

# **Does Total Knee Arthroplasty Reproduce Natural Knee Mechanics?**

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## **NOTES**

This thesis is assembled in article format with two independent articles. The first section of this thesis focuses on a general introduction to total knee arthroplasty, including a literature review followed by the methods used to conduct this study. The first article, entitled “*Gait dynamics on level and inclined surfaces following total knee arthroplasty*” reflects the gait analysis portion of this study. The second article, entitled, “*Biomechanical Analysis of Sit-Stand Tasks Following Total Knee Arthroplasty*” focuses on motion analysis for the sit-stand task. The final discussion encompasses a general discussion which briefly summarizes findings from each of the individual articles.

## **ACKNOWLEDGEMENTS**

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## **ABSTRACT**

As the number of total knee arthroplasty (TKA) procedures increases annually, the patient demographic is shifting to include younger patients with higher expectations for post-operative function. The aim of this study was to compare movement patterns during activities of daily living among TKA patients and a healthy, age-matched group using 3D motion analysis. Specifically, this analysis looked at walking on level and inclined surfaces, as well as sitting up and down from a chair. It was predicted that (1) TKA patients would exhibit reduced knee extension moments at the operated limb and increased adduction moments at the contralateral limb during gait, (2) walking downhill would result in greater differences between TKA and control groups, compared to level walking, and (3) TKA participants would have greater flexion angles, moments and power values at the hip, compared to controls, during the sit-stand tasks. Seventeen participants (age=62±6 years, BMI=30±3 kg/m<sup>2</sup>, time after surgery=11±5 months) were recruited from the Ottawa Hospital, having undergone unilateral TKA by the same surgeon. An age-matched control group was composed of 17 individuals (age=63±8 years, BMI=27±4 kg/m<sup>2</sup>) who were recruited from the local community. Three dimensional (3D) biomechanical assessment was conducted with all participants performing five trials of walking on level and inclined surfaces, stair ascent and descent as well as sit-stand tasks. Results from this study were focused on gait and sit-stand transitions, showing that TKA participants exhibited altered gait patterns on both walking surfaces, with significantly smaller knee flexion angles and moments, as well as reduced peak power at the knee. The TKA group also experienced reduced knee extension moments; however, this was only significant for downhill walking. Consistent with our hypothesis, downhill walking resulted in greater discrepancies between the groups compared to level walking. Contrary to our third hypothesis, TKA participants exhibited significantly smaller

peak hip flexion angles and moments during the sit-stand task, along with reduced hip abduction angles and knee abduction moments. The reduced knee flexion kinematics and kinetics observed during gait tasks, combined with the differences in frontal plane mechanics observed during the sit-stand task suggest that altered loading patterns persist six to twelve months after surgery. This may be a result of continued pre-operative movement patterns as well as the surgery itself, and should be kept in mind when developing rehabilitation programs for this patient population.

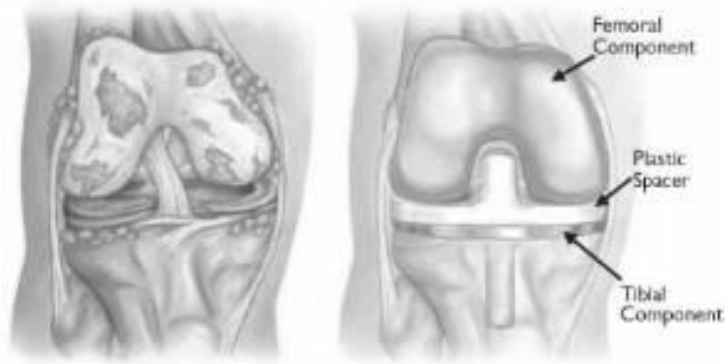
## INTRODUCTION

Osteoarthritis (OA) is the most common type of arthritis in Canada, affecting one in every ten Canadians and placing a great burden on the health care system (Hatfield et al., 2011). The knee is the most common joint to experience OA, and thus it is the most commonly replaced joint in severe cases of osteoarthritis or other degenerative diseases (Badley & DesMeules, 2003; Mizner et al., 2011; Thompson, 2011).



**Figure 1.** Physical effects of osteoarthritis on the structure of the knee joint.  
*Reprinted with permission from American Academy of Orthopaedic Surgeons (OrthoInfo, 2011)*

Most often, the joint is replaced by a prosthesis that mimics the complete tibiofemoral articular surface in a procedure termed total knee arthroplasty (TKA). This procedure has been proven to be effective in improving the quality of life for patients suffering from OA by increasing their ability to function independently and reducing chronic pain (CIHI, 2009; Harato et al., 2010; Levinger, P. et al., 2012; Milner & O'Bryan, 2008).



**Figure 2.** Replacement of an osteoarthritic knee joint with total knee replacement with tibiofemoral component.

*Reprinted with permission from American Academy of Orthopaedic Surgeons (OrthoInfo, 2011)*

Despite this success, functional limitations remain after TKA as prosthetic designs continue to fall short of being able to mimic the natural mechanics of a healthy knee during functional tasks (Alnahdi et al., 2011; Levinger, P., et al., 2012; Moonot et al., 2009).

With an aging yet active Canadian population, the average TKA patient is younger and has higher expectations for post-operative outcomes (Greene & Schurman 2nd, 2008). This patient demographic requires a knee prosthesis that provides greater stability, better functional outcomes, and a longer survivorship. Two of the most common prostheses used for TKA with this population include the Posterior Stabilized (PS) knee and the newer, Medial Pivot (MP) knee model. Studies that have looked at functional outcomes following TKA have generally focused on traditional prosthetic designs such as the PS knee; however, most studies fail to control for or to describe what prosthetic designs are used in their post-operative analyses of TKA patients. As a much newer prosthetic design, the MP knee has become a popular topic for studies looking at mechanical properties and design, yet there is limited research that has looked at functional outcomes for this prosthesis. Based on these limitations in the literature, our study proposes to examine functional outcomes for individuals with a MP or PS knee, following a unilateral TKA procedure to treat severe knee osteoarthritis. Our study was designed to examining function

based on the demands placed on the joint and the rest of the lower limb during activities of daily living, specifically for walking and transferring between sitting and standing positions. The mechanics of TKA participants during these activities were compared to those of healthy control participants that were matched with the TKA group based on age and BMI.

### OBJECTIVES

The purpose of this study is to examine how individuals who have undergone TKA with either a Medial Pivot or Posterior Stabilized knee implants, compare to their healthy counterparts when performing activities of daily living (ADL) such as walking on level and inclined surfaces, stair ascent and descent as well as sit-stand tasks. Specifically, the focus of this analysis is the range of motion, moments of force, and mechanical power experienced by the joints of the lower limbs during walking and transitioning between seated and standing positions. 3D motion analysis was used to examine patients' kinetics and kinematics of the lower limbs during the aforementioned gait and sit-stand tasks.

### RELEVANCY

Currently, there is a demographic transition happening in Canadian society which has resulted in TKA being offered to a younger patient population (Greene & Schurman 2nd, 2008). In the latest report from the Canadian Joint Replacement Registry, the largest increase in knee replacement surgeries was reported for individuals aged 45-54, with TKA rates more than doubling over the past decade for this age group (CIHI, 2009; Smith, P. N. et al., 2003). It can be suggested that younger, active patients will have higher expectations for a 'successful' surgery and, therefore, will place greater demands on their prosthesis than previous older patients. This growing younger patient population, combined with the ever increasing survival rates of modern implant designs, means that any residual limitations in function have become an important focus

for research (Yoshida et al., 2008). A better understanding of the effect of joint replacement on patients performing downhill walking and sit-stand tasks will provide new insight into how this procedure helps patients return to a 'normal' functional ability.

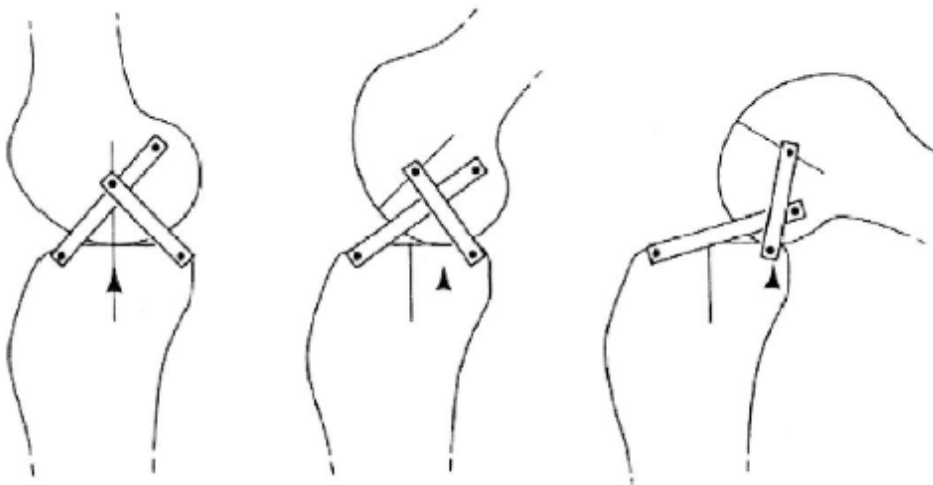
This information is valuable not only to the field of biomechanics, but also to those responsible for designing rehabilitation protocols and prosthetic models. Understanding the demands placed on the joints of the lower limbs during different functional activities could help health practitioners make appropriate decisions on how to design rehabilitation programs for TKA patients.

## REVIEW OF LITERATURE

Total knee arthroplasty is considered to be a safe and effective treatment for those with end-stage degenerative arthritis and severe functional limitations (Hatfield, et al., 2011; Jones et al., 2000). A successful TKA is expected to decrease or eliminate joint pain, provide stability, and improve functional ability (Rudan, 2003). This literature review looked at how previous studies have focused on gait mechanics and sit-stand performance of TKA patients, identifying gaps in these research areas.

### THE HEALTHY KNEE

Over the past few decades, there have been significant changes in the understanding of the natural biomechanics of the knee. Previously, the knee was thought to be guided by cruciate ligaments forming a four-bar linkage (Blaha, 2002; Smith, P. N., et al., 2003).



**Figure 3.**Representation of four-bar linkage model for the knee. Reprinted from (Smith, P. N., et al., 2003). Development of the Concepts of Knee Kinematics. *Archives of Physical Medicine and Rehabilitation*, 84(12), 1895-1902, with permission from Elsevier

The kinematics of this theorized four-bar model helped explain the combined glide and roll observed along the articulating surfaces of the knee joint, yet was limited to its basis on a two-dimensional perspective (Smith, P. N., et al., 2003). As a result, it was unable to account for factors such as the synchronous rotation of the tibia and the femur along the longitudinal axis, independent from flexion-extension. Progress in imaging technology has led to major advances in understanding the kinematics of the knee in a three-dimension model, recognizing the knee's six degrees of freedom. MRI technology has permitted separate analysis of the medial and lateral compartments of the knee, showing that longitudinal rotation can explain the motion of the femoral condyles on the tibial plateau (Smith, P. N., et al., 2003).

This has led to the development of a medial pivot model to explain knee motion patterns in flexion-extension. This pattern describes a synchronous rotation of the knee about its longitudinal axis and flexion about the flexion-extension axis (Smith, P. N., et al., 2003; Walker et al., 2009). Often these movements are associated with the "screw-home" mechanism of the knee, which describes the conjunct rotation of the tibia along the longitudinal axis and flexion at the epicondylar axis. (Neumann, 2010; Walker, et al., 2009). During these combined rotations, the medial aspect of the tibiofemoral joint maintains a larger contact force than the lateral aspect and undergoes a much smaller femoral roll-back. These specific movements along the longitudinal axis of the knee are possible due to the asymmetrical design of the joint (Neumann, 2010; Smith, P. N., et al., 2003). The medial aspect of the tibial plateau has a slightly concave shape, while the lateral aspect is convex or saddle-shaped (Neumann, 2010; Smith, P. N., et al., 2003). As a result, the medial femoral condyle remains relatively fixed, while the lateral condyle translates anteriorly and posteriorly throughout flexion-extension (Dennis et al., 2003). The asymmetry in femoral roll-back has been shown revealed through in vivo studies looking at

activities such as deep knee flexion, where the medial femoral condyle undergoes minimal posterior translation (2-3mm) whereas the lateral femoral condyle experiences 19-21mm of posterior translation (Dennis, et al., 2003; Dennis et al., 2005). Despite this knowledge of natural knee mechanics, many traditional prosthetic models have been developed with a symmetrical condylar design that could lead insufficient tibial rotation and femoral rollback (Liu et al., 2012).

## OUTCOMES FROM TOTAL KNEE ARTHROPLASTY

### BIOMECHANICAL PARAMETERS

#### *Gait*

Abnormal gait patterns after TKA, being either the result of prosthetic design or factors related to pain, must be identified and addressed, as this can cause abnormal loading and wear to the prosthesis and to other joints of the lower extremities (Hatfield, et al., 2011). For the purpose of this study ‘normal gait’ is defined as those patterns exhibited by healthy, age-matched controls who do not suffer from any lower extremity injuries or disease.

In general, gait parameters are described over a cycle that is defined by consecutive heel strikes. This defined gait cycle comprises two main phases of movement: stance phase and swing phase. Stance phase represents the first 60% of the gait cycle and is comprised of a period of double limb support followed by single limb support, with the contralateral limb in swing phase (Barr & Backus, 2001). The lower limb moves into a second brief period of double limb support (50-60% of gait cycle), just before the foot lifts off and enters swing phase. Analysis of stance phase is facilitated by breaking down this phase of the gait cycle into four periods: loading response, midstance, terminal stance, and pre-swing.

For the purpose of motion analysis, kinematic measures can include a variety of factors, including temporal parameters for gait such as velocity and stride length, as well as variables

such as range of motion and angular displacement (Winter, 2005). These measures are obtained with the use of cameras that record each participant's movement. Kinematic measures such as range of motion and angular displacement can be included in further analysis for kinetic measures. Kinetic analysis looks at the forces that are behind movement (kinematics) and is generally conducted through an inverse dynamics approach, incorporating vertical ground reaction forces (most often measured with force plates) and kinematic parameters to calculate moments of force and mechanical power (Lamontagne et al., 2009; Winter, 2005).

Previous studies that have incorporated kinematic and kinetic measures for gait analysis with OA patients have reported that individuals with knee OA exhibit reduced self-selected walking velocity and greater knee adduction moments at both the affected and unaffected knee, as well as at the contralateral hip, during gait (Hatfield, et al., 2011; Metcalfe et al., 2013). OA has also been associated with reduced knee flexion angles and moments in early stance phase, along with decreased knee extension moment at toe off. (Hatfield, et al., 2011). These gait alterations are of interest as continued patterns related to joint loading, such as knee flexion moment, are predictive of post-operative knee pain (Levinger, P., et al., 2012).

Gait analysis with TKA patients can help determine if these OA-related movement patterns persist after surgery, a principal concern for factors such as OA progression, component migration, and prosthetic loosening (Levinger, P., et al., 2012). An understanding of how OA influences gait patterns is important for monitoring post-operative outcomes as previous studies have reported that pre-operative gait patterns are often continued following surgery (Levinger, P., et al., 2012; Milner & O'Bryan, 2008).

In the past, most TKA gait studies have focused on motion in the sagittal plane, where angular kinematics are frequently associated with altered quadriceps function. Despite the

attention given to sagittal plane motion, previous studies have reported conflicting results following TKA. Hatfield et al. (2011) compared gait before and after surgery for patients with severe knee OA, reporting that knee flexion throughout gait was much higher following TKA; however, this comparison was made only to pre-operative gait and not to a healthy control cohort. Berth et al. (2002) compared post-op TKA patients to a healthy control group and found that TKA patients exhibited increased knee flexion at the operated knee during swing phase and stance, when compared to the non-operative or control knee. An increase in knee flexion, compared to healthy knee motion patterns, would be of concern as the resulting increase in muscular contraction may result in greater contact forces and load over the joint, which could negatively affect prosthesis longevity (Levinger, P., et al., 2012). Other studies (Benedetti et al., 2003; Ishii et al., 1998; Smith, A. J. et al., 2006; Yoshida, et al., 2008), however, have reported decreased knee flexion range of motion among TKA patients, compared to controls, during level walking tasks.

Abnormal gait patterns for flexion/extension moment at the knee have also been reported as part of altered mechanics observed before and after surgery with TKA patients (Levinger, P., et al., 2012). A normal gait pattern in the sagittal plane exhibits a biphasic flexion-extension curve, with an initial moment created from knee extension moving into a flexion moment at early stance, followed by a second phase where the knee extends and then flexes during late stance. Both Levinger et al. (2012) and Yoshida et al. (2008) reported decreased knee extensor moments at the operated limb during gait. At 4-months post-op, Levinger et al. (2012) noted that such a reduced extension moment was predictive of having an abnormal flexor moment pattern at 12-months following TKA. As the primary knee extensors, the strength and level of activation of the

quadriceps muscles may also play a role in such abnormal moment curves, as TKA patients have often been associated with quadriceps avoidance patterns (McClelland et al., 2007)

While sagittal plane mechanics have been a principal concern in previous TKA gait studies, movement and loading about the frontal plane has become a measure of significant interest in recent years due to its association with OA progression (Hatfield, et al., 2011; McClelland, et al., 2007). During level walking, the medial compartment of the knee sustains a load that is up to 1.7 times greater than that experience at the lateral compartment (Mundermann et al., 2008). Individuals who maintain a varus knee alignment may increase this rate of load imbalance, which can lead to tibial component failure (Green et al., 2002). A successful TKA should aim to reduce knee adduction moment and loading of the knee to prevent OA progression. Several studies that have looked at pre- and post-operative outcomes for TKA with individuals who suffer from severe knee OA have reported that TKA successfully reduced knee adduction angles and moments at the operated limb (Hatfield, et al., 2011; Mandeville et al., 2008). However, some studies have found that knee adduction moments are elevated at the non-operated limb (compared to the operated limb and controls) after surgery (Metcalf, et al., 2013; Milner & O'Bryan, 2008). This is likely a result of pre-operative gait patterns, placing a greater load at the non-operated limb to avoid pain at the affected limb. Maintenance of pre-operative gait alterations following surgery can further exacerbate this condition and lead to OA progression at the non-operated limb.

### *Walking on Inclined Surfaces*

While previous gait studies provide important insight into functional performance following knee replacement surgery, they do not provide a realistic test of how an individual navigates their environment. Recently, researchers have noted the need for more applicable gait analysis

(McIntosh et al., 2006; Prentice et al., 2004), looking at how healthy individuals navigate different inclined slopes. McIntosh et al. (2006) examined gait dynamics of healthy, young participants along an inclined walkway, reporting that flexion at the hip, knee and ankle joints at heel strike increased as the incline on the walkway increased from  $-10^{\circ}$  to  $+10^{\circ}$ . Redfern & DiPasquale (1997) also studied inclined walking, focusing on movement in the sagittal plane when walking downhill. The knee joint was reported to experience the most pronounced change in biomechanical parameters when moving from level to downhill walking, experiencing reduced peak flexion angles, as well as increased extension moments and peak power absorption during downhill walking (Kuster et al., 1995; McIntosh, et al., 2006; Redfern & DiPasquale, 1997). Similarly, Kuster et al. (1995) examined lower limb kinematics and kinetics during downhill and level walking with healthy participants at a controlled cadence. They reported that the most significant changes in gait patterns occurred at the knee and ankle, with greater moments and power at the knee, but smaller at the ankle, during downhill walking. An increase in hip flexion moment has also been reported during downhill walking (Redfern & DiPasquale, 1997; Kuster et al. 1995); however, changes in hip measures were described as being much smaller than those observed at the knee and ankle.

Each of these studies provides important information on how gait mechanics can change when negotiating different inclined surfaces; however, this research has been limited to focusing on young, healthy participants. To the best of our knowledge, no previous studies have examined the gait of TKA patients on non-level surfaces. As a result, it is difficult to predict any changes that may exist in movement patterns for these individuals, compared to matched control subjects. It could be suggested that changes in movement patterns between level and downhill walking would be similar to those observed with healthy participants in previous studies; however, the

magnitude of these values may differ due to muscle weakness or gait patterns adopted by TKA patients before or after surgery.

### *Sit- Stand*

The sit-stand task is commonly used in studies to examine lower-limb biomechanics in both healthy and pathologic populations, proven to be a valid test for measuring functional outcomes with TKA patients (Boonstra et al., 2008; Burnett et al., 2011; Christiansen et al., 2011; Farquhar et al., 2008). Despite this popularity, there remains limited information on the actual lower limb biomechanics used by TKA patients to complete this task as previous studies have placed a greater focus on weight-bearing symmetry (Abujaber et al., 2013; Boonstra et al., 2010; Burnett, et al., 2011; Christiansen, et al., 2011).

The ascension phase of sit-stand will be analyzed in this study as ‘sit-to-stand’. Schenkman et al. (1990) defined the movement pattern for sit-to-stand using four main phases of motion: flexion momentum, momentum transfer, extension, and stabilization. The flexion momentum phase characterizes the initiation of movement up to the point of seat-off, whereas all of the other phases focus on the period of movement between seat-off and the final standing position. This study focused on the period of movement after seat-off and, therefore, included these final three phases. The momentum-transfer phase is named based on the transfer of momentum from the upper body (occurring in preparation for seat-off) to the rest of the body in order to lift itself off the seat. During this phase, peak flexion occurs at the hip, knee, and ankle joints. Studies that have examined kinematics for TKA patients during sit-to-stand have reported no significant difference between these patients and controls for peak flexion at the knee and the ankle; however, Farquhar et al. (2008) did find that TKA participants exhibited significantly greater hip flexion angles at both 3-months and 1-year post-op, compared to controls.

The extension phase of sit-to-stand occurs between peak ankle dorsiflexion and the point at which hip extension begins to cease; characterizing the main period of lower limb extension during sit-to-stand (Schenkman, et al., 1990). Individuals suffering from knee OA have been reported to exhibit reduced hip and knee extension moments during this period of sit-to-stand (compared to controls), which was associated with WOMAC scores for reduced functional capacity (Turcot et al., 2012). Based on this link between sit-to-stand mechanics and functional capacity, it is of interest to examine how biomechanical patterns of TKA patients compare to healthy individuals. Farquhar et al. (2008) found that at 3-months post-op, TKA participants had significantly reduced extension moments at the operated knee compared to both the non-operated knee and controls; however, this difference was no longer significant at 1-year post-op. They also noted that the TKA group exhibited greater hip flexion moments at both the operated and non-operated limbs, compared to controls, at 3-months and 1-year post-op. While numerous studies have focused on the symmetry of loading at the lower limbs during sit-to-stand (Abujaber, et al., 2013; Boonstra, et al., 2010; Burnett, et al., 2011; Christiansen, et al., 2011), no references could be found for power generation or absorption during sit-to-stand for individuals with TKA.

There has been limited research focused on lower limb movement in the frontal plane for sit-to-stand with only one study, by Lamontagne et al. (2012), that examined hip angles, reporting peak hip abduction and adduction angles of  $8.5^{\circ}$  and  $7.6^{\circ}$ , respectively. We were unable to find any studies that describe knee angles in the frontal plane for the sit-to-stand task among healthy individuals or those with TKA/OA.

While sit-to-stand transfers have been a popular method of analyzing lower-limb function, there is limited information available regarding biomechanics of the stand-to-sit

movement, despite the fact that it is performed just as often as sit-to-stand during daily activities. Lamontagne et al. (2012) included a stand-to-sit analysis in their study looking at outcomes from total hip arthroplasty (THA), reporting that age-matched controls exhibited approximately 91° of hip flexion and 84±8.1° of knee flexion. This study was also the only one that could be found to describe moments of force and power at the lower limb during stand-to-sit, noting that the THA patients in their study exhibited significantly reduced extension moments and peak power absorption at the hip but not the knee, compared to controls. It is likely that this discrepancy reflects the nature of the surgery, impacting the structure around the hip joint. It may be suggested, then, that this type of discrepancy would be observed about the knee joint for TKA patients, although no studies could be found to confirm this assumption.

Wang et al. (2005) examined stand-to-sit performance among TKA participants, comparing single-axis and multi-axis PS knees. They reported peak flexion angles for the hip and ankle, but did not report overall peak knee flexion angles. It should be noted that one other study was found that described lower limb kinetics during stand-to-sit; however, results from this study were not applicable to this literature review. Kutzner et al. (2010) used instrumented knee prostheses to obtain in vivo measures; however, the authors reported joint moments based on percent of body weight and body height for each subject and, therefore, these results are difficult to compare to other TKA or control groups for the purposes of this study.

There is a limited amount of research that has focused on frontal plane biomechanics for stand-to-sit. Lamontagne et al. (2012) reported abduction angles and adduction moments at the hip during stand-to-sit, but did not include kinematic/kinetic measures for the knee in the frontal plane. Wang et al. (2012) conducted a study looking at kinematics and kinetics during stand-to-sit among a group of males who underwent bi-compartmental knee replacements, reporting no

significant differences between TKA participants and matched controls for any kinematic or kinetic measures. Based on these findings for frontal plane parameters, it may be expected that the TKA participants in our study would exhibit frontal plane kinematic and kinetic parameters that were similar to controls. However, the findings from Wang et al.'s study (2012) may have been influenced by the selection of only male participants within their testing groups. The study done by Wang et al. (2005) comparing single-axis and multi-axis PS knees included both males and females within test groups and reported much lower knee adduction angles.

### *Summary*

In summary, research and technological innovation in recent years has transformed the understanding of knee mechanics from two-dimensions to a three-dimensional model. In an effort to keep pace with these findings research has sought to understand how and to what degree TKA can return individuals back to natural movement patterns during activities of daily living. Gait analysis is a popular approach used to measure this type of functional performance; however, it has focused exclusively on level walking surfaces for TKA and OA populations. In this type of environment, individuals with knee OA exhibit increased adduction moments at the hip and knee, as well as reduced knee flexion angles and extension moments during the stance phase of the gait cycle. TKA has been shown to effectively reduce adduction moments and increase flexion angles at the operated knee, compared to pre-operative values. However, TKA patients continue to exhibit alterations in knee flexion/extension angles as well as reduced knee extension moments (compared to controls); the non-operated limb has also been shown to exhibit increased adduction moments, which may lead to OA progression. These changes in loading for both the operated and non-operated limbs are important to understand how effective TKA can be

in returning patients to normal, healthy biomechanical patterns that would prevent OA progression to other joints in the lower limbs.

Gait changes between level and downhill walking have been shown to impact the knee more than the ankle or hip joints and, therefore, it could be expected that further discrepancies in knee kinematic and kinetic measures (compared to control subjects) may be observed during downhill walking. The differences in peak power values for downhill walking may also result in discrepancies between TKA and control participants as the operated knee may struggle to meet the demands for greater power absorption that is associated with downhill walking. As most environments are not made up exclusively of level walking surfaces, it is important to understand how TKA patients negotiate surfaces that better replicate those encountered in their daily life.

The sit-stand task includes transitions between sitting and standing positions, divided into sit-to-stand and stand-to-sit, and has been proven as a valid method for measuring functional outcomes following TKA. While it has been a popular method for looking at loading ratios between operated and non-operated limbs (with TKA patients), there are a limited number of studies that have looked at lower limb biomechanics during these two movements. At 3-months and 1-year post-op, TKA participants have been shown to exhibit greater hip flexion angles compared to controls, but no significant differences between the operated and non-operated limbs. With respect to lower limb kinetics, TKA patients appear to exhibit significant differences in hip and knee extension moments at 3-months post-op; however, the majority of these differences are no longer significant at 1 year after surgery. The only discrepancy that appeared at both the 3-month and 1-year assessments was a significantly larger hip flexion moment at the operated and non-operated limbs, compared to controls. No studies could be found that addressed frontal plane kinematics/kinetics or power absorption/generation during sit-to-stand

among a TKA population, highlighting the need for this research. While Wang et al. (2005) did report sagittal plane kinematics/kinetics during stand-to-sit, their analysis was based on comparing single- and multi-axis PS knees, without including controls. Of the two studies that were found to address stand-to-sit for TKA patients, one study included only male participants in their TKA and control groups, while the other included both males and females but had no control group. Therefore, no comparisons were made for both sexes across surgical and control groups, despite the fact that sex has been shown to significantly influence comparisons for frontal plane biomechanics. The lack of a comprehensive analysis addressing both transitions (sit-to-stand and stand-to-sit) in the sagittal and frontal planes, including both males and females, warrants further analysis to understand how TKA patients compare to healthy, age-matched controls. Therefore, this study sought to include this type of comprehensive analysis, including TKA patients with both MP and PS knees at approximately 1 year after surgery.

## RESEARCH QUESTION

The overall question this study sought to answer was: Do TKA patients exhibit similar lower limb biomechanics during walking and sit-stand compared to healthy, age-matched counterparts? More specifically, with respect to walking, we wanted to find out if TKA patients and controls exhibit similar gait patterns on both level and inclined surfaces.

## RATIONALE

Walking and sit-stand are two activities that are performed numerous times on a daily basis as a part of an active, independent lifestyle for most adults. For individuals recovering from TKA, it is important to understand whether they are able to perform these activities, using similar mechanics to healthy, age-matched populations, to measure the success of their surgical procedure in restoring function. While some studies used gait analysis and sit-stand performance

to measure TKA outcomes in the past, they have been limited in their approach. No previous study, to our knowledge, has included inclined walking surfaces in gait analysis with TKA populations. With respect to sit-stand and TKA populations, no literature could be found that examined frontal plane mechanics for rising from a seat, while few studies could be found that addressed any biomechanical parameters for sitting down. This study sought to address these gaps in the literature using level and downhill walking tasks, as well as transitions between sitting down and standing up from a seat.

## HYPOTHESES

- In accordance with findings from previous studies, it was anticipated that TKA participants would exhibit reduced knee extension moments at the operated limb and increased adduction moments at the hip and knee of the non-operated limb, compared to the operated limb, during level walking.
- It was anticipated that TKA participants would exhibit reduced knee flexion angles and extension moments, compared to controls, during downhill walking. This prediction was based on two factors: (1) previous findings from comparisons of TKA patients and control groups during level walking, and (2) results from previous studies that noted reduced knee flexion angles and greater extension moments during downhill walking (compared to level walking) among healthy populations. It was believed that these increasing demands for downhill walking would be reflected in greater discrepancies between the TKA and control groups.
- Based on findings from previous studies noting reduced knee power absorption and ankle power generation during downhill (vs. level) walking, it was anticipated that TKA participants would have reduced peak power values for these measures when compared to

controls. Once again, this prediction was based on the assumption that increasing demands would bring out greater discrepancies between TKA participants and controls.

- In accordance with previous studies that examined the sit-stand task with TKA patients, it was predicted that the TKA participants in this study would exhibit greater hip flexion angles and moments during sit-to-stand (compared to the control group). This difference was also expected to carry over for stand-to-sit performance.
- No significant differences were anticipated between TKA and control groups for frontal plane kinematics/kinetics during sit-to-stand and stand-to-sit. This null hypothesis was based, in part, on the lack of literature related to these measures for TKA populations. With no previous studies addressing these measures for outcomes from TKA, there was no basis to assume there would be differences between these groups. This prediction was also based on findings from Wang et al. (2012) who noted no significant difference between their male participants in TKA and control groups.
- TKA participants were predicted to exhibit increased peak power values during sit-to-stand and stand-to-sit, compared to controls. This was predicted based on the assumption that TKA participants would also experience greater hip flexion angles and flexion moments, which contribute to calculated power values. No significant differences were predicted for peak power values at the knee or ankle for the sit-stand task. This assumption was once again based on the lack of literature that would lead the authors to assume any differences may exist.

## **METHODS**

This study compared two groups: individuals who had undergone unilateral TKA with a MP or PS knee and an age-matched control group. The TKA patients were assessed at 11±5 months following surgery. This research design allows for comparison of post-operative outcomes for TKA with MP or PS knees, to healthy age-matched control knees, with the control group serving as a baseline for what is considered to be a ‘normal’ knee.

### **PARTICIPANTS**

Seventeen participants between the ages of 45 to 75 years old were recruited through the Ottawa Hospital’s Division of Orthopaedic Surgery, on a voluntary basis. All TKA patients underwent surgery by the same surgeon; of the 17 surgical patients, 10 had received Medial Pivot knees, while the others had received a Posterior Stabilized prosthesis. Exclusion criteria for TKA subjects included any degenerative conditions (other than OA at the affected knee) impacting joints of the lower extremities, a BMI >35 kg/m<sup>2</sup>, bilateral knee replacement, previous joint replacement at the affected knee, and any other past or present conditions that may impact lower limb biomechanics. Seventeen healthy control subjects were recruited from the Ottawa-Gatineau region, with similar exclusion criteria to the TKA group.

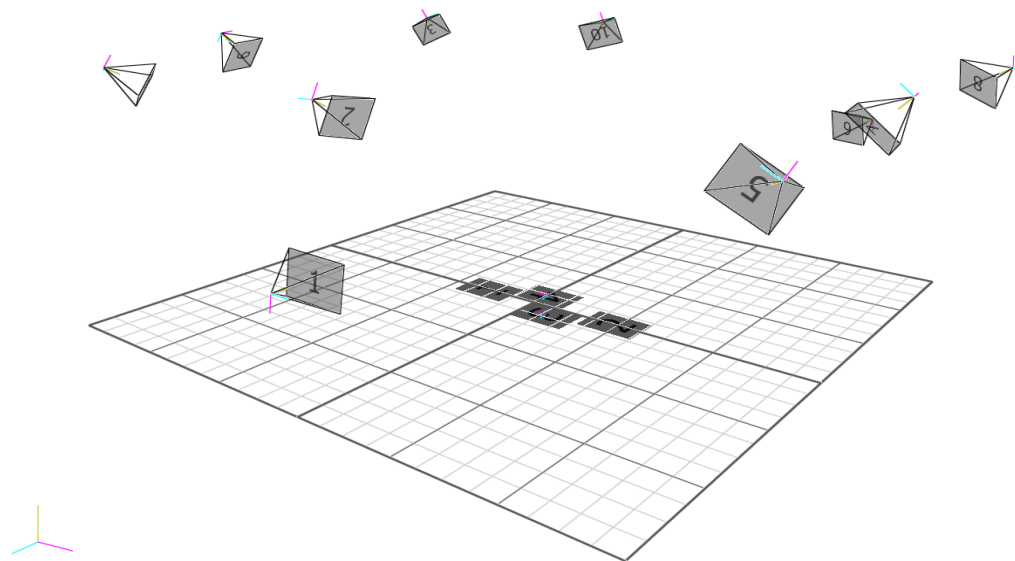
### **MATERIALS AND EQUIPMENT**

#### **KINEMATICS**

Walking trials were captured using a 10-camera motion analysis system (Vicon MX-13, Oxford Metrics, Oxford, UK) recording at 200 Hz. Prior to data collection, dynamic calibration of the capture volume was conducted with a 240mm T-shaped marker wand. Following a

successful dynamic calibration, a static calibration was performed with an L-shaped frame (ErgoCal 14mm) to set the origin for the global coordinate system within the capture volume.

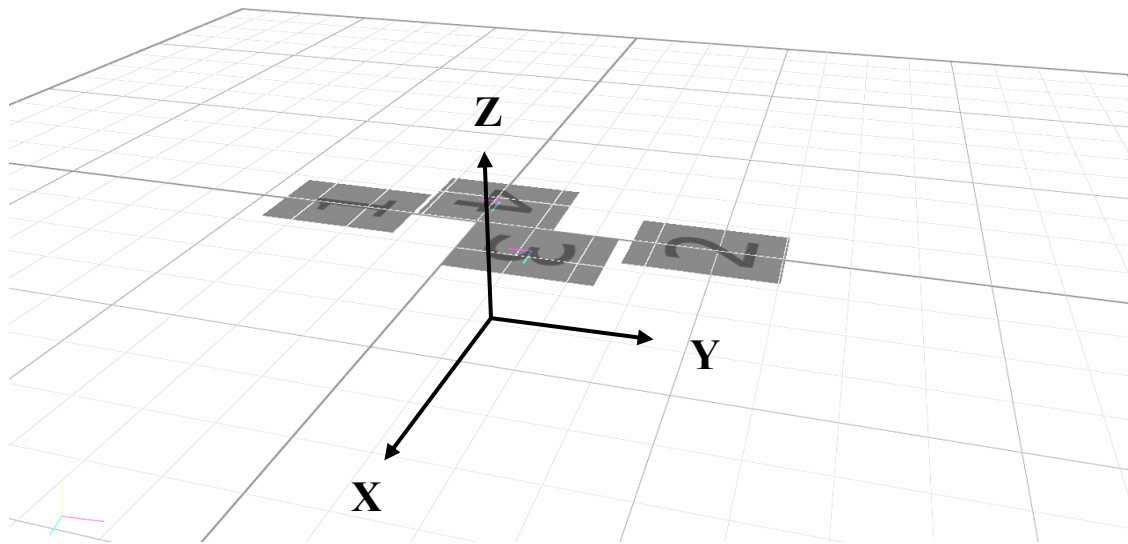
Each participant was outfitted with a tight-fit motion analysis suit and 45 reflective markers (14mm diameter) placed according to a modified Plug In Gait model; an outline of this marker placement can be found in Appendix A and Appendix B.



**Figure 4.** Capture volume for walking trials.

### KINETICS

During level walking trials ground reaction forces were recorded by four force plates, (Bertec Corporation, models FP4060-8 & FP4060-10, Columbus, Ohio; Kistler Instruments Corp, models 9286A & 9286AA, Winterthur, Swtz) which were embedded within the floor, recording at 1000 Hz. Two of these force plates (Kistler Instruments Corp, models 9286A & 9286AA, Winterthur, Swtz) were also used to measure ground reaction forces during downhill walking trials, with the plates fitted within the ramp base, flush with the surface.

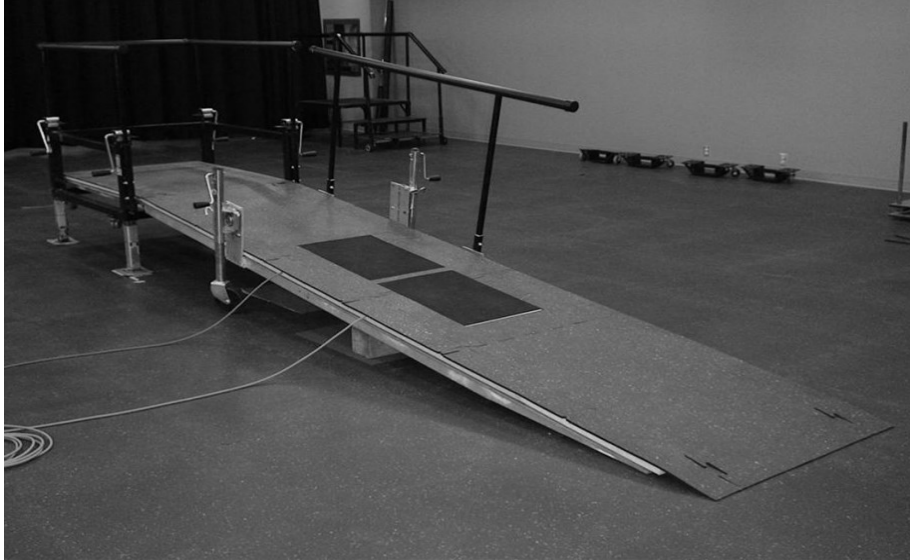


**Figure 5.** Force plate setup for level walking trials.

Platforms 1 and 2 represent the Kistler force platforms that were also used in inclined walking trials. Platforms 3 and 4 represent the Bertec force plates.

### *THE INCLINED WALKWAY*

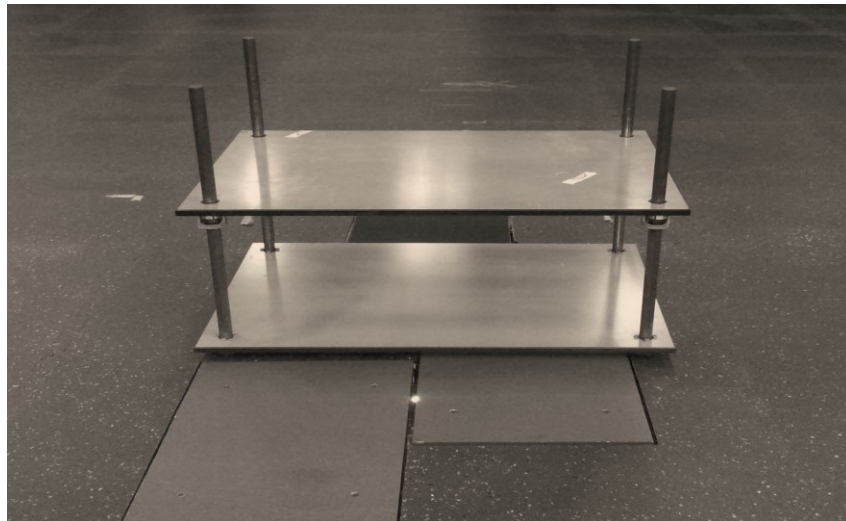
Inclined walking was analyzed using an instrumented ramp at the Human Movement Laboratory at the University of Ottawa. The ramp walkway was four metres long and one metre wide with a 2.5m x 1.0m horizontal platform at the end, and was set at a 9° incline (12.5% slope). This ramp setup is shown in Figure 6, including the two Kistler force plates that were embedded within the ramp's frame.



**Figure 6.** Ramp used for inclined gait

### ADJUSTABLE BENCH

The sit-stand task was performed using an instrumented seat that was set at 45cm height (Figure 7), which was selected based on the permitted toilet seat height under the Ontario Building Code Act. Two Bertec force plates (Bertec Corporation, models FP4060-8 & FP4060-10, Columbus, Ohio) were placed in front of the instrumented seat, flush with the floor, to collect ground reaction forces.



**Figure 7.** Adjusted bench used for sit-stand task





**Figure 9.** BTS FreeEMG wireless electrode used for recording muscle activation.

## PROTOCOL

A research coordinator from the orthopaedics division of the Ottawa General Hospital contacted TKA patients that met the inclusion criteria. Each patient was given a general explanation of what the testing involved and information on how to get to the lab. On their respective testing day, the participant was met in the main entrance and escorted to the lab, where the protocol was explained in greater detail. Once all questions and concerns were addressed, the participant was given the chance to review and sign the consent form prior to beginning testing. They were then guided to the changing area of the lab where they traded their clothes for fitted spandex shorts and t-shirt. At this point, anthropometric measures (outlined in Table 1) were taken and recorded in their Vicon Nexus subject file. All trials were conducted with participants wearing their own shoes.

**Table 1.** Anthropometric measurements of participants

Measurements (units)	Measurement Description
Mass (kg)	Weight of the participant
Height (cm)	Height of the participant
Leg Length (cm)	Distance between ASIS and medial malleolus taken for each limb while standing
Knee Width (cm)	Width of the knee taken between medial and lateral condyles while standing
Ankle Width (cm)	Width of the ankle between medial and lateral malleoli while standing

Following anthropometric measurements, 16 wireless electrodes were placed on both lower limbs, to measure muscle activation for the following 8 muscles: rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), biceps femoris (BF) semitendinosus (ST), tibialis anterior (TA), and the lateral (GAL) and medial (GAM) heads of the gastrocnemius muscle. These placements were done in accordance with the SENIAM protocol\*\*.

Once the electrodes were placed on the appropriate muscle locations, maximal voluntary isometric contractions (MVICs) were conducted to record maximal activation of the muscle, to be used in future analysis to normalize activation signals during dynamic activities. The positions for each isometric contraction are shown in Table 2; force outputs during each contraction were recorded using a handheld dynamometer (Lafayette Instrument Company, Lafayette, IN). Two MVICs were performed for each muscle, with a 30 second rest provided between each effort.

**Table 2.** Protocol for maximal voluntary isometric contractions

Muscle Group (Muscles)	Position
Quadriceps (RF, VL, VM)	Sitting with knee flexed at 60°
Hamstrings (BF, ST)	Lying prone with knee flexed at 60°
Dorsiflexors (TA)	Lying prone with ankle flexed at 100°
Plantar Flexors (GAM, GAL)	Lying prone with ankle flexed at 100°

Following MVICs, the participant was outfitted with forty-five reflective markers, placed on bony landmarks according to a modified Plug In Gait marker set named the University of Ottawa Motion Analysis Model (UOMAM) (see Appendices A & B). The participant then performed a static trial to define the coordinate system for each limb segment and to define joint centers. They were asked to stand in the middle of the capture volume with their arms held straight in front of them, parallel to the ground, with their feet shoulder width apart and their head facing forward, holding this position for a five-ten second period. Along with the static trial, the participant performed four tasks: walking up and down a ramp, walking across a level

floor, standing up and sitting down with an instrumented seat, and walking up and down a staircase. Inclined gait was performed with the individuals starting two steps away from the first force plate, with the operated leg striking this plate first. After reaching the end of the walkway, the participant would turn to face down the ramp, pausing two steps away from the force plate closest to the top of the ramp. They would then walk down the ramp with the operated limb striking this top force plate. Level walking trials were performed with the subject walking across a level walkway in one direction, stepping on two force plates embedded in the floor. For all of the walking trials participants were asked to walk at a self-selected pace and not to modify their stride length in an effort to step on the force platforms. If it was suspected that they altered their gait to target the force plate or if their foot did not land in the middle of the force plate, the trial was discarded. To avoid modifications in normal stride patterns, several practice trials were performed before each task to determine an appropriate starting position for participants to reach the force platforms at their normal pace. In total, a minimum of five trials were recorded for each of the two gait conditions as well as each of the sit-stand conditions. Stair trials were conducted in a similar fashion to the inclined walking task, with participants starting two steps away from the first step. They then proceeded to strike the force plate on the first step with the operated limb, climbing the stair apparatus until they reached the end of the platform on the top step. They then turned to face down the stairs, pausing before the top stair, walking down the stairs with the operated limb striking the force plate on the top step.

## DATA ANALYSIS

### KINEMATICS

Kinematic analysis focuses on the range of motion occurring about the joints of the lower-limb, calculated by combining three-dimensional images of movement with time. For this study, kinematics of the knee, hip, and ankle joints were calculated for two planes of motion (sagittal and frontal) during each of the tasks. These calculations were done with Vicon's Nexus software (version 1.7) and the UOMAM marker set. The UOMAM model was used for motion analysis as it provides an additional marker at both the left and right lateral iliac crests to compensate for the potential disappearance of an ASIS marker. With previous marker sets, if the ASIS marker became covered during any point of a dynamic trial, it would impede calculations for hip joint center. With the UOMAM model the missing ASIS marker is created based on the location of the marker on the respective iliac crest.

### KINETICS

The kinetic parameters of interest for this study were the moments of force and the power generated and absorbed at the hip, knee, and ankle joints. These values were calculated using inverse dynamics based on the kinematics and the ground reaction forces measured with the 3D motion capture system and Kistler and Bertec force plates respectively. Moment of force and power values were normalized based on body mass to give a value in N·m/kg and W/kg, to facilitate comparisons between different participants.

## GAIT

Gait cycle was normalized based on consecutive foot strikes for the limb of interest. Foot strike events were determined based on angular velocity of the ankle marker, while the foot-off event was defined as the first frame in which there was no force being applied to the force plate following stance phase of gait. These events were all determined by the same reviewer for all of the trials used in this analysis.

## SITTING AND STANDING MANOEUVRES

The sit-to-stand task was defined as the period beginning at seat-off, continuing until a full standing position was reached. Seat-off was defined as the point at which the participant lost contact with the seat, measured based on the peak vertical ground reaction force (vGRF) recorded by the force plates. A full standing position was defined as the point at which there is no further forward velocity of the ASIS markers (indicating full hip extension).

The stand-to-sit task began with initiation of flexion, defined as the point where velocity of the ASIS marker rose above zero. The cycle was completed at the seat-contact event, defined as the point just after peak vGRF (similar to the measure for seat-off).

## DATA PROCESSING

A Woltring filter (predicted mean square error value of  $15 \text{ mm}^2$ ) was applied to the raw three-dimensional (3D) marker trajectories while ground reaction forces were filtered with a 4<sup>th</sup> order, zero-lag Butterworth filter (10Hz cut-off), performed with Vicon Nexus software. Trials were time-normalized based on their task-specific movement cycle prior to being modelled and were then analyzed using a customized Matlab script (MathWorks, Natick, United States). This allowed for visualization of each trial, where outlier trials could be removed and peak values

could be automatically extracted and saved in a spreadsheet (Excel, Microsoft Office, Redmond, Washington) for further statistical analysis. Separate analysis was conducted for EMG data; however, this process will be discussed in future articles as it falls outside the scope of the articles for this thesis project.

## STATISTICAL ANALYSIS

Statistical analysis for each task began with testing for parametric assumptions related to outliers and normal distribution. With each task, many of the kinematic and kinetic measures in the frontal plane were found to have a non-normal distribution (based on results from Shapiro-Wilk tests). Outliers for these measures were examined; however, no participants were removed from the analysis as each frontal plane movement pattern was considered to be representative of the individuals' movements. It was believed that the non-normal distribution was likely reflective of the greater variability that is inherent between individuals for movement in the frontal plane, and that the small sample size may have contributed to these significant findings. As a result, frontal plane kinematics and kinetics were analyzed using non-parametric tests.

### PARAMETRIC TESTS

After this initial analysis, we wanted to examine whether the combination of males and females within the same groups would influence outcomes for gait and sit-stand parameters. To assess this potential influence, sagittal plane kinematics and kinetic measures, as well as power values, were analyzed using a 2-way ANOVA to determine if there was a significant interaction between group (TKA or control) and sex. For comparisons between the operated and non-operated limbs, this test was conducted using a between-within model, where the between factor was sex and the within factor was the leg (operated vs. non-operated).

### *NON-PARAMETRIC TESTS*

To determine if sex influenced variables in the frontal plane, a series of Mann-Whitney U tests were conducted for each leg (operated, non-operated, control) to determine if there was a significant difference between males and females for any of the variables of interest. If sex was found to have a significant effect of a kinematic or kinetic measure, comparisons across groups were done separately for males and females across groups. Comparisons between operated and control limbs were conducted using Mann-Whitney U tests, while Wilcoxon Ranked tests were used to compare the operated and non-operated lower limbs (assuming they were not independent groups).

As the focus of this thesis project was limited to kinematics and kinetics for downhill and level walking, as well as for the sit-stand task, the remaining data collected from this thesis project will not be discussed in the articles included for this thesis. Data pertaining to muscle activation, uphill walking, and stair climbing will be analysed and reported in future articles.

### *NORMATIVE RANGE*

Based on group averages calculated for the control group, a normative range was established for movement cycles. This normative range was included to reduce oversampling of data and to focus on differences that are more likely to be clinically relevant, as opposed to reporting significant differences that may have very small effect sizes. This range was calculated to be within one standard deviation of the control group's mean kinematic or kinetic measure at each frame of the movement cycle. If the movement pattern of the TKA group fell outside this normative range for any of the kinematic/kinetic measures, it was included in statistical analyses. Kaufman et al. (2001) included a similar normative range within their study of TKA patients;

however, they set their range to be within two standard deviations of the mean. This would make sense initially for our study, as we have set our confidence level at 95% ( $\alpha=0.05$ ), and statistical analysis of normal curves dictates that 95% of all data points should fall within two standard deviations of the mean for a given population. However, this assumption is based on the data set having a normal distribution along with a large sample population. Since our study included data that were not normally distributed along with smaller sample sizes, we believed that a smaller normative range should be used to minimize risks for Type II error. As a result, we designed our normative range to be within one standard deviation above and below the mean for each data point.

# **ARTICLE 1**

## Gait dynamics on level and inclined surfaces following total knee arthroplasty

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## ABSTRACT

Gait analysis is an accepted measure for functional outcomes following total knee arthroplasty (TKA) procedures; however, research has been limited to gait on level surfaces. This study looked at gait mechanics on both level and inclined walkways for TKA patients 6-12 months after surgery. Participants for the TKA group (n=17) were recruited after undergoing unilateral TKA with the same surgeon, and were matched with a control group (n=17) based on age and BMI. During both walking tasks, TKA participants had reduced range of motion, peak flexion angles and power generation at the knee, as well as altered peak knee flexion/extension moments for the operated limb. Similar discrepancies were observed for the operated limb during downhill walking, with an increase in discrepancies for power absorption and extension moment at the knee. These findings indicate that altered gait persists after TKA and that non-level walking not only replicates altered mechanics from level walking, but brings about further differences in gait patterns.

**KEYWORDS:** Total Knee Arthroplasty, biomechanics, kinematics, kinetics, knee, level and inclined walking, osteoarthritis

## INTRODUCTION

Osteoarthritis (OA) is a chronic condition that affects over 10% of the population (PHAC, 2010), resulting in significant costs to health care and labour forces. As the joint most commonly affected by OA (PHAC, 2010), the knee is often treated by replacing the tibiofemoral joint in a procedure known as total knee arthroplasty (TKA). This procedure has been proven to be effective in improving function and reducing pain for many individuals suffering from severe OA (Hatfield, et al., 2011; Levinger, Pazit et al., 2013). While the number of procedures increases every year, the largest rise in surgeries is being seen among 45-54 year old individuals (CIHI, 2009). This growth reflects a changing patient demographic for joint replacements, where the majority patients in the past have been over the age of 65 (CIHI, 2009).

This targeted patient population will have different expectations for surgical outcomes, seeking greater mobility and function along with greater prosthesis survivorship. With this in mind, understanding the functional outcomes for TKA is more important than ever to improve patients' quality of life. Previous studies looking at functional outcomes for TKA have noted altered gait patterns that include decreased knee extension moment during stance (Levinger, P., et al., 2012; Yoshida, et al., 2008), as well as reduced knee flexion excursion (Benedetti, et al., 2003; Ishii, et al., 1998; Smith, A. J., et al., 2006; Yoshida, et al., 2008). An additional area of concern for TKA gait patterns is related to knee adduction moments due to an association with OA progression. While TKA has been shown to reduce knee adduction moments at the operated limb, compared to pre-operative values, some studies have found significantly greater adduction moments at the hip and knee of the non-operated limb after surgery (Milner, 2008; Metcalfe, 2013).

While previous gait studies have helped to establish a better understanding of functional outcomes following TKA, they have limited their approach to gait analysis with level surfaces. Due to the fact that most daily environments are not limited to this type of surface, we believe that gait analysis should include non-level (inclined) surfaces to better reflect the demands placed on the lower limbs. While no previous study, to our knowledge, has looked at gait for TKA patients on inclined surfaces, studies that have included healthy individuals have noted that this task places different demands on the body (compared to level walking) (Kutzner, et al., 2010; Lay et al., 2006; McIntosh, et al., 2006; Redfern & DiPasquale, 1997).

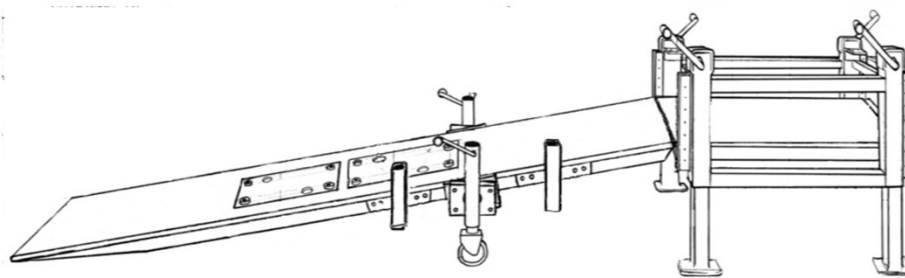
The purpose of this study was to examine lower-limb joint mechanics of TKA patients during gait on both a level and inclined surfaces, with comparison to healthy, age-matched controls. Specifically, this gait analysis focused on walking downhill since this type of task requires greater dynamic stability as opposed to walking uphill.

It was anticipated that TKA participants would exhibit reduced knee extension moments at the operated limb and increased adduction moments at the hip and knee of the non-operated leg, consistent with findings from previous studies with level walking. It may be argued that further differences between the TKA and control groups would be expected since downhill walking requires greater physical demands. Previous investigations have studied healthy individuals during downhill walking and reported reduced flexion angles, as well as increased knee extension moments and peak power absorption at the knee (Kuster, et al., 1995; McIntosh, et al., 2006; Redfern & DiPasquale, 1997). These findings led us to predict that TKA participants would exhibit significantly reduced knee flexion angles and knee extension moments, compared to controls, during downhill walking. The increased demands for related to mechanical power

during downhill walking also guided us to predict TKA participants would exhibit reduced power absorption at the knee during downhill walking, compared to controls.

## MATERIALS AND METHODS

Motion analysis was conducted using a 10-camera system (Vicon MX-13, Oxford Metrics, Oxford, UK) and 45 retro-reflective markers placed according to a modified Plug In Gait model (Appendix A, Figure 18). During level walking trials, ground reaction force was recorded using four force plates (Bertec Corporation, models FP4060-8 & FP4060-10, Columbus, Ohio; Kistler Instruments Corp, models 9286A & 9286AA, Winterthur, Swtz) embedded within the floor. Trials for downhill walking were conducted using a 4-metre inclined ramp, set at a 12.5% angle (Figure 10); ground reaction forces were recorded by two Kistler force plates (same models as above) that were embedded within the frame of the ramp. Vicon Nexus software (Oxford Metrics, Oxford, UK) was used to collect all motion trials and filter data.



**Figure 10.** Inclined ramp used for downhill walking trials.

## PARTICIPANTS

TKA participants (n=17) were recruited from the Ottawa Hospital, having undergone surgery with the same surgeon at least 6 months prior to beginning participation in the study. Inclusion criteria for this group consisted of having undergone unilateral total knee arthroplasty to treat severe knee osteoarthritis, without suffering from osteoarthritis or any other diseases/conditions at any of the other joints in the lower limbs. Participants were also selected based on being between the ages of 50-75 and having a BMI below 35 kg/m<sup>2</sup>. All participants in the surgical group underwent TKA procedures with the same surgeon; ten participants received a Medial Pivot (MP) prosthesis, while seven had a Posterior Stabilized (PS) knee. Control participants (n=17) were recruited from the local community, using the same inclusion criteria, with the restriction of no lower limb joint disorders/disease. Ethics approval was granted from the university and hospital governing boards, and all participants provided informed consent prior to beginning the motion analysis study.

## PROTOCOL & DATA ANALYSIS

Each participant performed a minimum of five trials for both walking tasks. Level walking was performed on a 6 metre level surface, while downhill walking trials were performed on an inclined ramp, beginning on a level platform at the top of the ramp. To maintain consistency between trials, participants began each trial two steps away from the first plate, stepping with the operated limb first, so that the operated leg struck the first force plate. This two-step distance was determined individually for each participant based on their natural walking speed. Joint kinetics, moment of force and power, were calculated using an inverse dynamics approach and were normalized for body mass and for gait cycle. The raw three-dimensional (3D) marker trajectories were filtered using a Woltring filter (predicted mean square error value of 15

mm<sup>2</sup>), whereas a low pass, 4<sup>th</sup> order, zero-lag Butterworth filter (cut-off frequency of 10 Hz) was applied to the ground reaction forces. Individual averages for each control participant were calculated across both lower limbs; these averages were then used to calculate an overall group average for the control group with respect to each variable. Based on this group average, a normative range was established for each measure, calculated using the group means, as seen with Kaufman et al. (2001). The area calculated for this normative range was set as one standard deviation above and below the mean at each point. If the movement pattern of the TKA group fell outside this normative range for any of the kinematic/kinetic measures, it was considered to be an area of interest that warranted further analysis.

### STATISTICAL ANALYSIS

Statistical analysis was conducted using SPSS 20.0 software (IBM, Armonk, U.S.A). All measures were first analyzed for outliers and distribution patterns, where it was observed that the greater variability inherent in frontal plane mechanics resulted in a non-normal distribution for most kinematic and kinetic measures, based on the Shapiro-Wilk test. As this distribution violated a basic assumption for parametric tests, kinetic and kinematic measures from the frontal plane were analyzed using non-parametric tests (described below). For each statistical analysis, the confidence interval was set at 95%, with an alpha=0.05.

#### *Parametric Tests*

Due to the combination of males and females within the same groups, we felt it was appropriate to determine whether sex would influence any of the outcomes used in our analysis. As a result, a sex\*group comparison was included in our statistical approach. Comparisons between the operated limb and controls were conducted using two-way ANOVAs for joint angles, moments of force, and power. Comparisons between operated and non-operated limbs of

TKA participants were done using 2x2 mixed ANOVAs with ‘sex’ as the between-subject measure and ‘leg’ (operated vs. non-operated) as the within-subject measure. Leg type was selected as a within-measure as it was anticipated that the lower limbs of the TKA participants were not independent of each other. Outcomes from these tests revealed no significant effect for sex; therefore, no further comparison was made between males and females for these kinematic and kinetic measures.

### *Non-Parametric Tests*

The influence of sex for frontal plane measures was analyzed using Mann-Whitney U tests for each measure. No significant effect was found for sex with any of the variables of interest; as a result, no further comparisons were made between males and females for these measures. Between-group comparisons for frontal plane mechanics were conducted using Mann-Whitney U tests for independent groups (operated limb vs. control) while Wilcoxon Rank tests were used to compare related groups (operated vs. non-operated limbs).

As this study did not seek to make comparisons between TKA patients with MP or PS knees, no direct analysis was made between these groups for the analysis of post-operative outcomes. Due to the small sample size of this project, it was not deemed statistically sound to make comparisons between these subgroups; however, it should be noted that Mann-Whitney U tests were used initially in this study to determine if the combination of prosthetic designs would skew our results. This analysis revealed no significant difference between the MP and PS participants for any of the kinematic or kinetic measures during both level and downhill walking. Based on these findings, it was deemed appropriate to combine the two subgroups to form one ‘surgical’ group for the purpose of this study.

## RESULTS

Despite efforts to match TKA and control participants, the TKA group had a higher average body weight and BMI compared to the controls (Table 3). This is not unexpected as TKA participants may have had to reduce physical activity in the final stages prior to surgery due to pain and lack of functional ability with their knee.

**Table 3.** Physical parameters of participants for gait tasks

Group	N (Male/Female)	Age (years)	Height (cm)	Weight* (kg)	BMI* (kg/m <sup>2</sup> )
TKA	17 (13/4)	62±6	172±9	90±16	30±3
Control	17 (12/5)	63±8	169±9	77±11	27±4

\*statistically significant (p<0.05)

To account for this potential influence of body mass, BMI was run as a covariate in the analyses across groups; however, no significant effect was found. As all kinetic measures in this study were normalized for body mass prior to group comparisons, we do not believe these group differences have impacted the results from our study. A breakdown of the surgical group is shown in Table 4 below, outlining physical characteristics for participants with either of the two knee prostheses. It can be seen that there was no significant difference between groups for any of the previously described physical parameters.

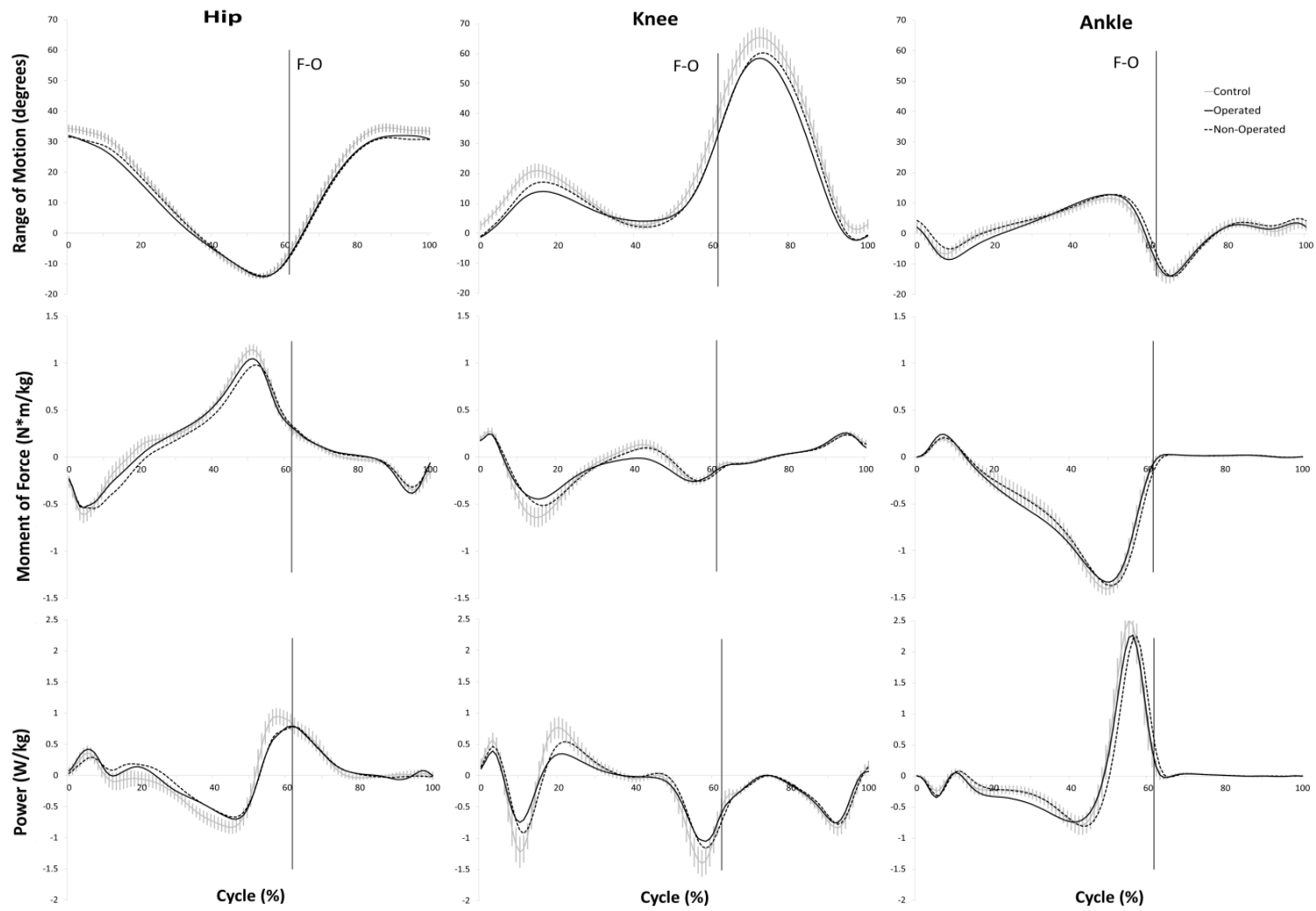
**Table 4.** Physical parameters of participants within surgical group.

Group	N (Male/Female)	Age (years)	Height (cm)	Weight (kg)	BMI (kg/m <sup>2</sup> )
MP	10 (8/2)	60±5	171±1	91±19	30±4
PS	7 (5/2)	64±6	173±0.8	88±11	30±3

## LEVEL WALKING

Group averages for lower limb kinematic and kinetic parameters occurring in the sagittal plane during level walking are shown in Figure 11. Each figure includes a normative range, representing the range of the standard deviation for the control group. Any values for the TKA group that fell outside this normative range were further investigated for group differences. As can be seen in Figure 11, only two measures (peak knee flexion and hip flexion) fell outside the normative range during the swing phase of gait. Of these two variables, only the reduced peak knee flexion was found to be significant ( $p=0.001$ ), with an average peak flexion of  $59.0^{\circ}\pm 4.5$  at the operated limb,  $80.9^{\circ}\pm 5.4$  at the non-operated limb, and  $65.6^{\circ}\pm 4.8$  at the control limb. As no other deviations (from the normative range) were observed, the remainder of our analysis for gait mechanics focused on measures in the stance phase of gait.

TKA participants exhibited significantly reduced range of motion, peak flexion angle, peak flexion moments and power generation at the operated knee, compared to the controls, during stance phase (Table 5). Two peaks for knee power generation were analyzed, with the first peak (PG1) occurring during the loading response period of gait (0-12% of gait cycle), coinciding with a reduced knee flexion angle that was observed throughout stance phase. This discrepancy in power generation reached statistical significance ( $p=0.01$ ) compared to controls (Figure 11, Table 5).



**Figure 11.** Kinematic and kinetic patterns in the sagittal plane during level walking. F-O indicates the point of foot-off which occurred at approximate 62% of the gait cycle.

**Table 5.** Kinematic and kinetic measures at the knee during level walking

	Op	Non-Op	Control	p-value
Kinematics (°)				
Knee ROM	16.0±4.2*	18.3±5.1	20.3±5.9	0.013
Knee Flexion	14.1±5.4*	17.2±7.0	20.6±6.4	0.009
Moments (N*m/kg)				
Knee Flexion	0.00±0.15* <sup>€</sup>	0.11±0.13	0.14±0.13	0.003
Power (W/kg)				
Knee PG1	0.40±0.2*	0.51±0.3	0.66±0.3	0.01
Knee PG2	0.38±0.3* <sup>€</sup>	0.57±0.3	0.80±0.5	0.004

PG1 – power generation during 0-12% of gait cycle, PG2- power generation at 12-30% gait cycle

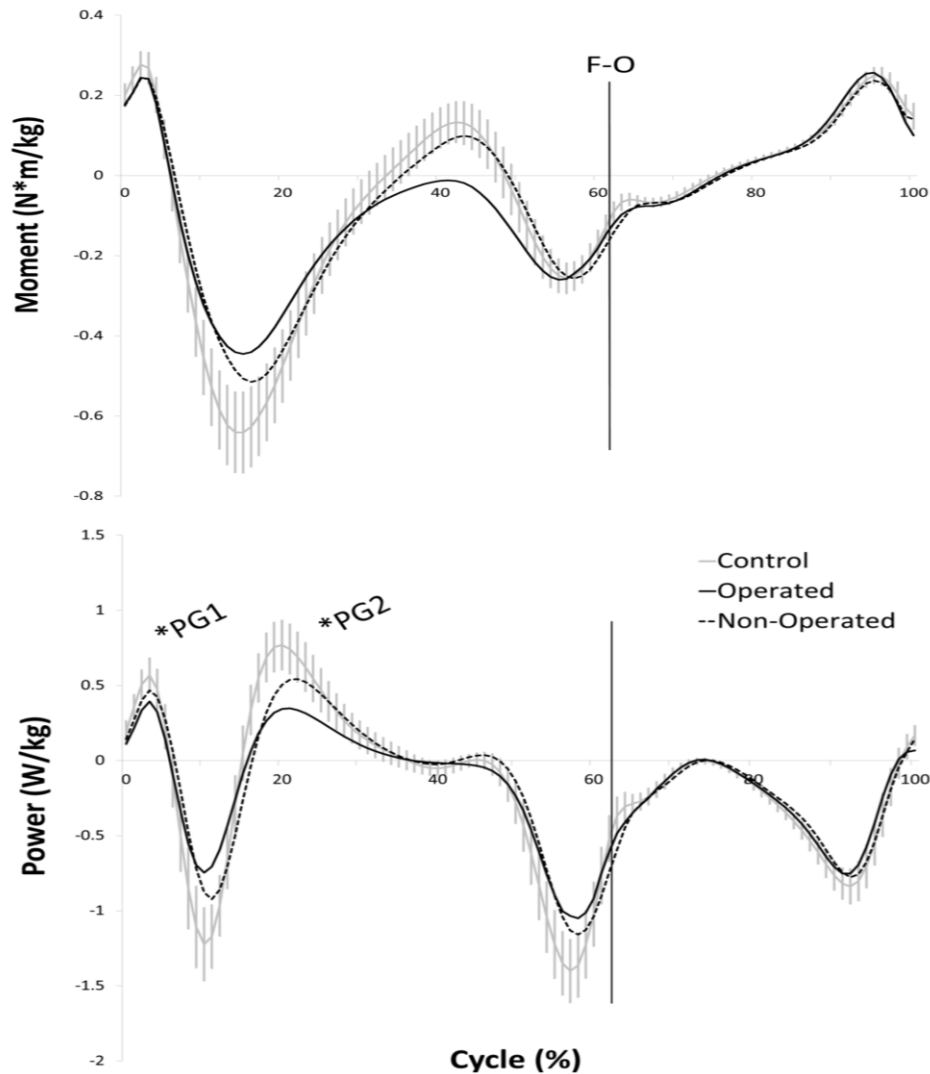
\* significant difference between operated and control ( $p < 0.05$ )

<sup>€</sup> significant difference between operated and non-operated lower limbs ( $p < 0.05$ )

The reduced knee flexion moment for the operated limb reached a group mean of 0.004 N·m/kg, remarkably smaller than the 0.11 N·m/kg and 0.14 N·M/kg peak values for the non-operated and control limbs. The small flexion moment observed at the operated limb was due to some participants remaining in an extension moment throughout the gait cycle. Further analysis revealed that differences between male and female participants may have largely contributed to this small peak flexion moment. All female participants exhibited reduced knee flexion moments compared to their male counterparts; however, this difference is most pronounced at the operated limb as the female participants tended to remain in an extension moment while males had a positive flexion moment. It should be noted, however, that differences between male and female participants did not reach statistical significance for the operated, non-operated, or control limbs, nor was there a significant interaction between sex and group as independent variables for the ANOVAs. At the operated limb, male participants had a peak flexion moment of 0.05±0.14 N\*M/kg, significantly lower than the non-operated (0.12±0.25 N\*m/kg) and control limbs (0.11±0.1 N\*m/kg) for all males. For female participants, the operated limb also had a significantly smaller moment, with a peak value of -0.12±0.13 N\*m/kg, compared to 0.08±0.16 N\*m/kg for the non-operated limb and 0.11±0.1 N\*m/kg for the controls. The lack of a

statistically significant difference between male and female participants, despite a large discrepancy in group means, may be a result of the small sample size (particularly for female participants). This should be kept in mind for future research that includes male and female TKA patients.

The reduced knee flexion angle observed during stance phase continued from loading response through to mid-stance, coinciding with the second peak in power generation (PG2) at 12-30% of the gait cycle, where the operated limb once again experienced significantly lower power generation at the knee compared to both the non-operated and control limbs ( $p=0.004$ ). The operated limb also exhibited lessened power absorption and extension moment at the operated knee during this stage of the gait cycle; however these differences were not found to be statistically significant (Figure 12).

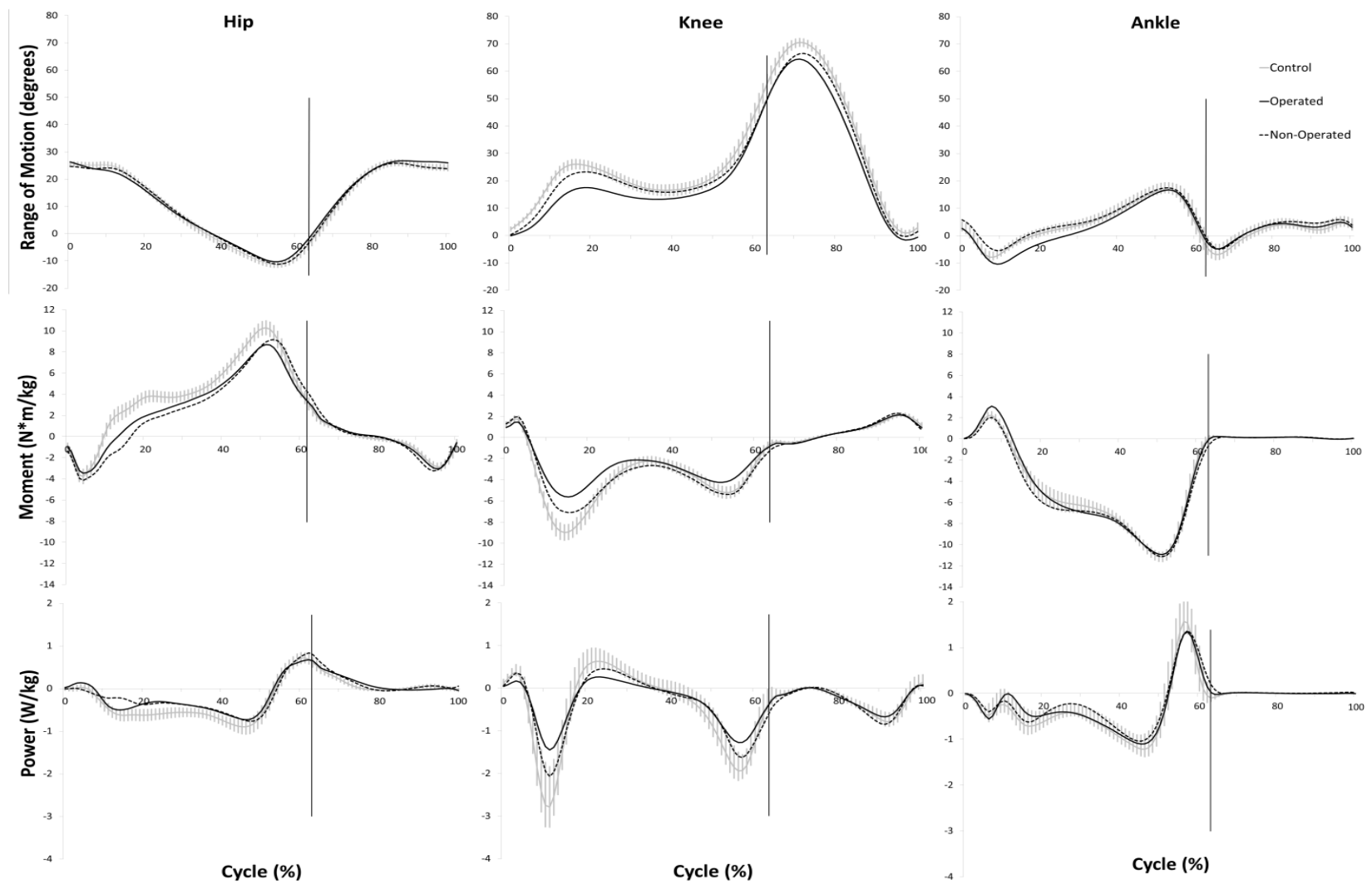


**Figure 12.** Knee flexion/extension moment and power during level walking. PG1= power generation during loading response period; PG2=power generation during mid-stance. \* $p < 0.05$ .

### DOWNHILL WALKING

Figure 13 shows gait parameters for the sagittal plane during downhill walking. Similar to the level walking task, the only peak value that was outside the normative range during swing phase was peak knee flexion. As a result, all further group comparisons related to peak value were isolated to focus on stance phase of gait. It should be noted that the difference in peak knee flexion during swing phase was significant ( $p=0.009$ ), where the operated limb of TKA participants experienced significantly reduced peak knee flexion (mean= $65.5 \pm 6$ ) when

compared to both the non-operated limb (mean=67.4°±6) and controls (mean=70.9±4). Variables that were found to fall outside the normative range are shown in Table 6 for downhill walking.



**Figure 13.** Kinematic and kinetic measures throughout gait cycle during downhill walking. F-O indicates the point of foot-off which occurred at approximate 62% of the gait cycle.

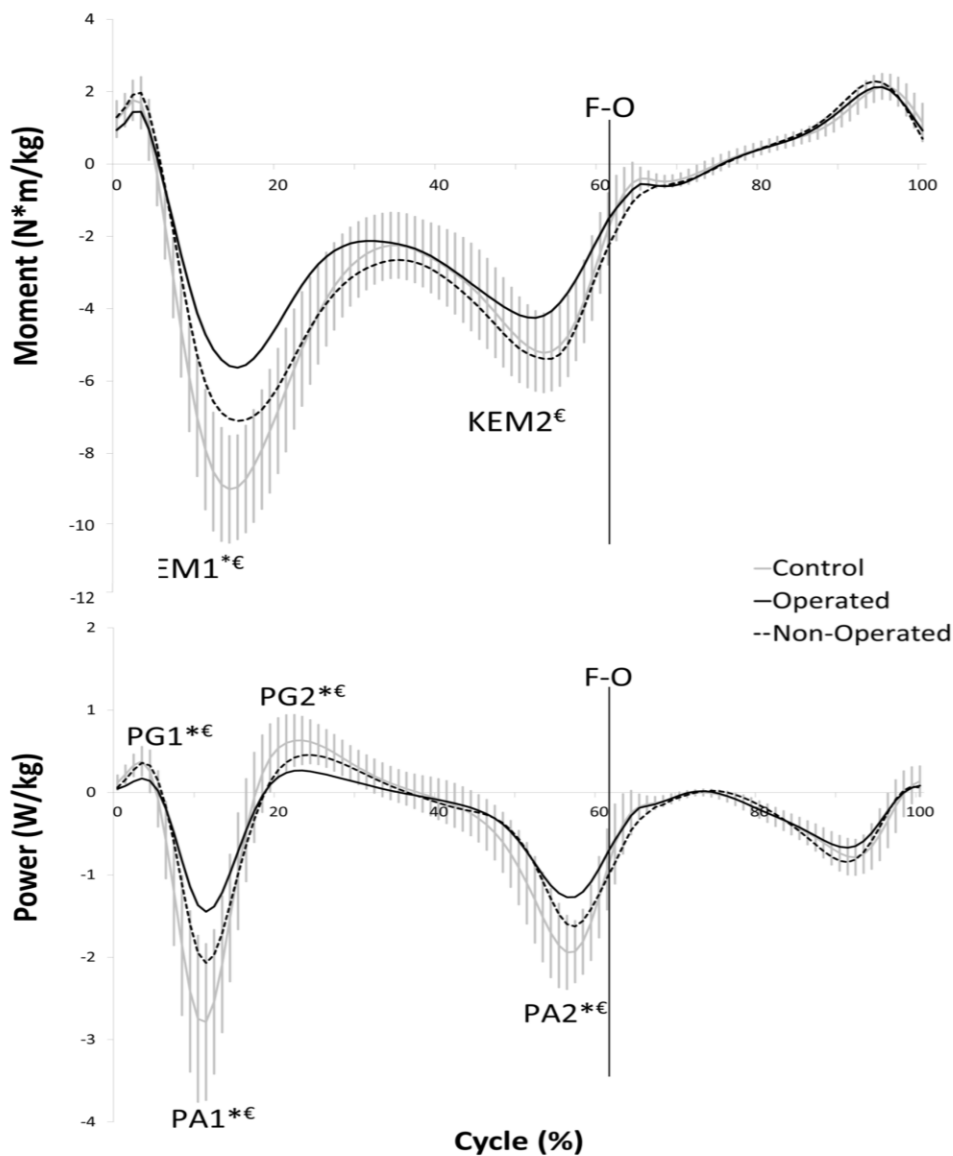
**Table 6.** Kinematic and kinetic measures during downhill walking

	Op	Non-Op	Control	p-value	
				Op vs. Control	Op vs. Non-Op
<b>Kinematics (°)</b>					
Knee ROM	42.8±7.0 <sup>*</sup>	42.4±5.3	48.6±4.7	0.01 (0.76)	0.537 (0.091)
Knee Flexion	18.1±5.8 <sup>*€</sup>	23.2±8.8	26.3±6.6	0.004 (0.857)	0.008
Dorsiflexion	2.6±4.4 <sup>€</sup>	5.9±4.8	2.8±4.7	0.865	0.006
<b>Moments (N*m/kg)</b>					
KEM1	-0.58±0.33 <sup>*€</sup>	-0.75±0.47	-0.91±0.35	0.022 (0.645)	0.009
KEM2	-0.45±0.23 <sup>€</sup>	-0.58±0.34	-0.53±0.15	0.322	0.007
PFM	-1.11±0.16 <sup>€</sup>	-1.13±0.40	-1.13±0.22	0.502	0.015
<b>Power (W/kg)</b>					
Knee PG1	0.20±0.2 <sup>*€</sup>	0.39±0.3	0.43±0.3	0.004	0.002
Knee PA1	-1.50±1.0 <sup>*€</sup>	-2.20±1.6	-2.90±1.5	0.02 (0.661)	0.012
Knee PG2	0.32±0.3 <sup>*€</sup>	0.56±0.5	0.72±0.4	0.003	0.001
Knee PA2	-1.40±0.7 <sup>*€</sup>	-1.70±1.0	-2.00±0.7	0.016 (0.698)	0.022

KEM1 – knee extension moment at 12-30% of gait cycle, KEM2 – knee extension moment during 50-62% gait cycle, PFM – peak plantar flexion moment, PG1 – power generation during 0-12% of gait cycle, PG2- power generation at 12-30% gait cycle, PA1- power absorption at 12-30% gait, PA2 – power absorption during 50-62% of gait cycle

<sup>\*</sup> significant difference between operated (TKA) and control; <sup>€</sup> significant difference between operated and non-operated (TKA)

The two designated values for knee extension moments represent the peak observed during mid-stance (KEM1, at 12-30% of gait cycle) and the peak that continues from late-stance to pre-swing periods (KEM2). Peak power values at the knee were labelled based on order of occurrence throughout the gait cycle, where PG1 and PG2 represent the first two positive peaks for knee power, consistent with level walking measures. The two peaks described for power absorption refer to those observed during mid-stance, described as PA1, and the peak that occurs during pre-swing, described as PA2. The peak values for each of these measures are shown in Figure 14.



**Figure 14.** Knee flexion/extension moment and power during downhill walking. KEM1=knee extension moment during early stance, KEM2=knee extension moment during late-stance and pre-swing, PG1=power generation during loading response; PG2= power generation during midstance, PA1=power absorption in midstance, PA2=power absorption in pre-swing. \*significant difference between operated and controls ( $p<0.05$ ); €significant difference between operated and non-operated limb ( $p<0.05$ )

Based on results shown in Figure 13 and Table 6, it can be seen that some of the altered gait patterns observed with the TKA group during level walking carried over to the inclined walking task. During loading response, TKA participants continued to exhibit reduced knee power generation at the operated limb, which was significantly different from both the non-operated ( $p=0.002$ ) and control limbs ( $p=0.004$ ). During mid-stance, the operated limb once

again exhibited smaller knee flexion angles and peak extension moments, along with reduced power absorption at the knee. As the non-operated limb entered the second phase of double support, during pre-swing, the operated limb exhibited significantly lower peak power absorption (PA2) compared to both the non-operated and control limbs. In addition to the differences, the non-operated limb experienced significantly greater dorsiflexion angles during loading response ( $p=0.006$ ), as well as higher knee extension moments and hip power generation during terminal-stance and pre-swing, compared to the operated limb.

Analysis across the frontal plane revealed no significant differences for kinematic and kinetic measures between the operated, non-operated, and control limbs during level or downhill walking.

## DISCUSSION

Contrary to our hypothesis, no significant difference was observed between the TKA and control groups (or between the operated and non-operated limbs) for adduction moments, or any other frontal plane kinematics/kinetics during either level or downhill walking. Based on the association between increased knee adduction moments and OA severity, these findings are promising for the ability of TKA to reduce overloading of the medial knee compartment for individuals suffering from knee OA.

Consistent with our hypothesis, TKA participants did exhibit reduced extension moments at the operated limb during stance phase for walking on both the level and inclined walkways; however, this difference was only significant during downhill walking. During both walking tasks, these smaller knee extension moments coincided with reduced peak power absorption values at the operated limb. During this phase of gait, the knee was flexed and the extensor muscles were undergoing eccentric contraction as they created an extension moment. The

significantly reduced knee flexion angles experienced by the operated limb are likely associated with the reduced extension moments observed during stance phase. This is similar to the findings of Benedetti et al. (2003) who associated reduced knee flexion angles and extension moments with prolonged co-activation of the rectus femoris-hamstring and gastrocnemius-tibialis anterior. Benedetti et al. (2003) suggested that this prolonged muscle activity may be associated with a “stiff knee pattern”, aimed at providing greater control of knee kinematics(during stance phase), in the absence of severe flexion contractures or quadriceps weakness. Further analysis that includes muscle activations patterns would therefore be of great interest to provide insight into these observed discrepancies for flexion/extension moments among the TKA participants.

The smaller peak knee flexion angles observed at the operated limb were consistent with our predictions for downhill walking, but were not anticipated during level walking. Previous studies looking at level walking with TKA participants have reported decreased knee flexion among TKA patients (Benedetti, et al., 2003; Ishii, et al., 1998; Smith, A. J., et al., 2006; Yoshida, et al., 2008), compared to non-operated limbs or controls; however, other studies have reported conflicting results with TKA patients exhibited greater knee flexion (Berth, 2002).

At the midstance period of the gait cycle the operated limb would be in a single support phase, with the knee extensor muscles serving to stabilize and support the limb to prevent it from collapsing as it supports the body’s weight. The fact that discrepancies for knee extensor moment and power absorption appear to be greater during downhill walking, with peak values almost doubling for all groups, indicates that this more demanding task brings about greater differences between the groups, with quadriceps function being further hindered. Further investigation is warranted to include EMG measure to understand the role of muscle activation (or muscle strength) with these gait patterns.

Two outcomes that were not anticipated were the decreased knee flexion moments observed at the operated limb during level walking, as well as the reduced power generation observed at the operated limb during both walking tasks. The lower values for power generation observed at the knee during both walking tasks coincided with periods of reduced knee flexion. It is possible that the reduced range of motion exhibited by TKA patients during this phase resulted in a diminished ability to generate power at the knee and may also have contributed to the discrepancies in knee power absorption. During level walking, the operated limb exhibited significantly lower peak knee flexion moments compared to both the non-operated and control limbs. In fact, at this phase in the gait cycle, several TKA participants remained in a negative moment indicating that they continued to undergo an extension moment at the knee. While it is unclear why this difference occurred for some participants, it may be related to knee extension angles. The operated limb appeared to exhibit reduced knee extension during stance phase, although this difference was not statistically significant. It is possible that individuals who experienced reduced knee extension would thereby have a smaller knee flexion moment at this time.

While overall patterns for lower limb mechanics were similar between the two walking tasks, downhill walking required greater knee flexion excursion, as well as greater peak knee flexion/extension moments and greater knee power absorption. On the other hand, peak values for kinematics and kinetics and the hip and ankle were relatively similar between walking tasks. These changes are consistent with findings from Redfern & DiPasquale (1997) who compared level and downhill walking with healthy individuals, reporting that the most pronounced changes in lower limb mechanics (between walking tasks) occurred at the knee. These findings illustrate

why inclined walking tasks are warranted for future research examining gait mechanics with TKA individuals.

During downhill walking, the non-operated limb experienced increased ankle dorsiflexion angle during the period of loading response. This timing coincided with the first phase of double support for the non-operated limb, while the operated limb was in the pre-swing period of gait. Levinger et al. (2008) also reported greater dorsiflexion angles at the non-operated limb during initial phases of foot contact for gait; however, their analysis was conducted on level walking and the discrepancy between limbs did not reach statistical significance for these kinematic measures. As previously discussed, the operated limb experience reduced knee extension moments and power generation during this phase; therefore, the increased dorsiflexion may reflect compensation at the non-operated limb as the body is lowered down the ramp. The increased knee extension moment observed at terminal stance for the non-operated limb occurred during the second period of double support for stance phase in the gait cycle. At this time the operated limb is in the first 10% of its own gait cycle, undergoing a period of reduced power generation and flexion at the knee (previously discussed). This greater extension moment at the non-operated limb may, therefore, be a means of compensation to absorb mechanical power and support the limb as it flexes and prepares for swing phase.

When considering the results from this study, some limitations should be noted, including the smaller sample size and the use of two different prosthetic designs. While a comparison of medial pivot and posterior stabilized knees would be of interest in this research, it was believed that the sample size was too small to avoid Type 1 and Type 2 errors. Future studies with potential for larger sample sizes should keep this comparison in mind, to better understand outcomes for the newer medial pivot design.

In conclusion, our study found that functional discrepancies remain for lower limb gait mechanics one year after TKA. Our results have shown that patterns of reduced knee flexion angles, knee flexion/extension moments, and lower generation and absorption of power at the knee persist with the operated limb during level and downhill walking tasks. Our results also show that downhill walking exacerbates some discrepancies observed during level walking, such as lower extension moments and power absorption at the operated knee. These findings warrant the inclusion of inclined walking surfaces to be included in future gait analysis with TKA participants to provide a more realistic understanding of the daily demands placed on the lower limbs during walking.

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## **ARTICLE 2**

# Biomechanical Analysis of Sit-Stand Tasks Following Total Knee Arthroplasty

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## ABSTRACT

Total knee arthroplasty (TKA) has been proven to effectively reduce pain and improve function for those suffering from severe osteoarthritis (OA); however, functional discrepancies remain. The sit-stand task is often used to measure functional capacity and symmetry for healthy and diseases populations, yet limited research has focused on biomechanical patterns used by TKA patients to transition between sitting and standing. The purpose of this study was to examine biomechanical parameters of standing up and sitting down on a seat for individuals at 1-year after undergoing TKA. The TKA group (N=15, age=62±6 years, BMI=30±3 kg/m<sup>2</sup>) consisted of individuals who underwent unilateral TKA with a Medial Pivot (MP) or Posterior Stabilized (PS) knee, and suffered from no other lower limb conditions/diseases. Controls (N=15, age=63±5 years, BMI=26±4 kg/m<sup>2</sup>) were recruited from the local community and matched to TKA participants based on age and BMI. Each participant performed 5 trials of standing up (sit-to-stand) and sitting back down (stand-to-sit) from an instrumented 18" seat. Compared to controls, TKA participants had reduced knee flexion angles, hip flexion moments, hip abduction angles, and knee abduction moments during sit-to-stand and stand-to-sit. These differences were only statistically significant during sit-to-stand. In addition to these differences, TKA participants exhibited reduced power absorption at the hip and knee during stand-to-sit. The small sample size and combination of two prosthetic designs within the TKA group may have limited findings from this study, where larger sample sizes would allow for comparisons across prosthetic designs. In conclusion, individuals who have undergone unilateral TKA with a MP or PS knee continue to exhibit functional discrepancies, compared to controls, at one year after surgery. The increased physical exertion required to lift the body during sit-to-stand may explain greater discrepancies observed between groups, compared to stand-to-sit.

## INTRODUCTION

As a proven effective treatment for severe knee osteoarthritis (OA), over 22,000 total knee arthroplasty (TKA) procedures are performed annually (CIHI, 2009). As this number continues to grow every year, a greater amount of research is needed to understand the impact this surgery has on improving functional capacity of patients. Numerous studies have looked at post-operative function using gait and stair climbing following TKA; however, the ability to transition from seated and standing positions precedes the needs to either of those activities. This movement has been proven to be a valid measure for knee function (Boonstra, et al., 2008), used in many studies to understand biomechanics of healthy participants, and to determine functional limitations amongst older or pathologic populations (Flanagan et al., 2003; Fotoohabadi et al., 2010; Schenkman, et al., 1990; Turcot, et al., 2012). Previous research looking at the sit-stand task among TKA patients has focused primarily on weight bearing symmetry, reporting asymmetry that exists at pre-operative and early post-operative outcomes but disappears by 6-12 months (Boonstra, et al., 2010; Christiansen, et al., 2011; Farquhar, et al., 2008). Outside of these symmetry measures, studies that have looked at sit-to-stand movement patterns among TKA populations have reported reduced extension angles and moments at both the hip and knee joints at the operated limb during acute phases of recovery (3-6 months). Farquhar et al. (2008) examined TKA participants performing sit-to-stand at 3 months and 1-year post-op, reporting that discrepancies for hip/knee kinematics and kinetics persisted, even though weight bearing symmetry was established. Based on findings that discrepancies in loading of the lower limbs disappears between 6 to 12 months after surgery, we believe that further research examining lower limb biomechanics during this period is warranted. In addition, none of these studies have examined biomechanics of the lower limbs in the frontal plane during sit-stand. Of the two

studies that were found to examine stand-to-sit performance among TKA patients, only one included a comparison between TKA participants and controls (Wang, et al., 2012; Wang, et al., 2005). Therefore, further research is warranted to understand how TKA patients are able to perform functional tasks, compared to age-matched counterparts, as part of their daily activities. The purpose of this study was to examine lower limb biomechanics during both sit-to-stand and stand-to-sit movements at approximately 6-12 months following TKA.

Based on findings by Farquhar et al. (2008), we hypothesized that TKA participants would exhibit greater peak hip flexion angles and moments during sit-to-stand, compared to controls. As a similar degree of hip flexion is required during the stand-to-sit movement task, it was anticipated that this discrepancy between TKA participants and controls would carry over for peak hip flexion angle in this task as well. Given that no previous studies have discussed frontal plane mechanics during sit-to-stand for TKA participants, no specific prediction could be made. As a result, no significant differences were anticipated between TKA and control participants for frontal plane kinematics and kinetic measures during sit-to-stand and stand-to-sit. Based on the prediction of altered hip kinematics and kinetics in the TKA group, it was also predicted that TKA participants would exhibit greater power values at the hip, compared to controls, during sit-to-stand and stand-to-sit. However, no significant differences were anticipated for power at the knee and ankle during either sit-to-stand or stand-to-sit movements.

## METHODS

### PARTICIPANTS

Participants for the surgical group were recruited from the Ottawa Hospital after having undergone unilateral total knee arthroplasty by the same surgeon. Criteria for inclusion were no

injuries/conditions impacting any other joints in the lower limbs, being between the age of 50-75 years, and a BMI < 35 kg/m<sup>2</sup>. Control participants were recruited from the local community and individually matched to each TKA participant based on age and BMI. Inclusion criteria were similar to that of the surgical groups, with no injuries or neuromuscular conditions/disease impacting the lower limbs. Based on these criteria 34 participants were recruited (17 TKA and 17 controls). However, due to equipment-related error, the trials from four of these participants (2 TKA and 2 controls) were excluded from our analysis for the sit-stand task. This left 15 participants for each group. The TKA group was composed of 6 individuals with a Posterior Stabilized (PS) knee and 9 individuals with a Medial Pivot (MP) knee; TKA participants were recruited between 6 to 12 months following surgery. Ethics approval was granted from the research ethics boards at both the Ottawa Hospital and the University of Ottawa, and each participant provided informed consent prior to beginning data collection.

### PROTOCOL

Motion analysis was conducted at the Human Performance Laboratory at the University of Ottawa, using a 10-camera Vicon system (Vicon MX-13, Oxford Metrics, Oxford, UK) and 2 Bertec force plates (Bertec Corporation, models FP4060-8 & FP4060-10, Columbus, Ohio) that were embedded within the laboratory floor. The sit-stand task was performed using a seat set at a 45cm height, in accordance with provincial building codes for standard toilet seat heights.

Each participant performed five successful trials moving from a seated to a standing position, with arms placed in front of the body, then returning to a seated position. Timing for each transition was done based on the instruction of the main researcher performing all data collection. Sit-to-stand tasks were time-normalized from initiation of ‘seat-off’, defined by peak vertical ground reaction force (vGRF), to completion of standing, defined by a plateau in the

velocity of the ASIS markers. The same events were used for the stand-to-sit task but in the opposite order, ending with 'seat-on' (defined by same parameter as seat-off).

### DATA ANALYSIS

The 3D marker trajectories were filtered using a Woltring filter with a predicted mean square error value of  $15\text{mm}^2$ , to obtain kinematic measures. Ground reaction forces were filtered using a 4<sup>th</sup> order, zero-lag Butterworth filter (10Hz cut-off), after which these forces were used to determine kinetic measures through an inverse dynamics approach. Kinetic measures were normalized for body mass and for movement cycle (seat-off to standing/standing to seat-on). Processed trials were then analyzed using a custom-made Matlab script (MathWorks, Natick, United States) that allowed for visualization of trials and extraction of peak values. Individual averages for each control participant were calculated across both lower limbs; these averages were then used to calculate an overall group average for each measure. Based on this group average, a normative range was established for movement cycles, as seen with Kaufman et al. (2001). This normative range was calculated to be within one standard deviation above and below the control group's mean kinematic or kinetic measure at each frame of the cycle. If the movement pattern of the TKA group fell outside this normative range for any of the kinematic/kinetic measures, it was considered to be an area of interest for further statistical analyses. Analysis of sit-to-stand included phases of movement described by Schenkman et al. (1990): momentum transfer, extension, and stabilization.

### STATISTICAL ANALYSIS

Statistical analysis was conducted using SPSS 20.0 software (IBM, Armonk, U.S.A). Prior to conducting group comparisons all data were analyzed for outliers and a normal distribution pattern. Based on results from the Shapiro-Wilk test, the majority of frontal plane

parameters failed to exhibit a normal distribution. As a result, non-parametric tests were used for all further comparisons using measures from the frontal plane. All statistical analyses included a confidence interval of 95% and an alpha value of 0.05.

### *Parametric Tests*

Kinematic and kinetic parameters from the sagittal plane, as well as joint powers were found to have a normal distribution and therefore were analyzed using parametric tests. Based on the combination of male and female participants within the same testing groups, two-way ANOVAs (independent variables: sex, leg) were run for between-group (control vs. operated limb) analyses to determine whether sex would have a significant interaction with group differences. Comparisons between the operated and non-operated limbs of TKA participants were done using repeated-measures ANOVAs with the same independent variables, assuming outcomes from each lower limb were not independent of the other limb. No significant interactions were observed between sex and the selected leg for any of the kinematic or kinetic variables of interest; therefore, no further comparisons were made based on sex.

### *Non-Parametric Tests*

To determine potential influence of sex on statistical comparisons, Mann Whitney U tests were used to compare peak values between male and female participants. Hip abduction angles were found to be significantly different between males and females within each group; therefore, separate group comparisons were made for males and females. No other kinematic or kinetic measures were found to be significantly different between males and females within the TKA and control groups; therefore, no further sex-specific tests were used for these measures. Comparisons between the TKA and control groups were done using Mann Whitney U tests, with leg type (operated vs. control) as the only independent variable. Wilcoxon Ranked tests were

used to compare kinematic and kinetic measures between the operated and non-operated limbs for each TKA participant.

## RESULTS

Physical parameters for participants are shown in Table 7; no significant differences were found for age, height or leg length. The TKA group had a significantly higher average BMI; however, this was not believed to significantly impact group comparisons as all kinetic measures were normalized for body mass.

**Table 7.** Physical parameters for participants in sit-stand task

Group	Sex (Male/Female)	Age (years)	Height (m)	Leg Length (m)	BMI* (kg/m <sup>2</sup> )
TKA (N=15)	12/4	62±6	1.73±0.1	0.86±0.1	30±3
Control (N=15)	9/6	63±5	1.67±0.1	0.87±0.5	26±4

\*significant difference between groups  $p < 0.05$

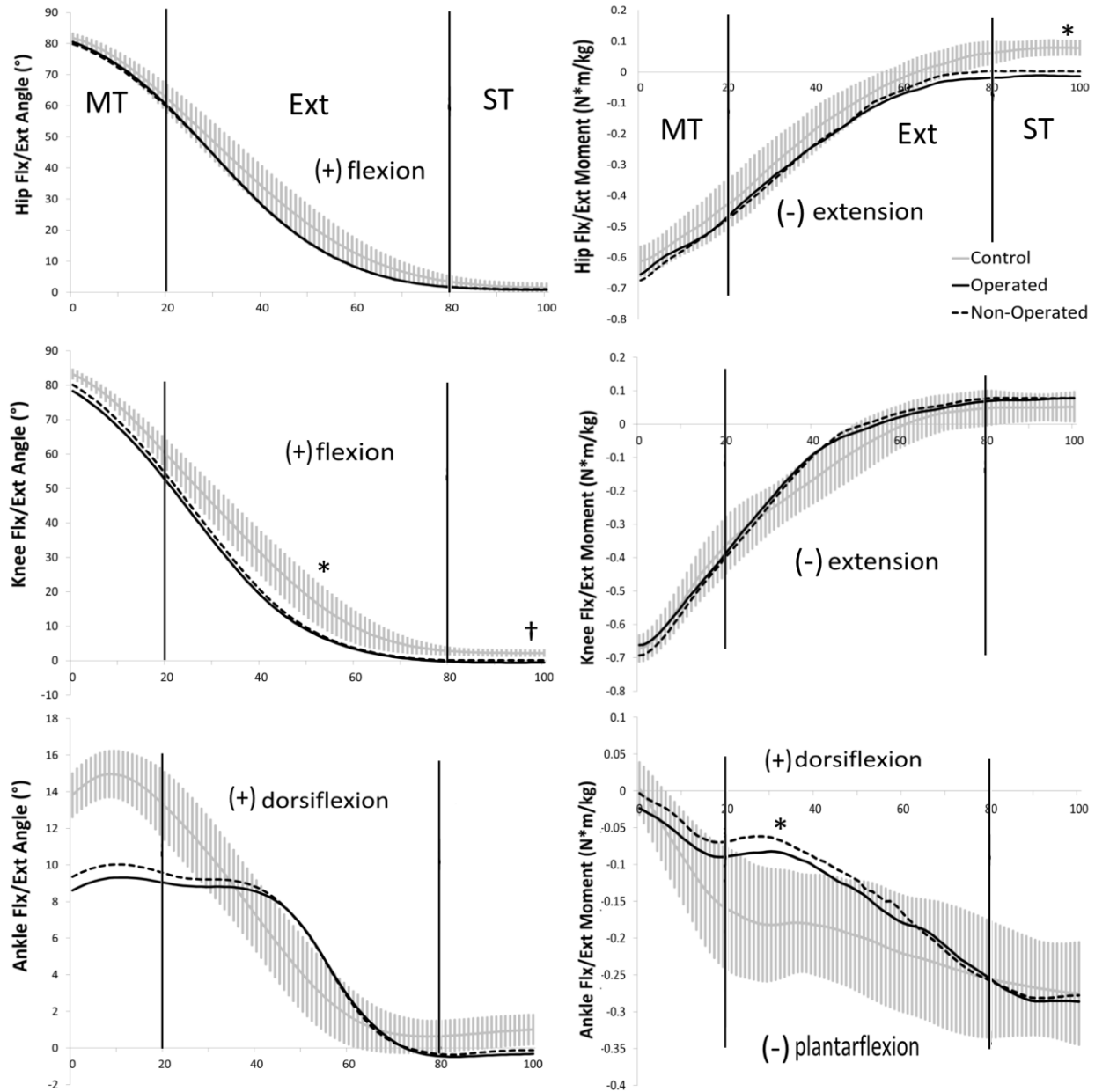
Table 8, below, provide physical characteristics of the MP and PS patients that made up the TKA group for this study. No significant differences were observed for physical characteristics between the two surgical subgroups.

**Table 8.** Physical parameters for participants within surgical group

Group	Sex (Male/Female)	Age (years)	Height (m)	Leg Length (m)	BMI (kg/m <sup>2</sup> )
MP (N=15)	7/2	60±6	1.71±0.1	0.89±0.6	29.5±3
PS (N=15)	4/2	64±6	1.73±0.08	0.82±0.2	30.3±3

Kinematic and kinetic patterns during sit-to-stand are shown in Figure 15; the normative range set by the control group can be seen as the grey shaded area in each graph. Hip and ankle flexion/extension angles are shown to reflect events defining phases of movement. Peak hip flexion and ankle dorsiflexion occurred between 0-20% of the cycle, representing the momentum

transfer (MT) phase. Following peak ankle dorsiflexion, the extension phase (Ext) was defined from 30% of the cycle until hip flexion velocity approached zero at approximately 80% of the movement cycle. During this extension phase, knee flexion angles and dorsiflexion moments appeared to move outside the normative range. As a result, these values were included in the statistical analysis, along with peak dorsiflexion angle and peak hip flexion moment.



**Figure 15.** Kinematic and kinetic measures that fell outside the normative range during sit-to-stand. MT=moment transfer phase, Ext= Extension phase, ST=stabilization phase. \*significant difference between operated and control limbs, †difference between operated and control limbs approached significance (p=0.051)

TKA participants exhibited reduced knee flexion throughout the sit-to-stand cycle; however, this difference only reached statistical significance (p=0.04, CI=95%) during the ‘extension’ phase of the movement cycle, with a minimum knee flexion of  $3.3 \pm 6.2^\circ$  at the operated limb and  $8.4 \pm 6.5^\circ$  at the control limb. It should be noted that differences in minimum

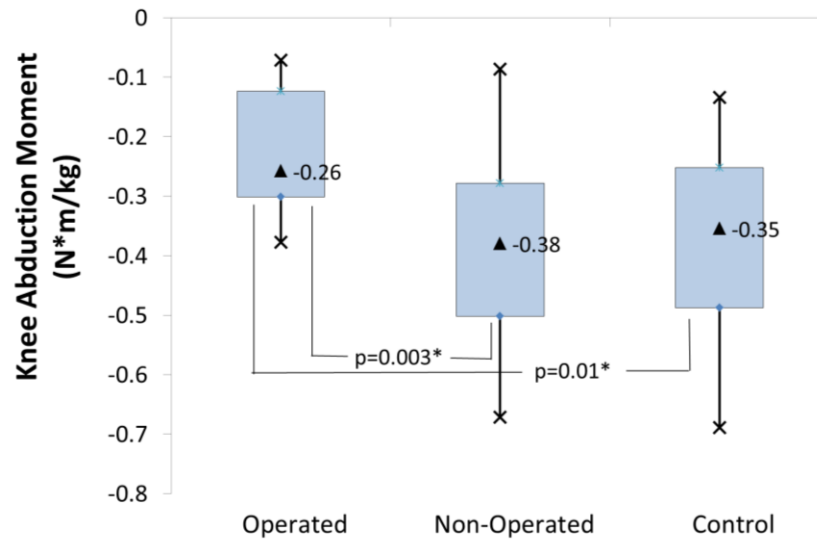
peak values for knee flexion approached significance ( $p=0.051$ ,  $CI=95\%$ ) in the final, stabilization (ST) phase of sit-to-stand as well. Minimum peak values for angles at the knee were  $-0.72\pm 3.2^\circ$  at the operated limb and  $-0.36\pm 5.4^\circ$  at the non-operated limb (negative values indicate knee extension), while peak minimum value for control participants was  $2.0\pm 4.0^\circ$  with the positive value indicating that the knee remained in flexion.

The TKA group also exhibited significantly smaller ( $p=0.02$ ,  $CI=95\%$ ) plantar flexion moments during the extension phase of movement, with minimum peak moments of  $-0.04\pm 0.1$  N\*m/kg at the operated limb,  $-0.03\pm 0.1$  N\*m/kg at the non-operated limb, and  $-0.14\pm 0.1$  N\*m/kg for controls.

In the stabilization phase of sit-to-stand (80-100% of cycle), the TKA participants were observed to have significantly smaller hip flexion/extension moments compared to controls. As can be seen in Figure 15, an initial extension moment is observed at seat-off, followed by a transition towards a final hip flexion moment in standing. The control group reach a maximum peak hip moment of  $0.09\pm 0.1$  N\*m/kg, with the positive value indicating a flexion moment. On the other hand, maximum peak moments at the operated and non-operated limb were  $-0.34\pm 0.1$  N\*m/kg and  $-0.02\pm 0.1$  N\*m/kg, respectfully; negative values indicate that TKA participants remained in an extension moment even at their final standing position. The difference between the peak hip moments was found to be significant ( $p=0.032$ ,  $CI=95\%$ ) between the operated limb and controls.

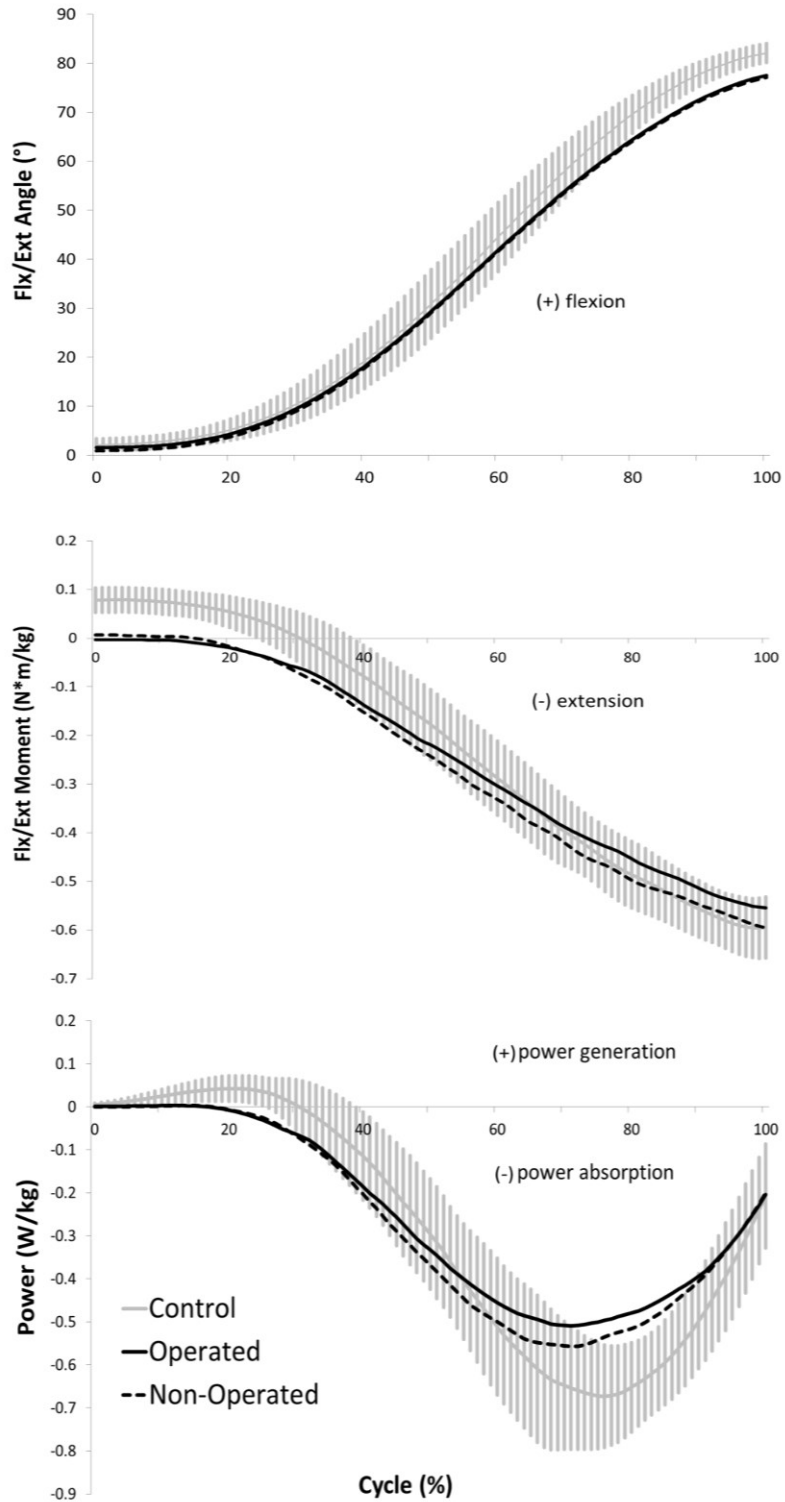
Analysis of peak values in the frontal plane revealed that the operated limb for TKA participants experienced significantly lower peak knee abduction moments than the non-operated ( $p=0.01$ ) and control limbs ( $p=0.003$ ), see Figure 16. Median values (25<sup>th</sup> percentile, 75<sup>th</sup>

percentile) for these groups were operated= -0.26 N\*m/kg (-0.3,-0.12), non-operated= -0.38 N\*m/kg (-0.5,-0.28), and control= -0.35 N\*m/kg (-0.49,-0.25).



**Figure 16.** Median knee abduction moment (25<sup>th</sup> quartile, 75<sup>th</sup> quartile) during sit-to-stand  
\*statistically significant (p<0.05)

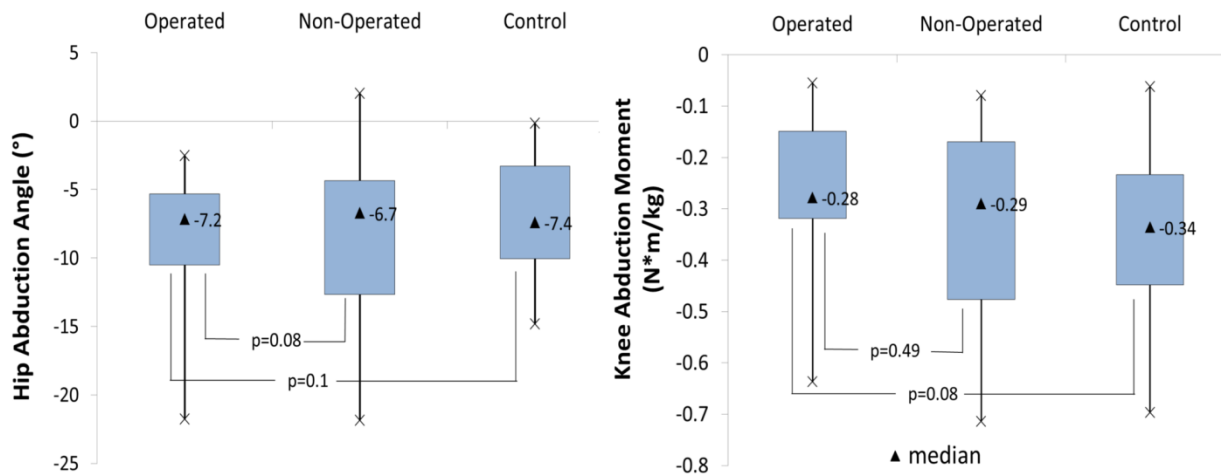
Analysis of stand-to-sit mechanics for the sagittal plane revealed several peak values that fell outside the normative range, primarily at the hip and knee joints. Kinematic and kinetic patterns for the hip joint are shown in Figure 17, with TKA participants experiencing reduced flexion angles and flexion moments, as well as reduced power absorption/generation. However, these differences were not statistically significant.



**Figure 17.** Hip mechanics during stand-to-sit (sagittal plane).

The only other measure that fell outside the normative range was power absorption at the knee, as TKA participants exhibited reduced power absorption at the operated ( $-0.62 \pm 0.2$  W/kg) and non-operated limbs ( $-0.7 \pm 0.34$  W/kg), compared to controls ( $-0.81 \pm 0.4$  W/kg); these differences did not reach statistical significance ( $p=0.14$ , CI=95%).

Examination of frontal plane mechanics revealed that TKA participants exhibited greater peak hip abduction angles and reduced knee abduction moments just prior to seat-on. These discrepancies approached statistical significance ( $p < 0.1$ ) between the operated and control limbs (Figure 18).



**Figure 18.** Median (25<sup>th</sup> percentile, 75<sup>th</sup> percentile) peak hip abduction angle and knee abduction moment during stand-to-sit.

Group medians for frontal plane mechanics that fell outside the normative range during both the sit-to-stand and stand-to-sit tasks are shown as additional tables in the Appendix (Table 9 & 10). While no significant differences were observed (other than those mentioned for sit-to-stand) it was deemed important to include peak values as a reference for future studies.

## DISCUSSION

The principal findings of this study are that TKA participants exhibited significantly different biomechanics during sit-to-stand compared to age-matched controls. Contrary to findings from previous studies, TKA participants did not exhibit greater hip flexion angles or moments during sit-to-stand. In fact, both lower limbs for TKA participants failed to reach a hip flexion moment during sit-to-stand, remaining in an extension moment throughout the movement cycle. During stand-to-sit, TKA participants experienced reduced hip flexion angles and moments compared to control participants, contradicting our initial hypothesis. While it was predicted that TKA participants would exhibit greater power generation at the hip during the sit-to-stand task, our results showed that this group experienced no significant difference during sit-to-stand and reduced power generation and absorption values during stand-to-sit. These outcomes for power during stand-to-sit can be linked to the smaller values for peak hip flexion angles and moments observed with the TKA group during this movement. Contrary to our null hypothesis, TKA participants exhibited discrepancies in the frontal plane parameters for both sit-to-stand and stand-to-sit, including reduced knee abduction moments, as well as reduced hip abduction angles and moments compared to controls.

The reduced peak values for knee abduction moment observed at seat-off for the operated limb may indicate greater loading at the medial aspect of the knee, consistent with previous findings for TKA and OA individuals during gait. It has been widely reported that both these affected groups exhibit a greater external adduction moment during the stance phase of gait, associated with greater loading of the medial compartment and, consequently, progression of OA. As no studies could be found that discussed frontal plane moments for sit-to-stand, we were unable to find any references that might explain this altered moment pattern for the selected task.

While the majority of the sit-to-stand movement is performed with an abduction moment at the knee, the reduced peak abduction value may be related to similar mechanisms or joint loading as the reported increased adduction moment in gait.

Other unanticipated outcomes from this study included the reduced knee flexion angles and increased plantar flexion moments that were observed just after seat-off with TKA participants during sit-to-stand.

While no significant differences were observed between the TKA and control groups during stand-to-sit, some patterns for the operated limb appeared to carry over from sit-to-stand. The operated limb continued to exhibit reduced knee flexion angles, as well as reduced moments of force for hip flexion and knee abduction. Discrepancies for knee flexion angle and hip flexion moment occurred at the beginning of the stand-to-sit cycle, when the individuals were beginning to lower the body from the initial standing position. As the movement in this period is similar to the period at the end of sit-to-stand, it is likely a continuation of the altered movement patterns observed during the sit-to-stand trials. Similarly, the reduced knee abduction moments were observed at the end of stand-to-sit, prior to seat-on, and likely related to mechanics for knee abduction moments just after seat-off in sit-to-stand. Therefore, the sit-to-stand task brought out a greater degree of differences in lower limb biomechanics between TKA and control participants, compared to the stand-to-sit; however, the types of altered movement patterns were relatively similar between both tasks. One unique feature of the stand-to-sit outcomes was that discrepancies were observed for power absorption at the hip and knee between TKA and control groups. This was not observed during sit-to-stand and, therefore, may be a reflection of the different demands to stabilize and control the body as it is lowered to the seat.

Compared to the study by Farquhar et al. (2008) who looked at TKA patients at one year post-op, our subjects demonstrated lower hip flexion angles, along with higher values for peak hip and knee extension moments during sit-to-stand. As our participant groups had a similar mean age and BMI, these physical characteristics are not likely to be responsible for the observed differences between studies. It is possible that the differences in peak values are a result of different seat heights, as Farquhar et al. (2008) set the seat based on knee height. It may also be possible that prosthetic design played a role in our different results, as Farquhar et al. (2008) did not disclose what prostheses were used for their study.

With respect to prosthetic design, this study did include TKA participants with two different types of prostheses (MP and PS knees), which should be considered as a limitation when interpreting these results. Ideally, TKA studies should control for prosthetic design when measuring outcomes for TKA, particularly since limited information is available discussing functional outcomes with MP knees. However, due to the small sample size for this study, statistical comparisons between TKA participants with different prostheses was considered too prone to Type 1 errors. Out of general interest, Mann Whitney U tests were conducted to observe any potential differences that might have existed between TKA participants based on this parameter. No significant differences were observed between individuals with MP and PS knee for any of the kinematic or kinetic measures discussed in this study. Future research should aim to make comparisons across prosthetic designs during functional tasks such as sit-stand, obtaining larger sample sizes to allow for stronger statistical analyses. The smaller sample size used in this study leaves statistical comparisons vulnerable to Type 1 errors, but also inhibits the statistical power that can be applied with parametric tests. As the first study to look at frontal

plane biomechanics during a sit-stand task with TKA populations, this study was novel in its approach.

Seat height was held consistent for the purpose of this study to measure functional performance of a daily activity (standing up and sitting down from a toilet seat). However, it is acknowledged that this may have influenced outcomes for biomechanical measures, as chair height has been found to significantly impact lower limb kinematics and kinetics among individuals who suffer from OA or have undergone TKA (Su et al., 1998).

In conclusion, individuals who undergo TKA with a MP or PS knee both exhibit altered lower limb biomechanics when transferring between sitting and standing postures at 1 year after surgery. The statistical significance of these discrepancies during sit-to-stand, but not stand-to-sit, suggests that this task may place a greater demand on the knees, potentially due to greater demands for power generation to lift the body.

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APPENDIX

**Table 9.** Frontal plane measures during sit to stand; median (25<sup>th</sup> percentile, 75<sup>th</sup> percentile)

Group	Operated	Non-Op	Control	P-value	
<b>Kinematics</b>				Op vs. Control	Op vs. Non-Op
Hip Abd				0.08	0.35
Males	-2.0 (-4.5,-0.3)	-1.4 (-3.1,-0.7)	-8.4 (-11.9,-7.3)		
Females	-2.5 (-3.9,-1.8)	0.4 (-1.2,2.0)	-4.2 (-8.0,-2.0)		
Hip Add	-1.8 (-2.7,-1.1)	-1.1 (-2.0,0.5)	-1.2 (-2.1,- 0.2)		
Knee Add	23.9 (7.4,29.6)	14.7 (10.3,26.3)	16.5 (10.9,24.3)	0.92	0.18
<b>Kinetics</b>					
Hip Add M	0.33 (0.24,0.64)	0.31 (0.20,0.52)	0.22 (0.13,0.34)	0.34	0.07
Hip Abd M	-0.05 (-0.07,-0.01)	-0.05 (-0.10,-0.02)	-0.10 (-0.26,-0.01)	0.12	0.44
Knee Abd M	-0.26 (-0.3,-0.12)	-0.38 (-0.5,-0.28)	-0.35 (-0.49,-0.25)	<b>0.01</b>	<b>0.003</b>

\*confidence interval set at 95%

**Table 10.** Frontal Plane measures during stand to sit; median (25<sup>th</sup> percentile, 75<sup>th</sup> percentile)

Group	Operated	Non-Op	Control	P-value	
<b>Kinematics</b>				Op vs. Control	Op vs. Non-Op
Hip Abd	-7.2 (-12.0,-5.9)	-6.7 (-12.4,-3.5)	-6.2 (-9.7, -3.0)	0.1	0.08
Hip Add	-1.6 (-3.7,0.005)	-1.2 (-1.7, -0.04)	-1.2 (-1.7,0.4)	0.28	0.46
Knee Abd	-0.5 (-3.0,0.5)	0.23 (-1.3,0.8)	-1.0 (-2.1,-0.2)	0.46	0.06
<b>Kinetics</b>					
Hip Abd M	-0.01 (-0.06, 0.005)	-0.08 (-0.1,-0.02)	-0.08 (-0.09,-0.02)	0.09	0.32
Hip Add M	0.2 (0.08,0.4)	0.14 (0.11,0.26)	0.14 (0.1,0.24)	0.47	0.87
Knee Abd M	-0.28 (-0.32,-0.15)	-0.29 (-0.48,-0.17)	-0.34 (-0.45,-0.23)	0.08	0.49
Knee Add M	0.002 (-0.02,0.04)	-0.003 (-0.04,0.06)	0.02 (-0.006,0.05)	0.59	0.51

\*confidence interval = 95%.

## **GENERAL DISCUSSION**

While there has been a considerable amount of literature focused on gait analysis after TKA, no studies to our knowledge have looked at gait on inclined surfaces. With respect to sit-stand tasks, there are a limited number of studies that have focused on lower limb biomechanics during sitting down or standing up from a chair. No studies could be found that addressed frontal plane parameters during sit-to-stand, despite the fact that numerous studies have linked frontal plane kinematics and kinetics in stance phase of gait to OA progression and overloading of the knee compartments. While it may be assumed that this interest in the loading phases of gait would carry over to other weight-bearing activities, there seems to be a large gap in literature for this topic. Meanwhile, only two studies could be found that addressed biomechanical parameters for stand-to-sit, yet both were limited in their selection of participants. By developing a study that included both male and female participants across both TKA and control groups, our results can provide a better perspective on the biomechanical patterns employed by TKA participants, compared to a healthy, age-matched cohort.

During level walking, TKA patients have been noted to exhibit altered gait patterns, including reduced knee flexion excursion and extension moments. Findings from our study reflected these differences in the sagittal plane during both level and downhill walking. The reduced knee flexion excursions, as well as observed discrepancies in knee flexion moments, may coincide with previous findings for a “stiff knee” (Benedetti, et al., 2003). However, verification for the causes of this altered gait pattern requires further analysis to look at muscle activity and strength. Therefore, future studies should build on these findings by including EMG analysis during inclined gait. It is also of interest to note that downhill walking brought out differences between groups that were not found to be significant during level walking,

suggesting that the increasing demands of this task exacerbate discrepancies between the TKA and control participants. Based on these findings, future studies should consider the effects of this type of walking surface when looking at gait after TKA.

Previous studies have noted that both TKA and OA patients walking at slower self-selected paces than healthy, age-matched populations. It has also been reported that walking speed has a significant interaction with joint moments. A self-selected walking speed was included in the protocol for this study as our aim was to measure kinematic and kinetic measures experienced by the lower limbs during activities of daily living. While it is acknowledged that controlling for walking speed would have allowed for a stronger analysis of the impact of TKA on lower limb movement patterns, it would not necessarily have reflected the loading experienced by the lower limbs on a daily basis. For this reason, we believed that using a self-selected walking speed was more appropriate.

Of the two studies that examined performance of sit-stand with TKA populations, Wang et al. (2005) looked at kinematics and kinetics in the sagittal plane, but limited their comparison to TKA participants with single-axis or multi-axis PS knees, without including a control group in their analysis. The second study by Wang et al. (2012) included a control group, but focused primarily on biomechanical measures in the frontal plane and only included male participants. Contrary to what was found in the literature, TKA participants exhibited reduced hip flexion excursion and moments during sit-to-stand, resulting in reduced hip power generation. These discrepancies observed at the hip may reflect the nature of the task selected, rather than true functional discrepancies at the hip joint itself. Flanagan et al. (2003) noted that among older adults, rising from a chair targets the hip joint structure, whereas a squatting maneuver was a better measure for the knee. For the purposes of this study we believe that the chair squat was an

appropriate task as it fit within our focus on activities of daily living. With this in mind, raising and lowering the body from a seat was deemed to be more applicable than having participants perform squats. For this reason, a set chair height was decided at 45cm, in accordance with building code regulations for toilet seat height. However, the difference between the demands of the sit-stand (chair) task and a squatting maneuver is of interest for future research. It is recommended that future studies look at lower limb biomechanics during both the sit-stand task with a chair, as well as a squatting maneuver, to understand how these different demands are reflected in lower limb movement patterns.

The significantly lower hip abduction angles and knee abduction moments among TKA participants are of interest during the sit-stand task due to their potential link to joint loading and OA progression. As previously mentioned, adduction angles and moments at the hip and knee are often associated with medial compartment loading during the stance phase of gait and have been linked to OA severity. Despite this focus on frontal plane mechanics during gait, there is a lack of literature addressing these measures during weight-bearing activities such as sit-stand. When rising from the seat, participants remained in abduction angles and moments throughout the movement; however, the peak abduction angles and moments were much smaller at the operated limb, compared to the non-operated and control limbs. The reduced abduction angles and moments may bear a similar effect as adduction angles and moments in gait, reflecting a shift in loading away from the lateral compartment and towards the medial compartment of the knee. While further studies would be needed to verify this assumption, including in vivo measures, it can be concluded either way that TKA participants show altered movement and loading parameters at the hip and knee of the operated limb when rising from a chair.

As trunk motion has been shown to significantly impact movement strategies during sit-stand tasks, information that could have been gained from this analysis would likely have aided in interpreting results from lower limb biomechanics. This is an important measure to consider for further research and may help to explain altered movement patterns observed with TKA participants.

Another factor that should be considered when interpreting outcomes from this study is the combination of individuals with either a Medial Pivot or Posterior Stabilized knee within the same TKA group. Prosthetic design may have a significant influence on lower limb biomechanics and therefore may have contributed to variability within the TKA group, risking Type 2 errors. While it would have been ideal to separate TKA participants into groups based on the type of prosthesis, the small sample size did not allow for this. Having a TKA group of 15-17 participants we felt that statistical comparison across individuals with a MP or PS knee was deemed too vulnerable to Type 1 and 2 errors. However, it should be noted that Mann Whitney U tests were included to determine if a significant difference existed between the MP and PS participants in this study (prior to combining them for our 'surgical' group). These tests found no significant differences between participants for any of the gait or sit-stand tasks. To understand whether these findings are a result of similar functional outcomes across prosthetic knees or simply a product of the small sample size for the MP and PS groups, future studies should aim to recruit larger sample sizes so that a larger, statistical comparison is possible. This would allow for a greater understanding of how prosthetic design may influence comparisons between TKA and control groups.

Finally, when looking at overall outcomes from this study, it is of interest to note that two consistent discrepancies were observed between the two groups during each of the tasks. For

each activity, TKA participants exhibited reduced knee flexion excursion and peak knee power values compared to controls. Looking back at the literature, it can be seen that knee flexion and range of motion have been two principal concerns for measuring surgical success following TKA. Reduced knee flexion has been associated with lower WOMAC function scores, correlating to reduced patient satisfaction and perceived quality of life (Miner et al., 2003). However, the extent to which the observed discrepancy in knee flexion angles translated to an actual reduction in the physical capabilities of TKA participants is unclear. It is possible that a difference of  $<10^\circ$  may not have a significant effect on the performance of daily activities for these individuals. With this in mind, studies should consider qualitative measures such as questionnaires or physical activity scores to examine how differences in kinematics/kinetics correlate with patients' perceptions and level of activity. Similar considerations should be made when interpreting results for peak power measures as they appeared to coincide with periods of reduced knee flexion excursion.

In conclusion, our study found that individuals continue to exhibit altered movement patterns during walking as well as transitioning between seated and standing postures at 6-12 months following TKA. Gait analysis showed that downhill walking exacerbated alterations in gait patterns that were observed during level walking. Sitting up and down from a seat resulted in similar discrepancies for movement patterns, although these differences were only significant as participants stood up from the seat. Altered lower limb movement strategies should be considered when measuring outcomes from TKA procedures as they may result in greater loading of the joints and could contribute to wear of the prosthesis. These discrepancies should be kept in mind for establishing rehabilitation protocols before and after TKA.

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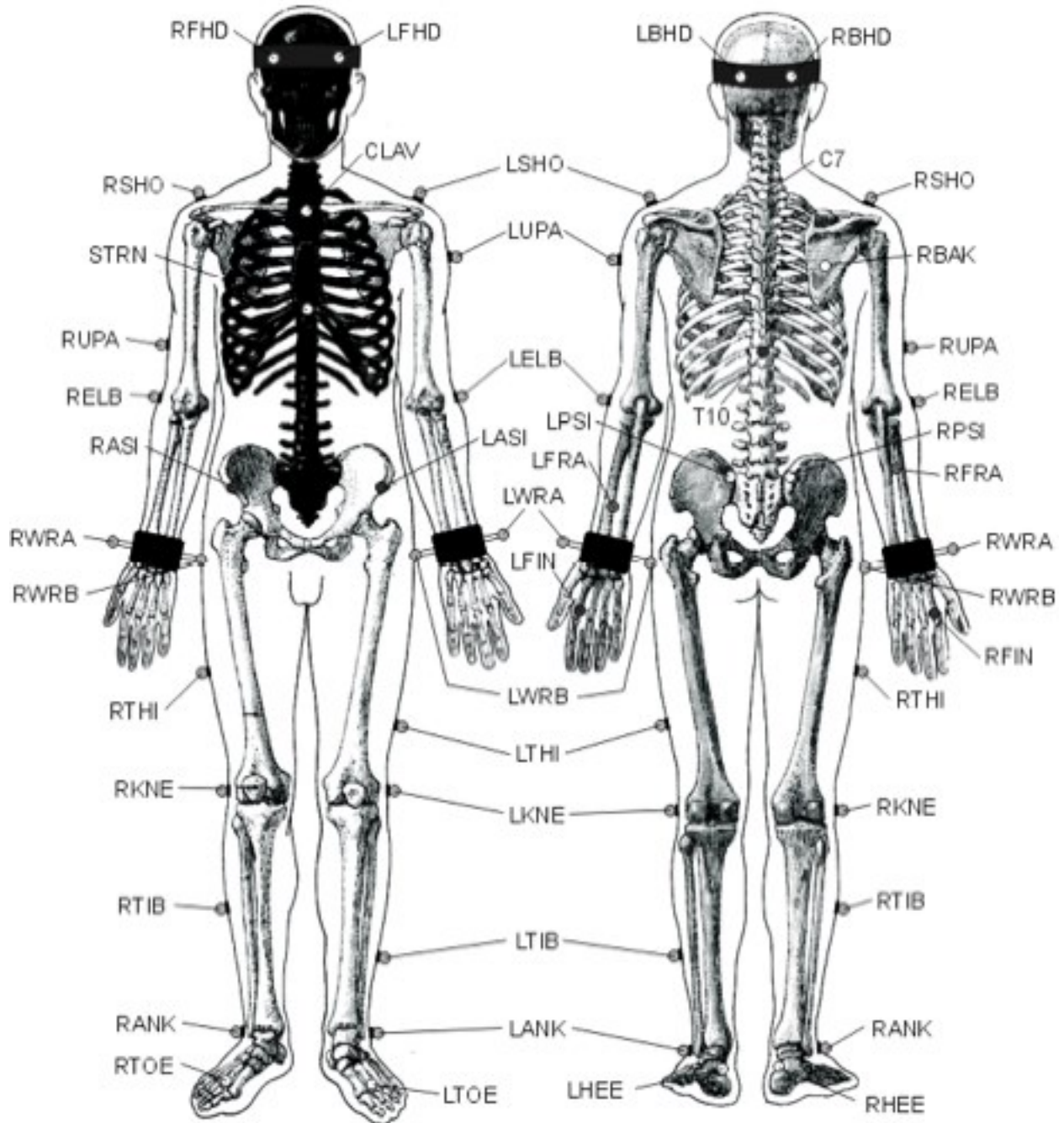
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## APPENDIX A

### UNIVERSITY OF OTTAWA MOTION ANALYSIS MODEL (UOMAM)



**APPENDIX B**  
**UNIVERSITY OF OTTAWA MOTION ANALYSIS MODEL (UOMAM)**  
**MARKER SET**

<b>Head</b>	
LFHD & RFHD	Left temple
LBHD & RBHD	Left back of head
<b>Torso</b>	
C7	7th cervical vertebrae
T10	10th thoracic vertebrae
CLAV	Jugular notch
STRN	Xiphoid process
RBAK	Middle of right scapula
<b>Arms</b>	
LSHO & RSHO	Acromio-clavicular joint
LUPA & RUPA	Upper arm
LELB & RELB	Lateral epicondyle
LFRA & RFRA	Forearm
LWRA & RWRA	Wrist bar thumb side
LWRB & RWRB	Wrist bar pinkie side
LFIN & RFIN	Dorsum of the hand head of 2nd metacarpal
<b>Pelvis</b>	
LASI & RASI	Anterior superior iliac crest
LPSI & RPSI	Posterior superior iliac spine
LLAS & RLAS	Iliac crest
<b>Legs</b>	
LTHI & RTHI	Lateral thigh
LMKN & RMKN	Medial epicondyle of the knee
LKNE & RKNE	Lateral epicondyle of the knee
LTIB & RTIB	Lateral shank
LANK & RANK	Lateral malleolus
LMAN & RMAN	Medial malleolus
<b>Feet</b>	
LTOE & RTOE	2nd metatarsal head of foot
LHEE & RHEE	Posterior calcaneus