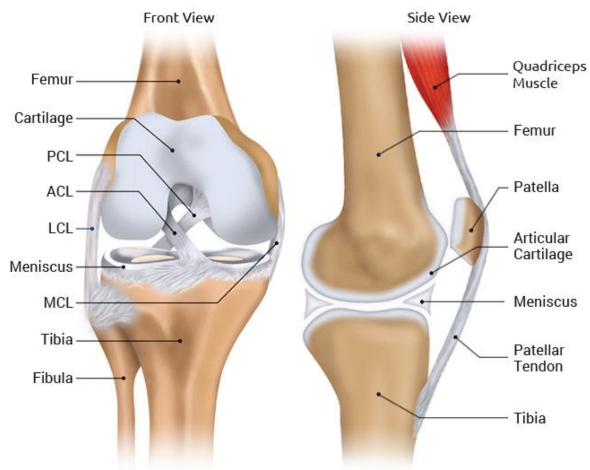


Figure 2.1 Knee Anatomy (http://www.stevenchudikmd.com/knee-surgeon-chicago-illinois/).

Human knee joint, if only considering the primary motion, is like a hinge joint in the field of mechanical engineering. It connects the thigh bone (femur) and shin bone (tibia) of our lower extremity (see front view in Figure 2.1). The knee can bend and straighten so it allows almost frictionless articulation between the femur and the tibia.

Except femur-tibia articulation, knee joint has one more articulation between the femur and the patellar bone (also known as the kneecap, see the side view in Figure 2.1). But to be convenient, we commonly refer tibiofemoral joint as the knee joint. The patellar articulates above the femur groove and is the core to the function of quadriceps. It acts as a pulley which transmits the pulling force of quadriceps muscle to the tibia through patellar tendon.

The knee joint can also be divided into three compartments: lateral compartment (the outside part of the tibiofemoral articulation), medial compartment (the inside part of the tibiofemoral articulation) and patellofemoral compartment (Figure 2.2). Total knee replacement often replaces all three compartments and perfect for patients who have arthritis in all compartments. Sometime knee arthritis is limited to only one compartment. Unicompartmental knee replacement is the best surgical option for such patients as it preserves other parts of the knee.

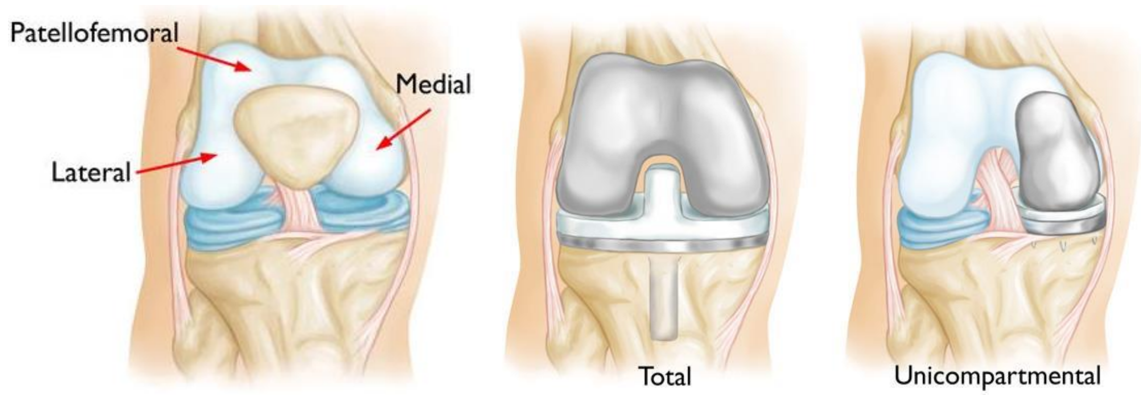


Figure 2.2 Diagrams showing three compartments of the knee joint. Total knee arthroplasty replaces all three compartments, while unicompartmental knee arthroplasty only replaces the one of the three compartments.

Bones, Cartilages and Meniscus

The bones are like rods and links in mechanical mechanism. They are the major structures to sustain/transmit external loading and provide sites for other connective tissues. Three major bones are involved in the knee joint, which are the femur, the tibia and the patellar. Both the femur and tibia are long bones, which are composed of cortical bone and cancellous (sponge) bone. Cortical bone is dense, stiff and constitute the outer shell of the long bone. The sponge bone is porous structure and adapts to external loading. The sponge bone is sandwiched in the cortical bone which helps to absorb external shock and resist fracture. The patellar bone is a small sesamoid bone. It covers and protects the anterior articular surface of the knee. All the bones provide attachment sites for ligaments and tendons. All the bones also provide fixation sites for knee implants, but some parts of the bones are cut into shapes and holes to conform to the shape of the implants.

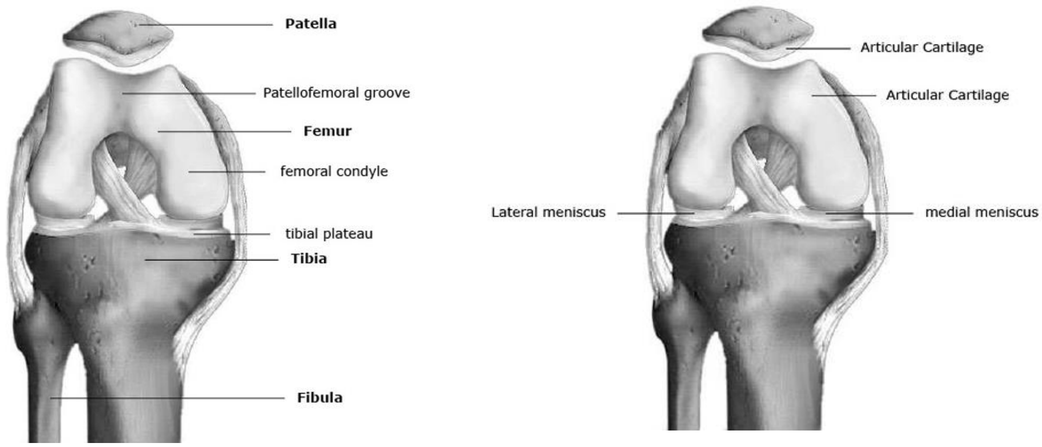


Figure 2.3 bones, articular cartilage and meniscus of the knee joint (from https://www.healthpages.org/anatomy-function/knee-joint-structure-function-problems/, accessed on 9/19/2018).

The shapes of the articulating surfaces are important for the knee kinematics. The top articulating surfaces of the tibiofemoral joint are curved and smoothed lateral and medial condyles of the femur (Figure 2.3). The lower articulating surfaces of the tibiofemoral joint are the lateral and medial meniscus together with the articular cartilage on the tibia plateau which is relatively flat. The articulating surface of the patellar bone is dome shaped and fits the groove shape the anterior articulating surface of the femur.

The articular cartilage is like the bearing in mechanical mechanism and acts joint lining. All articulating parts of the three bones are covered with cartilage. The articular cartilage articulates with the articular cartilage on the opposing bone. It is a very smooth and lubricated tissue. It is 2 to 4 mm in thickness and constitutes of water (80% of its wet weight) and organic matrix (60% dry weight is collagen). Though very thin, it provides almost zero friction coefficient for dynamic joint articulation and facilitates load transmission. It can adapt itself to mechanical environment. However, it has no blood vessel and it has limited intrinsic ability to heal and repair itself once damaged (Sophia Fox, Bedi, & Rodeo, 2009). One of the primary processes in osteoarthritis is the cartilage in the joint breaks down. The degeneration of cartilage damages and thins the tissue and may eventually lead to bone-on-bone contact.

A meniscus is a crescent-shaped (like the word C) structure that only partially divides the lateral or medial tibiofemoral joint cavity. The knee joint has two meniscuses, i.e. the lateral and medial meniscus (Figure 2.3). They served as pads to increase the congruity and the contact area of tibiofemoral articulation. The concave surface of meniscus on top of the tibia plateau enables effective articulation with convex femoral condyles. It can accommodate compression from body weight and other axial forces by deforming itself and slightly moving. The medial meniscus can move on average 2-3 mm while the lateral one can have greater anterior-posterior displacement (around 9-10 mm) during flexion (Fox, Bedi, & Rodeo, 2012). It will be illustrated in this thesis how the design of total knee replacement mimics the function of meniscus.

Muscles and Ligaments

The muscles are the primary movers (actuators) of the knee joint. It is like motor in a mechanical mechanism and drives the bones and joints to move. It also stabilizes the joint and balance external loading. For example, the quadriceps muscle group is the major knee extensor. The contraction of this muscle group generates a torque which can either balance external knee flexion moment or extend the knee. The hamstring is the major knee flexor. The contraction of hamstring generates a torque that either balances external knee extension moment or bends the knee.

The ligaments around the knee joint provide its internal passive stability and limit any excessive motion such as joint dislocation. There are four major ligaments around the knee joint (Figure 2.1): anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), medial collateral ligament (MCL) and lateral collateral ligament (LCL). Histological studies have found that mechanoreceptors (sensory nerve endings) are functioning in ACL and PCL. They can also act as sensors. They sense changes in ligament tension and length and may provide joint position information of the knee to the brain.

The mechanical function of ACL and PCL in guiding knee anterior-posterior motion has been illustrated with a simple four-bar linkage mechanism. Femur bone, tibia bone, ACL and PCL make up the four bars in the sagittal plane. The passive motion of the knee in the sagittal plane is regulated by this mechanism (Figure 2.4). The underlying assumption for this mechanism is that ligaments are inextensible link considering its high stiffness during passive motion.

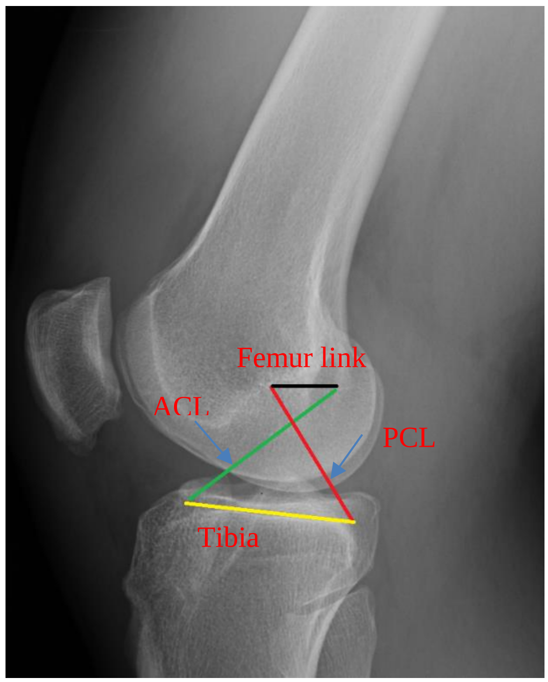


Figure 2.4 Four-bar linkage diagram illustrating the function of anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL) in guiding knee kinematics.

Knee Arthritis

Prevalence and Impact

Osteoarthritis (OA) is the most common chronic condition of the joints. The knee joint is one of the most commonly affected areas. In a 1991-1994 survey on the Americans, 37.4% US adults had radiographic knee osteoarthritis and 12.1% had both symptomatic and radiographic knee osteoarthritis, and the prevalence of knee osteoarthritis was higher among women than men

(Dillon, Rasch, Gu, & Hirsch, 2006). As the ageing and increased obesity of the US population, the prevalence of knee arthritis is increasing.

Knee arthritis impacts both the patient and the society. It limits the patient's physical function and quality of life. As knee arthritis is a chronic disease, it also brings economic burden to treat the disease for both the patient and the society.

Diagnosis and Grading

According to the definition given by Arthritis Foundation, osteoarthritis is also called degenerative joint disease or "wear and tear" arthritis. The wear and tear happen to the articular cartilage and the progression of OA gradually breaks down the cartilage. The degraded cartilage gradually loses its smoothness and eventually bone rubs on bone during articulation, and bone spurs (osteophytes) develop along bone edges (Figure 2.5). Besides cartilage degradation, subchondral bone also thickens during this process. At the same time, muscles around the knee become weak and inflammation may occur to the synovium tissue and tendon.

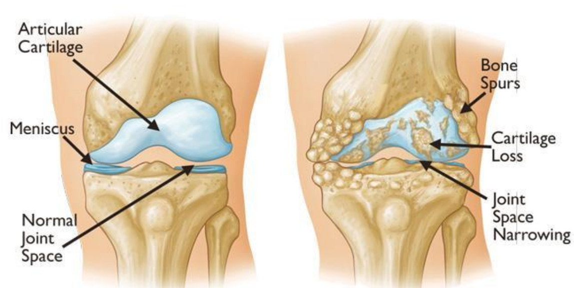


Figure 2.5 Knee Arthritis

The diagnosis of knee OA can be based on American College of Rheumatology criteria for the diagnosis of knee osteoarthritis. It largely depends on factors such as pain in the knee, clinical examinations, radiographic findings and laboratory findings. Symptomatic assessment is typically based on self-reported symptoms (such as knee pain, stiffness and swelling), simple clinical examinations (e.g. palpation, range of motion, pain assessing during motion) and clinical observations such as limb deformity and functional performance.

It is a progressive and irreversible disease. As the disease progresses and cartilage wears away, the joint space is becoming narrower and narrower (Emrani et al., 2008). Imaging test using X-rays is the standard method for assessing radiographic progression of knee OA. Bilateral weight bearing plain films are typically used to measure osteophyte presence, joint space narrowing, sclerosis (thickening of the subchondral bone) and bony deformity. The Kellgren and Lawrence (KL) scoring system is commonly used to classify the severity of knee osteoarthritis (OA) using 5 grades (grade 0 to grade 4) (Emrani et al., 2008). For example, grade 0 represents no radiographic feature of OA. If there are large osteophytes, marked joint space narrowing, severe subchondral sclerosis and definite bony deformity, the knee OA is the most severe one and is classified as grade 4.

Mechanical Pathway of Knee Arthritis

Knee osteoarthritis can be classified as primary (of unknown origin) or secondary to a known medical condition or injury. For example, the rupture of anterior cruciate ligament or the tear of meniscus is associated with a high risk of developing secondary knee osteoarthritis. Ageing, obesity, and Vitamin-D deficiency are typical risks for developing knee arthritis (Felson, 1990).

Biomechanical elements have been linked with the initiation and progression of knee OA. Mechanical quantities including force, force distribution, loading rate and knee joint kinematics have been explored (Wilson, Mc Walter, & Johnston, 2009).

Force on the cartilage of each compartment, which quantifies the direct mechanical loading to the tissue, is an ideal but hard-to-measure biomechanical element. Joint kinematics such as knee adduction angle and joint kinetics including hip adduction moment, knee adduction moment and knee flexion moment have been reported as relevant factors or surrogates of force on the cartilage. A lower hip adduction moment, although not directly related to the knee, has been correlated with a lower risk of knee OA progression (Chang et al., 2005). Knee adduction moment (KAM) adjusts the distribution of force between medial and lateral compartment. An increased KAM is correlated to a higher force on the medial compartment than the lateral compartment. KAM is substantially reduced in both knees by use of lateral wedge insoles which might protect the knees from OA progression (Jones et al., 2013). KAM is also associated with knee frontal plane alignment (hip-knee-ankle alignment on the frontal plane). Varus tibiofemoral mal-alignment (bow knee) is more correlated with the incidence of medial tibiofemoral OA, and lateral tibiofemoral mal-alignment (is more correlated with lateral tibiofemoral OA. With this factor in mind, wedged insoles have been designed with the aim to mediate knee adduction moment and slow down the progression of knee OA.

Loading rate of knee adduction moment during weight acceptance phase of walking have been investigated to assess the ability of the knee to absorb shock (Morgenroth, Medverd, Seyedali, & Czerniecki, 2014). Knee joint kinematics change especially knee axial rotation change has been illustrated as a shift of contact area to thin cartilage area which is not accommodated to loading (Andriacchi et al., 2004).

Compared to healthy people, patients accommodate knee OA with a reduced gait speed which may reduce joint loading. Knee OA patients may also have other kinematic strategies to reduce joint loading including increasing toe-out angle, leaning trunk toward the affected limb, and reducing knee flexion angle during walking.

Management of Knee Arthritis

The symptoms can be effectively managed using conservative treatments especially when the knee arthritis is not severe. Medications are typically employed to relieve knee pain. A variety of exercise programs that aim to increase knee muscles and range of motion have been developed by physical therapists. Life style changes such as losing weight, using assistive devices and adding a bench to the shower can also reduce the knee pain during standing.

When conservative treatments are not helping to relieve the symptoms, injection of steroid to the knee joint compartments is often utilized. If knee arthritis damages one side of your knee compartments more than the other side, the worn-out side (often medial compartment) has more weight bearing and aligning the bones by cutting/adding a bone wedge can adjust that asymmetry. However, when arthritis comes to the end-stage, TKA is the gold standard to treat it.

Total Knee Arthroplasty

Total knee arthroplasty is also known as total knee replacement. It is the golden standard to treat end-stage knee arthritis. The procedure replaces the bone-on-bone articulation of knee joint with metal-on-plastic articulation, so that the pain and abnormal mechanics of the knee could be fixed. Most patients receiving TKA are aged 50 to 80. It is estimated that 4.7 million Americans are living with artificial knees. In 2010 there were 4.7 million people (3 million women and 1.7 million men) with knee replacement. The 2010 prevalence of total knee replacement in the total U.S. population was 1.52% (Maradit Kremers et al., 2015). There is also trend toward a substantial rise over time and a shift to younger ages.

The Surgical Procedure

The first knee replacement was performed in late1960s and since then a lot of improvements have been made for both surgical techniques and implant designs. One TKA procedure takes about 1 to 2 hours. After admission and anesthesia, the patient is ready to receive knee replacement surgery. A brief procedure is shown in Figure 2.6. First, a straight midline skin incision line that starts from 2 to 4 cm above the patellar and ends at tibia tubercula. There are three standard approaches to evert patellar laterally and expose tibiofemoral joint: medial parapatellar approach, min-vastus approach, and sub-vastus approach. Before resecting tibia/femur, leg alignment is done to guide cutting and assess alignment. Soft tissue balancing assesses joint gap in medial and lateral compartments, and in knee flexion, mid-flexion and extension (Bottros, Gad, Krebs, & Barsoum, 2006). Symmetrical and balanced gaps ensure good alignment and are good for implant durability. The medial and collateral ligaments are persevered, and the one which is tight may be released during soft tissue balancing.

If patellar needs surfacing, the patellar resection is performed during any time of the surgery. Patellar bone section is measured to determine patellar component size and ensure same bone thickness after surgery. After preparing the bone and ensuring alignment, tibia component, femoral component and patellar button are pressed fit into the bone and this fixation can be done with cement or without cement.

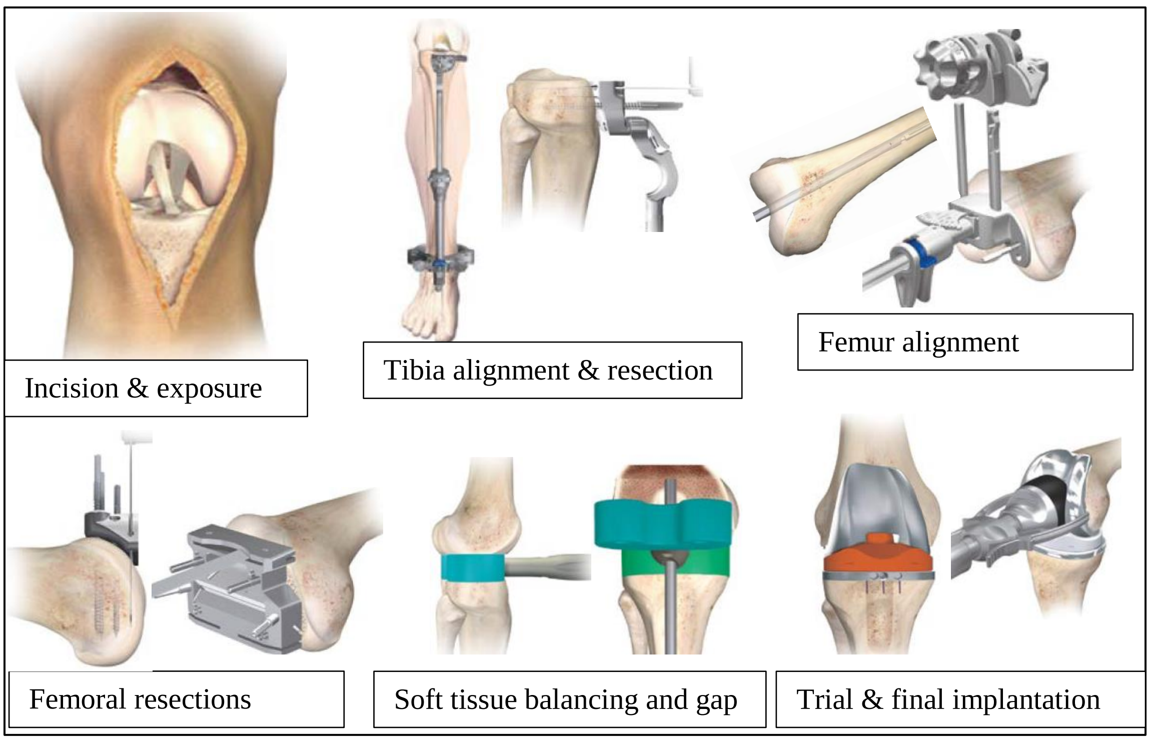


Figure 2.6 TKA surgical procedures (from SIGMA® primary knee system, balanced surgical technique guide of DePuy Synthes Joint Reconstruction)

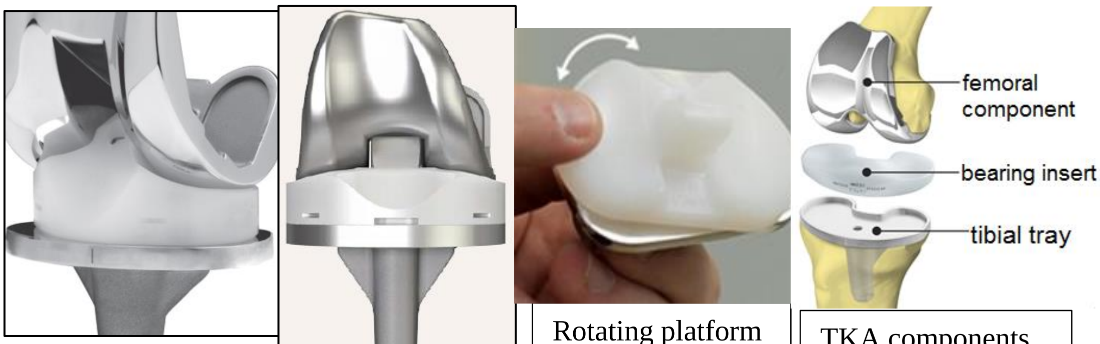
Modern Knee Implant designs

TKA implant is composed of four basic components: femoral components, bearing insert, tibial tray and patellar component. Components of tibiofemoral joint are shown in Figure 2.7.

Although the design intent of knee prosthesis is to replicate normal knee function during daily activities, there are structures and functions which knee implants cannot fully replicate. For example, ACL ligament at the time of total knee replacement is likely distorted by arthritis disease and may not provide normal function such as stabilizing the joint. Thus, ACL is often resected, and PCL is resected or preserved depending on both the chosen implant design and the integrity of the PCL. The resecting of ACL and PCL affects knee AP instability and rotatory stability.

Cruciate-retaining type TKA preserves only PCL or both ACL and PCL. The integrity of these ligaments may be questionable due to the process of knee arthritis. What is more, potential over-tensioning of the preserved ligament may limit the range of knee motion (Okada et al., 2018). On the other hand, cruciate-substituting type replaces the function of the resected cruciate ligaments. For example, posterior-stabilized TKA design mimics the function of PCL by engaging a post-cam mechanism to constrain AP translation and axial rotation. The post is the spine of the bearing insert while the cam is on the femoral component. The engagement of post-cam at high flexion angle (>70 degree) can promote natural roll-back. Bi-cruciate substituting TKA is a relative new design and designed to substitute both ACL and PCL (Murakami et al., 2018). All the TKA patients tested in this dissertation used posterior stabilized design.

Bearing insert is to replicate the articular cartilage and meniscus on top of tibia plateau. It is the only plastic component in modern TKA. Conventional TKA design is fixed-bearing TKA which locks the bearing insert on top of the tibial tray. TKA design aims at reducing contact stress by increasing tibiofemoral conformity during knee flexion-extension. However, knee rotatory laxity is reduced when conformity is increased. Mobile-bearing design is to resolve the conflict between conformity of the articular surfaces and knee rotatory/AP laxity. The plastic spacer (i.e. bearing) can rotate above tibial component in mobile-bearing TKA (Figure 2.7), while in fixed-bearing TKA the bearing is fixed or allowed to have very limited degree of axial rotation or AP translation.



Mobile-bearing TKA | Fixed-bearing TKA | Rotating platform | TKA components

Figure 2.7 Mobile-bearing and fixed-bearing TKA.

Rehabilitation and Training after TKA

Exercise is typically recommended to restore strength and mobility of the operated knee. Early post-operative exercises include straightening knee, straight leg raises, ankle pumps, bed-supported knee bends, and sitting-supported knee bends. Typical early activities including walking, stair climbing, and stair descending are also recommended for patients after surgery. Exercise and activity trainings could be at home and on machines.

Finding the right targets to improve and modify is important. Knee muscle weakness and limited range of knee motion have been well recognized in patients after TKA. It has been reported that post-op quadriceps strength was decreased by 62%, voluntary activation was decreased by 17%, and maximal cross-sectional area was decreased by 10% in comparison with the preoperative values. Failure of voluntary muscle activation and atrophy explained 85% of the loss of quadriceps strength (Mizner, Petterson, Stevens, Vandenborne, & Snyder-Mackler, 2005). So conventional post-operative rehabilitation programs aim to strength knee muscles and improve range of knee motion.

Strengths of the hip and ankle after TKA are also important, because there is strong coupling among the lower extremity joints. It has been shown early after TKA surgery (1-month) ankle plantar-flexors and dorsiflexors were respectively 17% and 18% weaker than pre-operative strengths (Judd, Eckhoff, & Stevens-Lapsley, 2012). Although at 3 and 6 months after surgery, ankle and knee strengths are restored to pre-operative level.

It has also been shown that hip abductor strength is impaired before surgery and worsens following surgery(Loyd et al., 2017). One recent study has shown that hip adductor strength contributes to physical function after TKA (Alnahdi, Zeni, & Snyder-Mackler, 2014).

Outcomes after TKA

Since there are definite differences between natural knee and TKA and the invasive nature of TKA surgery, it is of great value to evaluate and document patient outcomes after surgery.

Knee Scores

Giving patients a standardized questionnaire to fill out during their clinical visit is the easiest method to assess their outcome. The questionnaires mostly ask the patient to answer questions regarding the level of pain, satisfaction, expectation, and physical function of the knee. Some items such as range of motion, stiffness (limitation of motion), stability, and alignment are filled by physicians. Score is given for each checked item. An increase in the total knee score after an introduced intervention indicates an improvement. WOMAC, SF-36 and Knee Society Score are typical forms for TKA patients. These forms have been designed to be valid, reproducible and responsive to changes in a patient’s condition (Roos & Toksvig-Larsen, 2003).

One goal of knee replacement is to restore range of knee flexion-extension and reduce knee pain. These forms directly evaluate these aspects, and the advantage of such method is that it is simple and economic to ask patients to score themselves. The disadvantage of such method is that it is mostly subjective.

Timed Functional Tests

Functional tests such as 6-minute walking test, 30-second stair climb test and time-up-to-go test have also been used to assess functional capacity after TKA (Mizner et al., 2011). Six-min walking test measures the distance walked over a span of 6 minutes. This test evaluates all the physical systems including musculoskeletal, pulmonary and cardiovascular systems during walking. Timed up-to-go test tests how long it takes a patient to stand up, walk 3 meters, make a turn, walk back and sit down. This test evaluates several subcomponents of mobility (walking, turning and sitting/standing) and static/dynamic balance. These outcome measuring tools are objective and simple yet provide just the general information on the biomechanical outcome after knee replacement surgery.

Laxity and Stability Tests

Knee laxity and stability measures are important biomechanical outcomes after TKA. A poor knee stability can lead to failure in TKA. Knee stability and laxity are the extension of the concepts of stiffness and elastic modulus in mechanical engineering. Passive laxity is provided by soft tissues and contact of articular surfaces, and active stability is also contributed by the contraction of muscles. Instrumented anterior and posterior drawer tests (Figure 2.8) have been performed to quantify sagittal laxity after PCL-retaining TKA, and laxity is quantified as the displacement under a certain anteroposterior force to the proximal tibia or distal femur (Chouteau, Lerat, Testa, Moyen, & Banks, 2009). Knee laxity and stability have also been measured during surgery using navigation system. However, knee stability and laxity during dynamic motion are different from measures during static or passive motion.

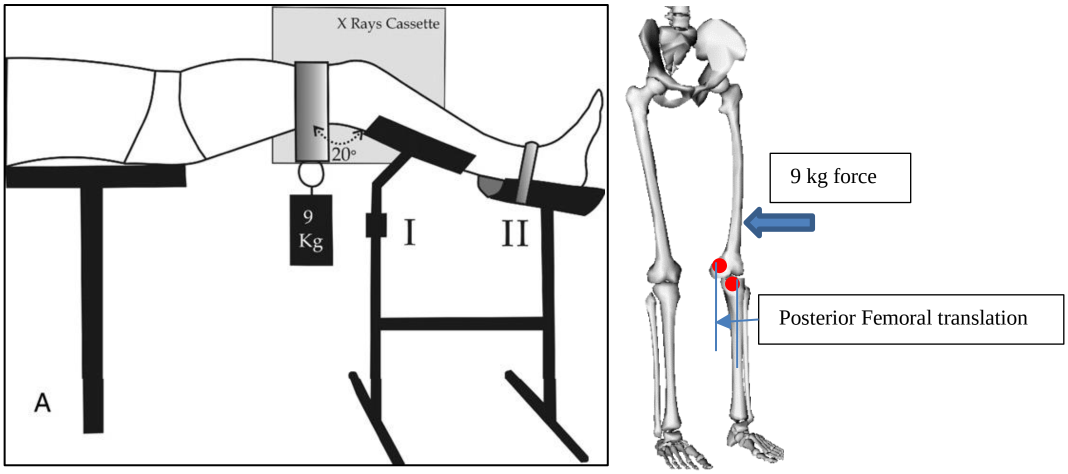


Figure 2.8 (Chouteau et al., 2009) measured sagittal plane laxity after total knee replacement. A weight is applied to the distal femur and X-ray measures anteroposterior translation.

Biomechanical Outcomes of the Knee

Knee kinematics and kinetics during daily activities provide measures of knee stability and laxity during daily activities. Knee (tibiofemoral) joint kinematics and kinetics after TKA have been reported in many studies. Knee joint kinematics is composed of three rotations and three translations. The three rotations are knee flexion-extension, knee internal-external axial rotation and varus-valgus rotation (Figure 2.9). The three translations are knee anterior-posterior (AP) translation (AP drawer, Figure 2.8), medial-lateral (ML) translation and superior-inferior (SI) translation (distraction/compression).

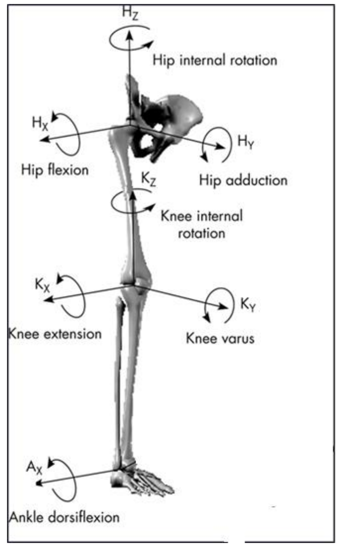


Figure 2.9 Ankle, knee and hip angular kinematics (a) and knee translational kinematics (b).

Knee flexion-extension is the primary motion. Range of flexion-extension is an important indicator of the success of TKA and can be as high as 150 degrees for some patients. Patients with post-operative hyper-extension of greater than 10 degrees or flexion contracture (knee flexion greater than 15 degree) are associated with poorer outcomes. Patients with greater range of motion tended to have higher score for stair climbing (Ritter, Lutgring, Davis, & Berend, 2008). Range of axial rotation are much smaller than knee flexion-extension. Mean axial rotation range of mobile-bearing TKA has been reported to be 13.4° during gait and 21.0° during sit-to-walk with turning steps. Mean axial rotation range of fixed-bearing TKA has been reported to be 9.7° during gait and 14.3° during sit-to-walk with turning steps (Zurcher et al., 2014).

AP translation of TKA during dynamic tasks reflects dynamic knee AP stability. Tibiofemoral AP contact positions have been quantified. Dennis et al. 2005 found that patients implanted with MB TKA tended to have a less AP translation of both the medial and lateral condyles than those implanted with FB TKA.

Knee kinetics is also important outcomes after TKA. External knee flexion moment (Figure 2.10) reflects the internal knee extension moment generated by knee extensors (quadriceps muscle). A well function knee extensor is required to ascent stairs and stand up. External knee adduction moment ((Figure 2.10) adjusts the distribution of medial and lateral compartments. External knee rotation moment applied to the tibia affects the fixation of knee implants.

It has been reported that TKA has improved knee biomechanics. Compare to before surgery, high peak knee adduction moment and knee adduction angle appear to be reduced, and low peak knee flexion moment during walking is likely increased, while increase in peak knee flexion angle during early stance of level walking is consistent in the literature (Sosdian et al., 2014).

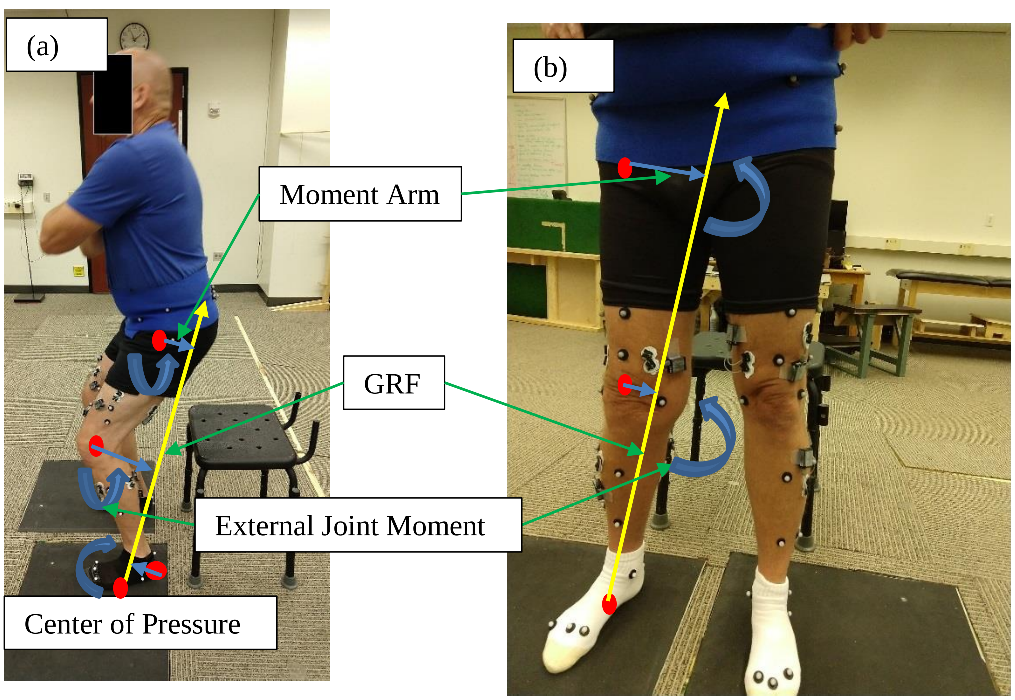


Figure 2.10 External joint flexion/extension moment (a) and external hip/knee adduction moment (b) during sit-to-stand.

Ankle Joint Kinematics and Kinetics

Ankle joint complex can include subtalar joint, talocrural joint, and tibiotalar joint. This study mainly focused on the sagittal plane motion and loading of the ankle joint. The distal tibia bone rests in talus bone. Talus bone is widest in anterior region and the joint is more stable in dorsiflexion. Ligaments in anterior and posterior of the ankle joint withstand tensile forces (shear forces at the joint) exerted on the ankle joint. Tibia anterior muscle contributes to ankle dorsiflexion. Muscles on the posterior of tibia contributes to ankle plantar-flexion. Internal ankle plantar flexion moment tends to plantar flexion the ankle and pushes off from the foot to propel the body forward during level walking.

While most literature on TKA biomechanics focuses on the knee joint itself, TKA is suspected to impact other lower extremity joints such as ankle and hip as well. Ankle, knee and hip are not independent and together contributes to overall function of an individual. It has been reported that gait speed is largely determined by hip and knee joint moments in healthy controls. Ankle and hip compensations to TKA have been reported.

After TKA, the ankle plantar flexion moment during level walking is increased with the increase in gait speed. The proportion of ankle proportion to the total support moment is higher than controls. C.D. Samaan 2010 reported that contralateral ankle compensated during walking and chair rising in patients after TKA. Levinger et al. 2013 found that knee kinematics and kinetics did not significantly change between pre-op and 12 months post-op. They found ipsilateral ankle increased plantarflexion moment, dorsiflexion moment, and push-off power (A3 power) during level walking. They argued that ankle compensated for knee impairment. Pozzi et al. 2016 suggested that patients after TKA compensated at the hip and ankle joints to step up and over. D. Thewlis 2009 followed patients before and after TKA. They found ipsilateral ankle increased external rotation, dorsi- and plantar-flexion moments during walking.

Biggs et al. 2016 reported that pre-operative ankle range of motion during walking was one of the predictors of post-operative function. Fenner et al. 2017 stated that ankle and hip biomechanics were important to understand biomechanical outcomes of TKA during stair descent.

Hip Joint Kinematics and Kinetics

Hip loading asymmetry has often been reported for patients after unilateral TKA. Most authors found lower hip loading on the operated side. Alnahdi et al. 2016 reported that hip loading was asymmetry after unilateral TKA during sit-to-stand. Operated limbs had a lower hip extension moment than non-operated limbs. Mizner et al. 2005 reported that quadriceps muscle activity and hip extension moment were smaller in operated limbs than non-operated limbs during sit-to-stand. However, Yoshida et al. 2008 found operated limbs had greater hip moments and contribution to support moment than non-operated limbs at 12 months after TKA.

Hip kinematics and kinetics have been reported to be abnormal after TKA and is explained as a compensation strategy. Standifird et al. 2016 showed that both limbs had a higher pushing-off peak hip adduction moments than controls, and operated limbs also had a higher loading peak hip adduction moment during stair ascent. Pozzi et al. 2016 found that patients with TKA reduced contribution from operated knee by increasing hip and ankle contributions to total support moment during step up and over task than healthy controls. Fenner et al. 2017 showed TKA patients had a more externally rotated hip during stance phases of both level walking and stair descent than healthy controls. Gaffney et al. 2016 found that patients with TKA compensated with both the knee and hip, with increased hip external rotation, decreased knee flexion, decreased quadriceps force, and decreased hip abductor force during inside/outside pivoting tasks and stair descending. Thewlis 2009 reported that patients with TKA increased hip external rotation and hip flexion moment. During walking, patients with TKA decreased coronal plane range of hip motion. Faquhar et al. 2008 found subjects with TKA still increased hip flexion and hip extensor moment to perform sit-to-stand task, although quadriceps muscle strength increased. Saari et al. 2004 reported that subjects with TKA tended to reduce hip extension angle and hip extension moment during stair ascent/descent compared with healthy controls. Byrne et al. 2002 reported that subjects with TKA increased hip extensor work of the lead limb compared with controls during step-up task. Su et al. 1998 reported that patients increased vertical hip joint forces, peak sound-side hip extension moment during chair rising compared with controls.

Chapter 3 Study Aims and Relevant Literature Reviews

Study Aim #1: Concurrent Osteoarthritis of the Contralateral Knee

Literature Background

Millions of individuals benefit from TKA in reducing the knee pain. Unfortunately, among patients after unilateral TKA, concurrent osteoarthritis of the contralateral (the opposite side of TKA) knee (CKOA) is highly prevalent and often progresses. It is estimated that about 87% of patients with knee OA have bilateral radiographic knee OA (Gunther et al., 1998). What is more, CKOA progresses to contralateral TKA with an overall 10-year risk of 37.2%, and the risk increases as CKOA severity increases (McMahon & Block, 2003; Tanavalee, Thiengwittayaporn, Ngarmukos, & Siddhiphongse, 2004).

The presence of arthritis of the contralateral knee is irreversible and may impact the biomechanics of the lower extremity joints on both sides. However, to our knowledge, most investigations did not examine the effect of increasing levels of CKOA severity on biomechanics of other lower extremity joints, and unilateral TKA has been used as a generic inclusion criterion and little attention has been paid to the severity of CKOA (McClelland, Webster, & Feller, 2007; C. E. Milner, 2009). Only a few studies considered the presence of CKOA. After unilateral TKA, the contralateral knee is reported to have higher knee adduction angle (more varus) and adduction moment than the treated knee during level walking (Alnahdi, Zeni, & Snyder-Mackler, 2011; Clare E. Milner & O'Bryan, 2008). Portia P.E. Flowers 2014 reported that sagittal plane asymmetries throughout mid-stance and frontal plane asymmetries throughout the stance remain up to 2 years after unilateral TKA surgery, and the pain and adduction moment of the non-operated knee were the primary predictors of contralateral TKA. These studies found asymmetry in knee adduction moment but did not assess the effect of CKOA progression on the function of other joints.

Previous studies have studied the effect of the progression of knee OA in people without TKA. Change et al. 2005 followed 57 persons with knee OA from baseline to 18 months and subgroup the patients into progressive and non-progressive group. They reported that a lower ipsilateral hip abduction moment during walking at baseline was associated with the progression of medial tibiofemoral OA. Hatfield et al. 2015 identified variables that were linked to future TKA were knee adduction moment magnitude and knee flexion/extension moment difference and stance-dorsiflexion moment at the baseline. But the effect of knee OA progression on gait mechanics was not known because they did not measure gait mechanics at 18 months. One study reported that biomechanics at ankle, knee and hip joints changed when knee osteoarthritis progressed from moderate phase to severe phase (Astephen, Deluzio, Caldwell, & Dunbar, 2008; Astephen, Deluzio, Caldwell, Dunbar, & Hubley-Kozey, 2008). It is not clear if unilateral TKA patients had similar changes as these people without TKA.

Classifying the severity of CKOA is the first step to perform this study and can be based on radiographic findings alone. However, there is substantial discordance between the radiographic classification, patient-reported knee pain, and physician's diagnosis (Hannan, Felson, & Pincus, 2000). A previous study classified knee osteoarthritis based on the combination of radiographic finding and clinical decision for treatment with (severe) or without (moderate) TKA surgery (Astephen, Deluzio, Caldwell, & Dunbar, 2008).

Hypotheses

The purpose of this study was to identify the difference in outcomes and identify biomechanical changes as moderate CKOA progresses to severe CKOA. We hypothesized the following:

(1) Severe CKOA patients have poorer performance than moderated CKOA patients.

(2) The progression of CKOA impacts the biomechanics of the operated knee (TKA).

(3) The progression of CKOA affects the biomechanics of both contralateral and ipsilateral ankles.

(4) The progression of CKOA affects the biomechanics of both contralateral and ipsilateral hips.

(5) The progression of CKOA decreases load on the contralateral leg and increases load on the ipsilateral (operated) leg.

(6) The progression of CKOA affects the joint contribution to dynamic support of the body.

(7) Severe CKOA patients had poorer asymmetry than moderate CKOA patients

Spatiotemporal variables such as gait speed, sit-to-stand time, stance duration of stair ascent/descent were used to represent the function of a patient. From the findings of previous knee OA studies, biomechanical changes of gait variables including vertical GRF, total support moment, joint angles, joint (intersegmental) forces, moments and powers were expected at hip, knee and ankle joints (Chang et al., 2005; Kean et al., 2012; Mills, Hunt, & Ferber, 2013). Hypothesis (7) was examined by comparing absolute asymmetry index between groups.

Study Aim #2: Bilateral versus Unilateral Total Knee Arthroplasty

Literature Background

Both bilateral and unilateral replacement are common options in TKA, although one is for patients with severe bilateral knee arthritis and the other for severe unilateral knee arthritis. There are two standard treatments of bilateral end-stage knee OA: staged bilateral TKA (BTKA) and same-day BTKA. According to a report published in April 2016 by Canadian Institute for Health Information, 86.7%, 10.6%, and 2.7% of all TKA patients received respectively unilateral TKA (UTKA), staged BTKA and same-day BTKA (Bohm et al., 2016). Staged BTKA stages the two surgical procedures, so it requires two hospitalization and two rehabilitation experiences. Same-day BTKA replaces two knees in one hospitalization which is more cost-effective than staged

BTKA. However, same-day BTKA requires careful selection of patients and could increase medical risk (Memtsoudis et al., 2013). What is more, staged BTKA places less burden to the patients during recovery than same-day BTKA, because staged BTKA patients have one “old” knee to stand on during each recovery.

The literature is not clear about the biomechanical outcomes after staged BTKA. It is usually assumed that the second TKA had similar ankle/knee/hip biomechanical outcomes as the first TKA in staged BTKA patients. However, few evidences exist to support this assumption. Only several studies in the literature reported that the two TKAs had similar range of movement and functional outcomes. Gabr et al. 2011 reported that the first and second knee had similar range of knee motion. Kumar et al. 2015 reported that the second knee replacement had similar functional outcomes (range of motion and outcome scores) to the first one in Asians undergoing staged bilateral TKA. Lizaur-Utrilla et al. 2018 reported that functional and patient-reported outcomes were similar between the first and second TKA. No studies have investigated the difference in ankle/knee/hip joint motion and moment between the first and second TKA.

UTKA replaces only the most affected knee while more than 80% of patients with knee arthritis have radiographical knee OA on both sides. Previous studies have also shown that the contralateral knee is prone to have radiographic arthritis (McMahon & Block, 2003). Although the contralateral knee arthritis is not severe, it can impact joint function and knee kinetic symmetry in the frontal plane (Berman, Zarro, Bosacco, & Israelite, 1987; Clare E. Milner & O'Bryan, 2008). Weight-bearing asymmetry during sit-to-stand has been reported for UTKA patients, and extension limitation during standing has been shown to affect this asymmetry (Harato et al., 2010). Lower hip and knee extension moments and lower vertical ground reaction forces during sit-to-stand in operated limbs than in non-operated limbs have also been reported (Alnahdi, Zeni, & Snyder-Mackler, 2016). Total support moment asymmetry and knee extensor moment asymmetry during 10° decline walking have been reported in UTKA patients and these asymmetries are related to quadriceps strength asymmetry (Christensen et al., 2018). In summary, UTKA patients tend to shift load to non-operated limbs although mild or moderated CKOA is present.

For patients after BTKA, since both knees are replaced, they may have better biomechanical symmetry and outcomes than UTKA patients. While there are a few studies have reported that staged BTKA can result in similar or even better outcomes compared to UTKA, no studies have compared biomechanical outcomes between them. Berman et al. 1987 compared spatiotemporal outcomes between UTKA and staged BTKA patients (Berman et al., 1987). They concluded that even asymptomatic arthritis in the contralateral knee can impair gait of UTKA patients and that staged BTKA can yield better gait. Mine et al. 2015 reported that UTKA and staged BTKA patients had similar step length, mean gait velocity and mean single support phase values at post-operative 3 months. However, mean step width was wider in BTKA patients than in UTKA patients. The authors suggested that contralateral TKA may not be necessary for bilateral knee OA patients who improved gait after UTKA. Bohm et al. 2016 reported that cumulative 3-year revision rates was higher in unilateral patients (2.3%) than staged bilateral patients (1.4%).

There are a few studies that compared outcomes between UTKA and same-day BTKA patients. A study that used questionnaires and functional tests concluded that two-year clinical outcomes between same-day BTKA and UTKA were similar (J. A. Zeni, Jr. & Snyder-Mackler, 2010). One-day BTKA patients put more weight on the dominant limb extremity than on the non-dominant one during standing up, while UTKA patients have similar weight ratio between the operated and non-operated extremity. However, these findings may not apply for the comparison between UTKA and staged BTKA because staged BTKA may have different biomechanics from one-day BTKA.

Although no evidence has shown that BTKA and UTKA have different biomechanical outcomes, the biomechanical differences between bilateral and unilateral knee OA have been reported in the literature. Bilateral knee pain is associated with gait symmetry in knee biomechanics while unilateral knee pain is associated with asymmetry (greater adducted angle and lower external flexion moment in the painful knee) (Creaby, Bennell, & Hunt, 2012). They concluded that pain played an important role in altering knee biomechanics. A later study recruited only patients with mild-to-moderate symptomatic knee OA, and found that patients with bilateral knee OA appeared to be more asymmetrical in hip adduction at initial contact and peak knee adduction during stance than patients with unilateral knee OA (Mills, Hettinga, Pohl, & Ferber, 2013). Another study also included only patients with mild-to-moderate symptomatic knee OA. However, they found no biomechanical difference between the most affect side of the bilateral knee OA group and the affected side of the unilateral knee OA group, and both groups had similar absolute biomechanical asymmetry during walking (Messier, Beavers, Herman, Hunter, & DeVita, 2016).

It is not known whether BTKA and UTKA are unique subsets of TKA from a biomechanical prospective. It is reasonable to conceive that BTKA patients would just replicate biomechanics of UTKA operated side on both sides of BTKA, but this assumption has never been examined scientifically. McClelland et al. 2007 reviewed gait analysis studies after TKA and found the included studies were not consistent in excluding or including bilateral TKA patients.

Hypothesis

Therefore, the purpose of this study was to examine the difference in outcomes between unilateral and bilateral total knee replacement. Several hypotheses were made:

(1) BTKA symmetry: the first TKA side in staged BTKA patients have similar ankle/knee/hip joint biomechanics as the second TKA side during tested activities.

(2) UTKA asymmetry: the operated side in UTKA patients have different ankle/knee/hip joint biomechanics as the operated side in UTKA patients during tested activities.

(3) BTKA patients had a lower asymmetry level than UTKA patients.

(4) BTKA patients have similar functional outcomes as UTKA.

(5) BTKA patients have similar ankle/knee/hip joint kinematics and kinetics as the operated side of UTKA patients.

If hypothesis (1) and (2) were supported, then staged BTKA patients have better biomechanical symmetry than UTKA patients. To further see the difference in degree of biomechanical symmetry, hypothesis (3) was also tested. Spatiotemporal variable such as gait speed and stance duration of stair ascent represented the overall mobility or function of a patient. Kinematic variables include ankle plantar-/dorsi-flexion angle, knee flexion/extension angle, knee internal/external rotation, knee varus/valgus angle, hip flexion/extension angle, hip add-/abduction angle, hip internal/external rotation angle were collected. Kinetic variables include ankle dorsiflexion moment, knee and hip moments in 3D (flexion/extension moment, rotation moment and adduction moment) were collected. Symmetry level could be quantified using the absolute symmetry index.

Study Aim #3: Mobile- versus Fixed-Bearing Total Knee Arthroplasty

Literature Background

Total knee arthroplasty (TKA) is an effective treatment in alleviating knee pain and restoring functional mobility for patients with end-stage arthritis. There are two basic options for the polyethylene bearing.

Over time, however, a knee replacement can fail for a variety of reasons such as implant wear and mechanical loosening. While both fixed-bearing (FB) and mobile-bearing (MB) are two basic options for the polyethylene insert (i.e. bearing or spacer) used in TKA, it is believed that MB can reduce implant wear (Callaghan et al., 2001) and achieve improved implant fixation (Henricson, Dalen, & Nilsson, 2006; Russo, Montagna, Bragonzoni, Zampagni, & Marcacci, 2005). As a matter of fact, MB is designed to allow the insert to rotate on top of the tibial tray. Therefore, a mobile insert can rotate relative to both the femoral and tibial components. The fixed insert in FB, however, can rotate relative to only the femoral component. Theoretically, a mobile insert can improve knee axial rotation and reduce rotation torque transmitted to the proximal tibia.

It is still not clear whether MB TKA could achieve better knee rotation when compared to FB TKA. Conflicting findings have been reported in the literature. A few in-vivo biomechanical studies reported similar knee rotation between FB and MB TKA during a variety of activities including deep knee bending, lunging, walking, stepping up and stair climbing(Banks & Hodge, 2004; Okamoto et al., 2014; Papagiannis, Roumpelakis, Triantafyllou, Makris, & Babis, 2016; X. Shi et al., 2014; Wolterbeek, Garling, Mertens, Nelissen, & Valstar, 2012). However, some in-vivo studies reported that MB TKA had a greater axial rotation (Delport, Banks, De Schepper, & Bellemans, 2006; Fantozzi et al., 2003; Ranawat, Komistek, Rodriguez, Dennis, & Anderle, 2004) or a different rotation pattern (K. Shi, Hayashida, Umeda, Yamamoto, & Kawai, 2008) during activities such as deep knee bending and stepping up. What's more, the in-vivo mobility of the insert has been confirmed in MB TKA (LaCour, Sharma, Carr, Komistek, & Dennis, 2014). Thus, more comparison studies are needed. It has been suggested that the advantage of MB in facilitating knee axial rotation may be evident in challenging tasks such as pivoting (Zurcher et al., 2014). An earlier study investigated pivoting but did not investigate peak knee rotation and rotation moment of the stance phase (Zurcher et al., 2014) during which knee implant wears and tears.

A torsional force is applied to the knee joint during functional tasks such as walking and pivoting, which contributes to tray loosening. MB design is designed to facilitate knee rotation and rely on the soft tissue to resist rotational stresses. Therefore, MB TKA should reduce rotation torque transmitted to proximal tibia when compared to FB TKA. To the best of our knowledge, only four studies have compared knee rotation torque between FB and MB. Two in vitro studies using composite or cadaveric specimens have found that MB TKA showed a significantly reduced torque and cortical strain in the proximal tibia in response to combined axial and torsional loading (Bottlang, Erne, Lacatusu, Sommers, & Kessler, 2006; Malinzak et al., 2014). Torque determined in in-vitro studies may not be applicable to in-vivo situations.

Although no studies have compared in-vivo measurement of knee rotation torque between FB and MB TKA, two in-vivo studies have compared intersegment knee rotation moment. Knee rotation moment is a surrogate measure of the torque transmitted to the proximal tibia and also appears to affect the implant longevity (Wimmer, Schwenke, Salineros, & Andriacchi, 2006). However, these two studies found that FB and MB TKA had similar intersegment rotation moment during walking (Papagiannis et al., 2016; Urwin, Kader, Caplan, St Clair Gibson, & Stewart, 2014). Whether the reduction of knee rotation moment would be evident in more demanding daily activities such as stair climbing requires further investigation.

Accordingly, it is of great interest to test if MB TKA had greater knee rotation and lower knee rotation torque than FB TK during daily activities of different mechanical demands than FB TKA. Three types of motor tasks which included level walking, stair climbing, and pivoting were of our interest. For patients after TKA, stair climbing is frequently encountered and good for strengthening lower body. Pivoting is necessary for changing direction which makes up 8% to 50% of daily locomotion (Glaister, Bernatz, Klute, & Orendurff, 2007). These motor tasks also constitute most of the impact the replaced knee suffers during daily life.

Hypotheses

Four null hypothesizes were proposed and tested in this study:

(1) FB and MB TKA had similar knee rotation angle during the stance phase of all tested activities;

(2) FB and MB TKA had similar knee rotation moment during the stance phase of all tested activities.

Chapter 4 Methods

Participants

This study recruited 89 subjects including 48 TKA patients and 20 healthy subjects. The 48 TKA patients included 13 patients after bilateral TKA and 35 patients after unilateral TKA. Informed consent was obtained from each subject before testing. All patients were tested at least 6 weeks after their latest knee replacement. None of the subjects had concurrent symptoms or previous injuries in joints including the hip, ankle and spine. All the tests were performed during the time from 2011 July to 2014 October at gait lab of Shanghai Ruijin Hospital. The study was approved by the Clinical Trial Ethics Committee at Shanghai Jiaotong University (#2015-97) and at University of North Carolina at Charlotte (#13-01-06).

Subgroups for Study #1: Concurrent Osteoarthritis of the Contralateral Knee

For study investigating the progression of contralateral knee arthritis, categorizing the patients into moderate and severe contralateral knee osteoarthritis (CKOA) groups were based on both the patient's indication for contralateral TKA and Kellgenren-Lawrence (KL) radiographic scores.

KL radiographic scores are scores that assess arthritic severity by measuring osteophyte presence, joint space narrowing, sclerosis and bony deformity. KL radiographic scores range from 0 to 4. Grade 0 represents no radiographic feature of OA. Grade 1 represents possible joint space narrowing (JSN) and osteophyte. Grade 2 represents definite osteophytes and possible JSN on weight-bearing radiograph. Grade 3 represents multiple osteophytes, definite JSN, bone thickening, and possible bony deformity. Grade 4 represents large osteophytes, significant JSN, severe bone thickening, and definite bony deformity. KL scores were obtained from orthopedic surgeons and based on patients' medical images before knee replacement.

Patients were grouped into severe CKOA group based on these two conditions: (1) patients planned to have contralateral knee replaced due to CKOA; (2) KL radiographical scores should be at least 3. Patients were grouped into moderate CKOA group based on these two conditions: (1) patients did not have evident deformity of the contralateral limb and had no plan to have contralateral knee replaced; (2) KL radiographical scores should be at most 3.

Among the 35 unilateral TKA patients, 13 of them met the inclusion criteria and were grouped into severe CKOA. The median KL score for the severe CKOA group is grade 4. Among the rest 22 unilateral TKA patients, one patient had contralateral TKA planned due to rheumatoid arthritis, and 8 patients had either evident pain, deformity or a KL grade score of 4. In total, 13 patients who had a KL score of the contralateral knee ranging from 2 to 3 were grouped into moderate CKOA group.

The detailed subject information for study #1 was shown in Table 4.1. The post-surgery time for the severe CKOA group was not matched with the moderate CKOA group. All moderate CKOA patients had a post-surgery time from 1.5 to 18 months, except that one patient had a post-surgery time of 72 months. All severe CKOA patients had a post-surgery time from 4 to 48 months, except that one patient had a post-surgery time of 84 months. After excluding the outliers (72 and 84) and examining the equality of variance using F test, one-tailed t-test showed that the sever CKOA group tended to have longer post-surgery time at the time of testing than the moderate CKOA group (P=0.055). A longer post-surgery time could allow the patient to recover better from the surgery. Since the purpose of this study was to examine the progression of CKOA, the severe CKOA group should have a longer post-surgery time than the moderate CKOA group. To match the gender in TKA groups, ten females and 3 males were chosen from the 20 healthy subjects (10 healthy male and 10 healthy female).

Table 4.1 Demography of the recruited subjects for CKOA study. Mean (SD or range).

Group	Severe CKOA	Control	Moderate CKOA
Age	71.9 ± 5.3	55.9 ± 4.4*	69.8 ± 6.6
Gender, Female/Male	9F/4M	10F/3M	11F/2M
TKA side, Left/Right	8R/5L	NA	4R/9L

mobile-/fixed-bearing	5FB/8MB	NA	4FB/9MB
Mass, kg	64.5 ± 4.9	63.9 ± 9.7	64.3 ± 13.2
Height, m	1.57 ± 0.06	1.59 ± 0.05	1.56 ± 0.09
Body Mass Index, kg/m^2	26.2 ± 1.6	25.1 ± 3.2	26.1 ± 3.6
Post-surgical time, months	23.7 (4-48,84)	NA	14.7 (1.5-18, 72)

*** Significant difference between control and CKOA groups (P <.001).**

Subgroups for Study #2: Bilateral versus Unilateral TKA

Among 13 bilateral TKA patients, 10 of them were truly staged with an interval of average 6 months (3-15 months) between two TKAs and 3 of them had an interval longer than 8 years (8, 11.5 and 18.5 years). These 10 staged bilateral TKA patients were grouped into staged BTKA group for bilateral versus unilateral TKA study.

Half of the 10 BTKA patients had first TKA on the left side and half on the right. To compare the outcomes between the first and second TKA in staged BTKA patients, the limbs that received the first TKA were further grouped in to the first TKA group and the rest limbs were grouped as the second TKA group. The detailed demography of BTKA patients was in Table 4.2.

To compare the outcomes between UTKA and BTKA, the included UTKA patients must satisfy these requirements: (1) post-surgery time matched that of BTKA, i.e. 3 to 36 months; (2) no contralateral knee replacement planned; (3) contralateral knee had no evident pain or deformity. Fifteen UTKA patients matched these criteria and were grouped into UTKA group in study #2.

Thirteen healthy subjects were used as the control group. It should be noted that control subjects were on average 14 years younger than TKA subjects which may affect its validation as an age-matched control group. The statistical companions did not include control group.

Table 4.2 Demography of the recruited subjects for BTKA study. Mean (SD or range).

Group	BTKA	Control	UTKA
Age	69.8 (60-76)	56.2 (50-62) *	70.1 (57-79)
Gender, Female/Male	7F/3M	10F/3M	12F/3M
TKA side, Left/Right	10L/10R	NA	9L/6R
mobile-/fixed-bearing	8MB/12FB	NA	12MB/3FB

Mass, kg	67.2 ± 15.1	63.9 ± 9.5	64.2 ± 11.4
Height, m	1.58 ± 0.08	1.60 ± 0.06	1.54 ± 0.08
Body Mass Index, kg/m^2	26.1 ± 5.6	24.9 ± 3.2	26.8 ± 3.8
Post-surgical time, months	12.9 (3-36)	NA	15.8 (3-27)

Note: * Significant difference between control and TKA groups ($P < .001$).

Subgroups for Study #3: Mobile-Bearing versus Fixed-Bearing Study

TKA Patients were included if they met all the following criteria: had primary unilateral or bilateral total knee replacement to treat end-stage osteoarthritis; underwent a rehabilitation program after surgery and achieved good clinical outcome; at the time of testing, one could perform level walking and stair climbing without aid; without history of musculoskeletal diseases or injuries other than primary knee osteoarthritis; the post-surgery time on the test day was in the range of 6 to 48 months.

Detailed subject information was listed in Table 4.3. Twenty-eight knees of 14 healthy subjects were included as the control group. MB group included 20 MB knees from 16 TKA patients. FB group included 17 FB knees from 11 TKA patients. The subjects in FB, MB and control groups were matched for height, weight, and body mass index. The FB and MB subjects were also matched for age and post-surgery time. Control subjects were 15.8 ± 1.5 years younger than the MB patients ($p < 0.01$), and 12.4 ± 1.4 years younger than the FB patients ($p < 0.01$).

Table 4.3 Demography for Mobile-Bearing versus Fixed-Bearing study, mean ± standard deviation

Parameters	FB group	MB group	Control group
Knees of interest	TKA (n=17)	TKA (n=20)	Asymptomatic (n=28)
Left/Right Knee	8/9	12/8	14/14
Male/Female	3/8	4/12	4/10
Age, years	72.2 ± 5.4	68.8 ± 5.5	56.4 ± 4.3
Weight, kg	63.4 ± 16.9	68.5 ± 13.1	65.1 ± 9.9
Height, m	1.56 ± 0.10	1.58 ± 0.08	1.61 ± 0.06
BMI, kg/m^2	25.8 ± 5.1	27.3 ± 3.5	25.1 ± 3.1
Post-surgery, months	16.1 ± 7.7	13.0 ± 5.5	NA

Movement Tasks

All tests were performed per the approved IRB protocol and consent forms were obtained before testing. A neutral standing posture was recorded to build calibrated anatomical reference frames for later kinematical calculations. The subject was then asked to perform a series of tasks in the following order: static trial (t-pose), level walking, stair ascent, stair descent, step turn and spin turn, sit-to-stand and stand-to-sit. All these tests followed previously reported procedures and methodologies to collect kinematic and kinetic data for level walking (Bo Gao & Zheng, 2010), stair climbing (B. Gao, Cordova, & Zheng, 2012) and pivoting (H. Wang & N. Zheng, 2010). The test scenarios were shown in Figure 4.1 and Figure 4.2.

The static trial was performed with the feet shoulder width apart and toes facing forward in a neutral standing pose. For stair climbing, a custom-built staircase of two steps was used and the step height is about 18 cm. The lower step matched the dimension of a force plate (46.4 cm x 50.8 cm) and was placed on one of the two force plates to measure ground reaction force during stair ascent or descent. The higher step was 70 cm in width and had an attached lower step for transition.

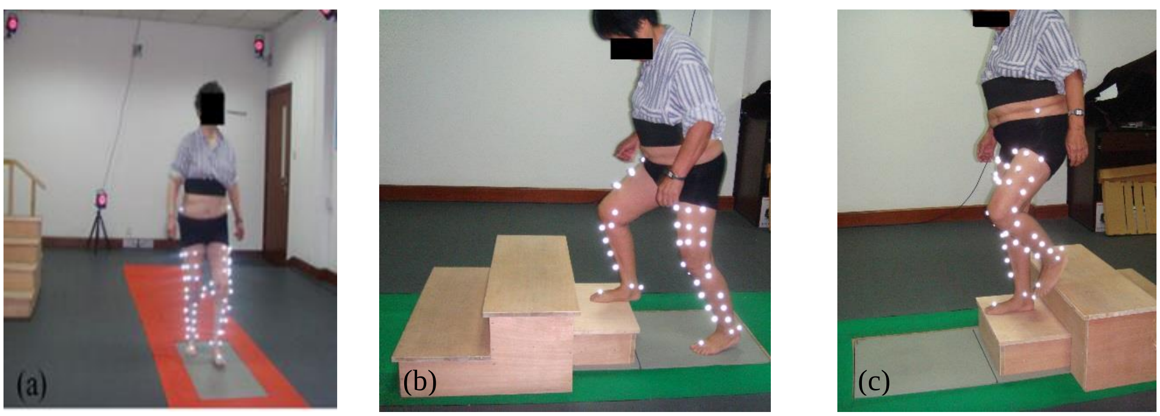


Figure 4.1 the scenario of a subjects during level walking test (a), stair ascending test (b) and stair descending test (c).

For turning tasks, subjects were asked to turn about 90 degrees after walking to a designated force plate. Subjects turned to the side of the supporting leg during spin turn, and to the opposite side during step turn. For sit-to-stand test,

To test both knees during stair climbing and pivoting, subjects alternated their starting leg. All these tasks were performed at the subject’s self-selected speed. Subjects had a rest of at least 2 minutes between different tasks. Five trials were collected for each task to ensure at least three successful trials. The trial was not considered successful when the foot was not completely on the force plate.

Figure 4.2 testing scenarios during step turn (a), spin turn (b) and sit-to-stand test (c)

The flow process for data capturing and analysis using motion capture system in a gait lab is shown in Figure 4.3. After placing markers on the participant, marker trajectory and ground reaction force data can be collected. A customized MATLAB program was written to filter the data and perform kinematic and kinetic calculation.

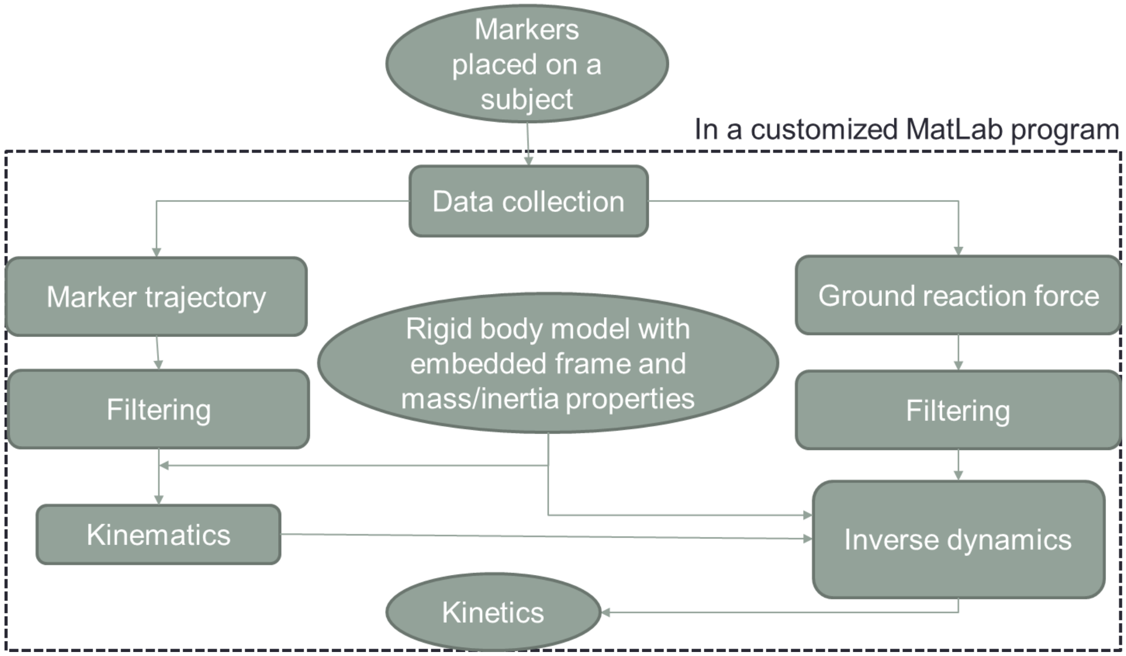


Figure 4.3 Process diagram for kinematic and kinetic analysis

Tracking Motion of Body Segments

Ten infra-red cameras (VICON, Oxford, UK) at 100 Hz were used to track the motion. To track the motion of a segment as rigid body, at least three markers should be placed. In total fifty-three reflective markers (10mm in diameter) were attached to the lower extremities of the subject following the design concept of a previous study (Figure 4.4) (H. Wang & N. Zheng, 2010). To reduce the effect of skin-motion-artefact on knee kinematics, redundant markers were attached to the anterolateral side of the shank and the thigh. Anatomical landmarks are used to define anatomical axes and joint centers. Detailed definition of segment coordinate systems is followed.

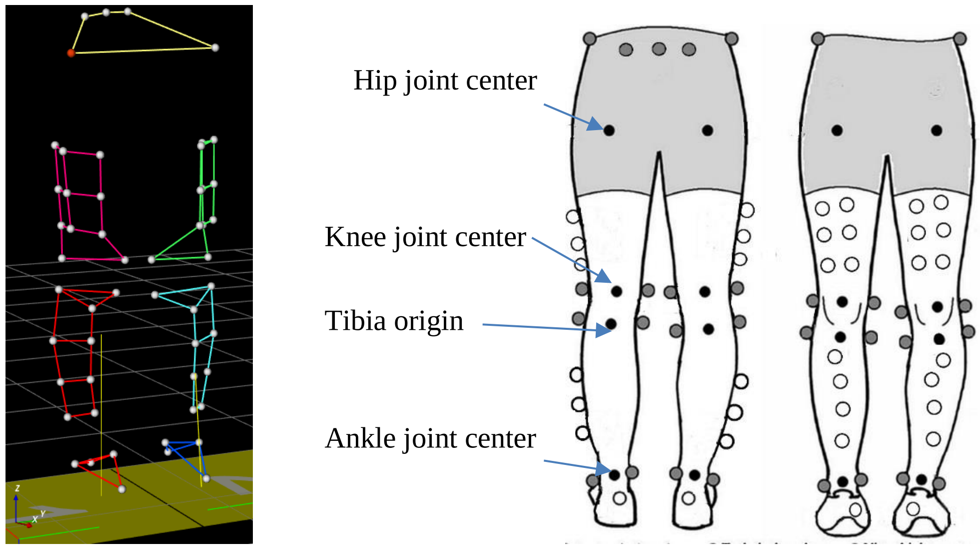


Figure 4.4 Marker set used in this study. Grey solid circle: bony landmark; black solid circle: virtual joint center.

Pelvic Segment

On the pelvic, five markers were placed onto the landmarks including left/right anterior superior iliac spine (ASIS), left/right posterior superior iliac spine (PSIS) and sacrum (Figure 4.5). The segment-embedded coordinate system (CS) was built according to the position of these five markers during static posture. First, the origin and orientation of an embedded reference frame were determined for each segment during the T-pose, following a previously published kinematic model (Bo Gao & Zheng, 2010). Specifically, the origin of the pelvic (the mid-ASIS in Figure 4.5) was set to be the mid-point of left ASIS and right ASIS. The mid-point of right PSIS and left PSIS was determined as mid-PSIS. A temporary vector TX was from mid-PSIS to mid-ASIS. The pelvic Y axis pointed from right ASIS to left ASIS. The Z axis of pelvic segment was the cross product of TX and Y axis of pelvic. Finally, X axis of pelvic segment was the cross product of Y and Z axis of pelvic. This gives the orientation matrix R = [X Y Z] of the pelvic CS during static posture.

Hip joint centers (O_h) on the left and right side were predicted from markers around the pelvic using the empirical method proposed by (Bell, Pedersen, & Brand, 1990). Two more pelvic-embedded CSs were created by translating the origin of pelvic CS to left/right hip joint centers

(Figure 4.5). The pelvic CS at mid-ASIS was used to track the kinematics of pelvic segment relative to lab CS. The pelvic CS at hip joint center was used to calculate hip joint kinematics. The kinetic calculation in this dissertation did not require the mass/inertia information of the pelvic segment.

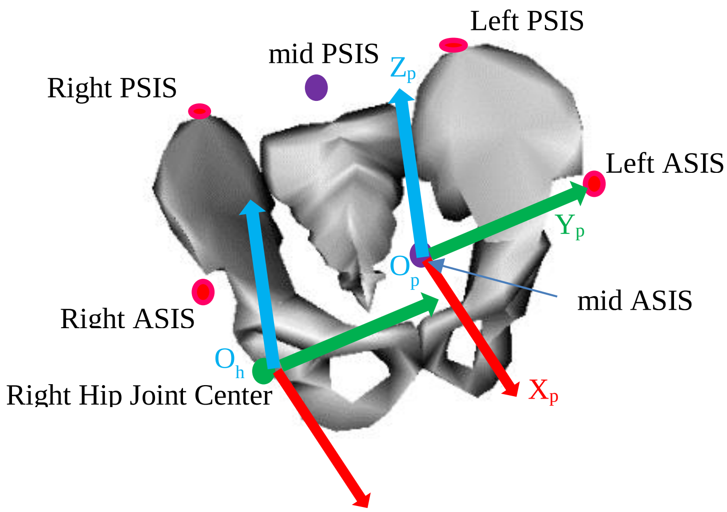


Figure 4.5 the pelvic coordinate system (X_p, Y_p, Z_p), and hip joint center. X_p is anteroposterior direction, Y_p is mediolateral direction and Z_p is vertical direction.

Thigh, Tibia and Foot Segments

The anatomical coordinate systems of thigh, tibia and foot segments were defined in a way similar to pelvic segment (Figure 4.6). At static and neutral posture, the knee joint center (O_f) was defined as the midpoint of lateral and medial femur epicondyle and used as the origin of femoral anatomical coordinate system (X_f, Y_f, Z_f). The femoral coordinate system was translated to hip joint center (O_h) when calculating hip joint kinematics. The definition femoral anatomical system started with defining Z axis which points from the femoral origin (O_f) to the hip joint center (O_h); Y axis (Y_f) parallels to the cross product of Z_f and the vector from the heel marker to the second metatarsal head; X axis (X_f) is the cross product of Y and Z axis (Y_f x Z_f).

The anatomical coordinate system of tibia (O_t) is set to be the midpoint of the medial and lateral ridges of tibia plateau, as is shown in Figure 4.6. To determine Z axis (vertical axis) of tibia, the ankle joint center was defined as the midpoint of lateral and medial malleoli. Z axis of tibia points from ankle joint center to tibia origin. Y axis parallels the cross product of Z axis and the vector from heel to the second metatarsal head. Finally, X axis is the cross product of Y and Z axis. The anatomical coordinate system of foot at neutral posture was set to be coincident with tibia anatomical coordinate system but with its origin at the ankle joint center.

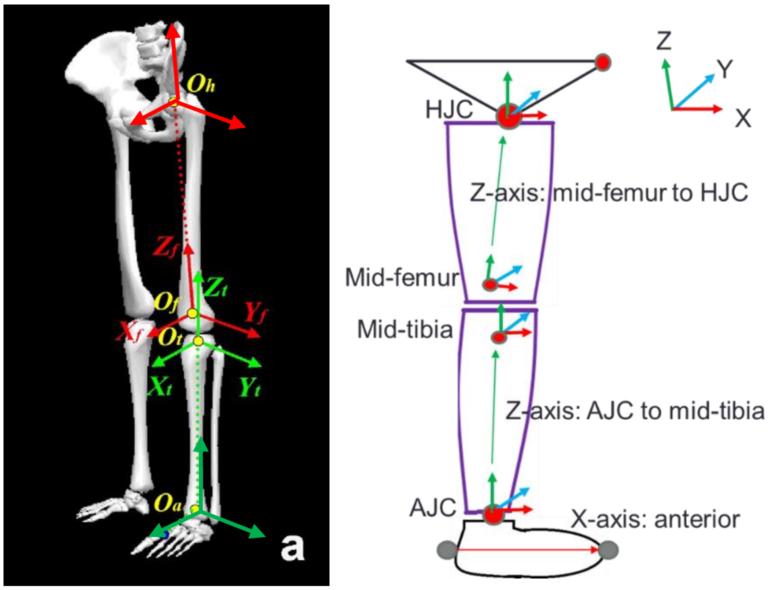


Figure 4.6 (a) Defining anatomical coordinate systems of the femur and tibia (B. Gao et al., 2012).

Segment Kinematics by Modified Least Square Method

Segment kinematics describes the motion of a body segment without considering the forces that causes its motion. The determination of joint kinematics accurately requires the kinematics of the articulating bones instead of the interconnected segments. In biomechanics, a body segment is treated as a rigid body which can represent the motion of its underlying bone.

After constructing segment-embedded reference frame at static posture, the posture of a segment needs to be determined at a dynamic instant. Given a certain rotation matrix R^j and translation vector v^j at a dynamic frame-j, the global position p_i of a marker on a segment is determined as in Figure 4.7:

$$p_i = R^j * a_i + v^j$$

Where a_i is the local vector of marker-i under segment reference frame.

In our lab, reflective markers are placed on the skin and the global position of a marker is given by the cameras. To avoid high-frequency noise, the kinematic data was low pass filtered at 6 Hz. Besides noise, the markers have relative translation and rotation relative to the underlying bone (Figure 4.7). Because joint kinematics (relative motion of the articulating bones) is desired, the relative motion between marker and bone is one kind of systematic error which is often called soft tissue artefact (STA). So, the real position of marker-i will be:

$$p_i = R^j * a_i + v^j + STA + noise$$

Because STA and noise are sources of error, segment kinematics cannot be determined without optimization the cost function which is a s summation of errors across markers at this frame-j:

$$f(\mathrm{v}, \mathrm{R}) = \frac{1}{n} \sum_{i=1}^{n} (Ra_i + v - p_i)^T (Ra_i + v - p_i)$$

Where n is the number of markers on the studied segment.

The orientation R and position v of each segment as a rigid body was computed in this dissertation by a modified least-mean-square algorithm (Spoor & Veldpaus, 1980; H. Wang & N. N. Zheng, 2010).

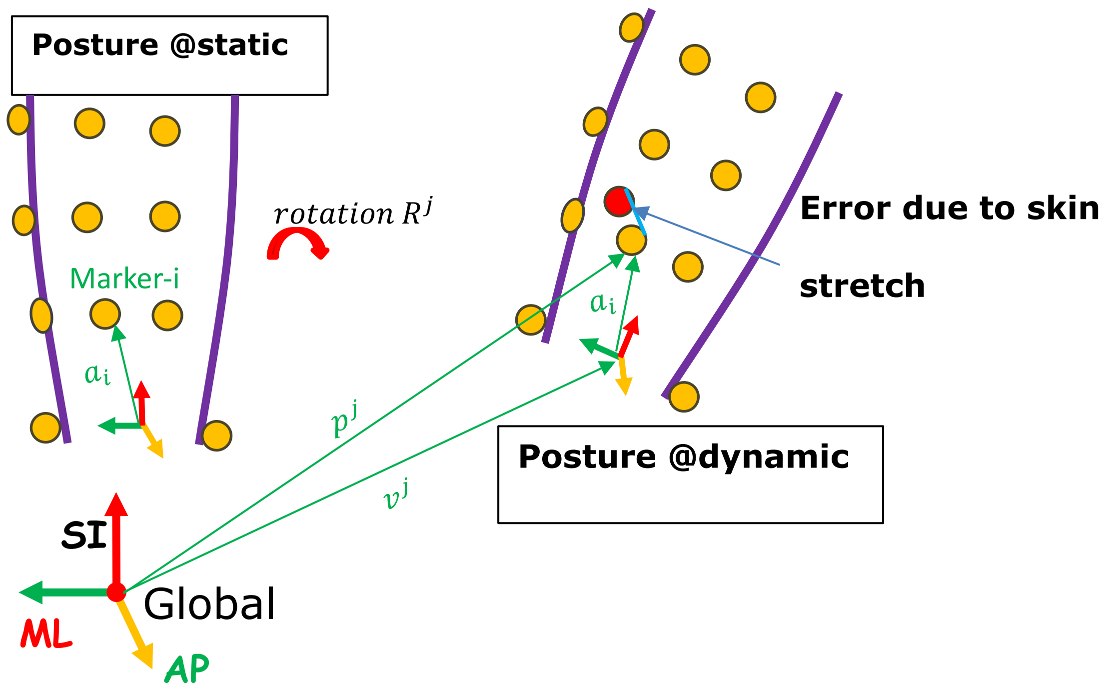


Figure 4.7 Rotation matrix R and translation vector v for defining segment kinematics. Soft tissue artefact (STA) for one of the markers due to skin stretch is also shown.

Joint Kinematics

After segment orientation was determined, three dimensional joint angles could be parameterized with several different methods: direction cosine matrix, Euler angle, quaternion, joint coordinate system method and projection method. Direction cosine matrix captures the transformation from one segment to the other, but it has many parameters and is not easy for clinical professions to understand. Euler angle is dependent upon the rotation sequence and suffers from Gimbal lock. Quaternion has four parameters which are not very useful in clinical applications. The joint coordinate system (JCS) is defined by two bone-fixed axes and one floating axis. Taking knee joint as an example, the two bone-fixed axes are long axis of tibia and femur bones. The floating axis is perpendicular to the two fixed axes. The joint motion, including three rotational and three translational components, is calculated relative to JCS axes and the translational reference point. The drawback of JCS method is that JCS is not necessarily an orthogonal coordinate system.

We adopted the projection method for calculating 3D ankle, knee and hip joint translation and rotation (Wang, Fleischli, & Zheng, 2013; H. Wang & N. Zheng, 2010). The projection method is like JCS method but is easier to understand and apply. Take $R_f = [X_f \quad Y_f \quad Z_f]$ as the orientation matrix of femur-embedded coordinate system and $R_t = [X_t \quad Y_t \quad Z_t]$ as the orientation matrix of tibia-embedded coordinate system. Both R_f and R_t were orientation matrix under lab global frame. The joint transformation matrix R_j described the orientation of femur and took the tibia as the fixed end (reference frame, identity matrix).

$$R_j = {R_t}^T \times R_f = [X_j \quad Y_j \quad Z_j]$$

Projecting X_f (anteroposterior axis of femur) onto the XZ plane of tibia (sagittal plane of tibia) defined the projected vectorX_{f-xz}. Since all vectors are now expressed in tibia reference frame, the y coordinate of X_{f-xz} is zero under tibia reference frame and the x and z coordinates of X_{f-xz} are the same as those ofX_j. That is:

$$X_{f-xz} = \begin{bmatrix} x_{X_j} & 0 & z_{X_j} \end{bmatrix}$$

The angle between X_{f-xz} and X_t is then defined as knee flexion angle (α in Figure 4.8). The equation to calculate the flexion angle is:

$$\alpha = \operatorname{arctan2}(z_{X_j},\ x_{X_j})$$

Positive α indicates knee flexion and negative α indicates knee extension. Similarly, Y_f is projected onto YZ plane of tibia. The angle between the projected vector Y_{f-yz} and Y_t is knee varus/valgus angle β. The x coordinate of Y_{f-yz} under tibia coordinate system is 0, and the y and z coordinates of Y_{f-yz} are the same asY_j.

$$X_{f-xz} = \begin{bmatrix} 0 & y_{Y_j} & z_{Y_j} \end{bmatrix}$$

$$\beta = \arctan2(z_{Y_j},\ y_{Y_j})$$

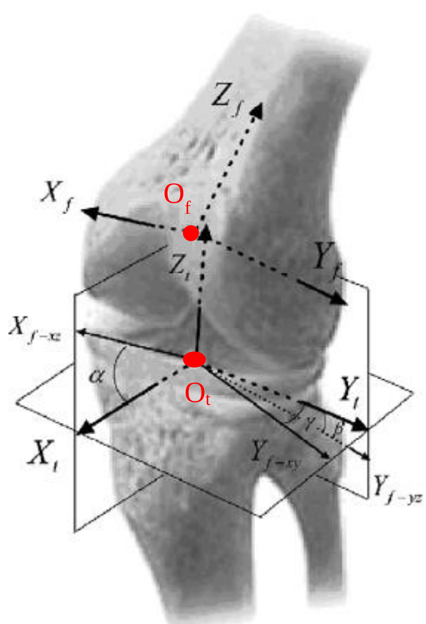
Where positive β means knee varus angle for left side and knee valgus angle for right side. Varus angle was defined to be positive in this study. So, the sign of β is flipped in code after β for right side is calculated.

For knee rotation angle γ,

$$Y_{f-xy} = \begin{bmatrix} x_{Y_j} & y_{Y_j} & 0 \end{bmatrix}$$

$$\gamma = \arctan2(x_{Y_j},\ y_{Y_j})$$

Where positive γ indicates femoral internal rotation (tibial external rotation) for left side and femoral external rotation (tibial internal rotation) for right side. Tibial internal rotation is defined as positive in this study. So, the sign of γ is flipped after γ is calculated for left side.



α is knee flexion angle

β is knee valgus angle

γ is femoral internal rotation angle.

Projecting X_f axis onto XZ plane of tibia makes X_{f-xz}.

Projecting Y_f axis onto XY plane of

Figure 4.8 Diagram showing the projection method for calculating 3-D knee joint angles of the left side.

This methodology could be applied to ankle and hip joint. For knee joint, the proximal segment is thigh (femur) and the distal segment is shank (tibia). For ankle joint, the proximal segment is replaced with shank (tibia) and the distal segment is replaced with foot. For hip joint, the proximal segment is replaced with pelvic and the distal segment is replaced with thigh (femur).

Since ankle and hip joint have very limited translation, only 3D knee joint translations were calculated. The translation vector from O_f to O_t (Figure 4.8) was tracked in tibia reference frame. The 3D knee translations were obtained by projecting the translation vector into tibia reference frame. The x, y and z components were respectively AP, ML and SI translation of femur relative to tibia.

Joint Kinetics

Joint Forces and Moments

Inverse dynamics in biomechanics field calculates force and torque based on kinematics of a body segment and the body segment's inertia properties (mass, center of mass, and moment of inertia). Subject's height and previously reported anthropometry relationships were used to construct segment model which includes segment size and location of center of mass (Winter, 1991). The mass and body inertial parameters of each segment were determined based on body weight of the subject (De Leva, 1996).

External loadings to body segments are ground reaction force (GRF) and moments (GRM) measured by force plates (OR-6, AMTI) at 1000 Hz. Only one foot stepped on a force plate at a time, and the reaction force between foot and force plate was measured by four force sensors at four corners of the force plate. The total forces (Fx, Fy, Fz) and moments (Mx, My, Mz) are output by the force plate and expressed at center of force plate (a, b, c) under global reference frame. Center of pressure (CoPx, CoPy, 0) was on top surface of the force plate whose surface was often set to be zero vertical position of a gait lab. CoPx and CoPy were computed using following equations:

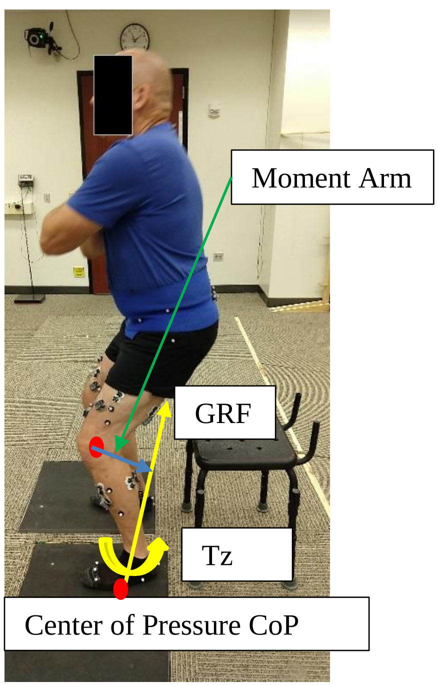
$$CoP_x = \frac{M_y + c \times F_x}{Fz} + x$$

$$CoP_y = \frac{M_x - c \times F_y}{F_z} + y$$

At center of pressure, the only moment left was about the vertical axis T_z:

$$T_z = M_z - (CoP_x - a) \times F_y + (CoP_y - b) \times F_x$$

Just as in kinematic model, the body segments are interconnected via joints from foot to pelvic. The inverse dynamics calculation is a bottom-up process which starts from the foot and ends at the thigh segment. Joint force and torque to the lower extremity joints are largely contributed by ground reaction force applied at the foot (Figure 4.9 a) and a small proportion contributed by the inertia force of body segments.



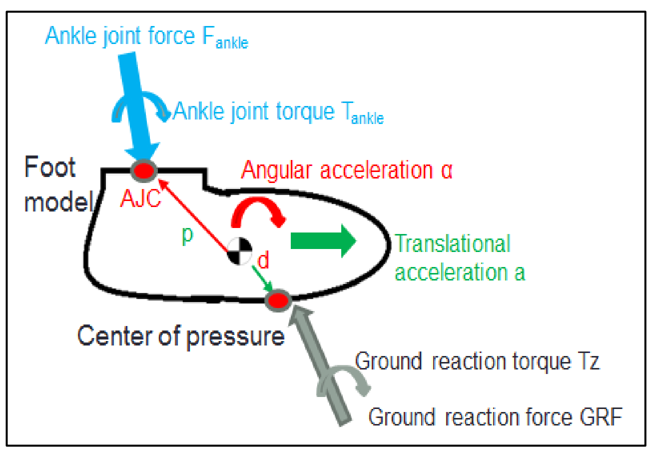


Figure 4.9 External knee flexion moment is contributed by external GRF (a), and inverse dynamic calculation of ankle joint forces and moments by free body diagram (b).

Ankle kinetics was calculated first. To calculate joint forces and moments at the ankle joint, a free body analysis was done on the foot. The inertia forces and moments were determined by these two simple equations:

$$F_{inertia} = m \times (a + g)$$

$$M_{inertia} = I \times \alpha$$

Where translational acceleration a and angular acceleration α were obtained from the motion of the foot. Foot mass and inertia information were based on anthropometry model. After taking ground reaction force GRF and free torque T as external loadings at CoP, external three-dimensional ankle joint force and moment at the proximal end of foot were derived based on Newton's equation of motion:

$$F_{ankle} + GRF - F_{inertia} = 0$$

$$M_{ankle} + T_z - M_{inertia} + d \times GRF + p \times F_{ankle} = 0$$

Where d is distal vector pointing from center of mass to distal point (center of pressure in this case), and p is proximal vector pointing from center of mass to proximal point (ankle joint center AJC in this case, Figure 4.9 b). These equations were solved in 3D. Similar procedures were applied on the tibia segment to solve for knee joint moment and forces, and on the thigh segment to solve for hip joint moment and forces.

Inverse dynamics derives the net forces and moments can be expressed from two different views. Externally, the net joint forces and moment represents external loading at joint center. Internally, net joint moment is created by muscles. The function of internal joint moment is to actuate joint motion or counter balance external joint moment. Muscle force is generated by the excitation of muscle fiber inside the muscle belly. Muscle strength, activation level, muscle fiber length and velocity, physiological cross section area of the belly, and muscle pentation angle all affect the generated muscle force. A non-zero moment arm of muscle at a joint is also important to produce internal joint moment.

Support Moment and Joint Power

David Winter 1980 showed that the summation of internal ankle, knee and hip extension moment was less variable than any of the three joint moments. He defined this moment as total support moment. The support moment describes the moment to prevent the collapse of the lower limb segments. The equation to calculate this moment was:

$$M_s = M_a + M_k + M_h$$

Where M_s, M_a, M_k, and M_h represents total support moment, internal ankle, knee and hip extension moments, respectively.

Different methods have been used to evaluate the relative contribution of ankle/knee/hip joint extensor moment to the total support moment. (Mandeville, Osternig, & Chou, 2007) calculated the percent contribution by dividing the mean joint value by the mean support moment from foot strike to the first peak vertical GRF of level walking or stair ascent. This may not be applicable to quantify joint contribution during pushing off, because hip suffers external extension moment (i.e. no internal hip extension moment). (Joseph A. Zeni & Higginson, 2011) took the joint moment at the time of peak support moment as its percentage to represent joint contribution. This study used the later method.

Joint power (muscle power) is defined as the scalar product of joint moment M and joint angular velocity ω on the sagittal plane (primary motion plane):

$$P = M \times \omega$$

Joint powers on other planes are typically not studies in biomechanics field, because they are heavily affected by the inaccuracies in angular measurements.

Gait Events Detection and Phases during Task

Level Walking

Both stance and swing periods of level walking, stair ascent, stair descent, step turn, and spin turn were of our interest. Gait events such as foot-strike (initial contact with ground or step) and toe-off events helped to segment the recorded motion into phases.

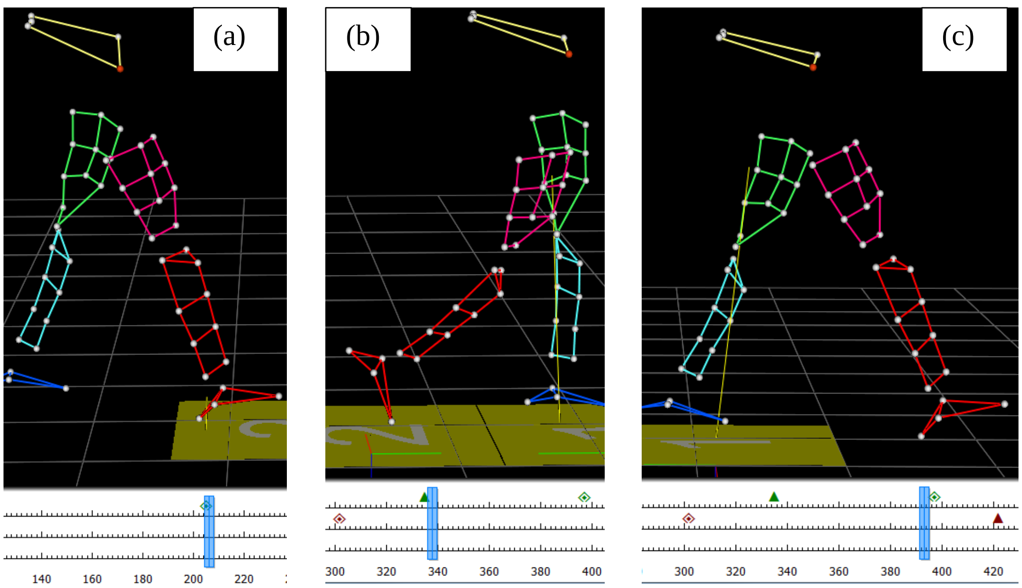


Figure 4.10 gait events in a gait cycle of right leg (red color). One gait cycle is from the first heel strike (a) to the second heel strike (c). Stance phase is from (a) to (b), and swing phase of right leg is from (b) to (c).

For level walking, one gait cycle is from a first heel strike to a second heel strike (Figure 4.10). A threshold of 10 Newton force was used to detect the first heel strike and toe off. Because the second heel strike was not on the force plate (Figure 4.10), its detection was based on the kinematics of heel marker. At heel strike, heel motion came to a stop and heel velocity was very small. These three gait events were detected for both sides and together decided different phases of level walking. The stance phases consist of two double support (DS) phases and one single support phase (Figure 4.11). Typically, stance phase makes up about 60% of a gait cycle.

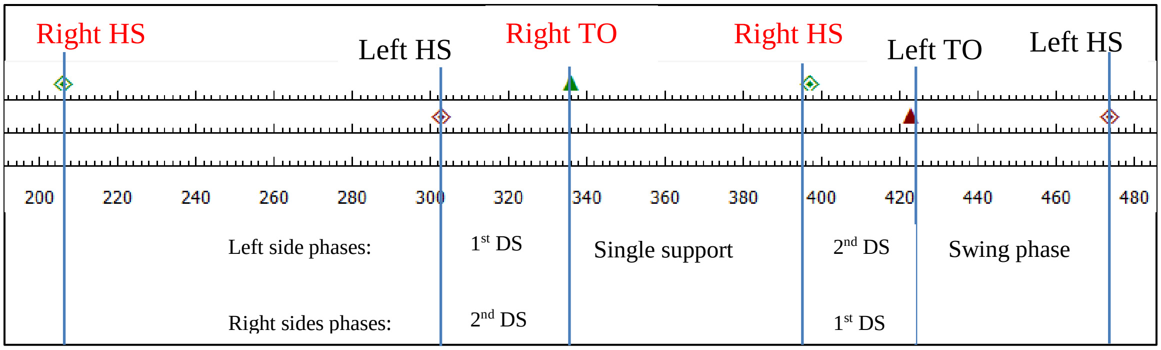


Figure 4.11 different phases in a gait cycle during level walking. Abbreviations: HS, heel strike; TO, toe-off; DS, double-limb support.

Stair Ascent and Descent

For stair ascent and descent, foot strike (FS) event and toe off (TO) event were also determined by a threshold of 10 Newton force. FS was used instead of heel strike (HS) as patients used different portions of foot to contact the steps. For stair ascent, the determined gait events for both left and right sides were shown in Figure 4.12. A complete cycle that combined both stance and swing phases were obtained by combing two motion trials that started with different legs.

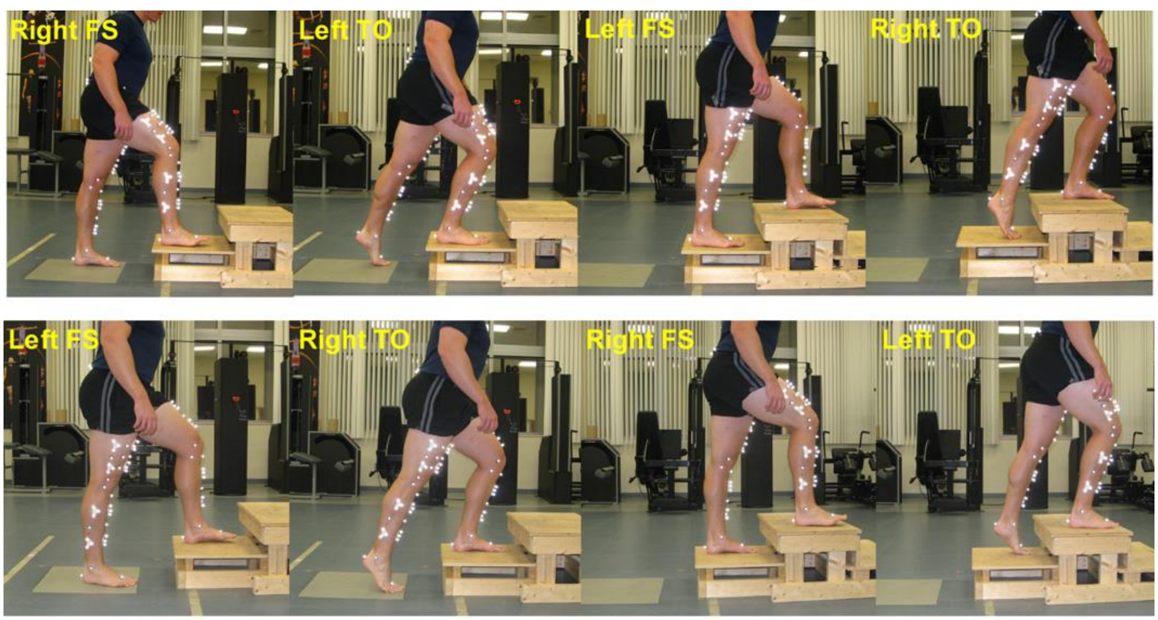


Figure 4.12 Stair ascent and key gait events (B. Gao et al., 2012). The top row of pictures shows a trial starting with the right leg and contains right stance phase and left swing phase; the bottom row of pictures shows a motion trial starting with the left leg and contains left stance phase and right swing phase. A complete gait cycle for each leg could be obtained by combining the two motion trials. Taking right side as an example, weight

acceptance and pull-up phase is from right FS to left TO. Forward continuance phase is from left TO until left HS. Push-up phase is from left HS to right TO. Swing phase is from right TO until right FS (bottom row). Abbreviations: FS: foot strike; TO: toe off.

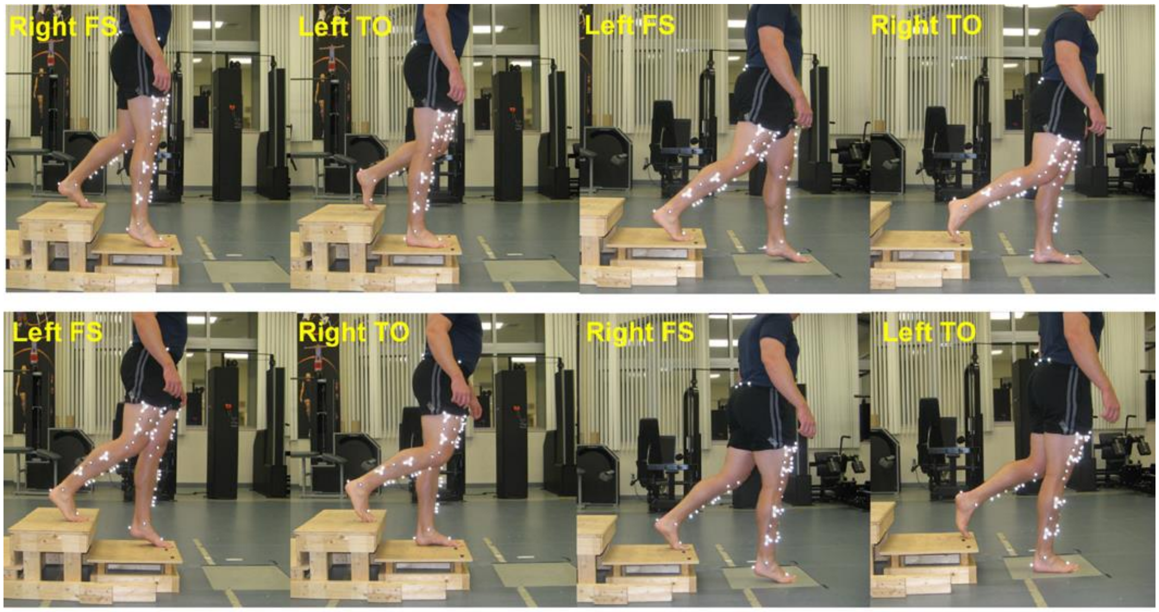


Figure 4.13 Stair descent and key gait events (B. Gao et al., 2012). The top row of pictures shows a motion trial that starts with the right leg and contains the whole stance phase of right leg and the swing phase of left leg; the bottom row of pictures shows a motion trial that starts with the left leg and contains left stance phase and right swing phase. A complete gait cycle can be obtained by combing stance and swing phases. Right Fs to left TO is weight acceptance phase on the leading leg. Left TO until left FS is forward continuance phase. Left FS to right TO is controlled lowering phase. Abbreviations: FS: foot strike; TO: toe off.

The stance phase of stair ascent consists of weight acceptance phase, pull-up phase, forward continuance phase and push-up phase (Aldridge Whitehead, Russell Esposito, & Wilken, 2016). For the leading leg, first double support phase is from FS to contralateral TO. During this phase, the leading leg accepts weight and pushes up the body while the trialing leg pushes up the body. After contralateral TO, the leading leg is in single support phase which is also called forward continuance. After contralateral FS, the leading leg enters second double support and begins to push up the body while contralateral leg accepts weight and pulls up the body.

For stair descent, the stance phase can be divided into several sub-phases: weight acceptance, forward continuance phase, and controlled lowering phases. For the leading leg, first double support phase is from FS to contralateral TO. During this phase, the leading leg accepts weight while the trialing leg controls the lowering of body mass. After contralateral TO, the leading leg is in single support phase which is also called forward continuance. After contralateral FS, the leading leg enters second double support phase and begins to control the lowering of body weight while contralateral leg accepts body weight.

Pivoting, Sit-to-Stand and Stand-to-Sit

The start of standing and sitting was defined as the initiation of hip flexion velocity. The end of standing and sitting was defined as when hip flexion velocity reached zero. Heel strike and toe-off of pivoting tasks were determined by a threshold of 10 Newton. Note that only stance phase of pivoting was of our interest.

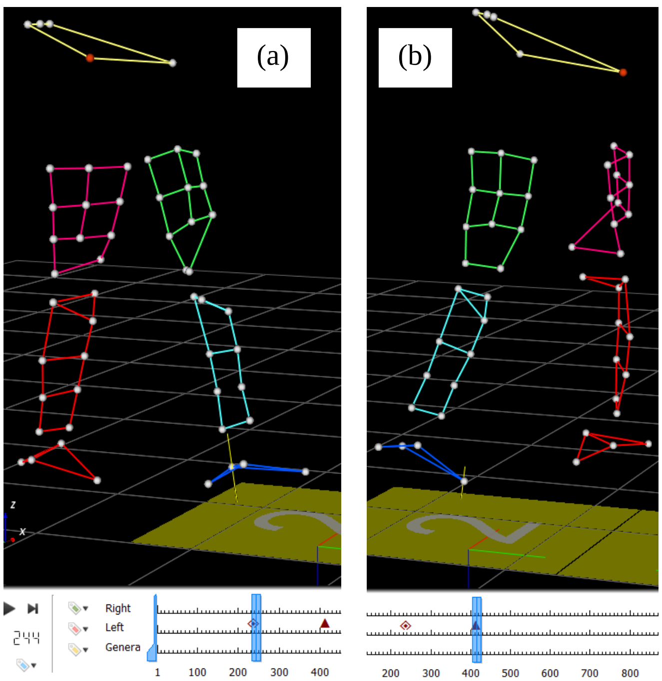


Figure 4.14 Heel strike (a) and toe-off (b) of left leg during spin turn.

Absolute Symmetry Index (ASI)

The following equation was used to assess the asymmetry level of TKA patients.

$$ASI = \frac{2 \times |x_L - x_R|}{|x_L| + |x_R|}$$

Where x_L and x_R are the variables of the left and right legs, respectively. This asymmetry index has been used in other biomechanical studies.

Data Normalization

All kinematic and kinetic data were normalized from HS to the next HS (0 to 100% gait cycle). Body mass and height affect the comparison of joint moment between subjects. Heavy people tend to have a higher joint moment because the ground reaction force especially the vertical component is proportional to body weight. Tall subject also tends to have a high joint moment due to an increase in moment arm.

There are several different normalization factors used extensively in the literature to normalize joint moment (muscle torque): the product of BW and H, the product of BW and leg length, and just body mass. Joint moment was usually expressed in the unit of Newton meter. This study used the first factor and the joint moment was in the unit of BW x H after normalization. This value is small in the unit of BW x H, so unit of %BW x H is often used instead. Joint power calculation used normalized joint moment. Joint force was normalized by BW.

A custom-developed MATLAB (Math Works Inc., Natick, MA, USA) program was used to perform all the data processing and normalization.

Statistical Analysis

Stride parameters including single/double limb support phases, stance phase, stride length and stride speed were also calculated for level walking. Stance, swing and stride duration of stair ascent/descent were collected and average for three trials. Sit-to-stand time and stand-to-sit time were also recorded and averaged for five trials. Sixteen function variables were obtained.

Kinematics and kinetics of ankle, knee and hip joints, and spatiotemporal parameters were calculated for both legs of every subject. The average of each subject's three digitized trials for each gait measure was used for analysis. These measures were represented as waveforms of 101 data points that changed throughout each task. A total of 256 discrete parameters were extracted from these gait measure waveforms. These parameters were chosen after visually checking the ensemble average waveform of each group, and included peak, valley, and range values. For example, peak ankle plantarflexion during early stance and valley ankle dorsiflexion during swing were extracted from ankle flexion waveform.

CKOA Progression

Variables were compared separately for both contralateral and ipsilateral legs to see changes on the operated side and non-operated side, respectively. Independent t-test was used to determine differences in these parameters between mCKOA and sCKOA groups (SPSS v18, Chicago, USA) for both sides. Healthy controls were 14 years younger and not included in comparison. But the mean and SD values of both sides were calculated and presented in results section.

UTKA versus BTKA

Spatiotemporal parameters, range of motion and joint biomechanical variables were collected and grouped. The following comparisons were made to examine the proposed hypotheses.

(1) Comparing biomechanical parameters between 1st TKA side and 2nd TKA side in BTKA patients, using both paired t-test and independent t-test;
(2) Comparing biomechanical parameters among UTKA-OP, UTKA-NOP, BTKA-OP groups using one-way ANOVA with post-hoc Bonferroni correction method
(3) Comparing absolute asymmetry index (ASI) among UTKA patients and BTKA patients using independent t-test.

If hypothesis (1) is supported, then first and second TKA sides can be combined as one group to represent the operated side of BTKA patients (BTKA-OP) in comparisons (2). Otherwise, separated groups (1st TKA and 2nd TKA) were used in comparisons (2).

A significance level of .05 was used. No correction was performed for multiple comparisons, which may lead to a higher type-I errors (false positive). To avoid this shortcoming, Cohen's d as effect size (ES) was calculated to determine the statistical power in detecting differences between groups. If ES is greater than 0.8, the effect of the independent factor is large.

MB versus FB TKA

One-way ANOVA with a Bonferroni post-hoc test was utilized to compare all the collected variables as well as patient characteristics among these three groups: FB, MB and control. Level of significance was set at 0.05 (IBM SPSS Statistics 24.0).

Chapter 5 Results

Study Results #1: The Progression of Contralateral Knee Arthritis

Demographic of the three subject groups were summarized in Table 4.1. The two TKA groups and healthy controls were matched for height, body weight and body mass index. While the two TKA groups were matched for age, healthy controls were on average about 15 years younger than the two TKA groups.

Changes of performance

Severe CKOA patients had lower gait speed, took longer to ascend/descend stair and stand up than moderate CKOA patients (Figure 5.1). During level walking, severe CKOA patients increased 2nd double support on the operated side (corresponding to 1st double support of the contralateral side) and decreased single-limb support side of the contralateral side (Table 5.1). Both groups took similar time to sit down.

Table 5.1 Mean ± SD of spatiotemporal parameters

Side	Ipsilateral side			Contralateral side			Both sides
Group	Moderate	Severe	P	Moderate	Severe	P	Control
stance phase, %GC a	61.6 ± 3.4	65.4 ± 3.1	0.007	61.8 ± 2.3	61.8 ± 2.4	NS	60.1 ± 1.6
1st double support, %GC	11.8 ± 2.2	12.7 ± 2.1	NS	11.2 ± 2.1	13.0 ± 2.0	0.031	10.6 ± 1.4
single limb support, %GC	41.1 ± 2.1	41.9 ± 2.8	NS	41.4 ± 3.2	38.3 ± 2.5	0.010	41.6 ± 1.8
2nd double support, %GC	11.2 ± 2.1	12.9 ± 1.7	0.036	11.7 ± 2.1	12.6 ± 1.9	NS	10.6 ± 1.4
stride length, m	1.05 ± 0.06	0.99 ± 0.08	0.043	1.06 ± 0.06	0.98 ± 0.09	0.017	1.14 ± 0.06
stride speed, m/s	0.85 ± 0.15	0.70 ± 0.13	0.010	0.86 ± 0.15	0.70 ± 0.13	0.009	1.02 ± 0.13
stance level walking, sec	0.78 ± 0.13	0.96 ± 0.18	0.010	0.79 ± 0.13	0.90 ± 0.17	0.064	0.68 ± 0.09
stride level walking, sec	1.26 ± 0.18	1.46 ± 0.22	0.022	1.27 ± 0.19	1.46 ± 0.25	0.044	1.14 ± 0.12
stance stair ascent, sec	1.19 ± 0.19	1.57 ± 0.36	0.003	1.21 ± 0.16	1.48 ± 0.37	0.028	1.00 ± 0.13
swing stair ascent, sec	0.82 ± 0.10	0.96 ± 0.18	0.026	0.82 ± 0.10	1.04 ± 0.21	0.003	0.74 ± 0.08
stride stair ascent, sec	2.01 ± 0.23	2.53 ± 0.49	0.002	2.03 ± 0.22	2.52 ± 0.47	0.003	1.74 ± 0.20
stance stair descent, sec	1.18 ± 0.31	1.92 ± 0.60	0.001	1.08 ± 0.18	1.99 ± 0.82	0.002	0.91 ± 0.15
swing stair descent, sec	0.92 ± 0.13	1.41 ± 0.44	0.002	0.91 ± 0.23	1.61 ± 0.52	0.000	0.77 ± 0.10
stride stair descent, sec	2.10 ± 0.41	3.33 ± 0.98	0.001	1.99 ± 0.33	3.60 ± 1.22	0.000	1.68 ± 0.23

stand-to-sit time, sec	1.80 ± 0.44	2.14 ± 0.67	NS	1.76 ± 0.44	2.14 ± 0.69	NS	1.51 ± 0.19
sit-to-stand time, sec	1.57 ± 0.40	2.00 ± 0.48	0.022	1.49 ± 0.34	1.99 ± 0.47	0.006	1.18 ± 0.17

Note: [a] gait cycle.

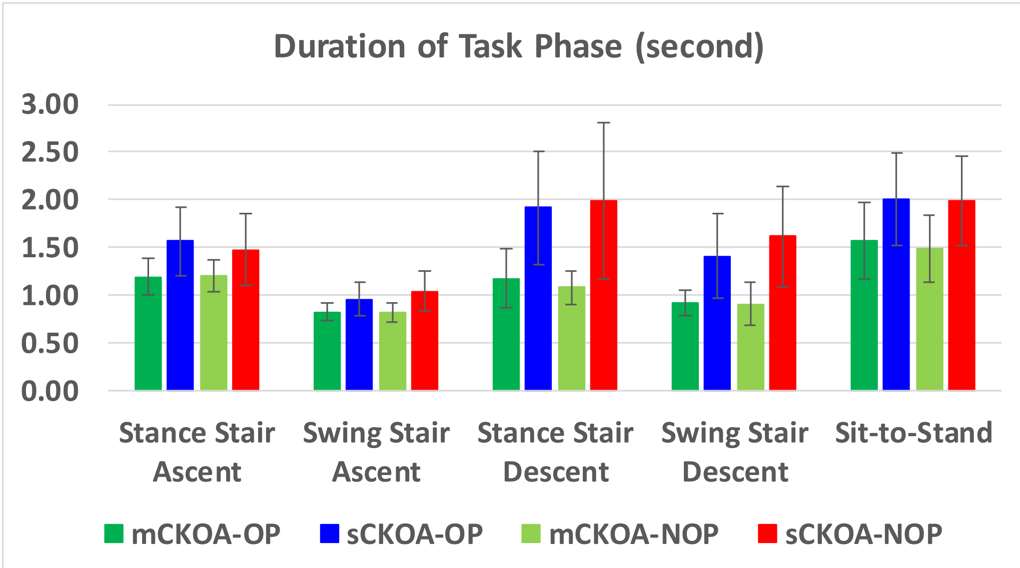


Figure 5.1 Duration of task phase on operated (OP) leg and non-operated (NOP) leg during stair ascent/descent and sit-to-stand. Severe CKOA patients took longer to perform these tasks than moderate CKOA patients.

Changes of the operated knee biomechanics

Table 5.2 shows the change of the operated knee. The progression of CKOA changed axial rotation of the operated knee and only during level walking (Figure 5.2). Operated knee increased external tibia rotation by 3.2° during level walking (P=.039, ES=.86) and range of knee axial rotation by 2.9° (P=.018, ES=1.00).

Table 5.2 Mean ± SD of kinematic and kinetic variables of knees on both sides

Side	Ipsilateral (OP) side			Contralateral (NOP) side			Both sides
CKOA condition	Moderate	Severe	P	Moderate	Severe	P	Control
KFM during stair ascent							
PKFM pull-up phase	-2.20 ± 1.80	-3.85 ± 1.61	0.024	-5.22 ± 1.42	-2.08 ± 1.53	<.001	-5.70 ± 1.52
P2P KFM stance phase	3.56 ± 1.35	5.08 ± 1.93	0.031	6.42 ± 0.96	3.32 ± 1.34	<.001	6.86 ± 1.53
Knee rotation stance LW							
Peak external rotation	-4.2 ± 3.8	-7.4 ± 3.8	0.039	-6.2 ± 5.1	-5.7 ± 3.1	NS	-4.3 ± 4.3
P2P knee rotation	7.8 ± 2.3	10.7 ± 3.4	0.018	10.9 ± 3.1	9.1 ± 3.5	NS	9.7 ± 2.9

Note: only changes of OP side were reported in this table. P2P: peak-to-peak; KFM: knee flexion moment; PKFM: peak knee flexion moment; LW: level walking

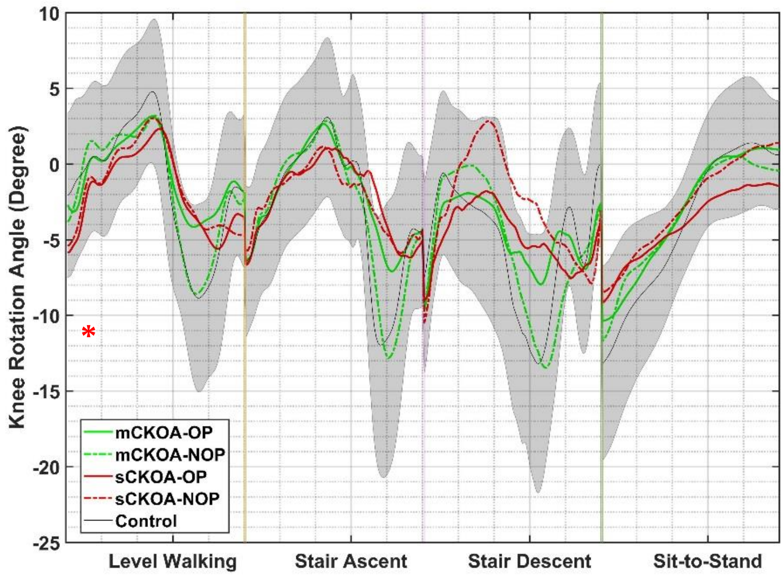


Figure 5.2 ensemble average curves of knee rotation during tested tasks. Tibia internal rotation is positive. Combined stance and swing phase of each task is shown. The shaded are represent mean +/- SD of healthy controls. *mCKOA-OP vs. sCKOA-OP, P<.05. The sCKOA patients had more externally rotated knee at heel strike of level walking than the mCKOA patients on the OP side.

The progression of CKOA changed knee flexion moment of the operated knee and only during stair ascent (Figure 5.3 and Table 5.2). During early stance (pull-up) of stair ascent, operated knee increased peak knee flexion moment (2.20 vs. 3.85 %BW x H, P=.024, ES=.97) while contralateral knee reduced peak knee flexion moment (5.22 vs. 2.08 %BW x H, P<.001, ES=2.14). As a result, operated knee increased peak-to-peak knee flexion moment (3.56 vs. 5.08 %BW x H, P=.031, ES=.92), while contralateral knee decreased peak-to-peak knee flexion moment (6.42 vs. 3.32 %BW x H, P<.001, ES=2.67).

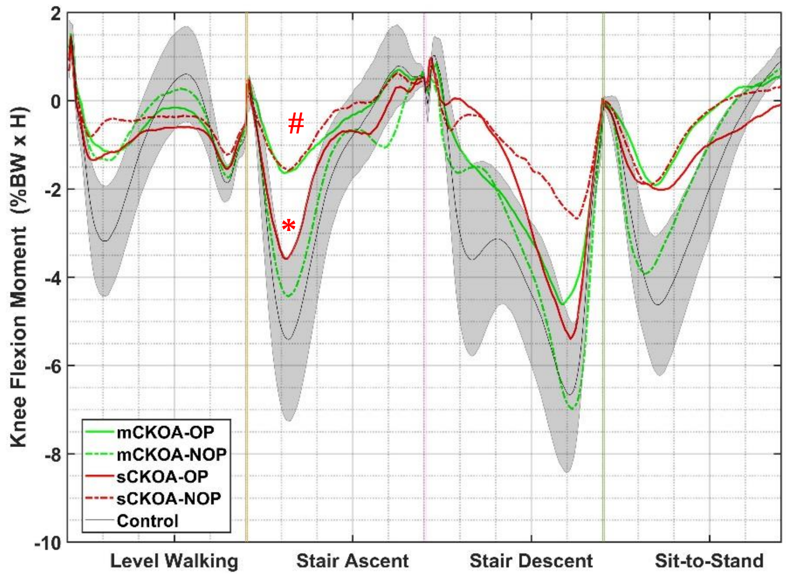


Figure 5.3 ensemble average curves of knee flexion moment during tested tasks. External knee extension moment (internal knee flexor moment) is positive. Only stance phase of each task is shown. The shaded are represent mean +/- SD of healthy controls. *mCKOA-OP vs. sCKOA-OP; #mCKOA-NOP vs. sCKOA-NOP, P<.05. With the progression of CKOA, patients increased peak internal peak knee extensor moment on the OP knee and reduced that on the NOP (contralateral) knee during pull-up phase of stair ascent.

Changes of both hips

Table 5.3 shows the biomechanical changes of both hips. Ipsilateral hip increased hip abduction angle by 6.2°during stair ascent and by 8.2° during stair descent. (Figure 5.4). The range of hip add-abduction during stance of stair descent was also increased by 12° on the ipsilateral side and 9.8° on the contralateral side. During sit-to-stand, only ipsilateral hip reduced external rotation by 7° (Figure 5.5).

Table 5.3 Mean ± SD of kinetic variables of hips on both sides, %BW x H.

Side	Ipsilateral (OP) side			Contralateral (NOP) side			Both sides
CKOA condition	Moderate	Severe	P	Moderate	Severe	P	Control
Hip flexion moment							
peak extension moment late stance level walking	-4.27 ± 1.13	-3.63 ± 0.67	NS	-4.48 ± 1.51	-2.58 ± 1.02	0.001	-5.24 ± 1.14
peak flexion moment early stance stair ascent	4.50 ± 1.85	4.77 ± 1.66	NS	3.27 ± 1.08	5.18 ± 1.45	0.001	3.77 ± 1.47
peak flexion moment early stance stair descent	1.66 ± 1.37	3.31 ± 1.34	0.008	1.93 ± 1.53	3.25 ± 1.37	0.041	0.94 ± 1.01
peak extension moment late stance stair descent	-2.18 ± 1.78	-1.06 ± 1.03	0.075	-4.80 ± 1.70	-1.56 ± 1.59	0.000	-4.52 ± 1.06

P2P moment stair ascent	5.51 ± 1.50	5.73 ± 1.54	NS	4.71 ± 0.87	5.73 ± 1.21	0.024	5.00 ± 1.21
P2P moment stair descent	4.08 ± 1.15	4.63 ± 1.48	NS	6.76 ± 2.20	5.11 ± 1.21	0.035	5.49 ± 1.04
Hip abduction moment							
max stair descent	0.64 ± 0.31	0.52 ± 0.37	NS	0.96 ± 0.45	0.34 ± 0.35	0.001	1.01 ± 0.37
min sit-to-stand	-1.68 ± 0.59	-2.09 ± 0.60	NS	-2.09 ± 0.85	-1.34 ± 0.73	0.026	-1.88 ± 0.67
P2P sit-to-stand	1.96 ± 0.63	2.31 ± 0.60	NS	2.42 ± 0.96	1.71 ± 0.76	0.048	2.05 ± 0.68
Hip rotation moment							
max stair ascent	0.99 ± 0.50	1.40 ± 0.60	0.072	1.68 ± 0.58	0.97 ± 0.44	0.002	1.85 ± 0.51
P2P stair ascent	1.33 ± 0.46	1.69 ± 0.58	NS	2.04 ± 0.54	1.19 ± 0.40	0.000	2.17 ± 0.48
max stair descent	1.09 ± 0.62	1.35 ± 0.46	NS	1.54 ± 0.69	0.77 ± 0.37	0.003	1.24 ± 0.47
P2P stair descent	1.30 ± 0.59	1.59 ± 0.38	NS	1.86 ± 0.62	1.01 ± 0.39	0.001	1.49 ± 0.45
Hip flexion power							
peak absorption power late stance level walking	-218 ± 147	-160 ± 80	NS	-225 ± 146	-107 ± 73	0.016	-288 ± 123
peak generation power late stance level walking	232 ± 68	209 ± 101	NS	277 ± 123	129 ± 58	0.001	375 ± 112
peak generation power early stance stair ascent	370 ± 151	327 ± 158	NS	226 ± 114	356 ± 159	0.027	364 ± 160
peak absorption power late stance stair descent	-102 ± 71	-28 ± 25	0.003	-154 ± 134	-75 ± 114	NS	-158 ± 86
peak generation power late stance stair descent	90 ± 76	47 ± 29	0.088	255 ± 158	102 ± 66	0.006	330 ± 188
peak generation power sit-to-stand	360 ± 157	302 ± 88	NS	403 ± 149	263 ± 71	0.006	360 ± 115

During weight acceptance phase of stair descent, both ipsilateral and contralateral hip increased peak hip flexion moment (Figure 5.6). The contribution of hip moment to peak total support moment was increased on both contralateral side (from 15% to 42%) and ipsilateral side (from 18% to 45%) during weight acceptance of stair descent phase.

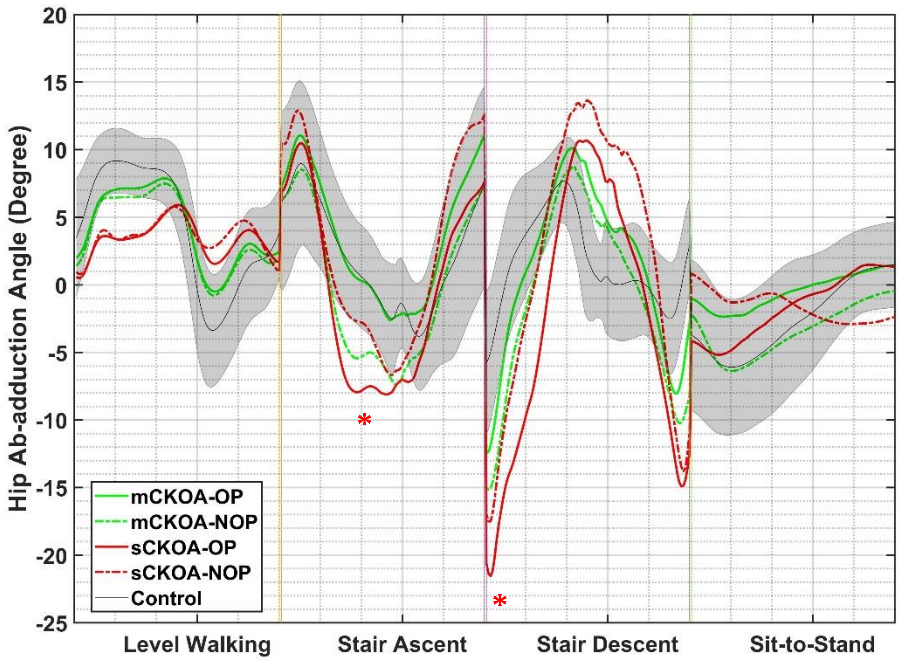


Figure 5.4 ensemble average curves of hip abduction (-) / adduction (+) angle during tested tasks. Combined stance and swing phases are shown. The shaded are represent mean +/- SD of healthy controls. *mCKOA-OP vs. sCKOA-OP, P<.05. With the progression of CKOA, patients increased hip abduction on the OP leg during push-off phase of stair ascent and weight acceptance phase of stair descent.

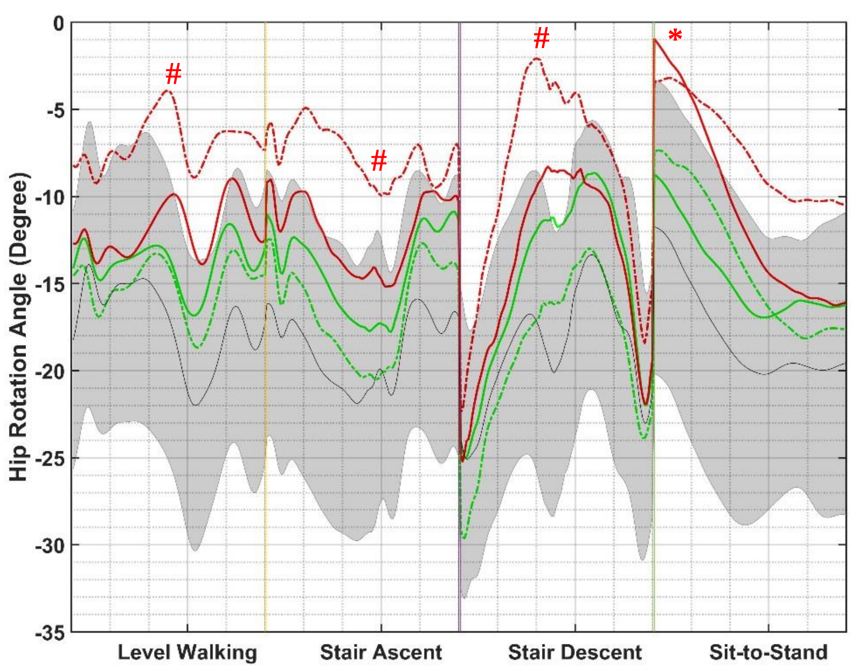


Figure 5.5 ensemble average curves of hip external (-) / internal (+) angle during tested tasks. Combined stance and swing phases (gait cycle) are shown for ambulation tasks. Curve legend is the same as other figures. The shaded are represent mean +/- SD of healthy controls. *mCKOA-OP vs. sCKOA-OP, P<.05. With the progression of CKOA, patients increased internal hip rotation.

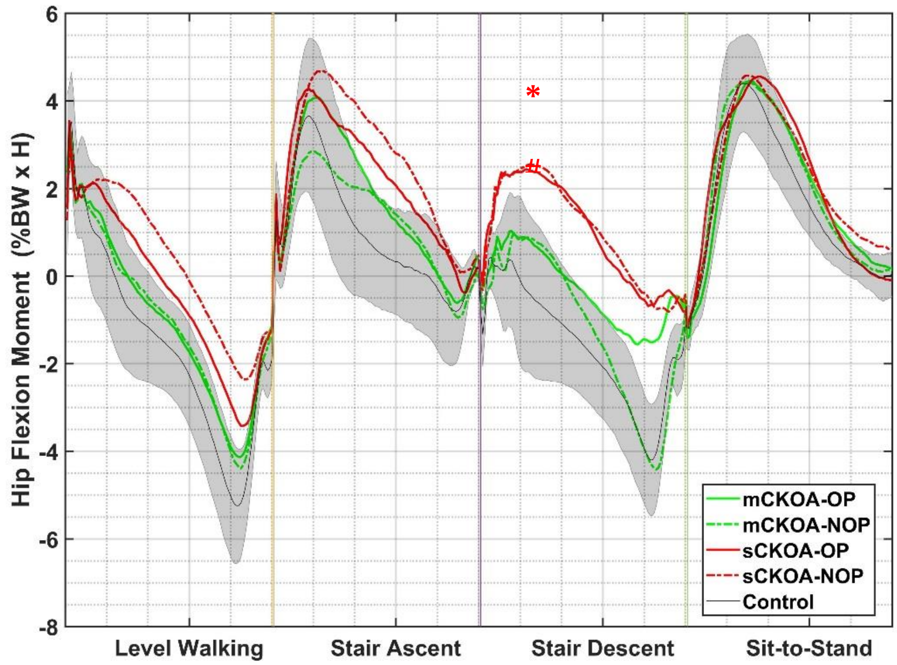


Figure 5.6 ensemble average curves of external hip extension (-) / flexion (+) moment during tested tasks. Only stance phase is shown. The shaded are represent mean +/- SD of healthy controls. *mCKOA-OP vs. sCKOA-OP; #mCKOA-NOP vs. sCKOA-NOP, P<.05. With the progression of CKOA, patients increased internal hip extensor moment on both OP and NOP legs during weight acceptance phase of stair descent.

Some evident changes were only on contralateral hip. Contralateral hip reduced range of flexion-extension during swing of stair ascent and sit-to-stand. Contralateral hip was more internally rotated and less externally rotated during level walking, stair ascent and descent (Figure 5.5).

Contralateral hip reduced peak hip extension moment during late stance of both level walking and stair descent. As a result, peak-to-peak hip flexion moment was decreased during stair descent. On the other hand, contralateral hip increased peak hip flexion moment during weight acceptance phase and peak-to-peak hip flexion moment during stance of stair-ascent (Figure 5.6). Contralateral hip's contribution was increased during both weight acceptance of stair ascent and sit-to-stand, while contralateral knee's contribution was reduced.

Contralateral hip decreased both hip rotation moment and peak-to-peak hip rotation moment during stance phases of both stair ascent and descent. Contralateral hip decreased hip abduction moment during pre-swing phase of stair descent and decreased also peak hip adduction moment and peak-to-peak hip adduction moment during sit-to-stand.

Contralateral hip reduced peak power generation during late stance of both level walking and stair descent, and during sit-to-stand. On the other hand, contralateral hip increased peak power generation during pull-up phase of stair ascent.

Changes of the both ankles

Changes of both ankles were summarized in Table 5.4. Ipsilateral ankle had only one change. Ipsilateral ankle reduced ankle power generation during late stance (controlled lowering phase) of stair descent (Figure 5.9). The contribution of ipsilateral ankle to peak total support moment was reduced (from 22% to 7%, P=.028) during early stance (pull-up) phase of stair ascent.

Table 5.4 Mean ± SD of kinematic and kinetic variables of ankles on both sides

Side	Ipsilateral (OP) side			Contralateral (NOP) side			Both sides
CKOA condition	Moderate	Severe	P	Moderate	Severe	P	Control
Ankle Angle, degree							
Plantarflexion early stance LW	11.1 ± 3.5	11.9 ± 4.1	NS	10.1 ± 3.0	13.3 ± 3.4	0.017	6.3 ± 2.8
Dorsiflexion swing LW	-2.3 ± 4.1	-1.3 ± 4.7	NS	-3.6 ± 2.9	0.2 ± 2.4	0.001	-5.1 ± 3.0
Dorsiflexion early stance stair ascent	0.2 ± 6.1	-2.1 ± 5.3	NS	-7.5 ± 2.8	2.7 ± 5.0	0.000	-10.6 ± 4.6
Dorsiflexion late stance stair ascent	-6.3 ± 2.8	-6.2 ± 2.9	NS	-10.3 ± 3.6	-2.4 ± 2.8	0.000	-9.7 ± 3.3
dorsiflexion swing stair ascent	-9.6 ± 5.6	-7.9 ± 7.1	NS	-12.4 ± 4.7	-5.8 ± 5.3	0.003	-16.8 ± 4.6
dorsiflexion stance stair descent	-20.9 ± 7.1	-22.3 ± 6.0	NS	-27.8 ± 4.8	-15.0 ± 5.8	0.000	-34.8 ± 4.5
dorsiflexion sit-to-stand	-5.6 ± 4.5	-4.9 ± 5.5	NS	-9.9 ± 4.4	0.2 ± 7.9	0.001	-15.6 ± 5.2
rom during stance stair descent	45.6 ± 9.0	51.8 ± 7.3	0.086	54.6 ± 11.1	43.8 ± 8.2	0.014	55.5 ± 5.0
Ankle moment, %BW x H							
dorsiflexion moment late stance level walking	7.10 ± 0.82	6.93 ± 0.44	NS	7.51 ± 1.02	6.67 ± 0.54	0.015	7.44 ± 0.74
dorsiflexion moment late stance stair ascent	6.47 ± 1.01	5.94 ± 0.67	NS	7.25 ± 1.41	5.90 ± 0.62	0.006	7.23 ± 1.05
dorsiflexion moment early stance stair descent	4.12 ± 1.01	4.76 ± 0.98	NS	6.05 ± 2.61	3.99 ± 1.00	0.019	5.12 ± 0.81

dorsiflexion moment late stance stair descent	6.32 ± 0.87	5.96 ± 0.53	NS	6.94 ± 0.82	5.27 ± 0.77	0.000	6.69 ± 0.72
Ankle power, % BW x H x degree/s							
generation power late stance level walking	713 ± 218	614 ± 165	NS	787 ± 272	587 ± 137	0.026	835 ± 192
generation power late stance stair ascent	894 ± 269	847 ± 321	NS	1333 ± 376	753 ± 149	0.000	1083 ± 291
absorption power early stance stair descent	-578 ± 179	-757 ± 327	NS	-1303 ± 682	-544 ± 276	0.002	-926 ± 281
absorption power late stance stair descent	-258 ± 132	-212 ± 55	NS	-340 ± 142	-180 ± 101	0.005	-403 ± 110
generation power late stance stair descent	406 ± 262	224 ± 114	0.040	645 ± 309	192 ± 167	<.001	767 ± 232

More changes were seen on the contralateral ankle. During loading response of level walking, contralateral ankle increased plantarflexion. During loading response phase of stair ascent and swing phase of both level walking and stair ascent, contralateral ankle also reduced peak dorsiflexion (Figure 5.7). During stance of stair descent, contralateral ankle reduced peak dorsiflexion and range of ankle motion.

Contralateral ankle also reduced moment and power (Table 5.4). Contralateral ankle decreased peak dorsiflexion moment during walking and stair ascent/descent. Contralateral ankle decreased ankle power generation during push-off phase of both level walking and stair ascent (Figure 5.9). Contralateral ankle also reduced ankle power absorption during early and late stance of stair descent.

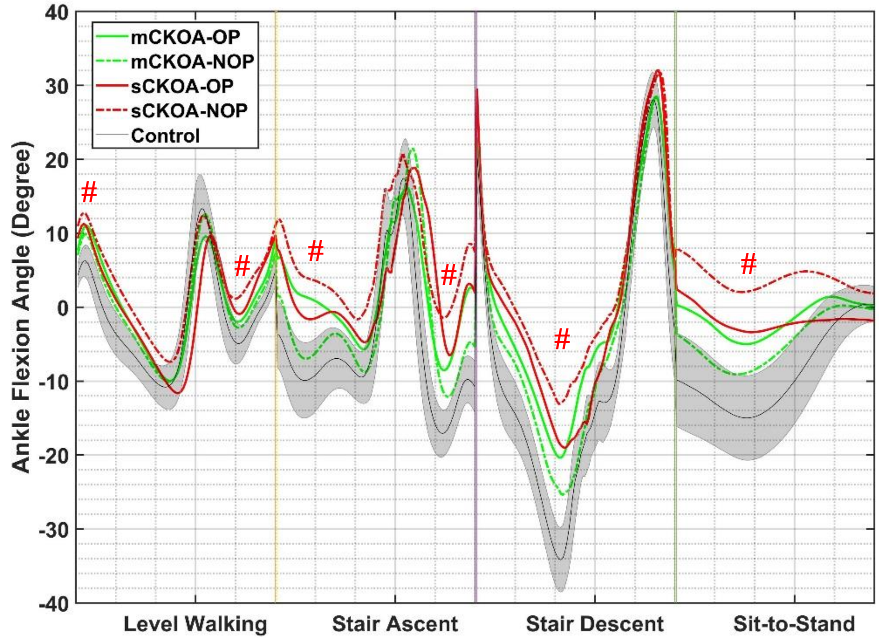


Figure 5.7 ensemble average curves of ankle dorsiflexion (-) / plantar-flexion (+) during tested tasks. Combined stance and swing phases are shown for each task. The shaded are represent mean +/- SD of healthy controls. *mCKOA-OP vs. sCKOA-OP; #mCKOA-NOP vs. sCKOA-NOP, P<.05. With the progression of CKOA, contralateral ankle reduced ankle dorsiflexion or increased ankle plantarflexion during all tested tasks.

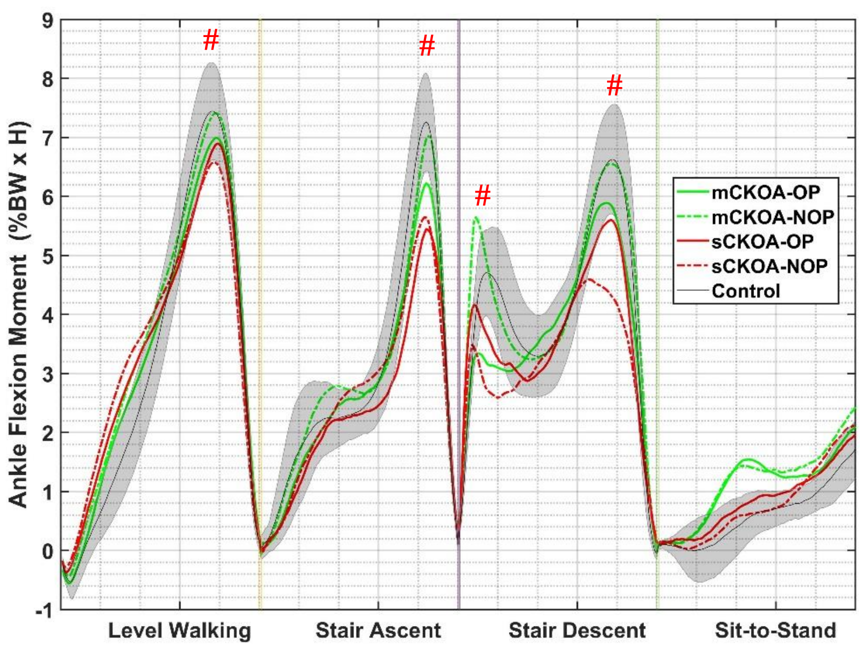


Figure 5.8 ensemble average curves of external ankle dorsiflexion (+) / plantar-flexion (-) moment during tested tasks. Only stance phase is shown for each task. The shaded are represent mean +/- SD of healthy controls. *mCKOA-OP vs. sCKOA-OP; #mCKOA-NOP vs. sCKOA-NOP, P<.05. With the progression of CKOA, contralateral ankle reduced ankle dorsiflexion moment during all tested tasks except standing.

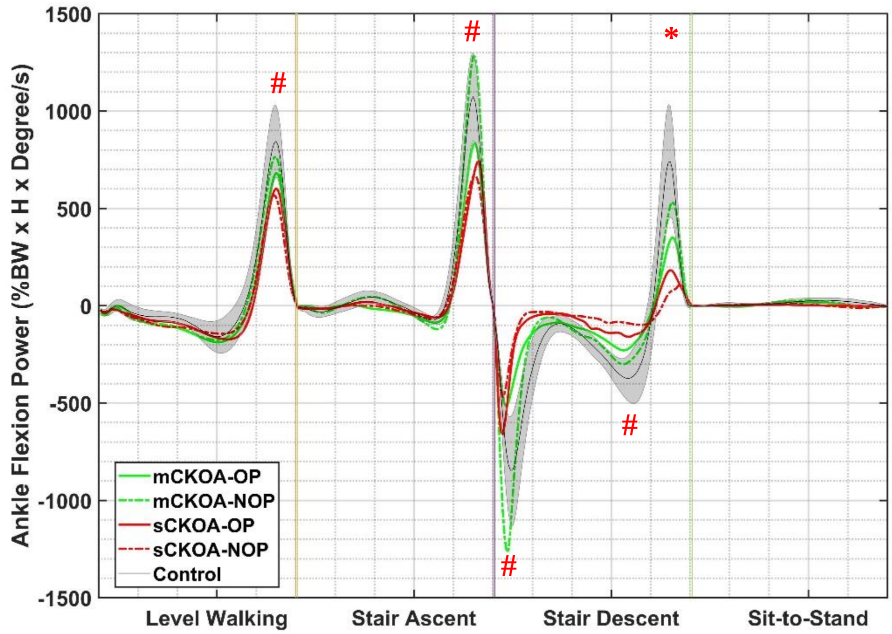


Figure 5.9 ensemble average curves of ankle power absorption (-) / generation (+) during tested tasks. Only stance phase is shown. The shaded are represent mean +/- SD of healthy controls. *mCKOA-OP vs. sCKOA-OP; #mCKOA-NOP vs. sCKOA-NOP, P<.05. With the progression of CKOA, contralateral ankle reduced ankle power in all tested tasks requiring ankle power, and ipsilateral ankle also reduced power generation during controlled lowering phase of stair descent.

Changes of loading on both legs

Table 5.5 shows changes of loading on both legs. During both pull-up phase of stair ascent and controlled-lowering phase of stair descent, peak total support moment (PTSM) was increased on the ipsilateral leg and decreased on the contralateral affected leg (Figure 5.10). During sit-to-stand, peak vertical GRF during standing was increased on the ipsilateral leg by 7% body weight (BW) and reduced on the contralateral leg by 10% BW (Figure 5.11).

Some changes were only evident on contralateral legs. Peak vertical GRF during push-off phase of stair ascent was reduced by 10% BW. Peak vertical GRF during weight acceptance phase of stair descent was reduced by 31% BW. PTSM during sit-to-stand was reduced while PTSM during early and late stance phase of level walking were increased.

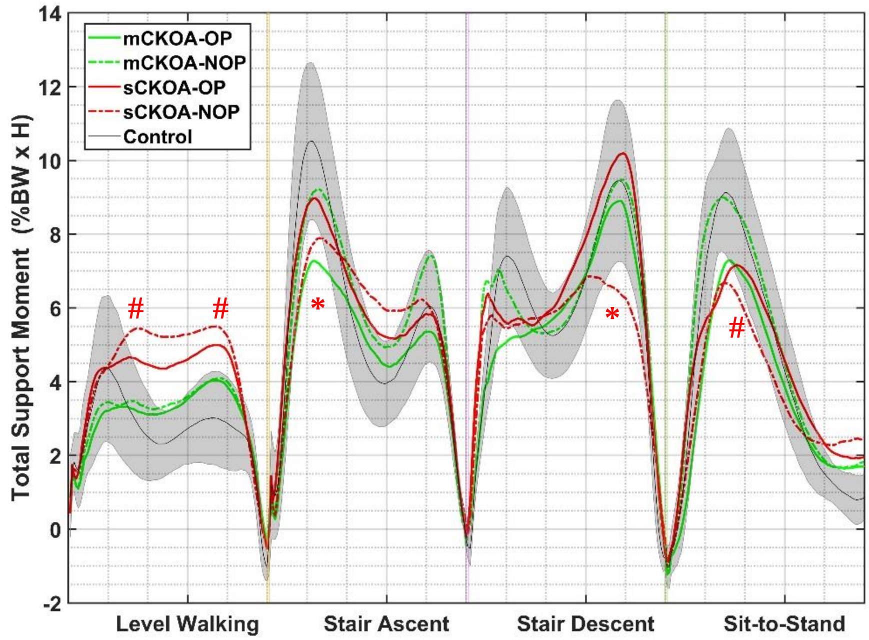


Figure 5.10 ensemble average curves of total support moment during tested tasks. Only stance phase is shown. The shaded are represent mean +/- SD of healthy controls. *mCKOA-OP vs. sCKOA-OP; #mCKOA-NOP vs. sCKOA-NOP, P<.05. With the progression of CKOA, ipsilateral leg increased total support moment during stair ascent/descent, while contralateral decreased that during stair descent and sit-to-stand.

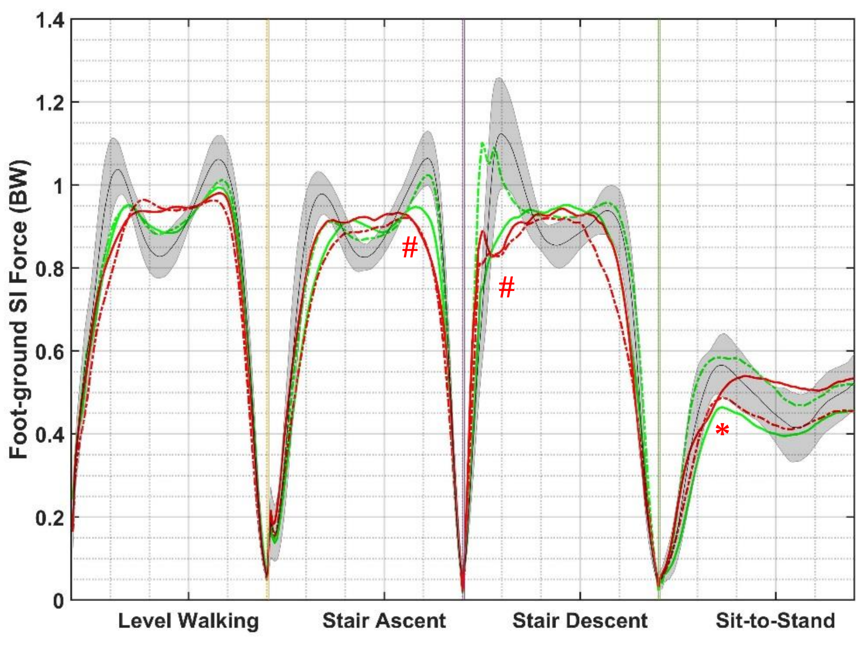


Figure 5.11 ensemble average curves of vertical GRF during tested tasks. Only stance phase is shown. The shaded are represent mean +/- SD of healthy controls. *mCKOA-OP vs. sCKOA-OP; #mCKOA-NOP vs. sCKOA-NOP, P<.05. With the progression of CKOA, ipsilateral leg decreased vertical GRF, while contralateral leg increased that.

Changes of joint contribution to dynamic support of the body

The contribution of hip moment to peak total support moment was increased on both ipsilateral side (from 18% to 45%) and contralateral side (from 15% to 42%) during weight acceptance of stair descent phase (Table 5.5). The contribution of ipsilateral ankle to peak total support moment was reduced (from 22% to 7%, P=.028) during early stance (pull-up) phase of stair ascent.

The contribution of contralateral hip was increased during both stair ascent (from 28% to 55%) and sit-to-stand (50% to 74%), while contralateral knee contribution wad decreased during stair ascent (from 50% to 18%) and sit-to-stand (from 44% to 24%).

Table 5.5 Vertical GRF, peak total support moment (PTSM) and joint contribution

Side	TKA (OP) side			Contralateral (NOP) side			Both sides
CKOA condition	Moderate	Severe	P	Moderate	Severe	P	Control
Vertical GRF, BW							
second peak vertical GRF stair ascent	0.98 ± 0.08	0.97 ± 0.03	NS	1.05 ± 0.12	0.95 ± 0.05	0.014	1.07 ± 0.08
first peak vertical GRF stair descent	1.04 ± 0.09	1.07 ± 0.12	NS	1.35 ± 0.33	1.04 ± 0.15	0.009	1.19 ± 0.13
peak GRF sit-to-stand	0.50 ± 0.07	0.57 ± 0.06	0.015	0.62 ± 0.08	0.52 ± 0.09	0.005	0.58 ± 0.06
PTSM, %BW x H							
early stance level walking	3.76 ± 1.78	4.95 ± 1.60	0.086	3.96 ± 1.49	5.54 ± 1.95	0.029	4.34 ± 1.76
late stance level walking	4.34 ± 1.65	5.16 ± 1.65	NS	4.37 ± 1.48	5.85 ± 1.67	0.025	3.32 ± 1.14
early stance stair ascent	7.84 ± 1.80	9.59 ± 1.64	0.018	10.20 ± 1.63	8.84 ± 1.82	0.062	10.86 ± 1.80
late stance stair descent	9.09 ± 1.76	10.50 ± 0.98	0.026	9.81 ± 1.91	7.79 ± 1.73	0.015	9.45 ± 2.18
sit-to-stand	8.14 ± 1.87	8.32 ± 1.64	NS	9.91 ± 1.98	7.32 ± 2.08	0.004	9.39 ± 1.30
Joint contribution to PTSM							
ankle contribution during gait	0.22 ± 0.07	0.15 ± 0.07	0.028	0.22 ± 0.07	0.28 ± 0.16	NS	0.17 ± 0.06
knee contribution during stair ascent	0.22 ± 0.28	0.39 ± 0.17	0.088	0.50 ± 0.14	0.18 ± 0.23	0.000	0.52 ± 0.15
hip contribution during stair ascent	0.56 ± 0.29	0.46 ± 0.16	NS	0.28 ± 0.12	0.55 ± 0.14	0.000	0.31 ± 0.14
hip contribution during stair descent	0.18 ± 0.26	0.45 ± 0.22	0.012	0.15 ± 0.18	0.42 ± 0.25	0.010	0.00 ± 0.19
knee contribution during sit-to-stand	0.28 ± 0.18	0.30 ± 0.16	NS	0.44 ± 0.08	0.24 ± 0.26	0.019	0.51 ± 0.12
hip contribution during sit-to-stand	0.61 ± 0.13	0.65 ± 0.15	NS	0.50 ± 0.09	0.74 ± 0.29	0.010	0.49 ± 0.11

Changes of absolute symmetry index (ASI)

Significant changes of ASI were listed in Table 5.6. Severe patients were more asymmetry in ankle dorsiflexion during late stance stair ascent, knee flexion angle during level walking, hip rotation during level walking and stair ascent, peak hip extension moment during late stance of level walking and ankle contribution to body support during stair ascent. On the other hand, severe patients were less asymmetry in second vertical GRF during stair ascent at stance phase, peak-to-peak hip flexion moment and first vertical GRF during stair descent at stance phase.

Table 5.6 Absolute asymmetry index

CKOA condition	Moderate	Severe	Moderate vs Severe, P	Control	Due to Change on OP or NOP side
Reduced Asymmetry					
range of hip flexion moment stair descent	49.2 ± 39.8	22.8 ± 14.1	0.043	11.1 ± 9.7	NOP decrease
second peak vertical GRF stair ascent	7.0 ± 5.2	3.4 ± 3.5	0.054	5.1 ± 3.9	NOP decrease
first peak vertical GRF stair descent	24.0 ± 16.9	10.8 ± 7.8	0.024	8.6 ± 6.2	NOP decrease
Increased Asymmetry					
ankle dorsiflexion angle late stance stair ascent	49.0 ± 48.5	103.1 ± 75.6	0.043	26.0 ± 14.9	NOP decrease
peak knee flexion swing level walking	8.1 ± 7.4	22.9 ± 24.5	0.048	4.8 ± 4.5	NOP decrease
Range of knee flexion level walking	14.1 ± 9.3	38.2 ± 39.0	0.041	3.7 ± 2.6	NOP decrease
peak hip extension moment late stance level walking	23.4 ± 17.1	45.6 ± 34.7	0.049	16.2 ± 8.5	NOP decrease
peak internal hip rotation angle level walking	101.6 ± 78.7	169.3 ± 61.3	0.022	81.4 ± 60.9	NOP decrease
peak external hip rotation angle level walking	55.1 ± 42.8	103.6 ± 55.9	0.020	41.5 ± 36.9	NOP decrease
peak external hip rotation angle stair ascent	48.1 ± 35.2	90.5 ± 61.6	0.044	39.6 ± 38.3	NOP decrease
ankle contribution to PTSM early stance stair scent	20.8 ± 16.8	59.5 ± 42.7	0.006	28.9 ± 23.1	OP decrease

Study Results #2: Bilateral versus Unilateral Total Knee Arthroplasty

Bilateral TKA Asymmetry: First TKA Side versus Second TKA Side

No significant difference in ankle/knee/hip sagittal plane joint angle and moment between the first and second TKA was found, except that second TKA side had a lower peak hip adduction moment than first TKA side (Table 5.7). Both sides were merged to represent bilateral TKA in later comparison study.

Table 5.7 BTKA asymmetry in hip adduction moment

Group	UTKA			BTKA			Control
Side	Operated	Non-operated (NOP)	Op vs. NOP, P	Second TKA	First TKA	First vs. second TKA, P*	Both sides
P2P hip adduction moment level walking	6.47 ± 1.00	6.94 ± 1.20	NS	6.51 ± 0.72	7.15 ± 0.81	0.077	7.35 ± 1.04
Peak hip adduction moment stair ascent	-5.81 ± 1.07	-5.77 ± 1.25	NS	-5.88 ± 0.68	-6.81 ± 0.94	0.020	-5.93 ± 1.00
P2P hip adduction moment stair ascent	6.35 ± 1.22	6.41 ± 1.17	NS	6.68 ± 0.78	7.69 ± 1.03	0.023	6.38 ± 0.99

Note: *independent t-test.

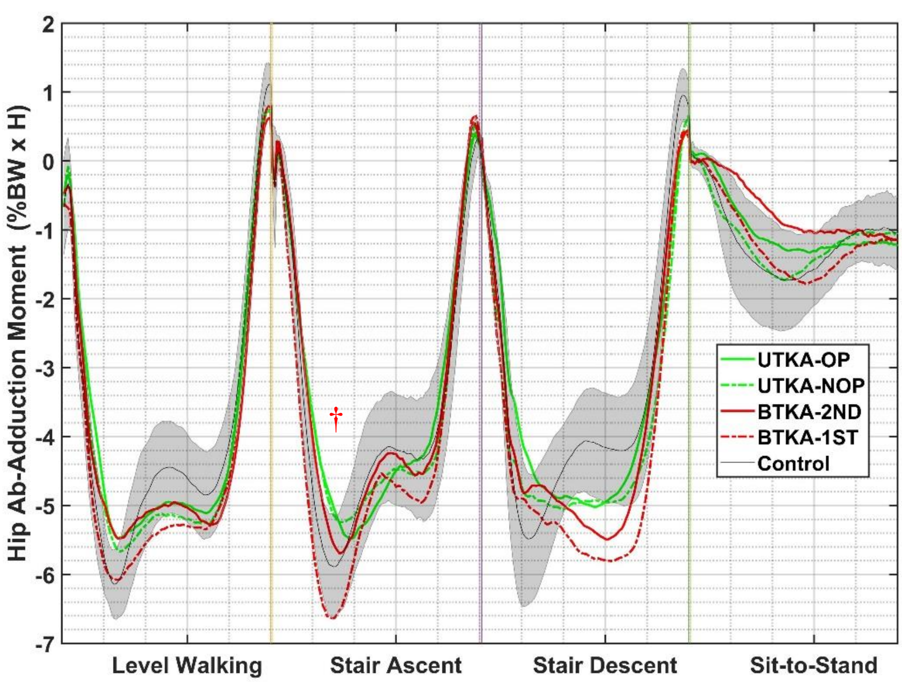


Figure 5.12 Normalized average waveform of hip adduction (-) /abduction (+) moment during multiple movement tasks. Only stance phase is presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. Significant differences: †BTKA-1ST vs. BTKA-2ND.

Bilateral versus Unilateral TKA

Speed and Duration Variables: Overall Performance

Results were summarized in Table 5.8. No difference was found between UTKA-OP, UTKA-NOP and BTKA groups (P>=.107), despite that BTKA group tended to have a shorter stance duration during stair ascent than UTKA-NOP group (P=.107, ES=.75).

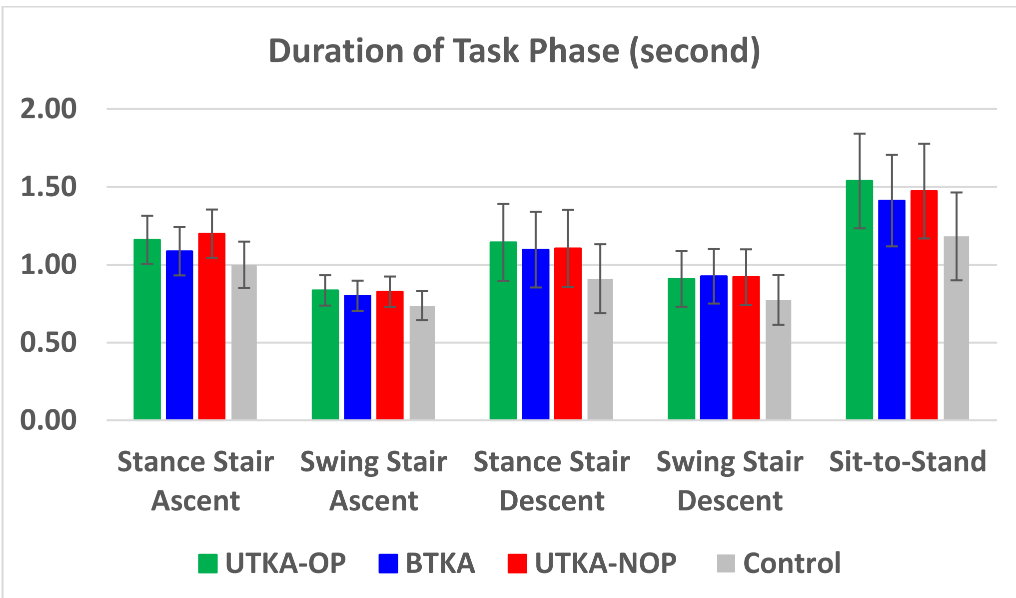


Figure 5.13 Performance comparison between BTKA patients and UTKA patients. No difference was found between them.

Table 5.8 performance comparison between UTKA and BTKA during multiple movement tasks. Mean ± SD

variable	UTKA-OP	UTKA-NOP	BTKA	Control
Side	Operated	Non-operated	Operated	Both
Stair ascent swing, sec	0.83 ± 0.10	0.83 ± 0.10	0.80 ± 0.10	0.74 ± 0.09
stair ascent stride, sec	1.99 ± 0.22	2.03 ± 0.22	1.89 ± 0.22	1.74 ± 0.21
stair decent stance, sec	1.14 ± 0.25	1.10 ± 0.25	1.10 ± 0.24	0.91 ± 0.22
Stair decent swing, sec	0.91 ± 0.18	0.92 ± 0.18	0.93 ± 0.17	0.77 ± 0.16
stair decent stride, sec	2.05 ± 0.39	2.02 ± 0.39	2.02 ± 0.38	1.68 ± 0.35
stand-to-sit, sec	1.66 ± 0.34	1.65 ± 0.34	1.68 ± 0.33	1.51 ± 0.31
sit-to-stand, sec	1.54 ± 0.30	1.47 ± 0.30	1.41 ± 0.29	1.18 ± 0.28

Note: no significant difference among UTKA-OP, UTKA-NOP and BTKA groups (P>=0.107);

Range of Joint Motion and Moment

UTKA asymmetry: UTKA group had symmetric ankle and hip range of motion. UTKA-OP group had a lower range of knee flexion during stair ascent in swing phase and a lower range of axial rotation during level walking than UTKA-NOP group.

BTKA group had similar range of motion as UTKA-OP group. BTKA group also had a lower range of knee axial rotation during level walking than UTKA-NOP group. Besides that, BTKA group had smaller range of ankle motion during level walking and swing phase of stair ascent.

Table 5.9 Joint range of motion (degree) during movement tasks, mean ± SD.

Variable\Group	UTKA-OP	UTKA-NOP	Asymmetry	BTKA	Control
Side	Operated	Non-operated	OP vs NOP	Operated	Both
Ankle flexion-extension					
level walking gait cycle	23.5 ± 3.4	25.6 ± 3.4	NS	20.9 ± 3.4[a]	25.8 ± 3.3
Stair ascent swing phase	32.3 ± 7.5	38.6 ± 7.5	NS	29.7 ± 7.5[a]	35.8 ± 7.2
Knee flexion-extension					
stair ascent swing phase	65.3 ± 10.3	76.7 ± 10.3	0.025	70.1 ± 10.3	74.5 ± 9.9
Knee rotation level walking	7.7 ± 2.6	10.9 ± 2.6	0.003	8.3 ± 2.6 [a]	9.7 ± 2.5

Note: Superscript indicates significance with P<.05; [a] BTKA vs UTKA-NOP.

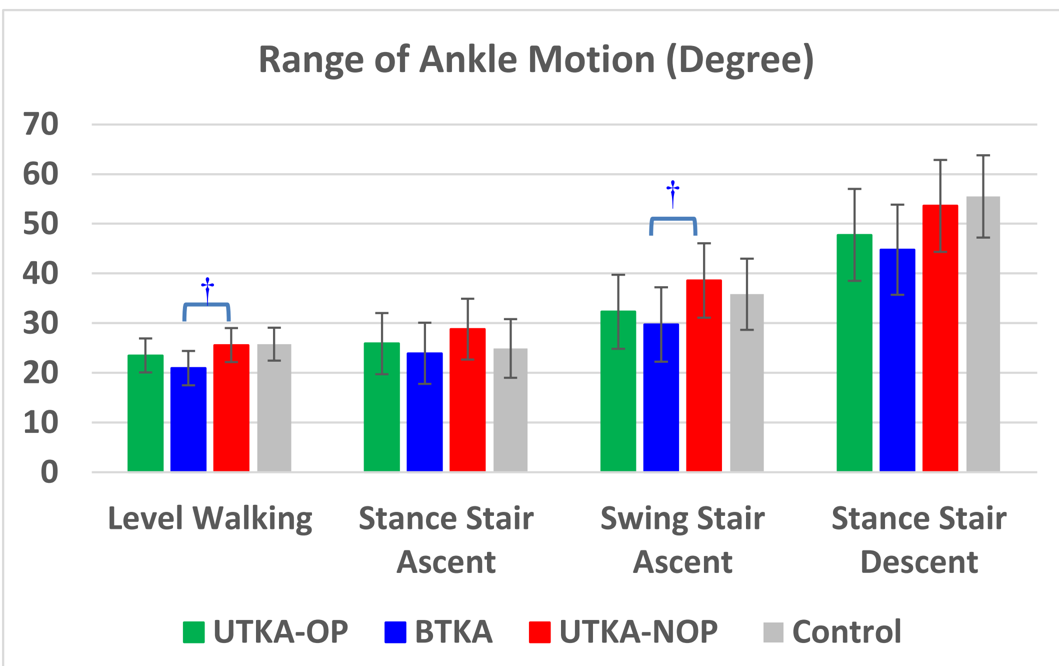


Figure 5.14 Range of ankle motion. † denotes significant difference P<.05, BTKA vs. UTKA-NOP side.

Joint Kinematics

UTKA asymmetry: UTKA patients had evident asymmetrical ankle (Figure 5.15) and knee motion (Figure 5.16 and Table 5.10). UTKA-OP had less dorsiflexed ankle than UTKA-NOP side during stair ascent in early stance, less flexed knee during stair ascent in swing phase.

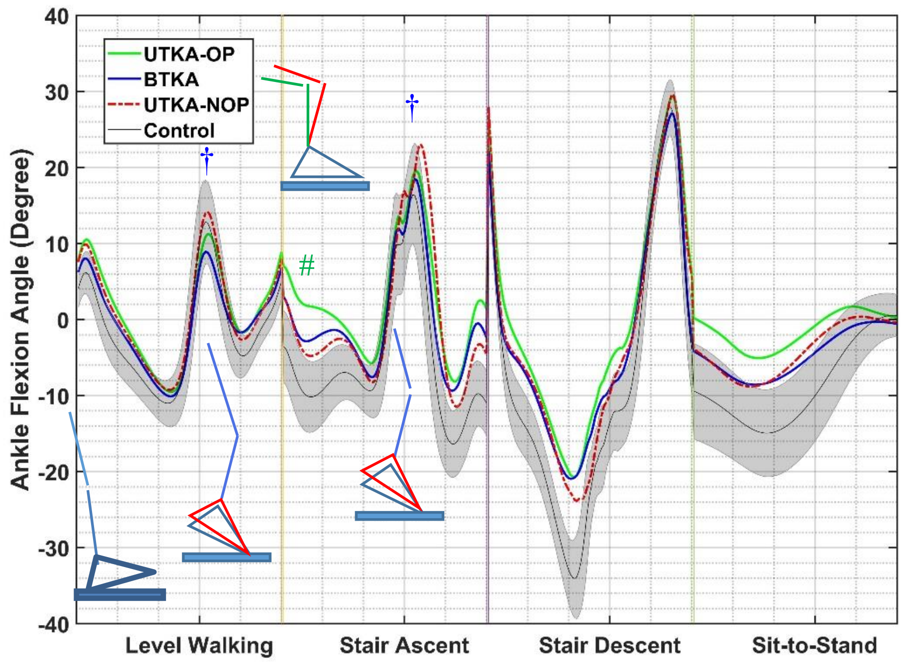


Figure 5.15 Normalized average waveform of ankle dorsiflexion (-) /plantarflexion (+) angle during multiple movement tasks. Both stance and swing phase are presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. Significant differences: †BTKA vs. UTKA-OP; # UTKA-OP vs. UTKA-NOP. Compared to BTKA patients, UTKA-NOP side had a more plantar-flexed ankle during early swing phase of both level walking and stair ascent (ES>=1.04). UTKA patients had evident asymmetrical ankle motion during stance phase of stair ascent (ES=1.27) and sit-to-stand (ES=.72), with a more plantar-flexed ankle on the operated side.

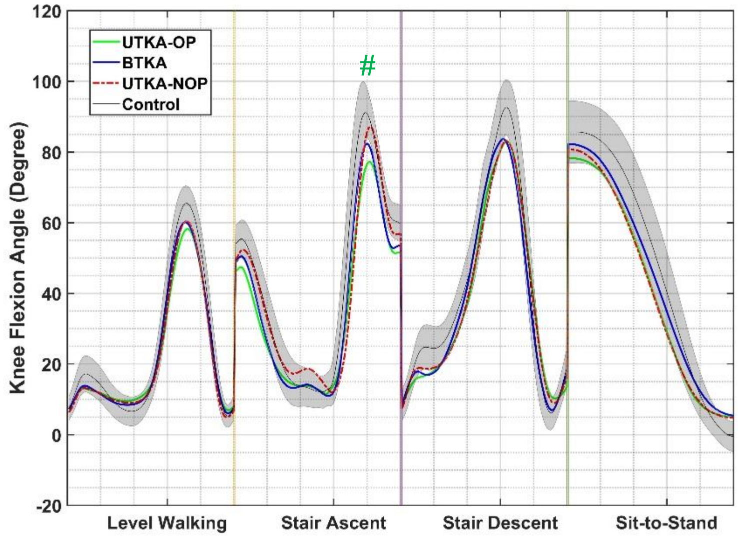


Figure 5.16 Normalized average waveform of knee extension (-) /flexion (+) angle during multiple movement tasks. Both stance and swing phase are presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. UTKA-OP knee had smaller flexion angle than UTKA-NOP knee during swing phase of stair ascent (P=.009, ES=1.14).

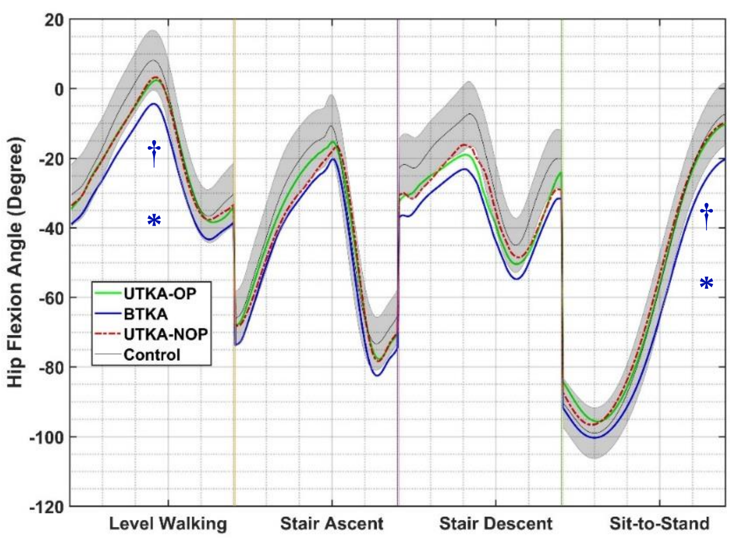


Figure 5.17 Normalized average waveform of hip flexion (-) /extension (+) angle during multiple movement tasks. Both stance phase and swing phase are presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. Significant differences: †BTKA vs. UTKA-NOP; *BTKA vs. UTKA-NP. BTKA patients had more hip flexion than both sides of UTKA patients during standing and level walking.

UTKA vs. BTKA: BTKA had a lower peak hip extension angle (Figure 5.17) and more anterior tilted pelvic (Figure 5.18) during level walking and standing than both sides of UTKA group. BTKA group had similar ankle and knee kinematics as UTKA-OP group. BTKA group had less plantarflexed ankle during early swing of both level walking and stair ascent than UTKA-NOP group.

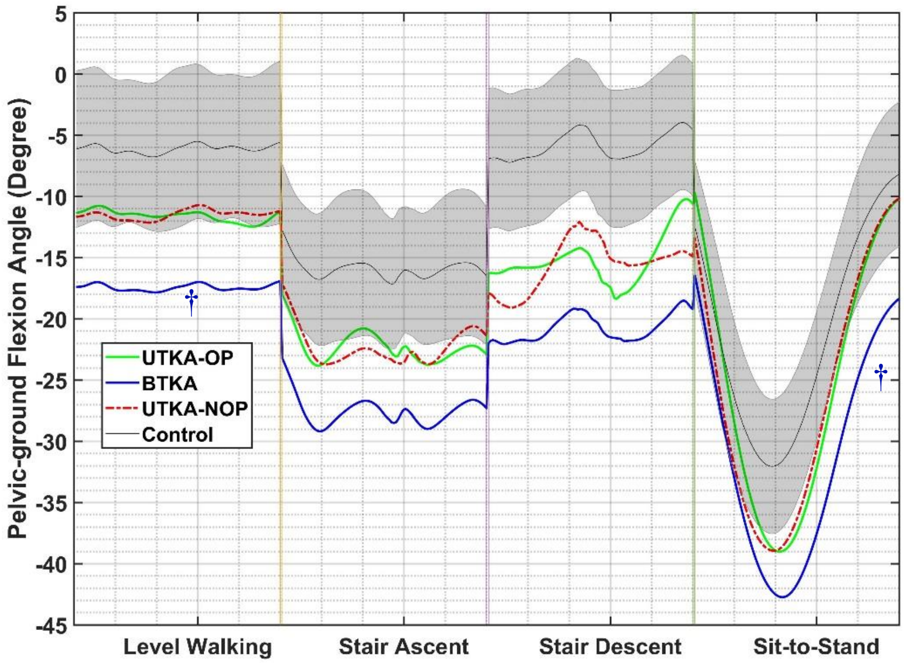


Figure 5.18 Normalized average waveform of pelvic anterior (-) /posterior (+) tilting angle during multiple movement tasks. Both stance phase and swing phase are presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. Significant differences: †BTKA vs. UTKA-NOP; *BTKA vs. UTKA-NP. BTKA patients had a more anteriorly tilted pelvic than both sides of UTKA patients during standing and level walking.

Table 5.10 Joint and segment angles (degree) on the sagittal plane, mean ± SD.

Variable	UTKA	UTKA-NOP	BTKA	Control
Side	Operated	Non-operated	Operated	Both
Ankle dorsi- (-)/plantar-flexion (+)				
peak plantarflexion early stance level walking	10.7 ± 3.0	10.1 ± 3.0	8.2 ± 3.0	6.3 ± 2.8
peak plantarflexion early swing level walking	12.7 ± 5.2	15.2 ± 5.2	9.6 ± 5.2[b]	14.0 ± 5.0
peak dorsiflexion early stance stair ascent	0.6 ± 4.9[c]	-5.6 ± 4.9	-3.1 ± 4.9	-10.6 ± 4.7
peak dorsiflexion late stance stair ascent	-6.4 ± 3.7	-9.3 ± 3.7	-8.2 ± 3.7	-9.7 ± 3.5
peak plantarflexion early swing stair ascent	23.6 ± 6.1	26.5 ± 6.1	20.3 ± 6.1[b]	18.8 ± 5.9
peak dorsiflexion sit-to-stand	-5.5 ± 5.6	-9.4 ± 5.6	-8.9 ± 5.5	-15.6 ± 5.2
Knee flexion (+)/ extension (+)				
peak knee flexion early stance stair ascent	48.7 ± 6.7	53.4 ± 6.7	50.6 ± 6.7	55.8 ± 6.5
peak knee flexion swing stair ascent	79.8 ± 8.2[c]	89.0 ± 8.2	83.2 ± 8.2	92.7 ± 7.9
flexion excursion early stance stair descent	8.1 ± 5.3	12.7 ± 5.3	10.5 ± 5.2	16.4 ± 4.7
Hip kinematics				
peak extension pre-swing level walking	3.2 ± 8.7	3.6 ± 8.7	-4.2 ± 8.7[a,b]	8.5 ± 8.4
peak flexion early stance stair descent	-17.3 ± 11.2	-12.9 ± 11.2	-21.5 ± 11.0	-6.5 ± 10.0
at end of sit-to-stand (standing)	-10.2 ± 8.7	-9.7 ± 8.7	-20.3 ± 8.4[a,b]	-7.3 ± 8.1
Pelvic anterior (-)/posterior tilt [d]				
at peak hip extension level walking	-11.3 ± 6.6	-11.1 ± 6.6	-17.4 ± 6.6[a,b]	-6.0 ± 6.3

at peak hip flexion stair ascending	-13.7 ± 7.5	-11.3 ± 7.5	-18.2 ± 7.3[b]	-15.8 ± 6.4
at end of sit-to-stand (standing)	-10.8 ± 6.4	-10.8 ± 6.4	-16.9 ± 6.4[a,b]	-7.2 ± 6.2

Note: Superscripts a-c indicate significant differences with P<.05; a BTKA vs. operated side of UTKA; b BTKA vs. non-operated side of UTKA; c UTKA-OP vs. UTKA-NOP; d pelvic kinematics is relative to the lab-coordinate system (ground).

Joint Moment and Power

UTKA asymmetry in vertical GRF and total support moment: UTKA-NOP group had a higher peak vertical GRF (Figure 5.19) during early stance of stair descent (1.32 BW vs. 1.10 BW, P=.039), and during sit-to-stand (0.62 BW vs. 0.50 BW, P<.001), and during late stance of stair ascent (1.04 BW vs. 0.99 BW, P=.078, ES=.85) than UTKA-OP group. UTKA-NOP also had a higher support moment during stair ascent in early stance and during sit-to-stand (Table 5.11).

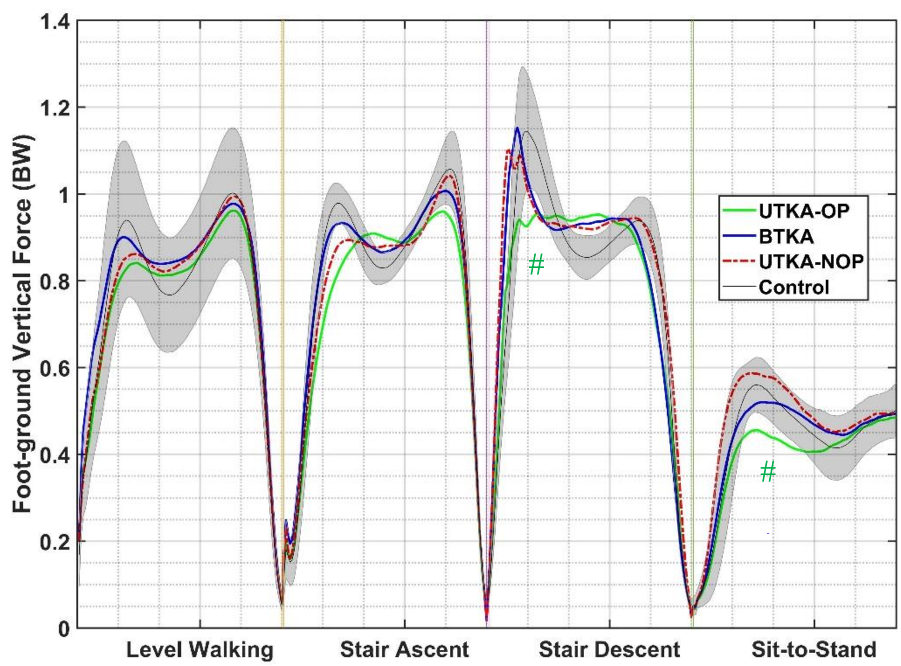


Figure 5.19 Normalized average waveform of vertical GRF during multiple movement tasks. Only stance phase is presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. Significant differences: †BTKA vs. UTKA-NOP; *BTKA vs. UTKA-NP; #UTKA-OP vs. UTKA-NOP.

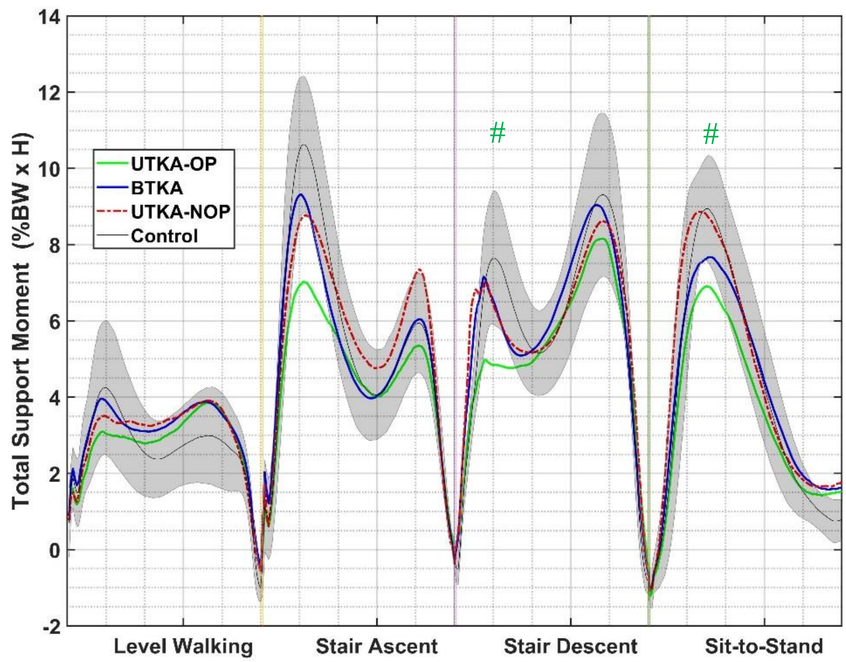


Figure 5.20 Normalized average waveform of vertical GRF during multiple movement tasks. Only stance phase is presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. Significant differences: †BTKA vs. UTKA-NOP; *BTKA vs. UTKA-NP; #UTKA-OP vs. UTKA-NOP.

UTKA symmetry in ankle moment/power: UTKA had symmetrical ankle moment, despite that UTKA-NOP group (6.08 %BW x H) tended to have a higher ankle dorsiflexion moment than UTKA-OP group (4.60%BW x H) during early stance of stair descent (P=.097). UTKA had greater ankle power generation during stair ascent in late stance (pushing up), and higher ankle power absorption during stair descent in early stance (weight acceptance).

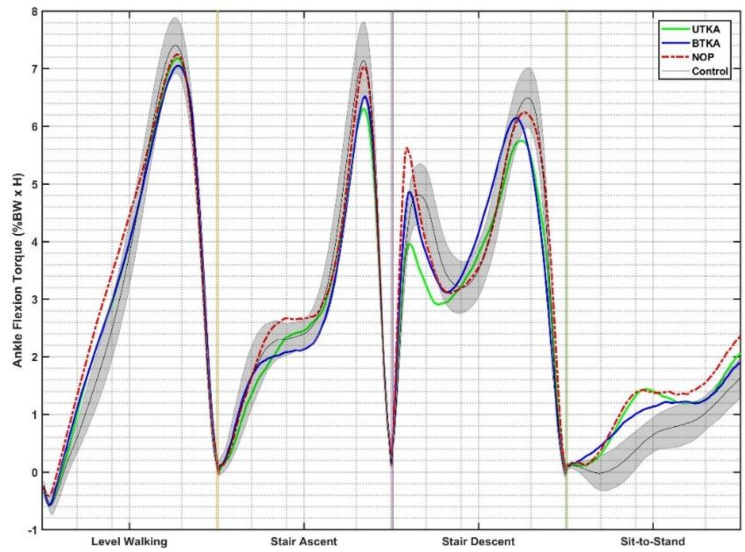


Figure 5.21 Normalized average waveform of external ankle dorsiflexion (+)/plantar flexion (-) moment during multiple movement tasks. Only stance phase is presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. Significant differences: †BTKA vs. UTKA-NOP; *BTKA vs. UTKA-NP; #UTKA-OP vs. UTKA-NOP.

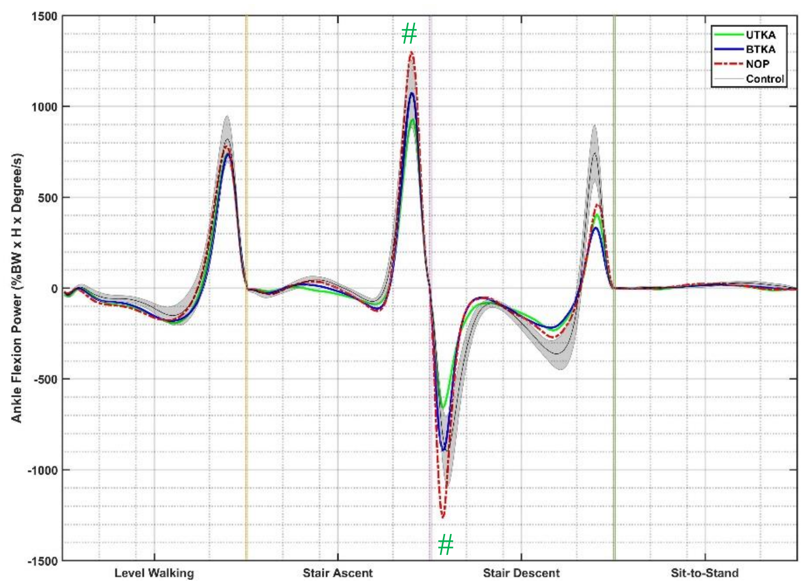


Figure 5.22 Normalized average waveform of external ankle power generation (+) / absorption (-) during multiple movement tasks. Only stance phase is presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. Significant differences: †BTKA vs. UTKA-NOP; *BTKA vs. UTKA-NP; #UTKA-OP vs. UTKA-NOP.

UTKA asymmetry in knee moment/power: UTKA-NOP group had a higher peak knee flexion moment during stair ascent in early stance, during stair descent in late stance, and during sit-to-stand than UTKA-OP group (all P<=.027). As a result, UTKA-NOP group also had a higher range of knee flexion-extension moment than UTKA-OP during stair ascent (P<.001) and stair descent (P<.032). UTKA-NOP group also had a higher peak knee generation power during stair ascent in early stance and sit-to-stand (P<=.009).

UTKA asymmetry in hip moment/power: UTKA-NOP group had higher peak hip extension moment and higher peak hip power generation than UTKA-OP group during late stance of stair descent. UTKA-NOP also had higher range of hip flexion-extension moment than UTKA-OP group.

UTKA asymmetry in joint contribution to body support: During sit-to-stand, UTKA-OP group had a lower knee contribution but a higher hip contribution to total support moment than UTKA-OP group.

BTKA vs. UTKA in vertical GRF and total support moment: like UTKA-OP group, BTKA group (.56 BW) had a lower peak vertical GRF than UTKA-NOP group (.62 BW) during sit-to-stand (P=.049). Like UTKA-NOP group, BTKA group also had a higher peak total support moment during stair ascent in early stance than UTKA-OP group (P=.017, Table 5.11).

BTKA vs. UTKA in ankle moment/power: BTKA group had similar ankle moment/power as UTKA-OP and UTKA-NOP groups.

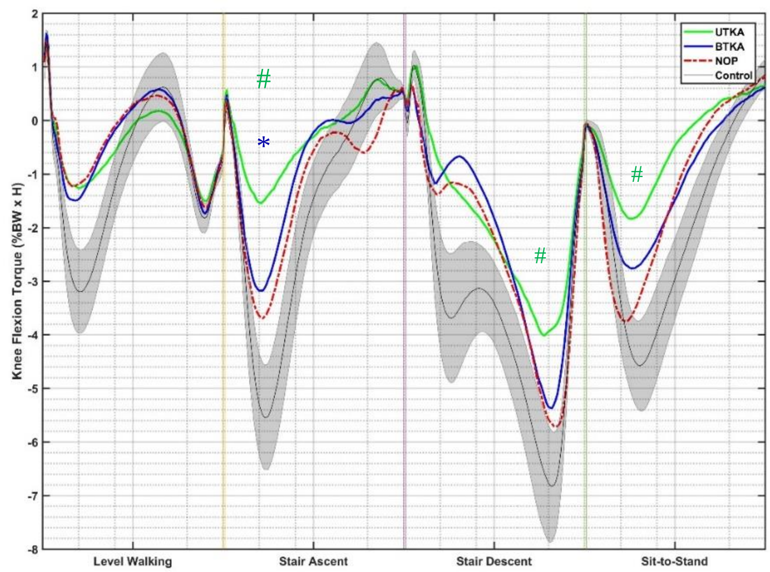


Figure 5.23 Normalized average waveform of external knee flexion (-) / extension (+) moment during multiple movement tasks. Only stance phase is presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. Significant differences: †BTKA vs. UTKA-NOP; *BTKA vs. UTKA-OP; #UTKA-OP vs. UTKA-NOP.

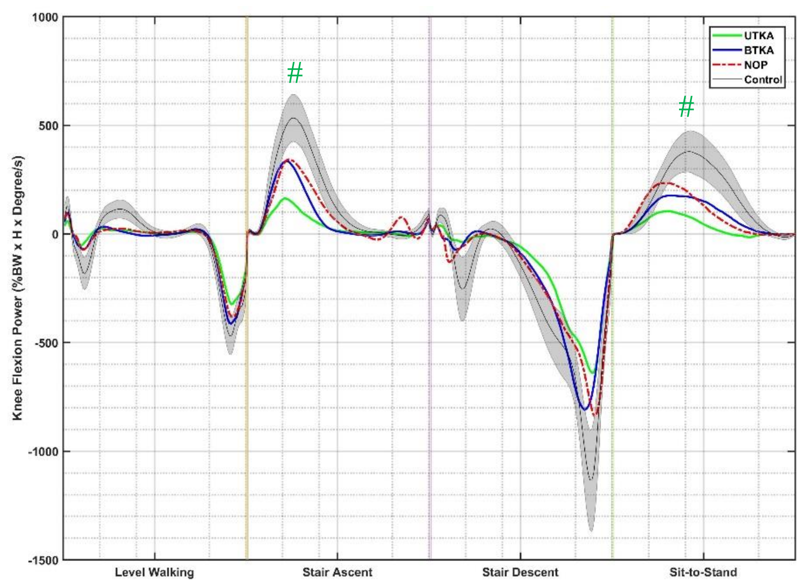


Figure 5.24 Normalized average waveform of knee power absorption (-) / generation (+) during multiple movement tasks. Only stance phase is presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. Significant differences: †BTKA vs. UTKA-NOP; *BTKA vs. UTKA-NP; #UTKA-OP vs. UTKA-NOP.

BTKA vs. UTKA in knee moment/power: Like UTKA-NOP group, BTKA group also had a higher peak knee flexion moment during stair ascent in early stance and higher peak knee power generation during sit-to-stand than UTKA-OP group. BTKA group had a higher range of knee flexion moment during stair ascent than UTKA-OP group but still lower than UTKA-NOP group. Like UTKA-OP group, BTKA group also had a lower peak knee flexion during stair descent in early stance than UTKA-NOP group.

BTKA vs. UTKA in hip moment/power: BTKA had similar hip flexion-extension moment/power as UTKA-OP group but tended to have a higher range of hip flexion moment during stair ascent than UTKA-NOP group (P=.052). Like UTKA-OP group, BTKA group also had a lower peak hip generation power during stair descent in late stance than UTKA-NOP group.

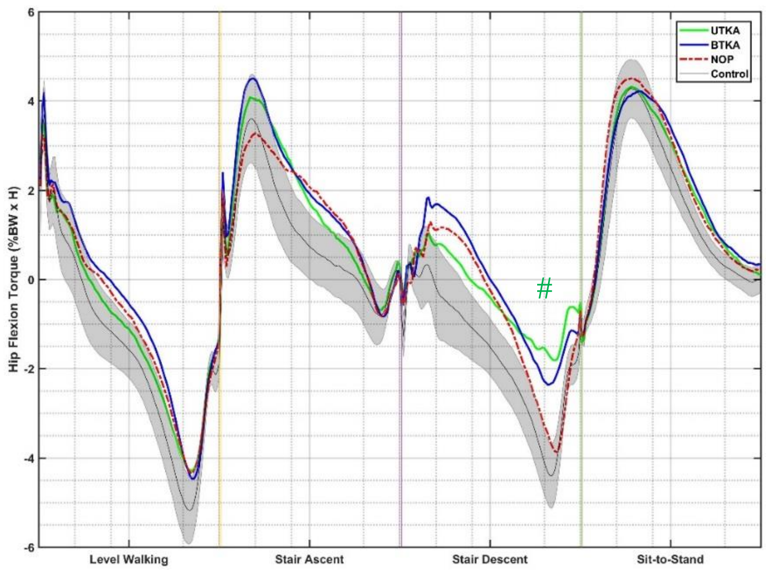


Figure 5.25 Normalized average waveform of external hip flexion (+) / extension (+) moment during multiple movement tasks. Only stance phase is presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. Significant differences: †BTKA vs. UTKA-NOP; *BTKA vs. UTKA-NP; #UTKA-OP vs. UTKA-NOP.

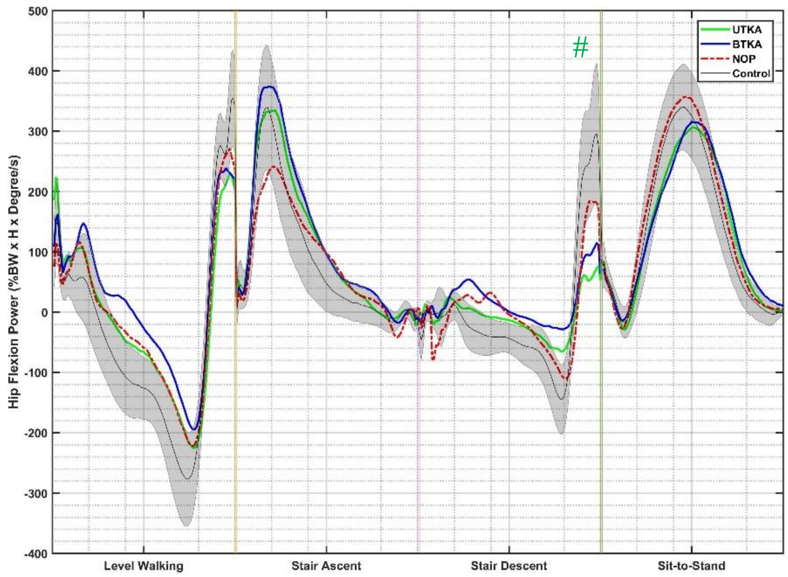


Figure 5.26 Normalized average waveform of hip power absorption (-) / generation (+) during multiple movement tasks. Only stance phase is presented. The shaded area represents mean +/- one standard deviation of the healthy control subjects. Significant differences: †BTKA vs. UTKA-NOP; *BTKA vs. UTKA-NP; #UTKA-OP vs. UTKA-NOP.

Table 5.11 comparing joint moment between UTKA and BTKA.

Variable	UTKA			BTKA	Control
Side	Operated (OP)	Nonoperated (NOP)	OP vs. NOP	Operated	Both
Ankle dorsi- (-)/plantar-flexion (+) moment					
peak dorsiflexion moment early stance stair decent	4.60 ± 1.54	6.08 ± 1.54	.031	5.08 ± 1.51[a]	5.12 ± 1.37
Knee flexion (+)/ extension (+) moment					
peak flexion moment early stance stair ascent	-2.00 ± 1.52	-4.32 ± 1.52	<.001	-3.30 ± 1.52	-5.70 ± 1.46
peak flexion moment late stance stair descent	-4.45 ± 1.90	-6.43 ± 1.90	.017	-5.66 ± 1.86	-7.00 ± 1.70
peak flexion moment sit-to-stand	-2.25 ± 1.25	-4.23 ± 1.25	<.001	-3.24 ± 1.21	-4.81 ± 1.16
range of flexion moment stair ascent stance	3.36 ± 1.37	5.65 ± 1.37	<.001	4.49 ± 1.37[b]	6.86 ± 1.32
range of flexion moment stair descent stance	5.84 ± 1.78	7.65 ± 1.78	.021	6.63 ± 1.74	8.16 ± 1.59
Hip flexion (-)/extension (+) moment					
peak flexion moment early stance stair ascent	4.44 ± 1.42	3.64 ± 1.42	NS	4.67 ± 1.42	3.77 ± 1.36
peak extension moment late stance stair descent	-2.39 ± 1.62	-4.30 ± 1.62	.006	-2.80 ± 1.59[b]	-4.52 ± 1.45
range of flexion moment stair ascent	5.50 ± 1.30	4.99 ± 1.30	NS	6.13 ± 1.30[b]	5.00 ± 1.25
range of flexion moment stair descent	4.44 ± 1.68	6.56 ± 1.68	.003	5.78 ± 1.65	5.49 ± 1.50
Total support moment					
peak total support moment early stance stair ascent	7.53 ± 1.89	9.56 ± 1.89	0.018	9.45 ± 1.89	10.86 ± 1.81

peak total support moment standing	7.94 ± 1.61	9.81 ± 1.61	0.016	8.55 ± 1.56	9.39 ± 1.50
knee contribution during sit-to-stand	0.28 ± 0.13	0.42 ± 0.13	0.012	0.36 ± 0.13	0.51 ± 0.12
hip contribution during sit-to-stand	0.62 ± 0.11	0.51 ± 0.11	0.022	0.55 ± 0.11	0.49 ± 0.11
Vertical ground reaction force [g]					
peak during late stance stair ascent	0.99 ± 0.09	1.06 ± 0.09	.078	1.04 ± 0.09	1.07 ± 0.09
peak during early stance stair descent	1.10 ± 0.24	1.32 ± 0.24	.039	1.32 ± 0.24[a]	1.19 ± 0.22
peak during sit-to-stand	0.50 ± 0.08	0.62 ± 0.08	<.001	0.56 ± 0.07[b]	0.58 ± 0.07

Table 5.12 comparing joint power (%BW x H x deg/s) between UTKA vs. BTKA.

Generation (+) /absorption power (-)	UTKA			BTKA	Control
	Operated (OP)	Nonoperated (NOP)	OP vs. NOP	Operated	Both
Ankle power					
peak generation power late stance stair ascent	985 ± 329	1337 ± 329	0.014	1116 ± 329	1083 ± 317
peak absorption power early stance stair descent	-726 ± 448	-1315 ± 448	0.002	-997 ± 440	-926 ± 401
peak generation power late stance stair descent	450 ± 270	586 ± 270	NS	397 ± 265	767 ± 241
Knee power					
peak generation power early stance stair ascent	218 ± 167	414 ± 167	0.006	350 ± 167	558 ± 161
peak generation power sit-to-stand	137 ± 125	294 ± 125	0.003	236 ± 121	407 ± 116
Hip power					
peak generation power early stance stair ascent	389 ± 168	272 ± 168	0.180	411 ± 168	364 ± 162
peak generation power late stance stair descent	121 ± 152	251 ± 152	0.066	145 ± 149	330 ± 136

Absolute Asymmetry Index

UTKA patients had a higher asymmetry level in knee moment and power, and total support moment than BTKA patients (Table 5.13).

Table 5.13 Comparison of absolute asymmetry index (ASI) between UTKA and BTKA patients.

Group	UTKA	BTKA	CONTROL	UTKA VS. BTKA, P	Higher loading on UTKA-NOP
Stair Ascent					
peak external knee flexion moment early stance stair ascent	96.5 ± 69.6	39.7 ± 22.5	28.7 ± 27.1	0.021	Yes

range knee flexion moment stair ascent	57.3 ± 27.9	22.5 ± 12.8	23.9 ± 16.8	0.001	Yes
Peak knee power generation stair ascent	89.1 ± 73.3	32.3 ± 28.2	31.0 ± 23.9	0.029	Yes
Peak total support moment early stance stair ascent	32.9 ± 22.8	13.6 ± 8.2	10.2 ± 7.4	0.018	Yes
Sit-to-Stand					
Peak knee power generation sit-to-stand	83.8 ± 61.9	35.9 ± 31.5	30.5 ± 23.7	0.036	Yes
Stair Descent					
P2P knee flexion moment stair descent	37.2 ± 18.3	22.0 ± 16.0	13.8 ± 10.9	0.049	Yes
peak external knee flexion moment early stance stair descent	79.6 ± 60.4	34.8 ± 28.8	23.4 ± 23.5	0.043	Yes

Study Results #3: Mobile- versus Fixed-Bearing Total Knee Arthroplasty

The subjects in FB, MB and control groups were matched for height (FB, 1.58 ± 0.08 m; MB, 1.56 ± 0.10 m; control, 1.61 ± 0.06 m), weight (FB, 68.5 ± 13.1 kg; MB, 63.4 ± 16.9 kg; control, 65.1 ± 9.9 kg), and body mass index (FB, 27.3 ± 3.5; MB, 25.8 ± 5.1; control, 25.1 ± 3.1).The FB and MB subjects were matched for age (FB, 68.8 ± 5.5; MB, 72.2 ± 5.4) and post-surgery time (FB, 13.0 ± 5.5 months; MB, 16.1 ± 7.7 months). Controls (age, 56.4 ± 4.3) were 15.8 ± 1.5 years younger than the FB patients ($p < 0.01$), and 12.4 ± 1.4 years younger than the MB patients ($p < 0.01$).

Stance duration

The discrete variables were normally distributed in each group. The FB and MB groups had similar stance duration for all the investigated tasks (Table 1, $p >= 0.13$). Both the FB and MB groups had a normal stance duration as the control group during 3 tasks ($p >= 0.126$), i.e. level walking (0.70 ± 0.09 s), step turn (1.17 ± 0.28 s) and spin turn (1.21 ± 0.24 s). The stance duration of the FB and MB groups were respectively 0.18 ± 0.06 s ($p<0.01$) and 0.13 ± 0.06 s ($p=0.058$) longer than the control group during stair ascent, and respectively 0.33 ± 0.10 s ($p<0.01$) and 0.26 ± 0.10 s ($p=0.018$) longer than the control group during stair descent.

Knee rotation

The FB knees had similar axial rotation as the MB knees during all investigated activities (Figure 1 and Table 1). Both the FB and MB knees had normal axial rotation range except during step and spin turns. During step turn and compare to controls, the FB and MB knees were respectively 3.9° ± 1.4° ($p=0.022$) and 3.7° ± 1.4° ($p = 0.024$) less internally rotated and had respectively 2.7° ± 1.3° ($p = 0.077$) and 5.3° ± 1.2° ($p < 0.01$) less axial rotation range. During spin turn, the FB knees were 5.1° ± 1.5° ($p < 0.01$) less internally rotated, and MB knees were 4.6° ± 1.5° ($p < 0.01$) less externally rotated than the control knees. In addition, the FB and MB knees had respectively 6.9° ± 1.5° ($p < 0.01$) and 6.8° ± 1.4° ($p < 0.01$) less axial rotation range than the control knees.

Knee rotation moments

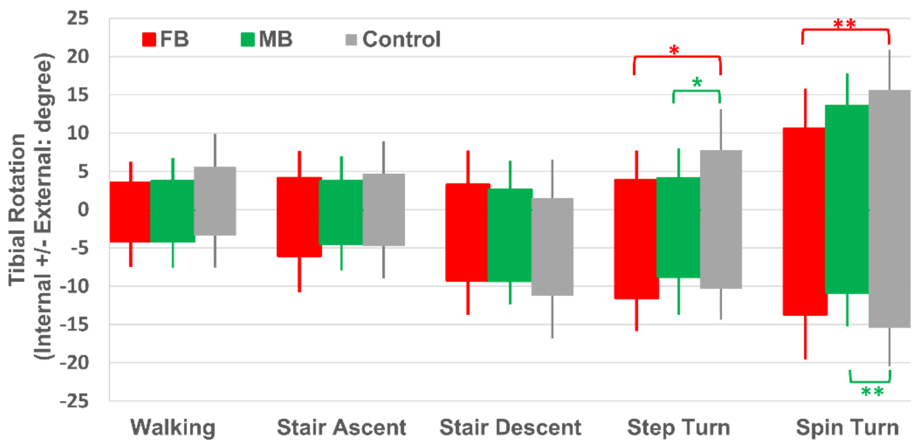
On the transverse plane, the FB knees had similar knee rotation moments as the MB knees during all investigated activities (Figure 2). When compared to the control knees, the MB knees had 0.044 ± 0.017 ($p = 0.035$) and 0.112 ± 0.034 ($p = 0.005$) %BW x H lower peak knee external rotation moment during level walking and spin turn, respectively. Except these differences between the MB and control groups, the three groups performed the other investigated tasks with similar peak external ($p > 0.07$), peak internal ($p > 0.18$) and peak-to-peak rotation moments ($p > 0.24$). Further correlation analysis indicated that peak knee external rotation moment was correlated to peak knee external rotation during spin turn ($p=0.001$, Pearson correlation $r = 0.409$).

Table 5.14 Mean ± standard deviation of stance time and rotation peaks, and the p value of each comparison.

Variable	FB	MB	Control	FB vs Control	MB vs. Control
Stance duration, second					
Walking	0.70 ± 0.08	0.71 ± 0.09	0.68 ± 0.09	NS	NS
Stair Ascent	1.18 ± 0.30	1.13 ± 0.12	1.00 ± 0.13	p=0.007	p=0.058
Stair Descent	1.24 ± 0.36	1.17 ± 0.42	0.91 ± 0.14	p=0.004	p=0.018
Step Turn	1.27 ± 0.35	1.21 ± 0.30	1.09 ± 0.20	NS	NS
Spin Turn	1.27 ± 0.27	1.26 ± 0.25	1.14 ± 0.19	NS	NS
Peak internal rotation angle, degree					

Walking	3.5 ± 2.8	3.7 ± 3.1	5.6 ± 4.3	NS	NS
Stair Ascent	4.1 ± 3.6	3.8 ± 3.2	4.7 ± 4.2	NS	NS
Stair Descent	3.2 ± 4.5	2.6 ± 3.8	1.5 ± 5.0	NS	NS
Step Turn	3.8 ± 3.9	4.1 ± 4.0	7.8 ± 5.4	p=0.022	p=0.024
Spin Turn	10.5 ± 5.3	13.5 ± 4.3	15.6 ± 5.2	p=0.004	NS
Peak external rotation angle, degree					
Walking	-4.1 ± 3.4	-4.1 ± 3.5	-3.4 ± 4.2	NS	NS
Stair Ascent	-6.0 ± 4.8	-4.4 ± 3.5	-4.8 ± 4.2	NS	NS
Stair Descent	-9.2 ± 4.5	-9.3 ± 3.1	-11.3 ± 5.6	NS	NS
Step Turn	-11.5 ± 4.3	-8.7 ± 5.0	-10.3 ± 4.1	NS	NS
Spin Turn	-13.7 ± 5.9	-10.8 ± 4.5	-15.4 ± 5.1	NS	p=0.009
Range of knee rotation, degree					
Walking	7.6 ± 2.0	7.8 ± 2.4	8.9 ± 3.0	NS	NS
Stair Ascent	10.1 ± 2.9	8.2 ± 2.5	9.5 ± 3.5	NS	NS
Stair Descent	12.5 ± 4.5	11.8 ± 3.8	12.8 ± 4.6	NS	NS
Step Turn	15.4 ± 3.9	12.8 ± 4.1	18.1 ± 3.6	p=0.077	p<0.001
Spin Turn	24.2 ± 5.7	24.3 ± 5.2	31.1 ± 3.8	p<0.001	p<0.001

Note: NS, not significant, p>0.1.



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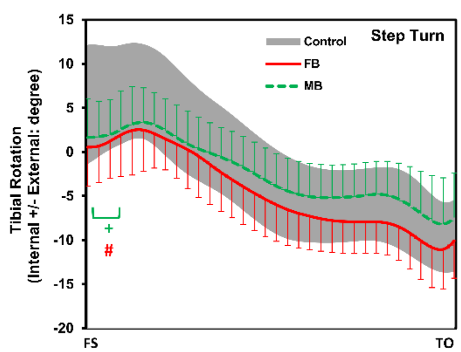
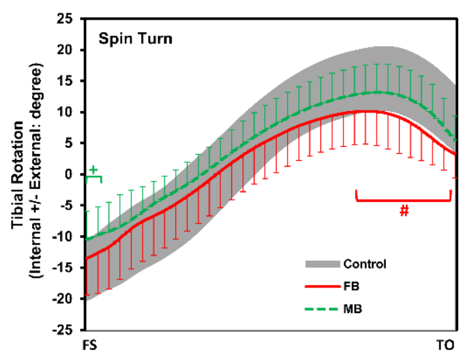
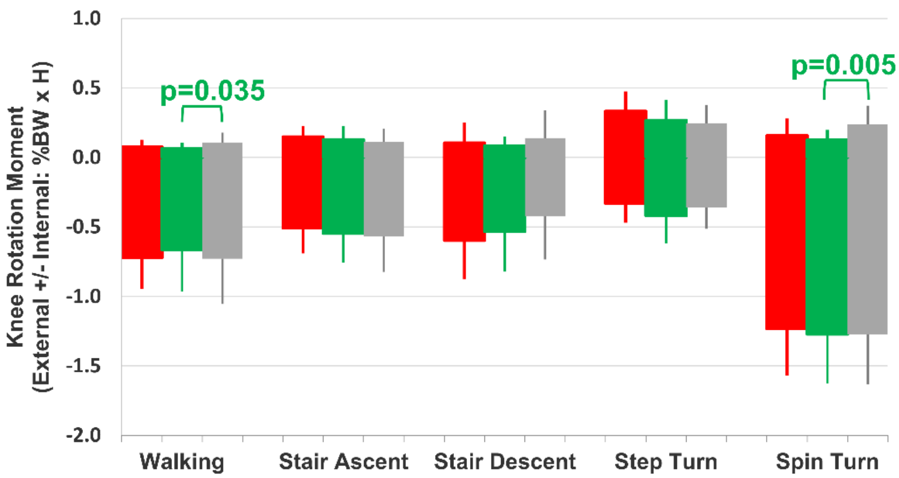


B)



C)

Figure 5.27 A) Range of Knee Rotation during 5 different tasks. The top and bottom of each box were respectively peak internal rotation and external rotation angles, whose one standard deviation was represented by extended whisker. Significant difference was denoted as *p<0.05 or **p<0.01. The group mean ensemble curves for knee rotation during B) step turn and C) spin turn. The shaded area represented mean ± standard deviation of the control group. FS: foot strike; TO: toe-off; +: MB vs. Control, p<0.05; # FB vs. Control, p<0.05.



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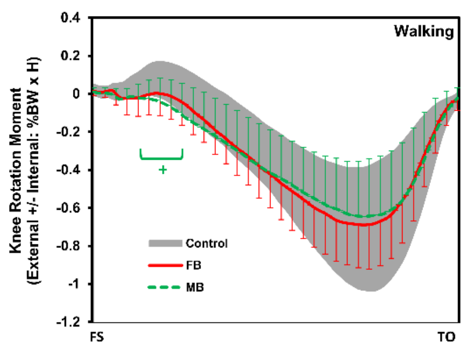

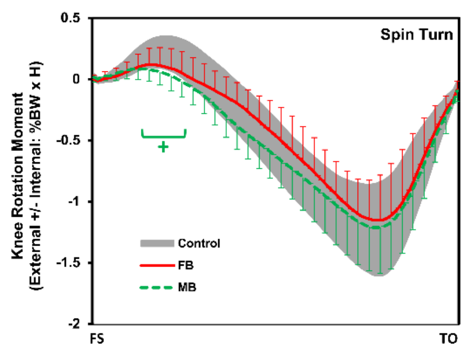

B) C)

Figure 5.28 A) Peak knee internal rotation moment and peak external rotation moment during 5 investigated tasks. The top and bottom of each box were respectively peak external rotation moment and internal rotation moment, whose one standard deviation was represented by extended whisker. Group mean ensemble curves of knee rotation moment during step turn B) and spin turn C). The shaded area represented mean ± standard deviation of the control group. +: MB vs. Control, p<0.05.

Chapter 6 Discussion

Discussion: Concurrent Osteoarthritis of the Contralateral Knee

The first hypothesis that the progression of CKOA will impact the performance of patients was supported. Besides reduced speed, single limb support phase was reduced on the contralateral affected side while stance phase was prolonged on the sound TKA side. These phase changes indicate a reliance of TKA side and avoidance of using painful side. Performance reduction with the progression of knee arthritis was expected and has been observed (Astephen, Deluzio, Caldwell, & Dunbar, 2008). The slower speed in walking, stair climbing and standing may affect the kinematic and kinetic comparison between groups. The selection of gait speed in gait study has been discussed elsewhere (Astephen, Deluzio, Caldwell, & Dunbar, 2008). Previous studies have demonstrated the effect of walking speed on gait mechanics especially sagittal plane kinematics (Joseph A Zeni & Higginson, 2009). But gait speed reduction is inherent in severe knee OA patients, and asking the patients to walk at a self-selected speed in this study enabled the identification of natural changes associated with the progression of CKOA.

The second hypothesis that the progression of CKOA impacted the biomechanics of the operated knee (TKA) was supported, although only two variables were significantly changed. The operated knee had increased external knee axial rotation at heel strike of level walking and suffered a higher external knee flexion moment during weight acceptance (pull-up phase) of stair ascent. The increased knee flexion moment could not be explained by stair-ascending speed, because severe CKOA patients climbed stair slower than moderate CKOA patients.

An increase in knee axial rotation during may increase the wear rate of the TKA bearings (Johnson, Laurent, Yao, & Gilbertson, 2001; H. M. J. McEwen et al., 2005). A increased peak knee flexion moment on the operated knee may indicate a higher demand on the TKA and risk of tibia component loosening (Hilding, Lanshammar, & Ryd, 1996). Future TKA studies that compare knee axial rotation during level walking or knee flexion moment during stair ascent should report the severity of CKOA. One subtopic of this dissertation, which was aimed to compare knee axial rotation and knee flexion moment between mobile- and fixed-bearing design, have excluded patients with severe CKOA in the study design.

The third hypothesis that both ankles were affected was partially supported, because the ipsilateral ankle only reduced ankle power generation during controlled lowering phase of stair descent. This change was bilateral change and could be partially explained by reduced ankle extension velocity. No other studies have reported the effect of CKOA on ipsilateral ankle. Ro et al. 2018 reported that gait speed is improved after TKA and ipsilateral ankle dorsiflexion moment is also increased (Ro et al., 2018).

The contralateral ankle reduced dorsiflexion during all tested tasks and reduced range of motion during stance of stair descent. The contralateral ankle kinematics might be altered to compensate the reduced kinematics of contralateral knee. (Astephen, Deluzio, Caldwell, & Dunbar, 2008) reported a reduction in ankle flexion range during level walking from average 30.7° of healthy controls to 27.6° of severe OA patients, while we found all three groups of this study had similar ankle flexion range (average 25°) during walking.

Contralateral ankle also reduced kinetics (moment and power) during level walking and stair ascent/descent. Reduced ankle dorsiflexion and task execution speed may account for the reduced ankle kinetics. It has been also reported peak ankle flexion moment of the same side during level walking is reduced as knee OA progresses (Astephen, Deluzio, Caldwell, & Dunbar, 2008). No other studies have reported how ankle of the affected side change its kinematics and kinetics with the progression of knee arthritis.

The fourth hypothesis that both hips were affected was supported. The progression of CKOA increased ipsilateral hip frontal plane kinematics during stair ascent/descent and more internally rotated ipsilateral hip during sit-to-stand. The ipsilateral hip also increased hip moment/contribution to support body during stair descent.

No study has reported the effect of knee OA on hip of the unaffected side. Previous study on knee OA has shown that hip of the unaffected side had greater extensor moment/contribution to support during level walking compared to healthy controls and hip of the unaffected side (Yoshida, Mizner, Ramsey, & Snyder-Mackler, 2008). On the other hand, it has been reported that TKA patients increased hip extensor moment and increased hip flexion to perform sit-to-stand task compared with healthy controls (Farquhar, Reisman, & Snyder-Mackler, 2008). Severe CKOA patients did not further increase hip flexion and moment during sit-to-stand.

Only contralateral hip reduced range of hip flexion-extension and only during swing phase of stair ascent and sit-to-stand. (Astephen, Deluzio, Caldwell, & Dunbar, 2008) reported that only severe knee OA patients (34.7 ± 6.2°) had a reduced hip flexion range compared with healthy controls (39.2 ± 4.8°). Severe CKOA patients (37.8 ± 3.3°) also had smaller hip flexion range than healthy controls (45.6 ± 3.3°) in this study, but both studies agreed that no progressive changes of the hip on the affected side during level walking. Contralateral hip increased frontal plane kinematics during stair tasks and had more internal hip rotation during all tested tasks.

The fifth hypothesis that the progression of CKOA will decrease load on the contralateral leg and increased load on the ipsilateral (operated) leg was supported. The results of vertical GRF and total support moment during stair ascent/descent and sit-to-stand provided strong evidence for this hypothesis. However, severe CKOA patients increased total support moment on the contralateral limb during early and late stance of level walking. This was unexpected and may possibly due to abnormal hip moment pattern during level walking.

The sixth hypothesis that the progression of CKOA will affect joint contribution to the dynamic support of the body was also supported. On the ipsilateral (TKA) side, ankle contribution to body support was reduced and hip contribution was increased. With the effect of OA and intervention of TKA, knee contribution to the total support moment during both level walking and stair ascent was lower at pre- and 6 months post-surgery than healthy controls, while hip contribution was increased (Mandeville et al., 2007). This study found that hip contribution on TKA side during stair descent was increased further with the progression of CKOA.

On the contralateral (OA) side, hip contribution was increased while knee contribution was decreased. It has been reported that patients with knee OA reduced knee joint contribution and increased ankle joint contribution during gait compared to healthy controls (Joseph A. Zeni & Higginson, 2011). Based on the results of this study, knee joint contribution was further reduced during stair ascent and sit-to-stand, while hip strategy was used instead of ankle strategy during stair ascent/descent and sit-to-stand.

The last hypothesis that severe CKOA patients had poorer asymmetry than moderate CKOA patients was partially supported. This might be due to the already existed asymmetry in moderated CKOA patients as they tend to rely on the contralateral limb. The progression of CKOA and contralateral knee pain reduced this reliance and severe patients had a better symmetry in vertical GRF during stair ascent/descent. However, ankle/knee/hip kinematic asymmetry was increased in severe CKOA patients.

This study is not without limitations. First, the small sample of patients in both groups, a greater proportion of females than males, and an unequal distribution of the side of the operated knee in each group may biased the findings. Hard inclusion and exclusion criteria associated with CKOA level, and a higher prevalence of knee arthritis in female over male led to a small sample size and an unbalance proportional in gender. The magnitude of most reported significant differences seems to be clinically meaningful. So, the generalization of these findings may still be good.

The second limitation is that the cross-section design of this study and that both CKOA groups had a wide range of times after TKA. In a longitudinal study, it may take up to 10 years for the moderate CKOA to progress to end-stage phase (McMahon & Block, 2003). The cross-section design of this study avoids long following-up time. This study intended to match post-surgery time between groups while allowing a wide range of time.

The third limitation is that healthy controls is about 15 years younger than both CKOA groups. Aging is known to affect joint mechanics (Cofre, Lythgo, Morgan, & Galea, 2011). The initial focus of this study was to identify the changes associated with the progression of CKOA. The younger age of healthy controls did not affect the comparison between the two CKOA groups but might affect the identification of abnormal mechanics. This was the reason why we did not include control group in statistical comparison.

The last limitation is that both groups contained MB and FB TKA (mCKOA, 9MB/4FB; sCKOA, 8MB/5FB) which may potentially confound the comparison. A two factor ANOVA analysis may be needed to exclude the effect of this confounding factor, but the sample size of each subgroup is small and highly unequal. The current design had similar implant designs in each group and this match may have reduced the confounding effect of implant design.

Discussion: BTKA versus UTKA

The asymmetry of BTKA patients

The first hypothesis that BTKA had symmetrical biomechanics was only partially supported, because the second TKA side had a lower hip adduction moment during level walking and stair ascent. The asymmetrical hip frontal plane kinetics was possibly related to the retained biomechanical adaptations of knee OA. A significantly lower hip adduction moment during level walking in early stance on the involved side than on the involved side for patients with medial knee OA. As knee OA progresses, ipsilateral hip adduction moment has been reported to be decreased (Astephen, Deluzio, Caldwell, & Dunbar, 2008). The CKOA study in this dissertation also suggested a decreased ipsilateral hip adduction moment. Hip abductor strength and activation pattern may be responsible for the asymmetry.

BTKA patient had symmetrical sagittal plane biomechanics. Both sides could be treated as a single group for comparison on sagittal plane biomechanics. Only Rossi et al. 2011 have reported that simultaneous BTKA patients achieved symmetrical knee extensor function during isokinetic strength testing.

The asymmetry of UTKA patients

The second hypothesis was supported. Although, UTKA patients restored biomechanical symmetry during level walking except in knee axial rotation, UTKA had asymmetrical ankle/knee kinematics and ankle/knee/hip kinetics during activities that placed more physical demand on limbs than level walking. For kinematics, UTKA-OP side had less ankle dorsiflexion and less knee flexion than UTKA-NOP side. For kinetics, UTKA-NOP side had higher vertical GRF, total support moment, ankle flexion power, knee flexion moment/power, and hip flexion moment than UTKA-OP side. The asymmetries were mainly limited to sagittal plane.

During early stance of stair ascent, UTKA-OP limb less dorsiflexed ankle and decreased knee moment and power generation and relied on ipsilateral hip and contralateral ankle (during its second double support phase) to generate the lacking moment and power to propel center of mass up and forward. The increased plantar-flexion of UTKA-NOP side during early swing is a result of increasing ankle generation power during late stance of stair ascent. The compensation from contralateral ankle to weak knee during stair ascent has also been observed for patients after anterior-cruciate ligament reconstruction (Kowalk, Duncan, McCue, & Vaughan, 1997). Operated knee weakness and asymmetrically lower loading-response knee flexion moment on the operated side during stair ascent following unilateral TKA have also been reported before (Standifird et al., 2016).

During stair descending (“controlled falling”), UTKA-NOP limb had an abrupt initial contact with stair and had first peak vertical GRF on average 22% body weight more than UTKA-OP limb. This was due to deficiency in internal knee extension moment (i.e. external knee flexion moment) and hip flexion moment/power of the UTKA-OP limb (during its second double support phase) to control the “falling” of the center of mass. As a result, UTKA-NOP limb suffered a higher peak external knee flexion moment and ankle dorsiflexion moment during weight acceptance, and UTKA-NOP limb also had an increased ankle power absorption. To absorb the shock, UTKA-NOP knee also tended to have 4.6° more knee flexion excursion during this weight acceptance than UTKA-OP knee (P=.062, ES=.91).

Like early stance of stair descent, UTKA patients shifted weight bearing (on average 12% body weight) to non-operated side during sit-to-stand. As a result, UTKA-OP knee also reduced knee flexion moment and power generation during standing, which was like early stance of stair ascent. However, the compensation of these reduced moment and power of UTKA-OP knee was not limited to contralateral ankle and ipsilateral hip as in stair ascent (ambulation task).

Previous studies have also reported asymmetries in knee adduction moment, quadriceps strength, knee extensor moment, total support, vertical GRF, after UTKA. These asymmetries may be related to the pre-operative asymmetry in loading before TKA and knee extensor strength asymmetry after TKA. Rehabilitation after unilateral TKA may be needed to fix these asymmetry biomechanics.

UTKA patients were shown to have a larger hip extensor contribution to the total support moment of the operated leg during both level walking (Mandeville et al., 2007; Yoshida et al., 2008) and stair ascent (Mandeville et al., 2007) than healthy controls. This study did not compare UTKA to healthy controls but asymmetrical higher hip compensation during sit-to-stand was found. An increased usage of hip to compensate the weakness of operated knee could explain that why about 5% of patients received total hip replacement after TKA.

BTKA vs. UTKA in asymmetry

The third hypothesis that BTKA patients had a lower level of biomechanical asymmetry than UTKA patients was supported, but mainly limited to knee sagittal plane kinetics during sit-to-stand, and stair ascent/descent. UTKA patients had higher loading on NOP knees while BTKA patients had similar loading on both replaced knees.

UTKA-OP and BTKA

BTKA patients had similar function as UTKA patients, except that BTKA patients tended to perform better in stair ascent. This might be due to that a higher peak knee extensor moment and total support moment.

BTKA patients had similar ankle biomechanics, but less extended hip and more anteriorly tilted pelvic than UTKA patients. Anterior tilt is the tilting of pelvic to the front of the body. Anterior pelvic tilt can be caused by chronic muscle imbalances due to increased work in some muscles but decreased work in other muscles. The difference in biomechanical outcomes between unilateral and bilateral TKA patients may be related the difference between the two groups before surgery.

Patients have staged bilateral TKA could expect similar or slight better outcomes of both operated knees as those having unilateral TKA. Rehabilitation after bilateral TKA should also focus on correcting pelvic anterior tilt and possible hip flexion contracture.

Study limitations

One limitation is that both groups contained MB and FB TKA (UTKA, 12MB/3FB; BTKA, 8MB/12FB). UTKA group had more MB patients than BTKA group. A two factor ANOVA analysis may be needed to exclude the effect of this confounding factor. Yet the sample sizes are unbalanced for level combinations, because UTKA-FB combination only has 3 patients. A further comparison that excluded FB patients in both groups revealed similar conclusion.

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Discussion: Mobile- versus Fixed-Bearing Total Knee Arthroplasty

The two hypotheses were accepted. Compared to the FB knees, the MB knees had neither a greater axial rotation nor a lower knee rotation moment during any of the investigated tasks including pivoting. Our results agree with previously observed similar axial rotation between FB and MB knees during walking(Banks & Hodge, 2004; Papagiannis et al., 2016; Zurcher et al., 2014) and stair ascent(Banks & Hodge, 2004). Different from our findings, an earlier study (Fantozzi et al., 2003) found a larger rotation in MB knees than in FB knees during stair ascent. However, the MB knees in that study retained the posterior cruciate ligament (PCL) and the FB knees sacrificed the PCL, while both the FB and MB knees in this study removed cruciate ligaments and were posterior stabilized. Except the mobility of the bearing, the retained PCL and the low conformity of articulating surfaces in a PCL-retaining design may also have contributed to a greater axial rotation in that study. It should be noted that the peak knee flexion angle reached during the stance phase of each investigated task in this study was not as large as the angle (90° or 120°) reached during deep knee bending (Delport et al., 2006; Ranawat et al., 2004). It is acknowledged that including swing phase and other motor tasks into investigation may lead to a different finding (Banks & Hodge, 2004; Zurcher et al., 2014).

The axial rotation of TKA knees was found to be limited only during the two pivoting tasks regardless of the bearing mobility. This contradict the common assumption that TKA knees can’t rotate as much as intact knees during level walking or stair climbing. Finding of normal rotation matched well with one optoelectronic study (Urwin et al., 2014) but contradicted a fluoroscopic study which observed a smaller rotation in TKA knees than in intact knees during walking (D. A. Dennis & Komistek, 2005). Although biomechanics of the operated knee on the transverse plane was normal during stair climbing, the stance duration was prolonged in TKA subjects which indicated poor function in challenging activities such as stair negotiation.

The results of this study also challenge the belief that a MB TKA would promote a closer to normal axial rotation than a FB TKA especially during demanding activities such as pivoting. The MB knees in this study couldn’t rotate more than the FB knees, and both demonstrated a limited axial rotation during the two pivoting tasks. Yet theoretically, the MB knees could rely on the femur-insert (top surface) rotation as a relay of the tibia-insert (under surface) rotation to achieve normal axial rotation in challenging tasks such as spin turn. Two reasons were proposed for this axial rotation reduction in the MB knees during the two pivoting tasks. First, this might be related to a possible decoupling loss or mobility reduction in the MB knees. Although some in-vivo (Delport et al., 2006; Douglas A. Dennis, Komistek, Mahfouz, Outten, & Sharma, 2005) and in-vitro(H. M. McEwen et al., 2001) studies have demonstrated that the mobile insert effectively decouples flexion and axial rotation of the knee joint, i.e. having axial rotation between the insert and the tibial tray (undersurface) and sliding between the femoral component and the insert (top-surface), at least two studies (Garling, Kaptein, Nelissen, & Valstar, 2007; Stiehl, Dennis, Komistek, & Keblish, 1997) showed that the mobility of the mobile insert could be as limited as a fixed-bearing in some MB knees. Secondly, it may be a common compensation strategy or self-protection mechanism for all TKA patients, regardless of bearing mobility, to intentionally reduce knee axial rotation during pivoting.

The observed similarity in peak internal rotation moment matched well with an earlier study (Papagiannis et al., 2016), though a lower mean value (FB, 0.11 Nm/kg; MB, 0.10 Nm/kg) of this parameter was found in this study than what was reported (FB, 0.236 Nm/kg; MB, 0.231 Nm/kg). Similar peak-to-peak rotational moment during walking might also indicate that MB TKA had no advantage over FB TKA regarding implant longevity (Wimmer et al., 2006). The correlation between external rotation and external rotation moment during spin turn suggested that a reduced external rotation moment in the MB knees might account for a reduced rotation angle during early stance (weight acceptance period) of spin turn. On the other hand, the reduced external rotation moment at the early stance phases of walking and spin turn also suggested that a mobile bearing might have lower rotation constraint than a meniscus with restraining structures such as horn attachments and transverse ligament (Guess, Thiagarajan, Kia, & Mishra, 2010).

Care should be taken when interpreting and generalizing the reported results of knee rotation moment in this study. Knee rotation moment was a surrogate measure of torsional load transmitted to the bone-implant interface. Technically, an instrumented knee prosthesis could provide a more direct measure than 3D gait analysis. In-vitro studies using cadavers or composite bones might also provide a direct measure. However, few studies that employed instrumented knee prosthesis have compared rotational moment between FB and MB knees. It has been shown that external flexion moments transmitted to the tibia tray are much lower than the intersegmental knee flexion moments, and the same goes for knee adduction moment (D'Lima, Patil, Steklov, Chien, & Colwell, 2007). It is argued that active and passive soft tissue around the joint absorbs much of the external moments (D'Lima et al., 2007). Similarly, the contribution of soft tissue in absorbing intersegmental rotation moment should never be overlooked. For example, resistance to pure torsional load (rotation moment) at full extension or 30° flexion has been reported to be provided mostly by knee ligaments, which include the medial collateral ligaments, the lateral collateral ligaments, and the cruciate ligaments (Seering, Piziali, Nagel, & Schurman, 1980). The two in-vitro studies, which have demonstrated a reduced load transmitted to proximal tibia in MB knees, did not provide any detail about if and how they simulated the soft tissue restraint in their experiments (Bottlang et al., 2006; Malinzak et al., 2014).

It also needs to be pointed out that soft tissue is not only possible to absorb the external (intersegment) rotation moment during a combined loading situation, but also possible to add rotation torque to the external rotation moment. This is because knee rotation moment is generally lower than knee moments in the sagittal and frontal planes (Kutzner et al., 2010), and balancing these higher external moments might be a top priority for soft tissue. Thus, the resulting torque from soft tissue can either absorb or strengthen the external rotation moment. An earlier study using instrumented knee prosthesis reported that peak external and internal rotation torques at the proximal tibia are -1.1%BWm and 0.53%BWm, respectively (Kutzner et al., 2010). The peak external/internal rotation moments observed in this study, however, are -2%BWm (-1.26 %BW x H) and 0.44%BWm (0.28 %BW x H), respectively. A further comparison of the knee rotation moment during the same motor task between this study and the earlier study (Kutzner et al., 2010) seemed to indicate that intersegmental knee rotation moment might overestimate the net internal rotation moment but underestimate net external rotation moment.

There were several limitations that possibly influenced the results of this study. Firstly, the control subjects were at least 10 years younger than the patients, which was due to the cost and availability of senior subjects. Younger control subjects might perform the tasks with a greater knee rotation angle and a higher rotation moment than age-matched control subjects, which might make the comparison between TKA patient group and controls invalid. Secondly, this study neglected the influence of the contralateral knee by including both unilateral and bilateral TKA patients. Yet a subgroup comparison revealed there was no significant differences between unilateral and bilateral knees in both the MB and FB groups. Thirdly, except the difference in mobility, the FB group also had some knees with a second implant design such that the geometry of the femoral component might slightly differ from those in the MB group. But all the femoral components used symmetric and multi-radius designs. Finally, the accuracy of a skin-based marker set in determining skeletal motion is a common limitation in this type of study. Based on previous validations done on a cadaver model, the accuracy of the adopted method for knee joint rotation was less than 2° in root-mean-square error (Bo Gao & Zheng, 2010). Bone-pin methods would achieve higher accuracy but is invasive to the subject. Radiography method such as single-plane fluoroscopy method can be as accurate as 0.5° to 1° for rotation, but it exposes the subjects to radiation and is technically difficult for pivoting task included in this study.

In conclusion, both FB and MB knees had abnormal transverse plane biomechanics during pivoting. No advantage of MB over FB was found in improving knee rotation or reducing knee rotation moment during the stance phase of the investigated tasks.

Chapter 7 Summary

The objective of this dissertation was to gain insights on biomechanical outcomes following total knee arthroplasty (TKA). The effect of contralateral knee arthritis (CKOA) progression, subgroups (unilateral vs. bilateral) and implant designs (mobile-bearing vs. fixed-bearing) on biomechanical outcomes were investigated in this dissertation.

The Novelties and Strengths

- To our knowledge, this is the first study to describe the effect of the progression of contralateral knee arthritis on other joints in unilateral TKA patients.
- For the first time, the question if staged bilateral TKA patients had symmetrical biomechanical outcomes is investigated.
- For the first time, the question if bilateral TKA patients have similar functional and biomechanical outcomes as unilateral TKA patients is investigated.
- Besides knee joint, this study also investigated ankle and hip biomechanics after TKA.
- Besides level walking, this study investigated physical activities that place more demand.
- For the first time, comparisons of knee rotation during both step and spin turns between mobile-bearing TKA and fixed-bearing TKA are reported.
- For the first time, comparisons of knee rotation moment during activities besides walking between mobile-bearing TKA and fixed-bearing TKA are reported.

Key Points Learned from This Study

CKOA progression affects both limbs (Figure 7.1 and Figure 7.2) and inter-limb asymmetry. TKA could suffer a higher loading due to CKOA progression. Both hips could suffer a higher loading to compensate for weakness in both knees. Before CKOA becomes severe, patients tend to rely on the non-operated leg for weight-bearing. With CKOA progression, gait asymmetry changes. Patients shifted weight bearing from the non-operated leg to the operated leg. Clinical interventions that delay the progression of CKOA may also benefit other joints especially the contralateral ankle joint.

Patients still had reduced ankle/knee kinematics and reduced kinetics on the operated side at 15 months after unilateral TKA. The non-operated side had higher vertical GRF, total support moment, ankle flexion power, knee flexion moment/power, and hip flexion moment than the operated side. Rehabilitation and training programs after TKA are needed to fix these asymmetries.

Staged bilateral TKA patients had symmetrical biomechanics, except a reduced hip adduction moment was still retained on the latest operated side. Staged bilateral TKA patients had a lower level of asymmetry in knee kinetics than unilateral TKA patients. Hip compensations were more evident in unilateral TKA patients. Function-wise, staged bilateral TKA patients tended to climb stairs faster than unilateral TKA patients. Thus, patients with indications for bilateral knee replacement were encouraged to have both knees replaced.

Different from unilateral TKA patients, staged bilateral TKA have abnormal hip flexion contracture and anterior tilted pelvic during standing and level walking. The reason and implication for these kinematic impairments need further investigation. Rehabilitation for bilateral TKA patients should consider such impairments.

Both mobile-bearing and fixed-bearing TKA had similar rotation and rotation moment. The decision to use one design over the other was not supported from current evidences.

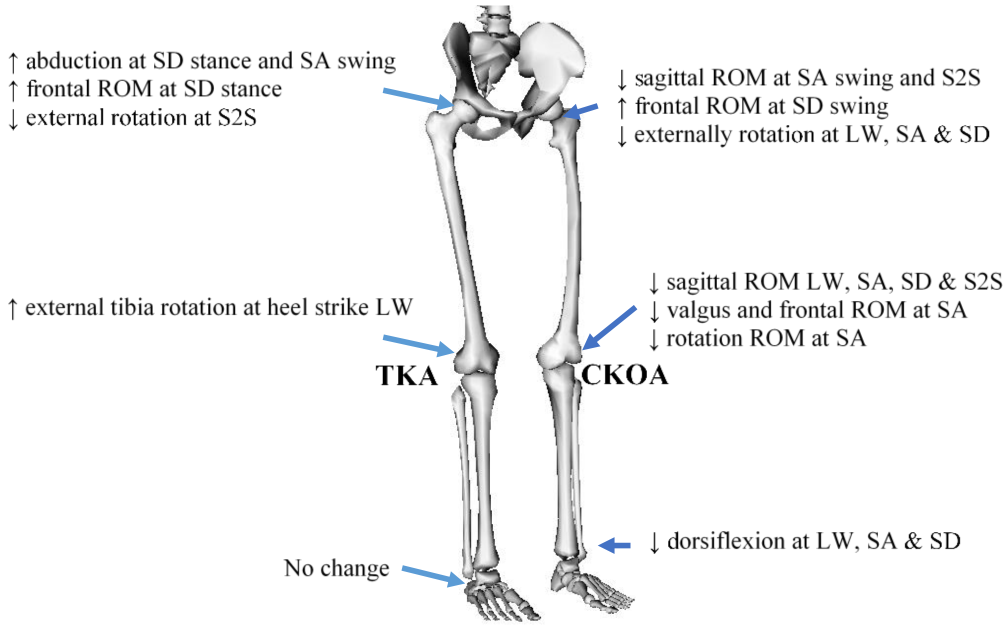

Figure 7.1 Summary of CKOA effect on joint kinematics. SA: stair ascent; LW: level walking; SD: stair descent. S2S: sit-to-stand.

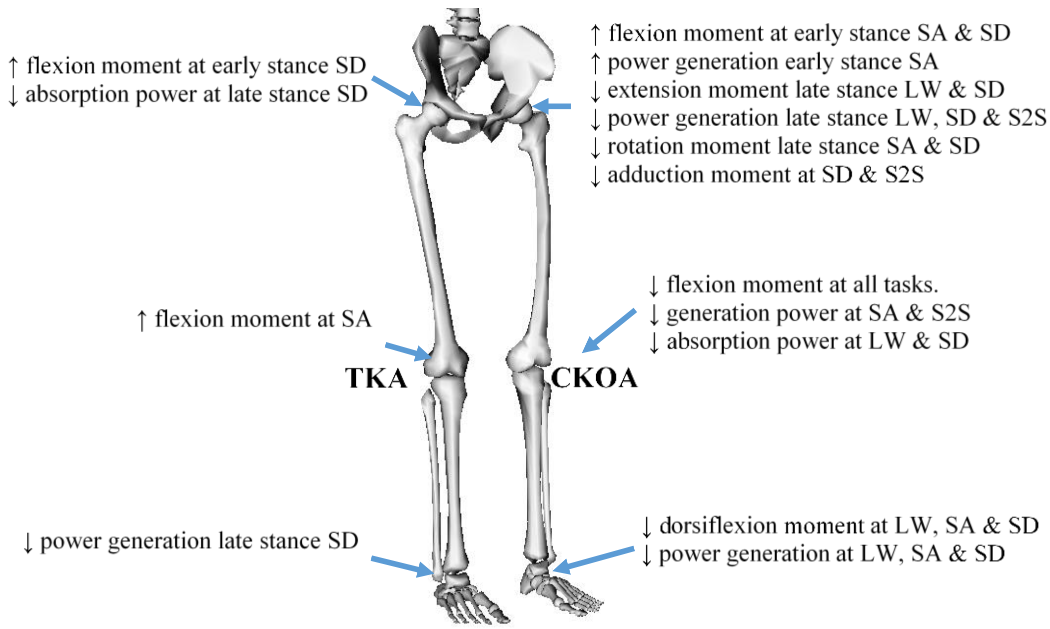

Figure 7.2 Summary of kinetic changes associated with the progression of CKOA. SA: stair ascent; LW: level walking; SD: stair descent. S2S: sit-to-stand.

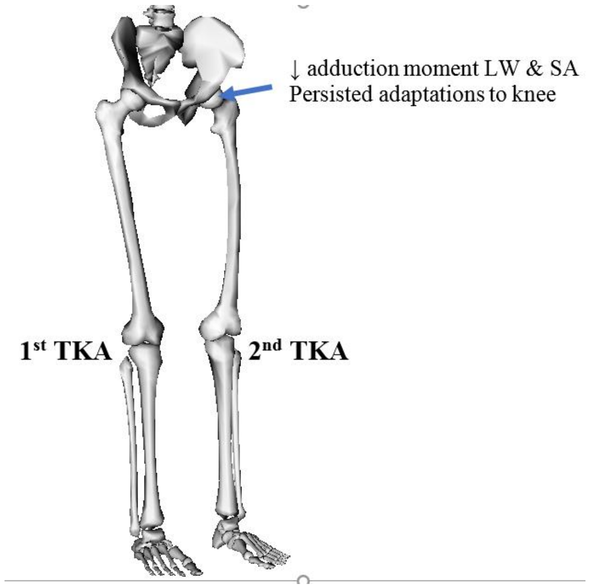

Figure 7.3 biomechanical asymmetry of BTKA patients. Note the side of 1st or 2nd TKA can be left or right.

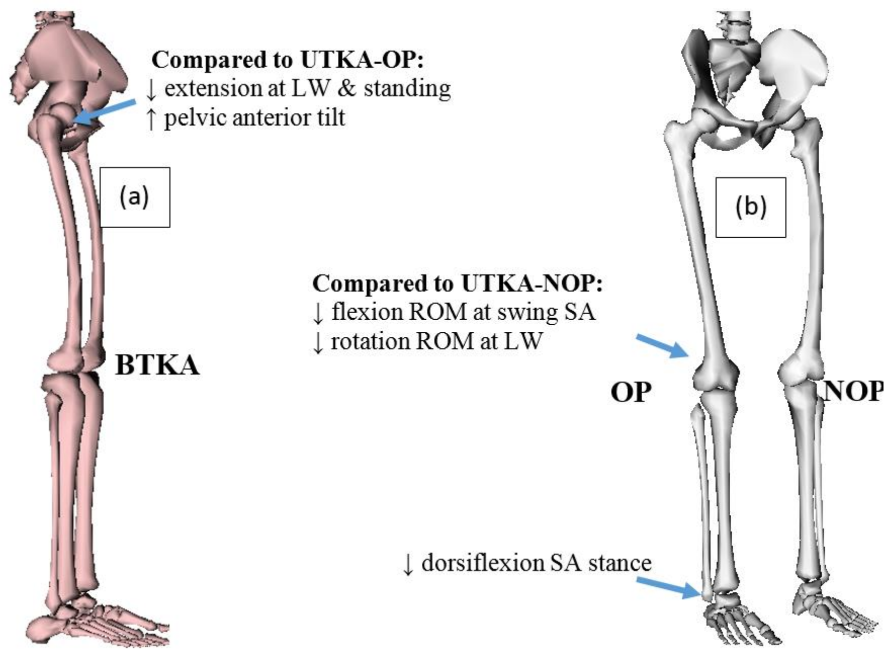

Figure 7.4 Kinematic differences between BTKA and UTKA-OP groups (a) and asymmetry of UTKA patients (b). Note that OP and NOP side can be left or right.

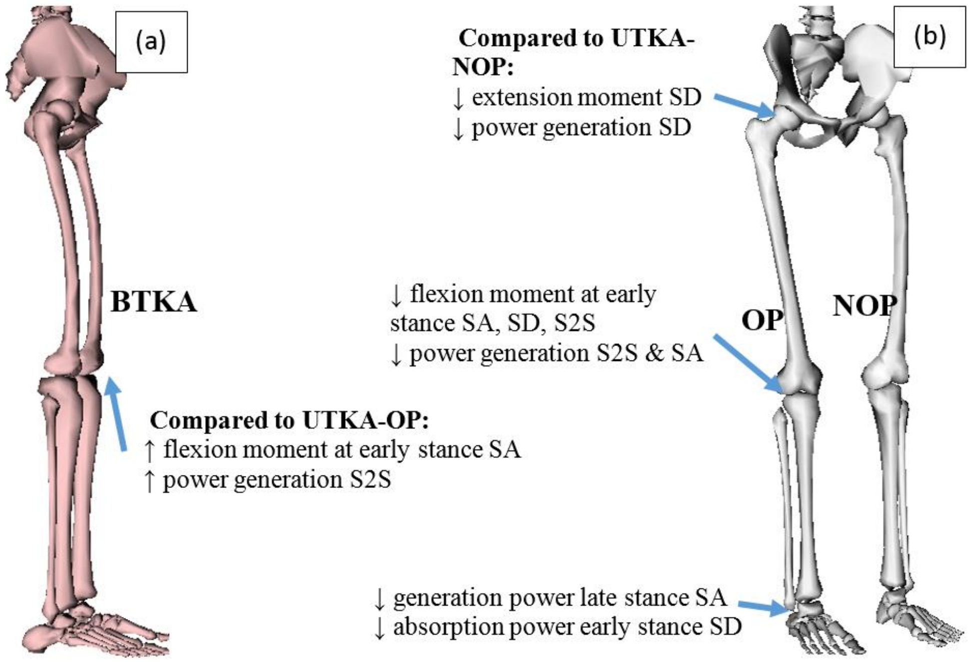

Figure 7.5 differences in moment between UTKA-OP and BTKA (a) and kinetic asymmetry of UTKA patients (b).

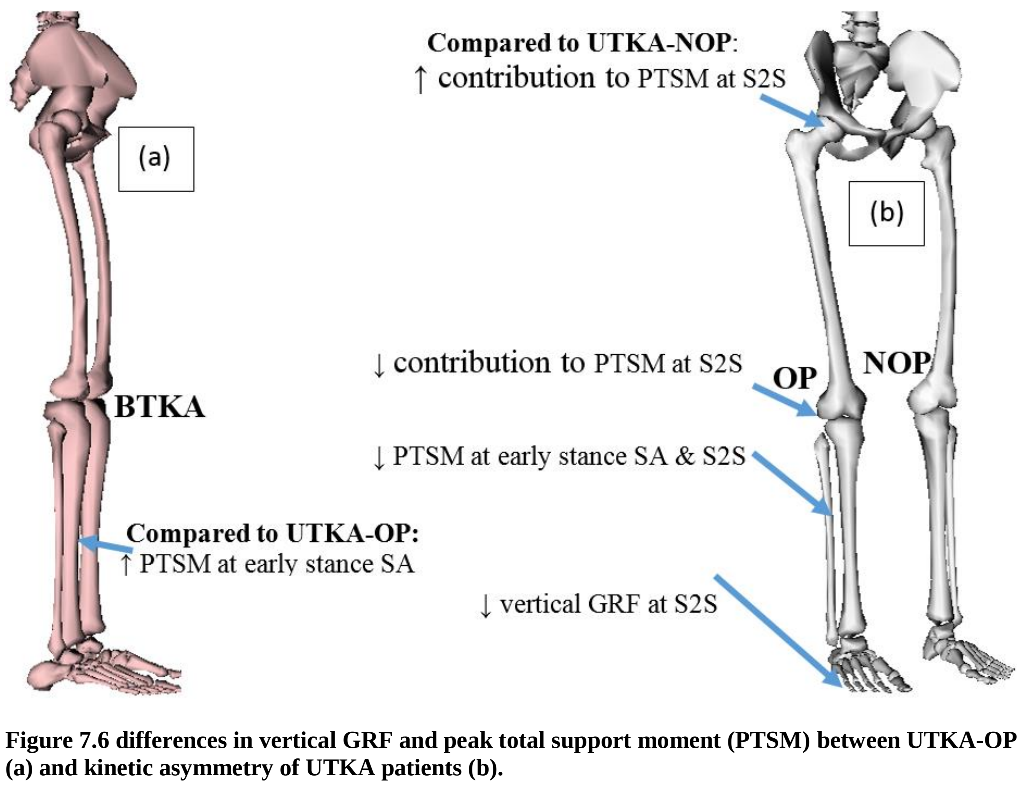

Figure 7.6 differences in vertical GRF and peak total support moment (PTSM) between UTKA-OP and BTKA (a) and kinetic asymmetry of UTKA patients (b).

Limitations and Future Directions

Neuromuscular control of TKA patients has been shown to be different from healthy controls. This study provided only muscle moment information and lacked detailed information on muscle synergy. Future study should include electromyography study and muscle strength tests to obtain muscle outcomes of TKA patients. Obtaining muscle activation time history during movement task may provide insight on any potential muscle co-contraction and failure to activation. For biomechanical study, muscle forces and internal bone-on-bone joint forces could be possible to estimate with the aid of muscle data. For example, a recent study has shown that a threefold quadriceps demand was on the TKA knee compared to the contralateral normal knee (Lester, Shantharam, & Zhang, 2013).

Knee rotation moment is a surrogate measure of rotation moment at bone-implant interface. A direct measure of this torque by instrumented implant or by finite element model that incorporates contribution of muscles, ligaments and articulating contact can provide a more accurate comparison between mobile- and fixed-bearing. However, instrumented implant can be expensive. Finite element modelling needs detailed geometry of implants, material properties, contact properties, muscle forces, anatomical model and experimental verification. Studies that used these two methods to investigated knee rotation angle and moment after TKA have not been reported.

This dissertation is a starting point and used cross-sectional design. A longitudinal follow up of the recruited patients could provide history of outcomes. Radio-graphical studies that evaluate implant migration and bone-mass-density could be performed and linked with biomechanical data. Prediction model could be used to identify modifiable targets that is highly related to poor clinical outcomes.

References

Aldridge Whitehead, J. M., Russell Esposito, E., & Wilken, J. M. (2016). Stair ascent and descent biomechanical adaptations while using a custom ankle-foot orthosis. *J Biomech, 49*(13), 2899-2908. doi:10.1016/j.jbiomech.2016.06.035

Alnahdi, A. H., Zeni, J. A., & Snyder-Mackler, L. (2011). Gait after unilateral total knee arthroplasty: frontal plane analysis. *J Orthop Res, 29*(5), 647-652. doi:10.1002/jor.21323

Alnahdi, A. H., Zeni, J. A., & Snyder-Mackler, L. (2014). Hip abductor strength reliability and association with physical function after unilateral total knee arthroplasty: a cross-sectional study. *Phys Ther, 94*(8), 1154-1162. doi:10.2522/ptj.20130335

Alnahdi, A. H., Zeni, J. A., & Snyder-Mackler, L. (2016). Quadriceps strength asymmetry predicts loading asymmetry during sit-to-stand task in patients with unilateral total knee arthroplasty. *Knee Surg Sports Traumatol Arthrosc, 24*(8), 2587-2594. doi:10.1007/s00167-015-3827-x

Andriacchi, T. P., Mündermann, A., Smith, R. L., Alexander, E. J., Dyrby, C. O., & Koo, S. (2004). A Framework for the in Vivo Pathomechanics of Osteoarthritis at the Knee. *Ann Biomed Eng, 32*(3), 447-457. doi:10.1023/B:ABME.0000017541.82498.37

Astephen, J. L., Deluzio, K. J., Caldwell, G. E., & Dunbar, M. J. (2008). Biomechanical changes at the hip, knee, and ankle joints during gait are associated with knee osteoarthritis severity. *Journal of Orthopaedic Research, 26*(3), 332-341. doi:10.1002/jor.20496

Astephen, J. L., Deluzio, K. J., Caldwell, G. E., Dunbar, M. J., & Hubley-Kozey, C. L. (2008). Gait and neuromuscular pattern changes are associated with differences in knee osteoarthritis severity levels. *J Biomech, 41*(4), 868-876. doi:10.1016/j.jbiomech.2007.10.016

Banks, S. A., & Hodge, W. A. (2004). 2003 Hap Paul Award Paper of the International Society for Technology in Arthroplasty. Design and activity dependence of kinematics in fixed and mobile-bearing knee arthroplasties. *J Arthroplasty, 19*(7), 809-816.

Bell, A. L., Pedersen, D. R., & Brand, R. A. (1990). A Comparison of the Accuracy of Several Hip Center Location Prediction Methods. *J Biomech, 23*(6), 617-621. doi:Doi 10.1016/0021-9290(90)90054-7

Berman, A. T., Zarro, V. J., Bosacco, S. J., & Israelite, C. (1987). Quantitative gait analysis after unilateral or bilateral total knee replacement. *J Bone Joint Surg Am, 69*(9), 1340-1345.

Bohm, E. R., Molodianovitsh, K., Dragan, A., Zhu, N., Webster, G., Masri, B., . . . Dunbar, M. (2016). Outcomes of unilateral and bilateral total knee arthroplasty in 238,373 patients. *Acta Orthop, 87 Suppl 1*, 24-30. doi:10.1080/17453674.2016.1181817

Bottlang, M., Erne, O. K., Lacatusu, E., Sommers, M. B., & Kessler, O. (2006). A mobile-bearing knee prosthesis can reduce strain at the proximal tibia. *Clin Orthop Relat Res, 447*, 105-111. doi:10.1097/01.blo.0000203463.27937.97

Bottros, J., Gad, B., Krebs, V., & Barsoum, W. K. (2006). Gap Balancing in Total Knee Arthroplasty. *J Arthroplasty, 21*(4), 11-15. doi:10.1016/j.arth.2006.02.084

Callaghan, J. J., Insall, J. N., Greenwald, A. S., Dennis, D. A., Komistek, R. D., Murray, D. W., . . . Dorr, L. D. (2001). Mobile-bearing knee replacement: concepts and results. *Instr Course Lect, 50*, 431-449.

Chang, A., Hayes, K., Dunlop, D., Song, J., Hurwitz, D., Cahue, S., & Sharma, L. (2005). Hip abduction moment and protection against medial tibiofemoral osteoarthritis progression. *Arthritis Rheum, 52*(11), 3515-3519. doi:10.1002/art.21406

Chouteau, J., Lerat, J. L., Testa, R., Moyen, B., & Banks, S. A. (2009). Sagittal laxity after posterior cruciate ligament-retaining mobile-bearing total knee arthroplasty. *J Arthroplasty, 24*(5), 710-715. doi:10.1016/j.arth.2008.05.028

Christensen, J. C., Mizner, R. L., Foreman, K. B., Marcus, R. L., Pelt, C. E., & LaStayo, P. C. (2018). Quadriceps weakness preferentially predicts detrimental gait compensations among common impairments after total knee arthroplasty. *Journal of Orthopaedic Research, 36*(9), 2355-2363. doi:doi:10.1002/jor.23894

Cofre, L. E., Lythgo, N., Morgan, D., & Galea, M. P. (2011). Aging modifies joint power and work when gait speeds are matched. *Gait Posture, 33*(3), 484-489. doi:10.1016/j.gaitpost.2010.12.030

Cram, P., Lu, X., Kates, S. L., Singh, J. A., Li, Y., & Wolf, B. R. (2012). Total knee arthroplasty volume, utilization, and outcomes among medicare beneficiaries, 1991-2010. *JAMA, 308*(12), 1227-1236. doi:10.1001/2012.jama.11153

Creaby, M. W., Bennell, K. L., & Hunt, M. A. (2012). Gait Differs Between Unilateral and Bilateral Knee Osteoarthritis. *Arch Phys Med Rehabil, 93*(5), 822-827. doi:https://doi.org/10.1016/j.apmr.2011.11.029

D'Lima, D. D., Patil, S., Steklov, N., Chien, S., & Colwell, C. W., Jr. (2007). In vivo knee moments and shear after total knee arthroplasty. *J Biomech, 40 Suppl 1*, S11-17. doi:10.1016/j.jbiomech.2007.03.004

De Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *J Biomech, 29*(9), 1223-1230.

Delport, H. P., Banks, S. A., De Schepper, J., & Bellemans, J. (2006). A kinematic comparison of fixed- and mobile-bearing knee replacements. *J Bone Joint Surg Br, 88*(8), 1016-1021. doi:10.1302/0301-620X.88B8.17529

Dennis, D. A., & Komistek, R. D. (2005). Kinematics of mobile-bearing total knee arthroplasty. *Instr Course Lect, 54*, 207-220.

Dennis, D. A., Komistek, R. D., Mahfouz, M. R., Outten, J. T., & Sharma, A. (2005). Mobile-Bearing Total Knee Arthroplasty. *Clin Orthop Relat Res, 440*(&NA;), 88-95. doi:10.1097/01.blo.0000185464.23505.6e

Dillon, C. F., Rasch, E. K., Gu, Q., & Hirsch, R. (2006). Prevalence of knee osteoarthritis in the United States: arthritis data from the Third National Health and Nutrition Examination Survey 1991-94. *J Rheumatol, 33*(11), 2271-2279.

Emrani, P. S., Katz, J. N., Kessler, C. L., Reichmann, W. M., Wright, E. A., McAlindon, T. E., & Losina, E. (2008). Joint Space Narrowing and Kellgren-Lawrence Progression in Knee Osteoarthritis: An Analytic Literature Synthesis. *Osteoarthritis and cartilage / OARS, Osteoarthritis Research Society, 16*(8), 873-882. doi:10.1016/j.joca.2007.12.004

Fantozzi, S., Benedetti, M. G., Leardini, A., Banks, S. A., Cappello, A., Assirelli, D., & Catani, F. (2003). Fluoroscopic and gait analysis of the functional performance in stair ascent of two total knee replacement designs. *Gait Posture, 17*(3), 225-234.

Farquhar, S. J., Reisman, D. S., & Snyder-Mackler, L. (2008). Persistence of Altered Movement Patterns During a Sit-to-Stand Task 1 Year Following Unilateral Total Knee Arthroplasty. *Physical Therapy, 88*(5), 567-579. doi:10.2522/ptj.20070045

Felson, D. T. (1990). The epidemiology of knee osteoarthritis: Results from the framingham osteoarthritis study. *Seminars in Arthritis and Rheumatism, 20*(3, Supplement 1), 42-50. doi:https://doi.org/10.1016/0049-0172(90)90046-I

Fox, A. J. S., Bedi, A., & Rodeo, S. A. (2012). The Basic Science of Human Knee Menisci: Structure, Composition, and Function. *Sports Health, 4*(4), 340-351. doi:10.1177/1941738111429419

Gao, B., Cordova, M. L., & Zheng, N. N. (2012). Three-dimensional joint kinematics of ACL-deficient and ACL-reconstructed knees during stair ascent and descent. *Hum Mov Sci, 31*(1), 222-235. doi:10.1016/j.humov.2011.04.009

Gao, B., & Zheng, N. N. (2010). Alterations in three-dimensional joint kinematics of anterior cruciate ligament-deficient and-reconstructed knees during walking. *Clinical Biomechanics, 25*(3), 222-229.

Garling, E. H., Kaptein, B. L., Nelissen, R. G., & Valstar, E. R. (2007). Limited rotation of the mobile-bearing in a rotating platform total knee prosthesis. *J Biomech, 40 Suppl 1*, S25-30. doi:10.1016/j.jbiomech.2007.02.013

Glaister, B. C., Bernatz, G. C., Klute, G. K., & Orendurff, M. S. (2007). Video task analysis of turning during activities of daily living. *Gait Posture, 25*(2), 289-294. doi:10.1016/j.gaitpost.2006.04.003

Guess, T. M., Thiagarajan, G., Kia, M., & Mishra, M. (2010). A subject specific multibody model of the knee with menisci. *Med Eng Phys, 32*(5), 505-515. doi:http://dx.doi.org/10.1016/j.medengphy.2010.02.020

Gundry, M., Hopkins, S., & Knapp, K. (2017). A Review on Bone Mineral Density Loss in Total Knee Replacements Leading to Increased Fracture Risk. *Clinical Reviews in Bone and Mineral Metabolism, 15*(4), 162-174. doi:10.1007/s12018-017-9238-4

Gunther, K. P., Sturmer, T., Sauerland, S., Zeissig, I., Sun, Y., Kessler, S., . . . Puhl, W. (1998). Prevalence of generalised osteoarthritis in patients with advanced hip and knee osteoarthritis: the Ulm Osteoarthritis Study. *Ann Rheum Dis, 57*(12), 717-723. doi:10.1136/ard.57.12.717

Hannan, M. T., Felson, D. T., & Pincus, T. (2000). Analysis of the discordance between radiographic changes and knee pain in osteoarthritis of the knee. *J Rheumatol, 27*(6), 1513-1517.

Harato, K., Nagura, T., Matsumoto, H., Otani, T., Toyama, Y., & Suda, Y. (2010). Extension limitation in standing affects weight-bearing asymmetry after unilateral total knee arthroplasty. *J Arthroplasty, 25*(2), 225-229. doi:10.1016/j.arth.2009.02.003

Henricson, A., Dalen, T., & Nilsson, K. G. (2006). Mobile bearings do not improve fixation in cemented total knee arthroplasty. *Clin Orthop Relat Res, 448*, 114-121. doi:10.1097/01.blo.0000224004.40883.ab

Hilding, M. B., Lanshammar, H., & Ryd, L. (1996). Knee joint loading and tibial component loosening. RSA and gait analysis in 45 osteoarthritic patients before and after TKA. *J Bone Joint Surg Br, 78*(1), 66-73.

Hoshino, A., & Wallace, W. A. (1987). Impact-absorbing properties of the human knee. *J Bone Joint Surg Br, 69*(5), 807-811.

Johnson, T. S., Laurent, M. P., Yao, J. Q., & Gilbertson, L. N. (2001). The effect of displacement control input parameters on tibiofemoral prosthetic knee wear. *Wear, 250*(1), 222-226. doi:https://doi.org/10.1016/S0043-1648(01)00650-0

Jones, R. K., Chapman, G. J., Findlow, A. H., Forsythe, L., Parkes, M. J., Sultan, J., & Felson, D. T. (2013). A new approach to prevention of knee osteoarthritis: reducing medial load in the contralateral knee. *J Rheumatol, 40*(3), 309-315. doi:10.3899/jrheum.120589

Judd, D. L., Eckhoff, D. G., & Stevens-Lapsley, J. E. (2012). Muscle strength loss in the lower limb after total knee arthroplasty. *Am J Phys Med Rehabil, 91*(3), 220-230. doi:10.1097/PHM.0b013e3182411e49

Kean, C. O., Hinman, R. S., Bowles, K. A., Cicuttini, F., Davies-Tuck, M., & Bennell, K. L. (2012). Comparison of peak knee adduction moment and knee adduction moment impulse in distinguishing between severities of knee osteoarthritis. *Clin Biomech (Bristol, Avon), 27*(5), 520-523. doi:10.1016/j.clinbiomech.2011.12.007

Kim, S. J., Bamne, A., Song, Y. D., Kang, Y. G., & Kim, T. K. (2015). Patients still wish for key improvements after total knee arthroplasty. *Knee Surg Relat Res, 27*(1), 24-33. doi:10.5792/ksrr.2015.27.1.24

Kowalk, D. L., Duncan, J. A., McCue, F. C., 3rd, & Vaughan, C. L. (1997). Anterior cruciate ligament reconstruction and joint dynamics during stair climbing. *Med Sci Sports Exerc, 29*(11), 1406-1413.

Kutzner, I., Heinlein, B., Graichen, F., Bender, A., Rohlmann, A., Halder, A., . . . Bergmann, G. (2010). Loading of the knee joint during activities of daily living measured in vivo in five subjects. *J Biomech, 43*(11), 2164-2173. doi:http://dx.doi.org/10.1016/j.jbiomech.2010.03.046

LaCour, M. T., Sharma, A., Carr, C. B., Komistek, R. D., & Dennis, D. A. (2014). Confirmation of long-term in vivo bearing mobility in eight rotating-platform TKAs. *Clinical Orthopaedics and Related Research®, 472*(9), 2766-2773.

Lester, D. K., Shantharam, R., & Zhang, K. (2013). Dynamic electromyography after cruciate-retaining total knee arthroplasty revealed a threefold quadriceps demand compared with the contralateral normal knee. *J Arthroplasty, 28*(4), 557-562. doi:10.1016/j.arth.2012.08.024

Loyd, B. J., Jennings, J. M., Judd, D. L., Kim, R. H., Wolfe, P., Dennis, D. A., & Stevens-Lapsley, J. E. (2017). Influence of Hip Abductor Strength on Functional Outcomes Before and After Total Knee Arthroplasty: Post Hoc Analysis of a Randomized Controlled Trial. *Physical Therapy, 97*(9), 896-903. doi:10.1093/ptj/pzx066

Malinzak, R. A., Small, S. R., Rogge, R. D., Archer, D. B., Oja, J. W., Berend, M. E., & Ritter, M. A. (2014). The effect of rotating platform TKA on strain distribution and torque transmission on the proximal tibia. *J Arthroplasty, 29*(3), 541-547. doi:10.1016/j.arth.2013.08.024

Mandeville, D., Osternig, L. R., & Chou, L. S. (2007). The effect of total knee replacement on dynamic support of the body during walking and stair ascent. *Clinical Biomechanics, 22*(7), 787-794. doi:DOI 10.1016/j.clinbiomech.2007.04.002

Maradit Kremers, H., Larson, D. R., Crowson, C. S., Kremers, W. K., Washington, R. E., Steiner, C. A., . . . Berry, D. J. (2015). Prevalence of Total Hip and Knee Replacement in the United States. *J Bone Joint Surg Am, 97*(17), 1386-1397. doi:10.2106/jbjs.n.01141

McClelland, J. A., Webster, K. E., & Feller, J. A. (2007). Gait analysis of patients following total knee replacement: a systematic review. *Knee, 14*(4), 253-263. doi:10.1016/j.knee.2007.04.003

McEwen, H. M., Fisher, J., Goldsmith, A. A., Auger, D. D., Hardaker, C., & Stone, M. H. (2001). Wear of fixed bearing and rotating platform mobile bearing knees subjected to high levels of internal and external tibial rotation. *J Mater Sci Mater Med, 12*(10-12), 1049-1052.

McEwen, H. M. J., Barnett, P. I., Bell, C. J., Farrar, R., Auger, D. D., Stone, M. H., & Fisher, J. (2005). The influence of design, materials and kinematics on the in vitro wear of total knee replacements. *J Biomech, 38*(2), 357-365. doi:https://doi.org/10.1016/j.jbiomech.2004.02.015

McMahon, M., & Block, J. A. (2003). The risk of contralateral total knee arthroplasty after knee replacement for osteoarthritis. *J Rheumatol, 30*(8), 1822-1824.

Memtsoudis, S. G., Hargett, M., Russell, L. A., Parvizi, J., Cats-Baril, W. L., Stundner, O., & Sculco, T. P. (2013). Consensus statement from the consensus conference on bilateral total knee arthroplasty group. *Clin Orthop Relat Res, 471*(8), 2649-2657. doi:10.1007/s11999-013-2976-9

Messier, S. P., Beavers, D. P., Herman, C., Hunter, D. J., & DeVita, P. (2016). Are unilateral and bilateral knee osteoarthritis patients unique subsets of knee osteoarthritis? A biomechanical perspective. *Osteoarthritis Cartilage, 24*(5), 807-813. doi:https://doi.org/10.1016/j.joca.2015.12.005

Mills, K., Hettinga, B. A., Pohl, M. B., & Ferber, R. (2013). Between-limb kinematic asymmetry during gait in unilateral and bilateral mild to moderate knee osteoarthritis. *Arch Phys Med Rehabil, 94*(11), 2241-2247. doi:10.1016/j.apmr.2013.05.010

Mills, K., Hunt, M. A., & Ferber, R. (2013). Biomechanical deviations during level walking associated with knee osteoarthritis: a systematic review and meta-analysis. *Arthritis Care Res (Hoboken), 65*(10), 1643-1665. doi:10.1002/acr.22015

Milner, C. E. (2009). Is gait normal after total knee arthroplasty? Systematic review of the literature. *J Orthop Sci, 14*(1), 114-120. doi:10.1007/s00776-008-1285-8

Milner, C. E., & O'Bryan, M. E. (2008). Bilateral Frontal Plane Mechanics After Unilateral Total Knee Arthroplasty. *Arch Phys Med Rehabil, 89*(10), 1965-1969. doi:http://dx.doi.org/10.1016/j.apmr.2008.02.034

Mizner, R. L., Petterson, S. C., Clements, K. E., Zeni, J. A., Irrgang, J., & Snyder-Mackler, L. (2011). Measuring Functional Improvement after Total Knee Arthroplasty Requires both Performance-Based and Patient-Report Assessments: A Longitudinal Analysis of Outcomes. *J Arthroplasty, 26*(5), 728-737. doi:10.1016/j.arth.2010.06.004

Mizner, R. L., Petterson, S. C., Stevens, J. E., Vandenborne, K., & Snyder-Mackler, L. (2005). Early Quadriceps Strength Loss After Total Knee Arthroplasty: The Contributions of Muscle Atrophy and Failure of Voluntary Muscle Activation. *J Bone Joint Surg Am, 87*(5), 1047-1053. doi:10.2106/JBJS.D.01992

Morgenroth, D. C., Medverd, J. R., Seyedali, M., & Czerniecki, J. M. (2014). The Relationship between Knee Joint Loading Rate During Walking and Degenerative Changes on Magnetic Resonance Imaging. *Clin Biomech (Bristol, Avon), 29*(6), 664-670. doi:10.1016/j.clinbiomech.2014.04.008

Murakami, K., Hamai, S., Okazaki, K., Wang, Y., Ikebe, S., Higaki, H., . . . Nakashima, Y. (2018). In vivo kinematics of gait in posterior-stabilized and bicruciate-stabilized total knee arthroplasties using image-matching techniques. *Int Orthop*. doi:10.1007/s00264-018-3921-z

Murphy, L., Schwartz, T. A., Helmick, C. G., Renner, J. B., Tudor, G., Koch, G., . . . Jordan, J. M. (2008). Lifetime risk of symptomatic knee osteoarthritis. *Arthritis Care Res (Hoboken), 59*(9), 1207-1213. doi:doi:10.1002/art.24021

Okada, Y., Teramoto, A., Takagi, T., Yamakawa, S., Sakakibara, Y., Shoji, H., . . . Yamashita, T. (2018). ACL Function in Bicruciate-Retaining Total Knee Arthroplasty. *J Bone Joint Surg Am, 100*(17), e114. doi:10.2106/jbjs.18.00099

Okamoto, N., Nakamura, E., Nishioka, H., Karasugi, T., Okada, T., & Mizuta, H. (2014). In vivo kinematic comparison between mobile-bearing and fixed-bearing total knee arthroplasty during step-up activity. *J Arthroplasty, 29*(12), 2393-2396. doi:10.1016/j.arth.2014.02.022

Papagiannis, G. I., Roumpelakis, I. M., Triantafyllou, A. I., Makris, I. N., & Babis, G. C. (2016). No Differences Identified in Transverse Plane Biomechanics Between Medial Pivot and Rotating Platform Total Knee Implant Designs. *J Arthroplasty, 31*(8), 1814-1820. doi:10.1016/j.arth.2016.01.050

Ranawat, C. S., Komistek, R. D., Rodriguez, J. A., Dennis, D. A., & Anderle, M. (2004). In vivo kinematics for fixed and mobile-bearing posterior stabilized knee prostheses. *Clin Orthop Relat Res*(418), 184-190.

Ritter, M. A., Lutgring, J. D., Davis, K. E., & Berend, M. E. (2008). The effect of postoperative range of motion on functional activities after posterior cruciate-retaining total knee arthroplasty. *J Bone Joint Surg Am, 90*(4), 777-784. doi:10.2106/jbjs.f.01022

Ro, D. H., Han, H. S., Lee, D. Y., Kim, S. H., Kwak, Y. H., & Lee, M. C. (2018). Slow gait speed after bilateral total knee arthroplasty is associated with suboptimal improvement of knee biomechanics. *Knee Surg Sports Traumatol Arthrosc, 26*(6), 1671-1680. doi:10.1007/s00167-017-4682-8

Roos, E. M., & Toksvig-Larsen, S. (2003). Knee injury and Osteoarthritis Outcome Score (KOOS) – validation and comparison to the WOMAC in total knee replacement. *Health and Quality of Life Outcomes, 1*(1), 17. doi:10.1186/1477-7525-1-17

Ruiz, D. J., Koenig, L., Dall, T. M., Gallo, P., Narzikul, A., Parvizi, J., & Tongue, J. (2013). The Direct and Indirect Costs to Society of Treatment for End-Stage Knee Osteoarthritis. *JBJS, 95*(16), 1473-1480. doi:10.2106/jbjs.l.01488

Russo, A., Montagna, L., Bragonzoni, L., Zampagni, M. L., & Marcacci, M. (2005). Fixation of total knee arthroplasty improved by mobile-bearing design. *Clin Orthop Relat Res*(437), 186-195.

Sanders, T. L., Maradit Kremers, H., Schleck, C. D., Larson, D. R., & Berry, D. J. (2017). Subsequent Total Joint Arthroplasty After Primary Total Knee or Hip Arthroplasty: A 40-Year Population-Based Study. *J Bone Joint Surg Am, 99*(5), 396-401. doi:10.2106/jbjs.16.00499

Schulze, A., & Scharf, H. P. (2013). [Satisfaction after total knee arthroplasty. Comparison of 1990-1999 with 2000-2012]. *Orthopade, 42*(10), 858-865. doi:10.1007/s00132-013-2117-x

Seering, W. P., Piziali, R. L., Nagel, D. A., & Schurman, D. J. (1980). The function of the primary ligaments of the knee in varus-valgus and axial rotation. *J Biomech, 13*(9), 785-794. doi:http://dx.doi.org/10.1016/0021-9290(80)90240-7

Shi, K., Hayashida, K., Umeda, N., Yamamoto, K., & Kawai, H. (2008). Kinematic comparison between mobile-bearing and fixed-bearing inserts in NexGen legacy posterior stabilized flex total knee arthroplasty. *J Arthroplasty, 23*(2), 164-169. doi:10.1016/j.arth.2007.01.009

Shi, X., Shen, B., Yang, J., Kang, P., Zhou, Z., & Pei, F. (2014). In vivo kinematics comparison of fixed- and mobile-bearing total knee arthroplasty during deep knee bending motion. *Knee Surg Sports Traumatol Arthrosc, 22*(7), 1612-1618. doi:10.1007/s00167-012-2333-7

Sophia Fox, A. J., Bedi, A., & Rodeo, S. A. (2009). The Basic Science of Articular Cartilage: Structure, Composition, and Function. *Sports Health, 1*(6), 461-468. doi:10.1177/1941738109350438

Sosdian, L., Dobson, F., Wrigley, T. V., Paterson, K., Bennell, K., Dowsey, M., . . . Hinman, R. S. (2014). Longitudinal changes in knee kinematics and moments following knee arthroplasty: a systematic review. *Knee, 21*(6), 994-1008. doi:10.1016/j.knee.2014.09.009

Spoor, C. W., & Veldpaus, F. E. (1980). Rigid Body Motion Calculated from Spatial Coordinates of Markers. *J Biomech, 13*(4), 391-393. doi:Doi 10.1016/0021-9290(80)90020-2

Standifird, T. W., Saxton, A. M., Coe, D. P., Cates, H. E., Reinbolt, J. A., & Zhang, S. (2016). Influence of Total Knee Arthroplasty on Gait Mechanics of the Replaced and Non-Replaced Limb During Stair Negotiation. *J Arthroplasty, 31*(1), 278-283. doi:10.1016/j.arth.2015.06.052

Stiehl, J. B., Dennis, D. A., Komistek, R. D., & Keblish, P. A. (1997). In vivo kinematic analysis of a mobile bearing total knee prosthesis. *Clin Orthop, 345*. doi:10.1097/00003086-199712000-00010

Tanavalee, A., Thiengwittayaporn, S., Ngarmukos, S., & Siddhiphongse, B. (2004). Contralateral total knee arthroplasty after unilateral surgery in bilateral varus gonathrosis. *J Med Assoc Thai, 87*(8), 902-909.

Urwin, S. G., Kader, D. F., Caplan, N., St Clair Gibson, A., & Stewart, S. (2014). Gait analysis of fixed bearing and mobile bearing total knee prostheses during walking: do mobile bearings offer functional advantages? *Knee, 21*(2), 391-395. doi:10.1016/j.knee.2013.10.007

Wang, H., Fleischli, J. E., & Zheng, N. N. (2013). Transtibial versus anteromedial portal technique in single-bundle anterior cruciate ligament reconstruction outcomes of knee joint kinematics during walking. *Am J Sports Med, 41*(8), 1847-1856.

Wang, H., & Zheng, N. (2010). Knee rotation and loading during spin and step turn. *Int J Sports Med, 31*(10), 742-746. doi:10.1055/s-0030-1261942

Wang, H., & Zheng, N. N. (2010). Knee joint secondary motion accuracy improved by quaternion-based optimizer with bony landmark constraints. *J Biomech Eng, 132*(12), 124502. doi:10.1115/1.4002856

Wilson, D. R., Mc Walter, E. J., & Johnston, J. D. (2009). The Measurement of Joint Mechanics and Their Role in Osteoarthritis Genesis and Progression. *Medical Clinics of North America, 93*(1), 67-82. doi:https://doi.org/10.1016/j.mcna.2008.08.004

Wimmer, M. A., Schwenke, T., Salineros, M., & Andriacchi, T. P. (2006). Can early post-op gait predict implant longevity? *J Biomech, 39*, S115. doi:10.1016/s0021-9290(06)83361-0

Winter, D. A. (1991). *Biomechanics and motor control of human gait: normal, elderly and pathological*.

Wolterbeek, N., Garling, E. H., Mertens, B. J., Nelissen, R. G., & Valstar, E. R. (2012). Kinematics and early migration in single-radius mobile- and fixed-bearing total knee prostheses. *Clin Biomech (Bristol, Avon), 27*(4), 398-402. doi:10.1016/j.clinbiomech.2011.10.013

Yoshida, Y., Mizner, R. L., Ramsey, D. K., & Snyder-Mackler, L. (2008). Examining outcomes from total knee arthroplasty and the relationship between quadriceps strength and knee function over time. *Clin Biomech (Bristol, Avon), 23*(3), 320-328. doi:10.1016/j.clinbiomech.2007.10.008

Zeni, J. A., & Higginson, J. S. (2009). Differences in gait parameters between healthy subjects and persons with moderate and severe knee osteoarthritis: a result of altered walking speed? *Clinical Biomechanics, 24*(4), 372-378.

Zeni, J. A., & Higginson, J. S. (2011). Knee Osteoarthritis Affects the Distribution of Joint Moments During Gait. *Knee, 18*(3), 156-159. doi:10.1016/j.knee.2010.04.003

Zeni, J. A., Jr., & Snyder-Mackler, L. (2010). Clinical outcomes after simultaneous bilateral total knee arthroplasty: comparison to unilateral total knee arthroplasty and healthy controls. *J Arthroplasty, 25*(4), 541-546. doi:10.1016/j.arth.2009.02.016

Zurcher, A. W., van Hutten, K., Harlaar, J., Terwee, C. B., Rob Albers, G. H., & Poll, R. G. (2014). Mobile-bearing total knee arthroplasty: More rotation is evident during more demanding tasks. *Knee*, *21*(5), 960-963. doi:10.1016/j.knee.2014.05.007