From knee osteoarthritis to post-operative total knee arthroplasty: understanding the role of muscle strength, activation, biomechanics and implant design on knee joint function

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Thesis submitted to the University of Ottawa
in partial fulfillment of the requirements for the degree of

DOCTOR OF PHILOSOPHY

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"Ability is what you're capable of doing.

Motivation determines what you do.

Attitude determines how well you do it."

- Lou Holtz

Acknowledgements

I would foremost like to thank my supervisor, mentor, and friend, Prof. Mario Lamontagne. You allowed me to work in your lab as a research coordinator after completing my Masters's. You continually provided unconditional support, guidance, and opportunities to expand my research and professional development. Even though I was coming into an established research project, you allowed me to explore my curiosities and develop my research questions to create this thesis I am genuinely proud of. I will always be grateful for all the skills and discipline you instilled.

To my co-supervisor, Prof. Jing Xian Li, none of this would be possible without your guidance and support. You guided me since I started my Masters's and helped me cultivate my curiosity and aptitude in biomechanics. You have always encouraged and supported me, so I thank you for all the guidance and advice you have provided me since I was just a first-year undergraduate student in your Anatomy class.

I want to thank Dr. Geoffrey Dervin for his continual support and insightful clinical aspects. Thank you for inviting me to your OR. I got to experience total knee arthroplasties firsthand in a way that no scientific article or video could replace. That was one of my favourite experiences throughout my training, and I am grateful for the opportunity you provided me.

I want to thank my Thesis Advisory Committee members, Prof. Ryan Graham and Dr. Gerd Melkus, for their guidance and support these past years. I appreciate the accommodations both of you allowed me for my thesis, as the Covid-19 pandemic halted surgeries and required me to change a significant part of my thesis. Your inputs were crucial for the thesis presented before you.

To my lab colleagues, a more than special thanks! The support and camaraderie we gave each other is something I will always cherish. Danilo, we met on the first day of graduate school and have supported each other since. Even though you are a continent away, you are always just a Zoom call out, whether it is for paper support or general advice. All the individuals that one day was (or still are) at my side, your collaboration was essential for this achievement: Geoff, Mike, Celine, Nok Yeung, Matt, Lisa, Nick, Ishi, Daphne, Alex, Nick, Bruno – Thank you all!

Without the help of my parents, Krystyna and Les Kowalski, I would not have been able to accomplish what I have. The support they gave me along the way will never be forgotten. I am grateful for all their hard work and dedication, as they instilled a strong work ethic in me and taught me that anything the mind perceives it can achieve. Thank you for always supporting me and encouraging me to aim for a goal outside my comfort zone.

I also would like to thank all funding agencies that helped with my academic and professional development by providing scholarship and travel awards that allowed me to present this research at various conferences in North America and Europe: The Hans K. Uhthoff Fellowship, FGPS, GSAÉD, CUPE 2626, School of Human Kinetics, and the Force & Motion Foundation.

Lastly, thank you to my loving and beautiful wife, Catherine! You encouraged me through the careful decision to start my Ph.D. and provided me with support, care and love every step of the way. I cannot wait for the following stages of our journey together!

Preface

The contents of this dissertation are organized into six chapters as follows:

I. *Introduction* introduces the research background, statement, and rationale.

II. *Literature review* provides an overview of knee osteoarthritis, total knee arthroplasty, and posterior stabilized and medial ball-and-socket implant designs.

III. *Study design* identifies the gaps in the literature, the thesis framework, and the general methods.

IV. *Results,* which are comprised of eight original research articles.

V. *Discussion* includes a summary of the thesis, the clinical implications of the findings, limitations, and future directions.

VI. *Conclusion*, which addresses the main highlights of the thesis.

This dissertation comprises eight original research articles, on which I was the lead author. I was fully involved with (1) the conception of the studies; (2) data acquisition, analysis and interpretation; (3) drafting, revising, and submission of the articles for peer review in scholarly journals. Three articles were published at the time of thesis submission (Chapters 4.1, 4.3, and 4.4), four are under revision (Chapters 4.2, 4.5, 4.6, and 4.7), and the final is prepared for peer review (Chapter 4.8) in journals specializing in clinical orthopaedics and biomechanics. Each of the eight articles within this thesis was formatted to the journals' requirements. Lastly, I do not have any conflicts of interest to report and certify that the research ethics of both institutions (the University of Ottawa and Ottawa Health Science Network Research Ethics

Board) approved the investigation protocols (Appendix – Chapter 8.1). All investigations were conducted in conformity with ethical research principles, and written informed consent for participation in the study was obtained from all participants.

Abstract

Knee osteoarthritis (OA) is a progressive disease that ultimately requires patients to receive a total knee arthroplasty (TKA) to replace the damaged structures within the knee with an artificial joint. Surgeons have many options when selecting an appropriate implant. Patients want a TKA that feels ‘normal’ and allows them to perform most activities without pain, stiffness, and other residual symptoms. However, 20% of patients remain unsatisfied with their surgery, regardless. This thesis aimed to examine the effect of implant selection during TKA on knee biomechanical function during various ADLs.

Several gaps were identified within the review of literature: 1) patient-reported outcome measures cannot differentiate between medial ball and socket (MBS) and posterior stabilized (PS) implants, 2) most biomechanical studies were performed only in postoperative patients, and 3) studies that compared MBS and PS implants were primarily focused on level walking conditions, and overlooked tasks that placed more demand on the knee joint.

Twenty-eight individuals with severe knee OA were randomized to receive either an MBS (n=14) or PS implant. They completed a biomechanical assessment within one month and one year after TKA and were compared to 14 controls of similar age, sex, and body mass index. They performed a variety of tasks which explored three main areas: 1) examine the alterations in gait variability among individuals with OA following a TKA procedure using either a PS or MBS implant; 2) enhance the understanding of the post-operative effects of TKA with either MBS and PS implants on knee biomechanics and muscle activities during level walking, as well as more demanding tasks such as descending a ramp or staircase; 3) simulate the dynamic knee

joint loads in post-operative TKA patients with either PS or MBS implants during closed-chain, bilateral tasks such as sit-to-stand.

Initially, a series of studies were performed to develop a new test called waveform-level variance inequality test (*eqvartest*), which had not been previously utilized in the literature. This test was used to identify discrepancies in gait variability pre and post-TKA in the gait cycle. Following TKA, patients showed decreased variability in knee moment and power at single-limb support. Neither the MBS nor PS implant provided the same level of variability as the control group, demonstrating reduced knee joint stability.

The MBS group had a gait pattern closer to the control group during level walking, whereas the PS group walked with a stiffer knee. However, during more demanding ADLs, the differences were less apparent. During ramp descent, knee joint stability issues became prominent as MBS and PS groups adopted a 'cautious gait pattern,' widening their base of support and stiffening their knee to reduce loading. During stair descent, the MBS implant provided increased stability as it required less muscle activity than the PS, requiring greater hamstring muscle activation.

During sit-to-stand, MBS and PS groups favoured their non-operated knee as they had reduced total vertical, medial, and lateral KCF on their operated knee compared to their non-operated side. This may be due to compensatory strategies developed through the progression of knee OA and may increase the risk of developing knee OA on the non-operated limb.

The outcomes of this thesis can assist clinicians in selecting the most appropriate implant for their patients and guide them in designing rehabilitation programs that can enhance patient function following TKA.

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List of Abbreviations

ACL: anterior cruciate ligament ADL: activities of daily living ANOVA: analysis of variance avg: average BF: biceps femoris BMI: body mass index BW: body weight CI: confidence interval CSA: cross sectional area CTRL: healthy control Deg: degrees DoF: degrees of freedom EMG: electromyography GM: medial head of gastrocnemius GL: lateral head of gastrocnemius GRF: ground reaction force JCF: joint contact forces KCF: knee contact forces KOOS: knee injury and osteoarthritis outcome score ID: inverse dynamics IK: inverse kinematics LCL: lateral collateral ligament MA: moment arm MCL: medial collateral ligament MSK: musculoskeletal	MSKM: musculoskeletal model MP: medial pivot MTU: musculotendon unit MVIC: maximal voluntary isometric contraction N/A: not applicable OA: osteoarthritis PeakLE: peak linear envelope PCL: posterior cruciate ligament Post-Op: postoperative Pre-Op: preoperative PROMs: patient reported outcome measures PS: posterior stabilized QOL: quality of life ROM: range of motion RRA: residual reduction algorithm SD: standard deviation SO: static optimization ST: semitendinosus SPM: statistical parametric mapping TKA: total knee arthroplasty UKA: unicompartmental knee arthroplasty UOMAM: University of Ottawa motion analysis model VL: vastus lateralis VM: vastus medialis

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1 Introduction

Background | Statement | Rationale

1.1 Background

Knee osteoarthritis (OA) is one of the most frequent causes of disability in the elderly (Srikanth et al., 2005) and occurs in 13% of women and 10% of men aged 60 years or older (Zhang & Jordan, 2010). Ultimately as knee OA progresses, patients require a total knee arthroplasty (TKA) to replace the damaged structures within the knee with an artificial joint. Before the Covid-19 pandemic, more than 70,500 TKAs and 5,825 unicompartmental knee arthroplasty (UKA) were performed annually across Canada. However, this number decreased to 55,285 in 2020-2021, and it was estimated that over 48,000 total hip and knee surgeries were not performed as expected if not for the pandemic (CIHI, 2022). Revision surgeries accounted for 7.5% of all knee surgeries performed in Canada, representing 4,161 total cases (CIHI, 2022). While infection (35.6%) and aseptic loosening (18.0%) are the top reasons for TKA revisions, 12.4% of performed revisions were due to instability (CIHI, 2022).

Canada’s population continues to age, and before the pandemic, the annual number of TKAs performed grew by 5% each year (CIHI, 2022). By 2031, it is forecasted that almost 6 million Canadians will have OA, up from 3.6 million in 2010 (64% increase) (Sharif et al., 2015). While not all will require surgical intervention, the number of TKAs will surely increase to treat the most severe cases. Most TKA recipients are satisfied with their surgery. However, approximately 20% of patients remain dissatisfied due to unresolved pain or poor knee joint function (Gunaratne et al., 2017). Over 11,000 Canadians who received a TKA in 2020-2021 will remain unsatisfied with their surgery. We must understand the factors contributing to poor function and satisfaction after TKA.

1.2 Statement

Maintaining mobility and quality of life into old age is a significant challenge, as it is estimated that more than two billion people will be over 65 years by 2050 (McGregor, Cameron-Smith, & Poppitt, 2014). While approximately 20% of TKA recipients remain dissatisfied (Gunaratne et al., 2017), the reasons for dissatisfaction were patient expectations before surgery, the degree of improvement in knee function, and pain relief following surgery (Gunaratne et al., 2017). Patients want their replaced knee to feel “normal,” resolve their pain and perform most activities of daily living (ADLs) without issue (Hofmann & Schaeffer, 2014). Walking and climbing a staircase or ramp are tasks individuals face daily, so patients expect to perform well during these tasks of daily living. Understanding the intricate role of muscle strength, activation, biomechanics, and the role of implant design on knee joint function after TKA is necessary if we want to reduce the number of dissatisfied patients.

The ultimate goal is to understand better how knee biomechanical function changes after a total knee replacement with either a medial ball and socket (MBS) or posterior stabilized (PS) implant after a one-year follow-up during activities of daily living.

This research aims to examine the *effect of implant selection during total knee arthroplasty on knee biomechanical function during various activities of daily living.*

This research proposes to:

- Evaluate how healthy individuals without knee osteoarthritis differ from symptomatic individuals with knee osteoarthritis.
- Examine the post-operative kinematics, kinetics, and muscle activation patterns during various activities of daily living in TKA patients with either an MBS or PS implant.
- Examine the muscle forces and knee contact forces during activities of daily living after TKA surgery with either an MBS or PS implant.

1.3 Rationale

Patients want a TKA that feels ‘normal’ and allows them to perform their ADLs without pain, stiffness, or other residual symptoms (Hofmann & Schaeffer, 2014). While several factors can affect the outcome of TKA, some factors that surgeons have partial control over include their surgical technique and the selected implant (Judge et al., 2012; Lingard, Katz, Wright, & Sledge, 2004; Mugnai, Digennaro, Ensini, Leardini, & Catani, 2014; Santaguida et al., 2008; Schwartz & Lange, 2017). The medial subvastus approach provides better earlier post-operative benefits than a medial parapatellar approach, such as better knee range of motion after one week and a shorter time to regain an active straight leg raise. Still, these differences do not exist after one year (Berstock, Murray, Whitehouse, Blom, & Beswick, 2018), suggesting surgical approach may have less influence on longer-term function. Surgeons have numerous implant design options for their patients that vary in constraint, fixation, and materials (Marques et al., 2021). The implant selection remains a shared decision between the patient and surgeon (Carlson & Sierra, 2020). Understanding the relationship between postoperative function and implant design could lead to more appropriate implants being selected for patients, leading to increased satisfaction.

Patient-reported outcome measures (PROMs) are regularly used within a surgeon's clinical practice and continue to become more prevalent within the TKA literature (Lan, Bell, Samuel, & Kamath, 2020). Although many different PROMs tools exist, they vary considerably in design, which makes comparisons between various studies challenging (Lan et al., 2020). A review study which compared over 4000 patients with different implant designs using PROMs found no differences between other implants two years after surgery. (Kahlenberg, Lyman, Joseph, Chiu, & Padgett, 2019). PROMs may not be sensitive to detect functional differences between implant groups as they only capture a subjective assessment of the pain and functional deficits that persist following TKA (Stevens-Lapsley, Schenkman, & Dayton, 2010). Biomechanical analysis may be more suitable for identifying functional differences between patients with different implant designs.

The biomechanical analysis involves analyzing a movement and the forces that produce it. Combining kinematics, kinetics, and electromyography analysis can identify the main contributors to joint moments of force during the movement (Lamontagne, Beaulieu, Varin, & Beaulé, 2009). Biomechanical gait studies have identified differences between various implant designs, but their mixed findings make it difficult to reach a consensus (Esposito, Freddolini, Marcucci, Latella, & Corvi, 2020; Gray et al., 2020; Kulshrestha et al., 2020). These differences included PS implants having more significant gain in knee flexion (Kulshrestha et al., 2020), MBS having more similar kinematic profiles to healthy controls than PS implants (Gray et al., 2020), or both MBS and PS groups walking with ‘stiff knee pattern’ (Esposito et al., 2020). The different designs between MBS and PS implants will influence knee joint stability, so it is necessary to evaluate knee joint function during various ADLs to understand better if one implant provides better functional benefits. Gaining better mobility or knee joint function using a particular implant design could improve patient satisfaction.

2 Literature Review

Knee Osteoarthritis | Total Knee Arthroplasty | Posterior Stabilized versus Medial Ball and Socket Implants

This review of literature will focus on the two general themes of this thesis – knee OA and TKA. The degenerative progression of knee OA is why patients require a TKA, so a basic understanding of the impairments caused by knee OA and the adaptation and changes of biomechanical function after TKA. The review on TKA will briefly describe the various implants and surgical approaches the orthopaedic surgeon can use. This will be followed by comparing MBS and PS implants to understand differences in patient-reported outcome measures, kinematics, kinetics, muscle activity, and joint loading between the two implant designs.

2.1 Knee Osteoarthritis

Knee OA is a degenerative disease affecting the knee joint's articular cartilage, bone, and soft tissues (Migliorini et al., 2022). It is typically the result of wear and tear and is diagnosed using radiographs and a simultaneous presence of pain, aching, or stiffness (Zhang & Jordan, 2010). It is one of the most frequent causes of disability in the elderly (Srikanth et al., 2005), and the yearly global incidence of developing knee OA is 203 per 10,000 people (Cui et al., 2020). There are primary and secondary types of knee OA depending on its cause. Primary knee OA results from articular cartilage degeneration without a known cause, whereas secondary knee OA results from a known cause (Martel-Pelletier et al., 2016). This thesis will focus on primary knee OA.

2.1.1 Epidemiology

Radiographic evaluation has long been considered the reference standard for defining knee OA (Zhang & Jordan, 2010) and is most often done with the Kellgren-Lawrence (K/L) grading scheme (Kellgren & Lawrence, 1957). This uses an anteroposterior radiograph and grades the OA in five levels from 0 to 4. Radiographs of knee OA show the formation of osteophytes, periarticular ossicles, joint space narrowing, sclerosis, cysts, or deformity (Kellgren & Lawrence, 1957). The first indication of radiographic knee OA occurs at Grade 2 and is the criterion in population studies to classify someone with knee OA (Cui et al., 2020). Although classified as minimal knee OA (Grade 2), many people may experience pain, aching or stiffness at this stage. Still, some may not experience symptoms until the later stages of knee OA where indications and symptoms become more moderate (Grade 3) and severe (Grade 4) (Kellgren & Lawrence, 1957 ; Zhang & Jordan, 2010). Ultimately individuals require surgical intervention after all non-surgical treatments are exhausted (Skou et al., 2016).

The predisposing risk factors for developing knee OA can be divided into modifiable and nonmodifiable. Unmodifiable risk factors are age, sex, and genetic factors (Roos & Arden, 2016). Knee OA can develop at any age but is more common at older ages. Globally, the prevalence of knee OA is 2.5% for women and men under age 40, but it continues to rise with age. For people in their 60s, the prevalence is 28.7%, but by age 80, it reaches 49.8% (Cui et al., 2020). Women have a higher prevalence (21.7%) than men (11.9%) (Cui et al., 2020). There is also a genetic component in the risk of knee OA as several genes linked to the development of knee OA have been identified. The probability of inheriting these genes is estimated at 45% (Spector & Macgregor, 2004).

Several modifiable risk factors include obesity, previous knee injury, biomechanical modifications, occupational, recreational, and socioeconomic factors. BMI is a significant risk factor as obese (BMI $\geq$ 30 kg/m^2) individuals have a lifetime risk of 60.5% in developing knee OA (Murphy et al., 2008). Previous knee injury or trauma is also a risk factor for developing knee OA (Silverwood et al., 2015). Young adults (25-34) who suffer a knee injury were diagnosed with knee OA at a much higher rate when they were older than those who did not sustain any injuries. Biomechanical modifications in the knee, such as valgus and varus deformities, also increase the risk of knee OA (Migliorini et al., 2022).

Heavy, physically demanding occupations, occupational activities, and sports are associated with greater rates of knee OA. The more at-risk occupational activities often involved heavy lifting, frequent climbing, prolonged kneeling, squatting, and standing (Wang et al., 2020). The role of sport and physical exercise is still debated, as not all sports carry the same risk for knee OA. As such, these sports get divided into three categories: endurance, mixed, and power (Kujala, Kaprio, & Sarno, 1994). Endurance sports, such as running, subject the knee joints to repetitive strain; mixed sports (soccer, hockey, basketball, etc.) have risks of impacts and joint sprains; and power sports (weightlifting, wrestling, boxing, etc.) apply large loads to the joints. In a recent systematic review, 74% (6859 patients) of the athletes suffered from OA. Of these, 41% were involved in team sports such as soccer (24%), handball (11%), or hockey (11%) (Migliorini et al., 2022). Conversely, 26% of the athletes did not report significant differences in the progression of OA compared to controls. These unaffected athletes participated in aerobic sports such as running (47%) (Migliorini et al., 2022).

2.1.2 Physiopathology

The development of knee OA depends on the interactions of many factors and is considered the product of an interplay between systemic and local factors (Zhang & Jordan, 2010). The disease becomes more progressive and disabling due to a combination of the previously mentioned risk factors, which include advanced aging, increased biomechanical loading of joints through obesity, genetics, trauma, knee malalignment, and an imbalance in physiological processes (Eaton, 2004).

There is a cascade of changes in the joint structure, including subchondral bone expansion, bone marrow lesions, meniscal tears and extrusion, to cartilage defects, which may ultimately lead to cartilage loss and radiographic OA (Heidari, 2011). Many structures in the knee joint are involved in the OA process and include the menisci, ligaments, periarticular muscles and joint capsule (Heidari, 2011). The infrapatellar fat pad in patients with knee OA has inflammatory cells, which can lead to pain in the anterior area of the knee (Clockaerts et al., 2010). These changes cause muscular impairments and lead to functional adaptations.

2.1.3 Muscular Impairments and Modifications

Quadriceps weakness is consistently found in patients with knee OA (Cunha et al., 2019; Slemenda et al., 1997; Slemenda et al., 1998), and it can be used as a predictor for whole knee and medial tibiofemoral cartilage loss after three years (Chin et al., 2019). Weakened quadriceps muscles fatigue more quickly and lead to poor muscle control, which may accelerate knee cartilage loss (Hortobágyi et al., 2005). A 14-year longitudinal study identified that having greater quadriceps strength reduced the risk of developing knee OA by 64% in women, with a similar but insignificant trend occurring in men (Sattler et al., 2012).

Individuals with knee OA had decreased quadriceps cross-sectional area (CSA) and increased intramuscular adipose tissue (Mohajer et al., 2022). For those with unilateral knee OA, quadriceps CSA was lower on their affected limb than non-affected limb. This also corresponded to a significantly lower maximal isometric knee extension force on the affected side (Sattler et al., 2012). However, the differences in muscle CSA and muscle strength appear to be limited to the quadriceps, as no differences existed for the hamstrings muscles CSA or knee flexion isometric force (Sattler et al., 2012).

Fatty infiltration of inter- and intramuscular adipose tissue in the thigh muscles of individuals with knee OA is greater than controls (Maly, Calder, Macintyre, & Beattie, 2013; Mohajer et al., 2022). Although the exact pathway of how fatty infiltration acts on the skeletal muscle is not fully understood, it is thought to create an unfavourable environment for muscle contraction (Childs, Sparto, Fitzgerald, Bizzini, & Irrgang, 2004).

2.1.4 Functional Adaptations

As mentioned above, many muscular impairments and structural changes occur with the progression of knee OA, which often lead to patients adopting their movement patterns (Hurley, Scott, Rees, & Newham, 1997). Patients move with reduced knee joint ROM (Benner, Shelbourne, Bauman, Norris, & Gray, 2019), altered knee joint stiffness (Zeighami, Dumas, & Aissaoui, 2021), or altered neuromuscular control strategies (Ghazwan, Wilson, Holt, & Whatling, 2022). Up to 72% of individuals with knee OA report knee joint instability, which is the reported feeling of the knee buckling or giving way. So these adaptations may try to increase joint stability, whereas others may adapt their movements to avoid pain (Wallace, Riches, & Picard, 2019).

Patient-reported instability is a common complaint with knee OA. However, no quantitative measure of knee instability exists (Wallace et al., 2019). As knee OA progresses, articular cartilage degeneration increases, which causes joint space narrowing (Slemenda et al., 1997), resulting in laxity in the cruciate and collateral ligaments, decreasing the passive joint stability, causing greater muscular co-contraction to provide joint stability (Heiden, Lloyd, & Ackland, 2009; Hubley-Kozey, Deluzio, & Dunbar, 2008; Wallace et al., 2019).

Individuals with unstable knees walk slower and with increased knee flexion (Esch, Knoop, & Leeden, 2012; Gustafson, Gorman, Fitzgerald, & Farrokhi, 2016). The increased joint laxity reduces passive stiffness in the frontal plane of the knee and the sagittal plane during walking, which suggests that an unstable knee is more difficult to control as it lacks the restraining characteristics to control perturbations even under small load (Esch et al., 2012; Gustafson et al., 2016; Wallace et al., 2019). The nature of knee joint instability has yet to be fully understood. Still, evidence suggests that there are confounding factors, such as the effect of pain and the influence of sex on knee joint instability, which need further exploration (Wallace et al., 2019).

Altered neuromuscular co-contraction has been identified in both stable and unstable osteoarthritic knees. Increased anteroposterior (Childs et al., 2004; Heiden et al., 2009; Hortobágyi et al., 2005; Hubley-Kozey, Hill, Rutherford, Dunbar, & Stanish, 2009; Metcalfe et al., 2013; Patsika, Kellis, Kofotolis, Salonikidis, & Amiridis, 2014; Schmitt & Rudolph, 2007; Smith, Allan, Marreiros, Woodburn, & Steultjens, 2019; Zeni, Rudolph, & Higginson, 2010) and mediolateral (Heiden et al., 2009; Schmitt & Rudolph, 2007; Smith et al., 2019) co-contraction has been identified in people with knee OA compared to healthy controls. During more demanding tasks, muscle co-contraction was greater, especially mediolateral co-contraction (Bouchouras, Sofianidis, Patsika, Kellis, & Hatzitaki, 2020; Smith et al., 2019).

These muscular impairments lead to functional limitations for individuals with knee OA. Compared to healthy individuals, those with knee OA walk slower (Astephen, Deluzio, Caldwell, Dunbar, & Hubley-Kozey, 2008; Hanlon & Anderson, 2006; Harkey et al., 2021), with a lower cadence (Chen et al., 2003; Hart et al., 2021), longer double support times (Chen et al., 2003; Li et al., 2022; Smith, Lloyd, & Wood, 2004), shorter stride lengths (Baliunas et al., 2002; H. Li et al., 2022), increased knee flexion at heel strike (Childs et al., 2004; Ismailidis et al., 2020; Mündermann, Dyrby, & Andriacchi, 2005), less knee flexion throughout stance phase (Astephen et al., 2008; Baliunas et al., 2002; Ismailidis et al., 2020), and altered knee abduction and extension moments (Asay, Erhart-Hledik, & Andriacchi, 2018; Huang et al., 2021).

2.1.5 Treatment and Management

There are currently no approved disease-modifying OA drugs to mitigate knee OA-related symptom worsening or delay knee replacement (Mohajer et al., 2022). The American Academy of Orthopedic Surgeons (AAOS) strongly recommends exercise (land and water-based) that targets thigh muscle strengthening and modifying muscle composition (Jevsevar, 2013). Additional recommendations include oral and topical non-steroidal anti-inflammatory drugs and weight management (Jevsevar, 2013). Initial treatment of knee OA should be conservative, and surgery should be considered only if symptoms persist after the appropriate use of nonsurgical treatments.

Surgical treatment options are arthroscopic debridement, cartilage repair surgery, osteotomy with axis-correction, and unicompartmental (UKA) or total knee arthroplasty (TKA) (Rönn, Reischl, Gautier, & Jacobi, 2011). TKA remains the gold standard treatment for patients with end-stage knee OA who have failed nonoperative management. UKA is an attractive alternative surgical option for end-stage knee OA patients limited to one knee compartment, most commonly on the medial side (Carlson & Sierra, 2020). UKA does have certain benefits over TKA, including decreased length of hospital stay, quicker recovery as it is a less invasive surgery, superior knee ROM and kinematics, and patient satisfaction equal to or higher than that of TKA patients (Carlson & Sierra, 2020).

Converting a failed UKA to a TKA creates more challenges for surgeons and produces inferior outcomes compared to primary TKA (Yun, Qutami, Chen, & Pasko, 2020). The selection of UKA over TKA remains a shared decision between the patient and surgeon. It may be a better option, especially in older (80+) patients with only knee OA in a single compartment. The decision with younger patients is less clear, as they may eventually need a TKA, creating a more challenging and expensive second surgery (Carlson & Sierra, 2020).

2.2 Total Knee Arthroplasty

The review on TKA will briefly introduce the various implant designs and surgical approaches surgeons can select for a patient's procedure. This will be followed by comparing the posterior stabilized (PS) and medial ball and socket (MBS) implants, highlighting the differences in patient-reported outcome measures and satisfaction, knee biomechanics, and neuromuscular control.

2.2.1 Implant Design

The development of TKA as we know them today began in the early 1970s with tibiofemoral condylar replacements (Ranawat & Ranawat, 2012). Advancements in the designs of TKA prostheses have led to the commercialization of at least 150 implants (Carr & Goswami, 2009). These knee implants are made from strong, long-lasting, biocompatible materials such as titanium alloys, cobalt chrome, and ultra-high molecular weight polyethylene (Carr & Goswami, 2009). Ten- and fifteen-year survival rates remain above 90% for TKA, regardless of implant design (Kim, Jin, Lim, Song, & Seon, 2021). However, a fifth of patients remain dissatisfied with their TKA (Gunaratne et al., 2017). Unresolved pain, longer than expected recovery times, and an inability to return to usual daily activities or recreation due to limited function are among the main reasons why many patients remain dissatisfied after surgery (Conner-Spady et al., 2020). Part of this dissatisfaction may be due to the patient's implant, as different design affects functional performance during daily tasks, such as walking and climbing stairs (Komaris, Govind, et al., 2021). A better understanding of the relationship between postoperative function and implant design may help surgeons select a more appropriate prosthesis for the patients, which could provide function and satisfaction.

A range of implant components and combinations make up a TKA implant, all of which have implications on function and survivorship (Table 2.2.1). Bearing mobility refers to the polyethylene liner between the metal femoral and tibial components which can be 'fixed' to the tibial component or 'mobile' with the movement of the liner permitted on the tibial component. Constraint varies between implant designs, as they can keep the posterior cruciate ligament (cruciate retaining) or not (posterior stabilized/cruciate sacrificing). Typically, designs sacrifice the ACL. However, some designs keep both the ACL and PCL (bicruciate stabilized). Fixation is done with or without cement on the femoral and tibial components. Patella resurfacing can also be done with a cemented or cementless design (Marques et al., 2021).

Table 2.2.1: Total knee arthroplasty implant construction options.

Bearing mobility	Fixed	Mobile			
Constraint	Cruciate retaining	Posterior stabilized	Bicruciate stabilized	Constrained condylar	Hinged
Fixation	Cemented	Uncemented	Hybrid	Inverse hybrid	
Construct	Monobloc	Modular			
Bearing materials	Metal bearing on conventional or highly cross-linked polyethylene tray	Metal femoral component on all-polyethylene or metal tibial component (monobloc)	Other materials (ceramic bearings and ceramicised metals)	Specialized or customized implants	
Patella resurfacing	Patella resurfacing cemented	Patella resurfacing uncemented	No patella resurfacing		

Mobile bearing implants were developed as an alternative to fixed bearing implants to reduce wear and improve ROM (Capella, Dolfin, & Saccia, 2016). Studies continue to show no differences in insert wear, risk of loosening, survivorship, or clinical outcomes between fixed- and mobile-bearing TKA (Fransen, van Duijvenbode, Hoozemans, & Burger, 2017; Kim, Park, & Jang, 2021). A systematic review could not identify a significant improvement in function with a mobile bearing over a fixed bearing, so their theoretical benefits still need to be substantiated (Capella et al., 2016). Both the posterior stabilized (PS) and medial ball and socket (MBS) implants are fixed designs that sacrifice the ACL and PCL.

Surgeons must consider the cruciate ligaments' health when selecting an appropriate implant. If the PCL is healthy enough to ensure knee joint stability, the surgeon could consider a cruciate retaining knee implant, and if both ACL and PCL are healthy, a bicruciate retaining implant could be used. These implants provide many benefits for patients. More bone is preserved as the implant does not require slot resection on the distal femur to accommodate a tibial post. More soft tissues remain, which preserves proprioception, and restores knee kinematics closer to the native knee (Castellarin, Bori, Rapallo, Pianigiani, & Innocenti, 2023; Song, Park, & Bae, 2019). However, these implants are often not the first choice for many surgeons (Moretti et al., 2022). Several contraindications include PCL insufficiency, posterolateral instability, or extensor mechanism deficiency, among others (Song et al., 2019). Approximately 20% of TKA patients are eligible to receive a bicruciate implant (De Faoite, Ries, Foster, & Boese, 2020). The surgery is also more challenging, requiring appropriate soft tissue balancing to achieve proper flexion and extension gaps. Soft tissue balancing is easier with implants that resect the PCL and ACL (Castellarin et al., 2023). Since many patients cannot receive a cruciate retaining implant, surgeons must consider an implant that requires resection of the cruciate ligaments.

The PS design incorporates a post-cam system to substitute the cruciate ligaments and provide anteroposterior stability (Castellarin et al., 2023). It does not restore normal kinematics of the knee and is associated with the paradoxical anterior translation of the femur with progressive knee flexion (Chang, Kayani, Moriarty, Tahmassebi, & Haddad, 2021). MBS implants were designed to restore the natural kinematics of the knee. The medial compartment of the MBS implant functions as a ball and socket joint, with minimal rollback of the medial femoral condyle on the tibial plateau. The lateral femoral condyle can move more, rotates around the medial compartment, and translates posteriorly with progressive knee flexion (Chang et al., 2021). This has reported advantages over PS designs, including better restoration of native knee kinematics, improved contact stresses, reduced polyethylene wear, and increased mid-flexion stability compared to PS implants (Chang et al., 2021; Mannan & Scott, 2009; Samy, Wolfstadt, Vaidee, & Backstein, 2018). A more detailed comparison between the MBS and PS implants will be provided in section 2.2.3. Before postoperative outcomes can be compared between the implants, the surgeon must perform the TKA, which can be completed using various surgical techniques.

2.2.2 Surgical Approach

Several common surgical approaches are used, including the midvastus, subvastus, medial parapatellar arthrotomy, or lateral approach, which cut through different soft tissues (Figure 2.2.1). Mixed findings have been found when comparing different surgical approaches, some studies have seen favourable outcomes for the midvastus approach (Cho, Kim, Umrani, & Kim, 2014), for the subvastus approach (Varela-Egocheaga et al., 2010), for the medial parapatellar approach (Varnell et al., 2011), or no differences (Heekin & Fokin, 2014; Pan et al., 2010; Sidhu et al., 2021). A minimally invasive incision uses a smaller cut with less esthetic impact, less blood loss, and greater respect for capsule, tendon and muscle structures, providing some theoretical benefits. However, long-term benefits still need to be clarified (Sanna, Sanna, Caputo, Piu, & Salvi, 2013; Sidhu et al., 2021). Therefore, the selected approach should be based on the surgeon’s skill and experience (Sanna et al., 2013). One surgeon (Dr. G. Dervin) exclusively performed the TKA surgeries for the patients involved in this thesis. He performed all surgeries using a minimally invasive subvastus approach using a mechanical alignment so the review will be focused on these methods.

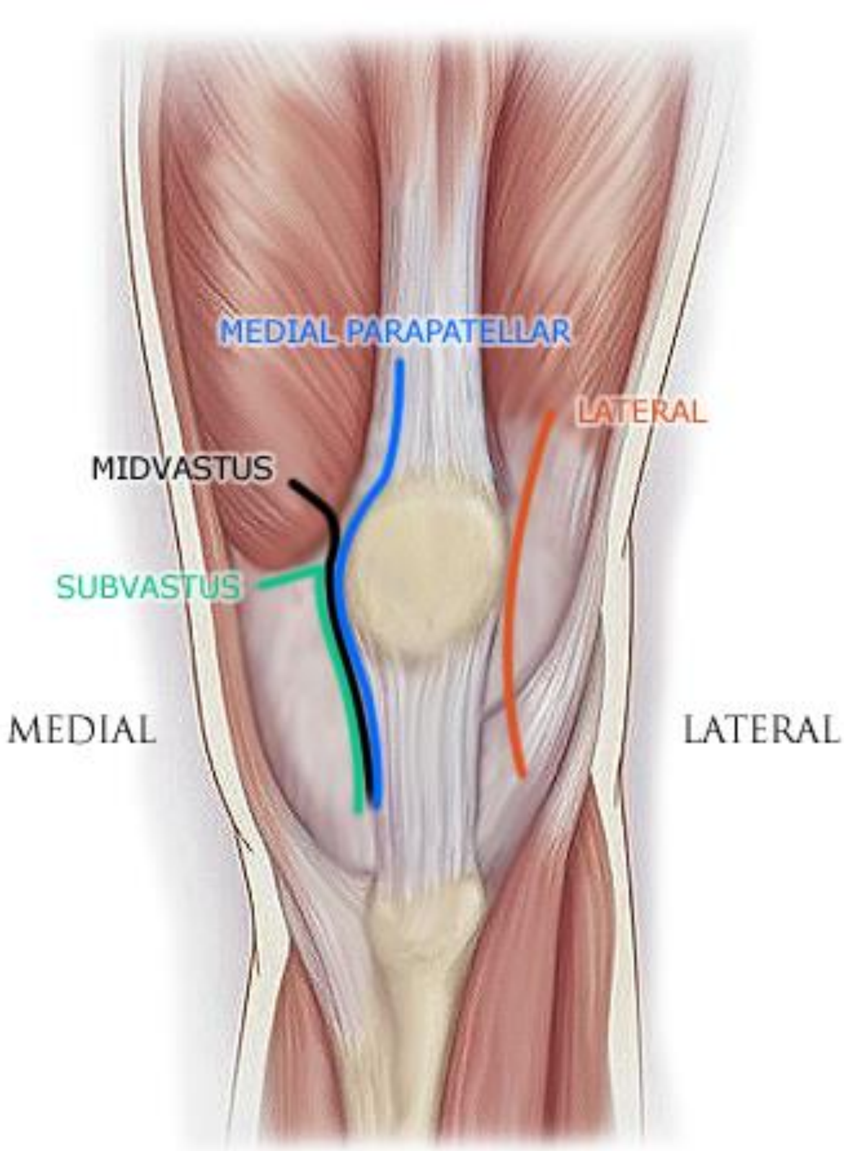


Figure 2.2.1: Incision lines for different surgical approaches. © 2016 Stefan Cristea, Vlad Predescu, Șerban Dragosloveanu, Ștefan Cuculici and Nicholas Mărăndici. Originally published in Surgical Approaches for Total Knee Arthroplasty under Creative Commons Attribution 3.0 License. Available from: 10.5772/62001 (Cristea, Predescu, Dragosloveanu, Cuculici, & Marandici, 2016)

A minimally invasive approach involves a 10 to 14 cm incision to expose the knee. It can be used for patients with pre-operative knee flexion greater than 90° and who are not obese. An anterior midline incision is required for patients who do not meet this criteria and involves a larger incision of 20 to 30 cm (Haas, Cook, & Beksac, 2004). Figure 2.2.2 shows the postoperative scar with the minimally invasive incision compared to the anterior midline incision. Once the knee is exposed, the surgeon can select their surgical approach.

Figure 2.2.2: Minimally invasive (left) compared to a standard anterior midline incision. © 2016 Stefan Cristea, Vlad Predescu, Șerban Dragosloveanu, Ștefan Cuculici and Nicholas Mărăndici. Originally published in Surgical Approaches for Total Knee Arthroplasty under Creative Commons Attribution 3.0 License. Available from: 10.5772/62001 (Cristea et al., 2016)

The subvastus approach requires specially modified instruments and is made by entering the knee joint through the inferior border of the vastus medialis and avoiding the quadriceps mechanism. This approach can only be achieved with the patella subluxed (Cristea et al., 2016). The subvastus approach is “quadriceps sparring”, but is more challenging to perform. It can only be performed on thinner patients (Engh, Holt, & Parks, 1997; A. A. Hofmann, Plaster, &

Murdock, 1991; Roysam & Oakley, 2001; Vaishya, Vijay, Demesugh, & Agarwal, 2016), and may have decreased accuracy in implant positioning (Yuan, Wang, Zhou, Yu, & Jiang, 2017).

Restoration of knee coronal alignment is a key foundation for a successful TKA. Mechanical alignment is the most common method used, partially due to its high reproducibility and easiness to perform (Insall, Binazzi, Soudry, & Mestriner, 1985). Mechanical alignment aims to create a neutral hip-knee-angle angle, which can optimize load distribution and prolong the prosthesis survival by reducing polyethylene wear and component loosening (Liu, Feng, & Tu, 2022). This changes native joint lines, Q-angle, and limb alignment. In contrast, kinematic alignment aims to restore the alignment and kinematics of the TKA to match pre-osteoarthritis anatomy. This restores native joint lines, Q-angle, and limb alignment. However, restoring natural varus can increase contact stress between tibiofemoral and patellofemoral joints, leading to an increased risk of early implant dysfunction or failure (Liu et al., 2022). The primary differences between mechanical alignment and kinematic alignment are in Table 2.2.2.

Table 2.2.2: Main key points in coronal alignments between mechanical and kinematic alignment.

	Mechanical Alignment	Kinematic Alignment
Distal femoral cuts	90°	Femoral resurfacing
Proximal tibial cuts	90°	According to extension gap
Femur external rotation to posterior condylar axis	3°	Femoral resurfacing
Overall alignment (Hip-knee-ankle)	0°	Native alignment
Ligament release	Yes	No
Type	Systematic	Patient specific

A recent meta-analysis found that patients reported higher knee satisfaction and scored higher on patient-reported outcome measures when they received a kinematic alignment (Liu et al., 2022). Some evidence exists suggesting patients achieve higher knee ROM with the kinematic alignment. However, some studies also found no differences between mechanical and kinematic alignment. The two alignment methods had similar radiographic parameters and complication rates (Liu et al., 2022). When compared using motion analysis, findings are mixed as some studies found no differences in temporospatial, knee angles, or knee moments during a gait task between patients who received TKA with either a mechanical or kinematic alignment (McNair et al., 2018; Yeo, Seon, Lee, & Song, 2019). Others found better biomechanical outcomes with the kinematic alignment (Blakeney et al., 2019). More research is necessary to determine if patients achieve better functional outcomes with the kinematic alignment.

Poor function after TKA has been related to rotational malalignment of the femoral and tibial components (Gonzalez & Mekhail, 2004). The rotational alignment goal of the tibial component is to achieve parallelism between the femoral transepicondylar axis and the medio-later axis of the tibial component on the coronal plane and avoiding errors in internal or external rotation between the two axes (Indelli, Graceffa, Marcucci, & Baldini, 2016). However, ideal coronal parallelism is hard to achieve during ROM as the transepicondylar axis has been demonstrated to be cylindrical (Hancock, Winston, Bach, Davidson, & Eckhoff, 2013) and the tibial plateau undergoes substantial internal rotation when patients perform various ADLs (Iwaki, Pinskerova, & Freeman, 2000). For the femoral component, rotational alignment establishes the symmetry of the flexion gap, and is key to achieving a stable knee during all ROM and correct joint kinematics (Castelli, Falvo, Iapicca, & Gotti, 2016). An asymmetric flexion gap can lead to varus or valgus instability during knee flexion and pain during mid-flexion. Misplacement of the femoral component can lead to complications with anterior knee pain, and patellar maltracking caused by patellar subluxation, or even patellar dislocation (Matsuda et al., 2001).

For proper rotational alignment of components, surgeons use various anatomical landmarks. The choice of a certain method depends on the surgeon's experience and the instruments used during TKA (Castelli et al., 2016). To obtain accurate rotational position of the tibial component, several landmarks are utilized, including the medial third of the tibial plateau (Lützner, Krummenauer, Günther, & Kirschner, 2010), the "Akagi" line (Akagi et al., 2004), the central third of the tibial tubercle (Bindelglass, 2001), and the postero-lateral tibial corner (Rossi, Bruzzone, Bonasia, Marmotti, & Castoldi, 2010). For femoral component rotational alignment, landmarks include the transepicondylar axis, posterior condylar line, or the trochlear anteroposterior axis, as known as whiteside line (Castelli et al., 2016).

The posterior tibial slope angle (PTSA) is the angle formed between the posterior slope of the tibial plateau and the horizontal axis. The PTSA influences the flexion gap, the tension of the PCL in cruciate retaining TKAs, and knee stability (Bae, Song, Yoon, Noh, & Moon, 2012). Depending on the type of implant, PTSA can vary between 3 to 9°. In PS TKA, an excessive posterior slope can lead to anterior post-cam impingement (Callaghan et al., 2002). For the Zimmer Biomet NexGen PS implant, the tibial cutting guide provided by the manufacturer uses a fixed PTSA of 7° (Lu, Yuan, Qiao, & Hao, 2021). For the MicroPort Advance Medial Pivot implant, the tibial cutting guide provided by the manufacturer uses a fixed PTSA of 3° (Macheras, Galanakos, Lepetsos, Anastasopoulos, & Papadakis, 2017).

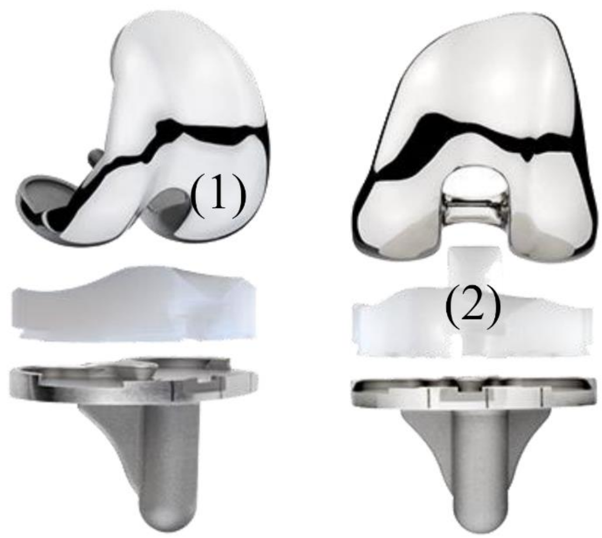
2.3 Posterior Stabilized versus Medial Ball and Socket Implants

Posterior stabilized (PS) implants use a post-cam system to achieve knee motion and stability instead of a PCL. This design improves knee flexion as it enhances the roll-back motion of the femur on the tibia (Samy et al., 2018). However , the femur will slide forward during mid-flexion and present a 'paradoxical anterior movement'(Dennis, Komistek, & Mahfouz, 2003).

Only when the post and cam collide is the ideal roll-back achieved, which may cause mid-flexion knee instability. The design of the multi-curvature radius can lead to unstable soft tissue tension and affect stability, and common complications include anterior knee pain or patellar clunk or crepitus (Shi et al., 2022; Z. Wang, Zhang, Ding, Wang, & Xu, 2021).

Medial ball and socket (MBS) implants are called by many names and are often referencing a specific company's implant: medial pivot (MicroPort Orthopedics), medial stabilized (DePuy Synthes), medial rotating (MatOrtho), or medial congruent (Zimmer Biomet) (Cacciola et al., 2021). To avoid bias, this thesis will refer to them by their generic name, MBS. The medial side of an MBS implant is designed as a ball and socket, which constrains the movement in the medial compartment, while the lateral compartment can move freely forward and backward (Shi et al., 2022). It has a single curvature radius which enhances the strength of the quadriceps femoris, ensures constant tension of the LCL throughout flexion, and has increased joint stability by the raised anterior and posterior lips on the polyethylene liner (Hunt et al., 2014). The design of the MBS reduces contact stress, polyethylene wear and improves survivorship by maximizing the contact area between the femoral component and the polyethylene liner (Sisko et al., 2017). During surgery with an MBS implant, there is less bone loss as implantation does not require an intercondylar box to accommodate the cam-post system of PS implants (Shi et al., 2022; Z. Wang et al., 2021).

Some MBS implants have options that preserve the PCL (Giustra et al., 2022). However, this thesis will focus on the PCL-sacrificing designs, which will be more in line with the PS design. The following subsections will compare PS and MBS implants in terms of patient-reported outcome measures (PROMs) and various biomechanical parameters. This will include differences in temporospatial, knee kinematics and kinetics, muscle activity, and musculoskeletal modelling. Figure 2.3.1 illustrates some of the primary design features of the MBS and PS implants evaluated in this thesis.



	MBS	**PS**
Manufacturer	MicroPort Orthopedics	Zimmer Biomet
Model	Evolution® Medial Pivot with CS tibial inserts	NexGen® PS TKA system with PS inserts
Feature:	(1) Medial compartment of femur functions as a ball-and-socket joint	(2) Post-cam system
Fixation:	Cemented	Cemented
Materials:	Metal-on-polyethylene	Metal-on-polyethylene
Fixation:	Cemented	Cemented
Constraint:	ACL/PCL sacrificed	ACL/PCL sacrificed
Coronal alignment:	Mechanical	Mechanical
Femoral component alignment:	3° external rotation relative to the posterior condylar axis, or set visually by referencing the A/P or epicondylar axis depending on varus/valgus deformity	0 to 7° external rotation relative to the A/P or epicondylar axis, depending on varus/valgus deformity
Tibial component alignment:	Coronal neutral angle	Coronal neutral angle
Tibial slope:	3°	7°
Patella:	Resurfaced	Resurfaced

Figure 2.3.1: Comparison of MBS and PS implants.

2.3.1 Patient-Reported Outcome Measures

Various tools to evaluate PROMs exist, such as the Knee Injury and Osteoarthritis Outcome Score (KOOS) (Roos, Roos, Lohmander, Ekdahl, & Beynnon, 1998), Western Ontario and McMaster Universities Arthritis Index (WOMAC), Oxford Knee Score (OKS), Hospital for Special Surgery scoring system (HSS) score and forgotten joint score (FJS) (Shi et al., 2022). While the different questionnaires vary in length and their questions, they all broadly assess pain and activity function and have been determined to be valid and reliable (Ramkumar, Harris, & Noble, 2015). PROMs have become increasingly popular in clinical practice and clinical trials (Kluzek, Dean, & Wartolowska, 2022). However, they suffer from ceiling effects and consequently cannot detect differences from well-performing groups (Eckhard et al., 2021).

A recent (2022) meta-analysis evaluated differences in PROMs and complication rates between MBS (1811 knees) and PS (1972 knees) implants (Shi et al., 2022). They found no differences in PROMs between the MBS and PS implants when evaluated with KSS, OKS, or FJS, but the MBS groups had lower WOMAC and HSS scores (lower indicating better outcomes) compared to the PS. No differences were observed in revision rates, but the MBS group had a lower complication rate. The authors concluded that the MBS implant has a better clinical effect and significantly lower complication rate than a PS implant (Shi et al., 2022). PROMs cannot capture the objective functional deficits that remain following TKA (Stevens-Lapsley et al., 2010). Therefore, a biomechanical analysis is necessary to understand the functional adaptations patients acquire after TKA with either an MBS or PS implant.

2.3.2 Biomechanical Differences

Surgeons routinely evaluate maximum knee flexion and ROM after TKA in their clinic, but more detailed comparisons require a biomechanical analysis. The biomechanical analysis provides the analysis of a movement and the forces which produce it and can give an objective assessment of function after a TKA with either an MBS or PS implant (Lamontagne et al., 2009). Gait is commonly evaluated (Vij, Leber, & Schmidt, 2022), but other activities of daily living, such as climbing a staircase (Komaris, Tedesco, et al., 2021) or ramp (Wen, Cates, Weinhandl, Crouter, & Zhang, 2022), are also compared. Implant groups are compared to determine differences in temporospatial measures or knee joint kinematics and kinetics (Lamontagne et al., 2009).

Several meta-analyses found that no clinical differences in knee ROM or maximum knee flexion exist between patients with either an MBS or PS implant (Kakoulidis et al., 2022; Nisar, Ahmad, Palan, Pandit, & van Duren, 2022; Shi et al., 2022). While these values do not provide insight into how patients move during functional tasks, they can at least help surgeons and rehab specialists determine which tasks patients can complete. For example, walking activities require the least knee flexion (60-90°), with level walking requiring approximately 67°, ascending ramps 64°, and descending ramps 72°. Tasks, including negotiating stairs and chairs require more knee flexion (90-120°). Achieving 110° of knee flexion is a reasonable goal for rehabilitation after TKA to perform most ADLs (Rowe, Myles, Walker, & Nutton, 2000), and patients who achieved high knee flexion (> 130°) after TKA reported increased ease in performing high-flexion activities and reported greater satisfaction with their TKA (Han, Kim, Lee, Won, & Lee, 2021). However, PS implants can cause mid-flexion knee instability (Shi et al., 2022), so evaluating how these different implants perform throughout different ADLs is necessary.

Knee biomechanics patients do not return to the levels of controls during gait. MBS and PS groups walk slower and with shorter steps compared to controls of similar age (Stolarczyk et al., 2022). When comparing between MBS and PS groups, findings are mixed, with some finding that the MBS group achieved a faster walking speed than the PS group (Kulshrestha et al., 2020) or no differences between them for any measured temporospatial parameter (Bianchi et al., 2021; Ghirardelli et al., 2021; Shi et al., 2020; Stolarczyk et al., 2022). Inconsistent findings for knee kinematics and kinetics have also been reported, with similar results between PS and MBS groups (Tan et al., 2021), MBS achieving knee kinematics closer to controls than PS (Gray et al., 2020), or PS achieving greater knee flexion during gait (Esposito et al., 2020). MBS and PS groups adopt a stiff knee pattern (Esposito et al., 2020), characterized by reduced knee joint angles and moments (Benedetti et al., 2003). This seemed to be accompanied by a compensatory mechanism called 'quadriceps avoidance' to limit knee pain by decreasing quadriceps force (Benedetti et al., 2003). These gait abnormalities often develop throughout the progression of knee OA and frequently remain abnormal after TKA (Metcalfe et al., 2017).

Few differences were found during level walking between MBS and PS groups, which may partially be to not being a difficult enough task to find differences. Walking requires less knee ROM than tasks such as descending ramps, negotiating stairs and standing up from a chair (Rowe et al., 2000). Knee joint loads measured *in vivo* with an instrumented TKA implant demonstrate that total peak knee contact force is 2.6 BW during level walking and highest during stair descent (3.5 BW) (Kutzner et al., 2010). Descending tasks are more demanding on the knee joint than level walking, they create larger knee extension moments (Pickle, Grabowski, Auyang, & Silverman, 2016), and TKA patients often report feeling uncomfortable performing these types of tasks (Collins, Misra, Felson, Crossley, & Roos, 2011). These tasks may be more appropriate for finding differences between PS and MBS groups.

Several studies have identified altered biomechanics in patients after a TKA while descending a ramp (Komnik et al., 2016; Simon, Della Valle, & Wimmer, 2018; Wen et al., 2022; Wiik, Aqil, Tankard, Amis, & Cobb, 2015) or staircase (Elkarif et al., 2021; Fenner, Behrend, & Kuster, 2017; Komaris, Govind, et al., 2021; Komaris, Tedesco, et al., 2021; Standifird, Cates, & Zhang, 2014; Trinler et al., 2016). However, no studies have directly compared MBS or PS implants during these tasks. In general, a review of the literature found that during stair descent, TKA patients walk with less knee ROM and extension moment (Standifird et al., 2014), with similar findings in more recent studies (Fenner et al., 2017; Komaris, Govind, et al., 2021; Trinler et al., 2016). Similar alterations in knee angle and extension moments were observed in TKA patients during ramp descent tasks when compared to healthy controls (Simon et al., 2018; Wen et al., 2022; Wiik et al., 2015).

The general findings during these more demanding tasks demonstrate that knee function does not return to the level of controls, however, it is still being determined if MBS or PS implants perform differently during these tasks. These tasks require greater stability as the centre of mass is lowered, so differences in design between MBS and PS implants may affect knee joint stability. Dynamic stability of the knee has been assessed by evaluating the variability of temporospatial and knee biomechanical measures during gait (Yakhdani et al., 2010; Lewek, Scholz, Rudolph, & Snyder-Mackler, 2006). Gait variability remains lower than healthy controls before and after a TKA (Fallah Yakhdani et al., 2010; Kiss, Bejek, & Szendrői, 2012; Smith, 2014), suggesting the knee is less stable and patients are less capable of adapting to perturbations, potentially increasing fall risk (Chan, Jehu, & Pang, 2018; Liu et al., 2020). The surrounding ligaments, muscles, and implant design maintain knee joint stability. The design difference should result in patients adopting different neuromuscular strategies to maintain knee joint stability.

2.3.3 Neuromuscular adaptations

Only a few studies have compared electromyography differences between patients with MBS and PS implants (Beach, Regazzola, Neri, Verheul, & Parker, 2019; Esposito et al., 2020). Neither of these studies evaluated patients during ramp or stair descent tasks. During level walking , MBS and PS groups had prolonged activations of the vastus medialis, biceps femoris, and rectus femoris muscles compared to the controls. The MBS had longer rectus femoris activation compared to the PS (Esposito et al., 2020). This study also reported knee kinematics and kinetics and found that both MBS and PS groups walked with a stiff knee pattern (Benedetti et al., 2003), characterized by reduced knee flexion angle and extension moments and increased co-contraction time (Esposito et al., 2020). Their study's lack of EMG amplitude normalization prevented comparisons of the magnitude of the muscle activations (Esposito et al., 2020), so it is uncertain if either MBS or PS requires greater peak or total muscle activity to maintain knee stability during various ADLs.

The other study did not identify differences between the MBS or PS groups during a level walking or step ascent for any of the muscles they evaluated (Beach et al., 2019). These studies were limited in the number of muscles they evaluated, and neither included the gastrocnemii muscles, which cross the knee joint and act as a knee flexor (Li, Landin, Grodesky, & Myers, 2002). More research comparing the neuromuscular adaptations after TKA with either a PS or MBS implant is required, especially during more demanding tasks.

2.3.4 Knee Contact and Muscle Forces

Unlike biological tissues, the materials in a knee implant do not regenerate or remodel. Therefore, understanding how forces are transmitted through the knee after TKA is essential, as knee contact forces (KCF) are directly implicated in wear and damage, especially to the polyethylene liner (D'Lima, Fregly, Patil, Steklov, & Colwell, 2012). While contact forces can be measured *in vivo*, they require specialized, instrumented implants, which are costly and limit their widespread use. Computational modelling is a non-invasive alternative that can estimate KCF and muscle forces.

Computational modelling of TKA can be divided into macro-mechanics studies by musculoskeletal (MSK) modelling and micro-mechanics studies by finite element modelling (Shu, Li, & Sugita, 2020). Finite element modelling is applied in cases in which localized structural deformations or soft tissues need to be described and analyzed in detail (Roupa et al., 2022). Although routinely used in biomechanics, finite element modelling is out of the scope of this thesis.

In vivo measurements of muscle forces are invasive, technically challenging, and ethically questionable (Komi et al., 1996; Pourcelot, Defontaine, Ravary, Lemâtre, & Crevier-Denoix, 2005). Several approaches have been proposed to estimate muscle forces, including static optimization (SO) and EMG-driven models. EMG-driven models combine joint kinematics and EMGs as inputs to estimate individual muscle forces and joint moments (Lloyd & Besier, 2003), whereas SO uses musculoskeletal geometry and joint moments calculated from inverse dynamics to estimate individual muscle forces at each instant in time. The muscle forces are resolved by minimizing the sum of squared muscle activations (Heintz & Gutierrez-Farewik, 2007). While direct validation of muscle forces is only possible with *in vivo* techniques, estimated muscle forces are often compared qualitatively against EMG measurements. This provides a broad indication of whether the estimated muscle force patterns are sensible. However, it offers limited use in the magnitude of the force estimates as there is no simple relationship between the EMG signal and the magnitude of force generation within the muscles (Trinler, Hollands, Jones, & Baker, 2018). EMG-driven and SO estimated muscle force profiles show broad agreement (Trinler et al., 2018). Force estimates are susceptible to model parameters, so subject-specific models may generate more physiologically plausible muscle forces (Akhundov et al., 2022).

Subject-specific MSK models use medical images to create individualized geometries and properties for each participant. This creates a more complex modelling workflow that is very time-consuming and creates additional uncertainties to the simulations due to the other parameters required (Kainz, Wesseling, & Jonkers, 2021; Modenese et al., 2018). A study which compared a generic model with SO, generic EMG-driven, subject-specific SO, and subject-specific EMG-driven found joint kinematics were similar between generic and subject-specific models, as the overall root mean squared difference was below 5°. Root mean squared differences in muscle forces were below 0.2 BW for all muscles between the different models. In contrast, the root mean squared differences in joint contact forces were up to 2.2 BW, with little differences occurring between generic SO and generic EMG-driven models. Their findings suggest that personalized geometry obtained from magnetic resonance imaging scans (i.e., subject-specific models) had a higher impact than the inclusion of personalized neural control (EMG) information (Kainz et al., 2021). Other studies also found that EMG-driven models presented limited improvements (Dumas & Moissenet, 2020; Marra et al., 2015; Zhang et al., 2022). Creating subject-specific models is costly and time-consuming, which often limits the sample size of a study which uses this workflow. Until the automated workflows to create highly subject-specific models improve, researchers will need to determine if the extra cost and time are worth it over the use of generic models (Akhundov et al., 2022). Generic MSK models that can accurately predict KCF provide a suitable solution as they can obtain accurate kinematic and dynamic outcomes and muscle forces in less time (Roupa et al., 2022).

Several generic MSK models, which vary in complexity, have been developed. Earlier MSK models were developed and validated to evaluate gait (Delp et al., 2007). However, gait does not utilize the knee's entire ROM, and consequently, this model would overestimate KCF at deep knee flexion angles (Imani Nejad et al., 2020; Schellenberg et al., 2018). KCFs from earlier MSK models, especially at both high and low knee joint flexion angles, should be interpreted carefully (Schellenberg et al., 2018). Later models were developed and modified to accommodate larger joint ROMs making them applicable to a larger range of tasks (Bruno Luiz Souza Bedo, Catelli, Lamontagne, & Santiago, 2020; Catelli, Wesseling, Jonkers, & Lamontagne, 2019; Lerner, DeMers, Delp, & Browning, 2015; Rajagopal et al., 2016). Some models can segment the vertical force into the medial and lateral compartments of the knee to better understand the distribution of the contact forces during various tasks (Curreli, Di Puccio, Davico, Modenese, & Viceconti, 2021; Fregly et al., 2012; Lerner et al., 2015). Researchers need to ensure their models provide accurate estimates of KCFs, which is often validated using *in vivo* measurements from instrumented knee prostheses (Curreli et al., 2021; Nejad et al., 2020; Schellenberg et al., 2018).

Various instrumented knee prostheses have been developed which can measure KCF *in vivo* during different ADLs (D'Lima, Patil, Steklov, Slamin, & Colwell, 2006; D'Lima, Steklov, Patil, & Colwell, 2008 ; Fregly et al., 2012; Heinlein et al., 2009; Kutzner et al., 2017; Kutzner et al., 2010; Moewis, Trepczynski, Bender, Duda, & Damm, 2022; Varadarajan, Moynihan, D’Lima, Colwell, & Li, 2008). However, none of the instrumented knee prostheses were PS or MBS designs, as they were either cruciate retaining (D'Lima et al., 2006; D'Lima et al., 2008) or ultra-congruent cruciate sacrificing designs (Heinlein et al., 2009; Kutzner et al., 2017; Kutzner et al., 2010; Moewis et al., 2022; Varadarajan et al., 2008).

While differences in implant designs exist, peak KCF measured *in vivo* was similar between the different studies, with peak KCF between 220 and 350% bodyweight (BW) for most dynamic activities. The lowest peak KCF occurs during stand-to-sit (225% BW). Sit-to-stand, single-leg stance and level walking caused similar peak KCF (246-261% BW), and the highest forces occurred during stair ascent (316% BW) and stair descent (346% BW) (Moewis et al., 2022).

The distribution of KCF between the medial and lateral compartments of the knee is activity dependent. During gait-type tasks (walking, climbing stairs), the medial KCFs are greater than lateral, whereas during sit-to-stand tasks, the lateral KCFs are greater than medial KCFs (Kutzner et al., 2017). This medial force ratio is dependent on both the tibiofemoral alignment (e.g., valgus/varus) and the activity (Kutzner et al., 2017; Moewis et al., 2022). This *in vivo* reference data is crucial for validating generic models.

Selecting a generic MSK model is an important step. Many generic models exist, some of which include Lower Limb Model 2010 (Arnold, Ward, Lieber, & Delp, 2010), Rajagopal2016 Full-Body Model (Rajagopal et al., 2016), and Lai2017 Full Upper and Lower Body Model (Lai, Arnold, & Wakeling, 2017). These models share the same body segment geometries, inertial properties (Ward, Eng, Smallwood, & Lieber, 2009), body coordinate reference systems, and knee joint kinematic definition (Walker, Rovick, & Robertson, 1988). The accuracy of these three models in predicting KCF is similar (Andersen, 2018; Chen et al., 2016; Curreli et al., 2021; Marra et al., 2015). Knee joint kinematic definitions in these models are based on healthy knees (Walker et al., 1988), so they cannot consider TKA implant geometries in their KCF estimates (Thorsen, Wen, & Zhang, 2021). More computationally demanding frameworks have combined MSK modelling and finite element analysis, which considers implant geometry. However, this involves using magnetic resonance imaging, computed tomography scans, and accurate implant geometries (Loi, Stanev, & Moustakas, 2021; Shu et al., 2018).

Several authors have used MSK modelling to estimate muscle forces and KCFs after TKA for various ADLs and obtain comparable results measured *in vivo*. Navacchia et al. (2016) estimated that KCFs in patients after TKA were greatest during stair descent (340% BW), lowest during level walking, and lateral KCF were greater than medial KCF during sit-to-stand (Navacchia et al., 2016). Li et al. (2013) found that patients with PS implants walked with a quadriceps avoidance pattern, as their vasti muscles contributed little to the knee extension moment during the stance phase of the gait cycle compared to healthy controls (Li et al., 2013). During stair ascent, patients with a PS implant had similar KCF as controls but had different muscle force compensatory strategies during the loading and push-off phases. TKA patients had reduced quadriceps muscle force but greater knee flexor muscle forces, which ultimately resulted in similar KCFs as controls (Rasnick, Standifird, Reinbolt, Cates, & Zhang, 2016). During more demanding tasks such as stepping down and pivoting, TKA patients adopted a “stiffening strategy” which included reduced knee flexion and varus angles, reduced knee extension moment. They had decreased quadriceps force compared to healthy controls. This strategy was likely the result of quadriceps avoidance, which may be due to instability or a persistent strategy developed with the progression of knee OA (Gaffney et al., 2016).

A recent study (Thorsen et al., 2021) compared medial and lateral KCF (Lerner et al., 2015) in operated and non-operated limbs in TKA patients during level and uphill walking. Their study included patients with PS, CR, and BiCR implants. Although they did not evaluate the effect of implant design, they did find that during uphill walking, total KCF was lower in the operated limb compared to the non-operated limb. Still, no between-limb differences existed during level walking. The decrease in total KCF during uphill walking was primarily due to decreased medial KCF (Thorsen et al., 2021). During stationary cycling, TKA patients had 23.5% less medial compartment loading on their operated limb than their non-operated limb (Hummer et al., 2022). TKA patients may have compensated more with their non-operated limb as a defensive mechanism whereby the operated knee was guarded at the expense of the non-operated limb, potentially increasing the risk of OA developing on the non-operated limb (Hummer et al., 2022; Shakoor, Block, Shott, & Case, 2002; Thorsen et al., 2021).

3 Study Design

Gaps in the literature | Framework | General methods

3.1 Gaps in the literature

From the review of the literature on knee OA and TKA (Chapter 2), it was apparent that most biomechanical studies were performed only in postoperative patients, and studies that compared MBS and PS implants were primarily focused on level walking conditions and overlooked tasks that placed more demand on the knee joint. Overall, the previous chapters of this thesis highlighted three main areas for further development:

1. Examine the changes of gait variability in patients with OA after they receive a TKA with either a PS or MBS implant.
2. Enhance the understanding of the post-operative effects of TKA with either MBS and PS implants on knee biomechanics and muscle activities during level walking, as well as more demanding tasks such as descending a ramp or staircase.
3. Simulate the dynamic knee joint loads in post-operative TKA patients with either PS or MBS implants during closed-chain, bilateral tasks such as sit-to-stand using MSK modelling.

Functional adaptations occur in patients with severe knee OA, but knee function does not readapt to reach the level of healthy controls after TKA (Stolarczyk et al., 2022). TKA is successful at reducing pain associated with knee OA. However, 20% of TKA recipients often remain dissatisfied due to the degree of improvement in knee function, unresolved pain, or unmet expectations (Gunaratne et al., 2017). PROMs continue to become more prevalent in the TKA literature (Lan et al., 2020). However, they may not be sensitive to detect functional deficits between implant groups as they only capture a subjective assessment of pain and functional deficits that persist following TKA (Stevens-Lapsley et al., 2010). Therefore, a more robust biomechanical comparison that evaluates knee kinematics, kinetics, muscle activity, muscle forces, and KCFs for various ADLs may indicate if an MBS or PS implant design is more appropriate to improve patients' function after TKA.

3.2 Framework

To better understand how biomechanical knee function changes after TKA with either an MBS or PS implant after a one-year follow-up during various ADLs, the purpose of this thesis was to address the question at large: *how will the choice of implants during TKA impact knee biomechanics during various activities of daily living?* With a cohort that includes pre-operative knee OA patients who underwent TKA with either a PS or MBS implant and healthy CTRL participants, this research program was separated into eight main studies, written as separated manuscripts:

1. **Limb-Matching in Healthy Controls**: this study examined the effect of different limb-matching methods on the statistical outcomes of biomechanical variables during a gait task. The aim was to identify the most appropriate methods for comparing CTRLs with OA and TKA participants (Chapter 4.1).
2. **Biomechanics Before and After TKA During Gait**: this study consisted of examining patient-reported outcome measures (PROMs) , temporospatial parameters,

and knee kinematics and kinetics during a level walking task before and one year after TKA using either an MBS or PS implant (Chapter 4.2).

3. **Waveform Test for Variance Inequality Development**: In this study, a test was developed to assess variability across an entire waveform. This test was utilized to identify the locations along the waveform the significant differences in variability amongst the groups. (Chapter 4.3).
4. **Variability Between Healthy Young and Older Adults**: this study compared the waveform variability to determine normal variability in healthy older adults (Chapter 4.4).
5. **Variability Before and After TKA During a Gait Task**: the aim of this study was to assess the waveform variability in patients with knee OA before and after TKA with either an MBS or PS implant during a level walking task (Chapter 4.5).
6. **Effect of Implant Design on Knee Biomechanics During Ramp Descent**: this study compared PROMs, temporospatial, and knee kinematics and kinetics in patients while descending a ramp (Chapter 4.6).
7. **Muscle Activity After TKA when Descending a Staircase**: The objective of this study was to compare the activity of quadriceps, hamstrings, and gastrocnemii muscles, as well as isometric strength, knee kinematics, and kinetics during a stair descent task. (Chapter 4.7).
8. **Sit-to-Stand Musculoskeletal Modelling in Patients After TKA**: this study examined knee muscle forces and KCF using MSK modelling during a sit-to-stand task (Chapter 4.8).

The general methods will be described in the following section, and an illustrative framework of the methodology of the eight studies in this thesis is presented in Figure 3.2.1.

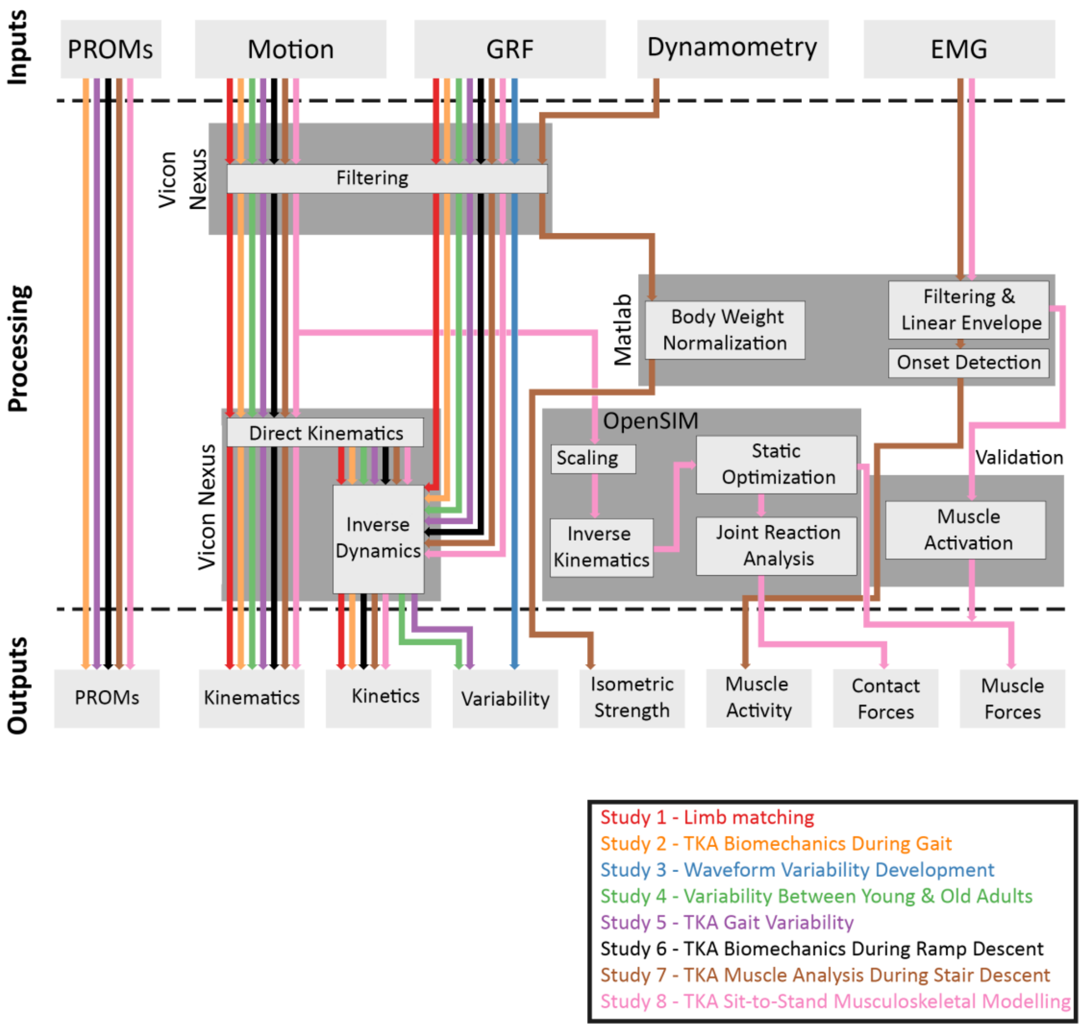


Figure 3.2.1: Thesis scheme: a methodological framework. The collected inputs were motion data (marker trajectories), ground reaction forces (GRF), isometric dynamometry, and knee muscles electromyography (EMG). Trajectories, GRF, and dynamometer signals were filtered in Vicon Nexus. Inverse kinematics and kinetics were completed, and EMG signals were filtered and processed in Matlab. The musculoskeletal model was scaled, and inverse kinematics, dynamics, static optimization, and joint reaction were run to estimate angles, torques, and muscle and contact forces. The muscle activations estimated from the static optimization were compared to those by EMG data for indirect validation. Isometric dynamometer data was calculated and normalized to body weight to determine isometric strength. The coloured lines highlight the parts of the methodological framework involved in the different studies.

All eight studies included in the thesis provide a cohesive understanding of joint biomechanics before and after TKA. Since patients were compared to a group of health controls, Study 1 (Chapter 4.1) was developed to determine if the selection of limb-matching method for the controls would affect the statistical outcome of future studies. Study 2 (Chapter 4.2) utilized a gait task to compare patients before and after TKA with controls. Gait is one of the most basic ADLs, so it was utilized as a comparison to all other studies. Studies 3 to 5 (Chapters 4.3 to 4.5) developed the Equality of Variance test (Chapter 4.3) and established normal variability within healthy participants (Chapter 4.4), before evaluating variability differences before and after TKA (Chapter 4.5). The ADLs performed in the remaining tasks were more challenging than level walking. In study 6 (Chapter 4.6) patients' knee biomechanics were evaluated as they descended a ramp. For study 7 (Chapter 4.7) muscle activations were evaluated during a stair descent task. Finally, study 8 (Chapter 4.8) estimated muscle and knee contact forces in patients after TKA while they performed a sit-to-stand task.

3.3 General methods

3.3.1 Participants

This thesis was primarily part of a large research program focused on knee OA patients before and after TKA. The initial goal was to recruit 20 participants in the MBS and PS groups. An effect size of 0.294 was extracted from previously collected TKA data from the Human Movement Biomechanics Laboratory for peak knee power absorption during ramp descent. Using G*Power, an apriori power calculation of 0.870 was calculated with total sample size of 60 in the 3 groups. However, due to Covid-19 related closures, twenty-eight participants completed pre- and post-operative assessments and were included in the thesis. Post-operative follow-ups were supposed to occur at 12±1 months. However, due to Covid-19 pandemic-related research closures, four PS participants had follow-up visits beyond 13 months (14.0 to 20.7 months post-operation). Using the previously mentioned effect size of 0.294, an apriori power calculation of 0.812 was determined with the final sample size of 42 participants within the 3 groups.

This research program also recruited 14 healthy individuals from the community of similar age and BMI to act as a control group. Studies 1, 3 and 4 required additional control participants, including a younger group. These other healthy participants were from previously collected studies, and their data were accessed from the Human Movement Biomechanics Laboratory research database. A summary of the demographic information of the participants who took part in the studies is found below in Table 3.3.1.

Table 3.3.1: Summary of the participant groups' mean (SD) demographics values used throughout the thesis.

	Pre-Operative			Post-Operative			
	Number (M/F)	Age (years)	BMI (kg/m^2)	Number (M/F)	Age (years)	BMI (kg/m^2)	Follow-up (months)
MP	6/8	62.7 (5.8)	27.9 (3.8)	6/8	63.7 (5.7)	27.4 (3.5)	12.4 (0.5)
PS	6/8	64.5 (8.1)	29.8 (3.4)	6/8	65.6 (8.1)	30.3 (3.9)	13.6 (2.2)*
Younger CTRLs (Studies 1,3,4)	13/15	31.3 (7.0)	24.4 (2.8)				
Older CTRLs (Studies 1,3,4)	17/18	63.8 (6.0)	25.5 (3.1)				
CTRL (Studies 2,5-8)	7/7	64.4 (5.6)	24.9 (2.1)				

* Four PS participants had follow-up visits beyond 13 months (14.0 to 20.7 months post-operation).

The Clinical Research staff of the Division of Orthopaedics at The Ottawa Hospital recruited knee OA patients on patients who were scheduled to undergo TKA by a single senior orthopedic surgeon (Dr. Geoffrey Dervin). The inclusion criteria required participants between the ages of 45 and 75 at the time of enrollment and needed to be willing to complete the required study visits. Participants were excluded if they had a BMI and waist circumference measurement > 35 kg/m^2 and 102 cm respectively for men, and > 35 kg/m^2 and 88 cm for women, or any past or present condition, which in the opinion of the investigators may impact gait; or previous joint replacement of the enrolled knee or other lower limb joint replacement. They were also excluded if they had a degenerative condition (other than OA in the enrolled knee) impacting joints of the lower extremities. Eligible participants underwent randomization to receive either an MBS

(MicroPort EVOLUTION® MP System with Cruciate Sacrificing (CS) tibial inserts) or PS (Zimmer® NexGen® PS TKA system with PS inserts) TKA. They visited the biomechanics laboratory within one month of surgery and 12 ± 1 months post-operatively (Figure 3.3.1). CTRL participants had the same inclusion criteria and were excluded if they had any degenerative condition impacting the joints of the lower extremities. CTRL participants completed a single visit.

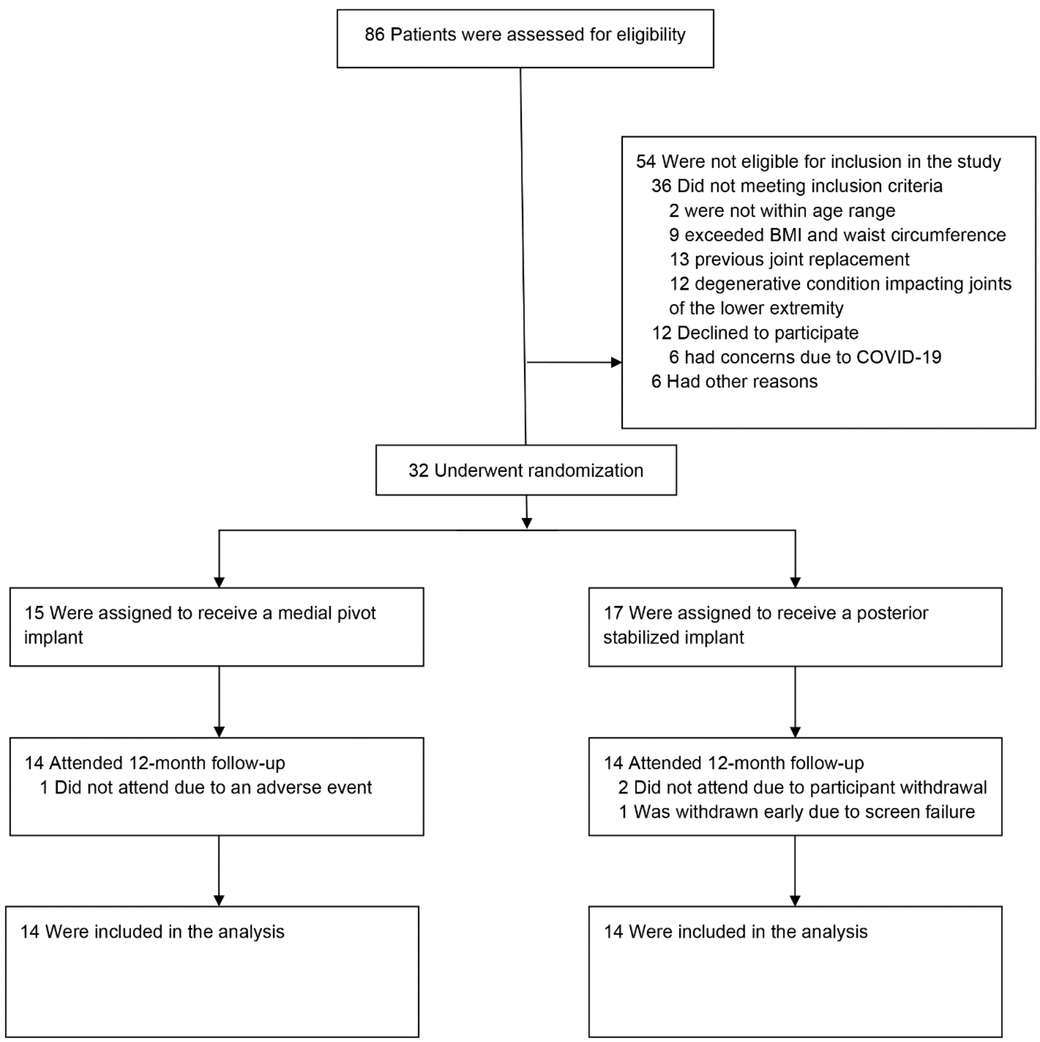


Figure 3.3.1: Consolidated Standards of Reporting Trials (CONSORT) flow diagram for enrolled patients.

MBS and PS groups underwent TKA with the same surgeon (GD) and same surgical approach involving a midline incision and subvastus approach (Hofmann et al., 1991). Manual instruments were used with the goals of mechanical neutral alignment with the femur-first technique and the tibial component at a coronal neutral angle. The protocol required patella resurfacing, and the PCL was released in all patients. All components were cemented, and tourniquet use was restricted to the time of cementation and deflated before closure. No patients required additional soft tissue release, and there were no complications or revisions with the surgical cohort.

3.3.2 Equipment

The motion capture visits occurred at the Human Movement Biomechanics Laboratory (University of Ottawa). The equipment included: a Monark cycle ergometer (Monark, Vansbro, SE), a Biodex isokinetic dynamometer (System 4 Pro, Biodex, Shirley, USA), an instrumented knee flexion/extension machine, ten infrared Vicon cameras (Vicon, Oxford, UK) sampled at 200 Hz, two fixed Bertec force plates (models FP4060-08, Bertec Corporation, Columbus, US) and two portable Kistler force plates (models 9286BA, Kistler Instruments Corp, Winterthur, CH) sampled at 1000 Hz, and a 16 channel Delsys Trigno (Delsys, Boston, US) surface EMG (sEMG) system sampled at 1000 Hz. Ground reaction forces and marker trajectories were recorded and synchronized through Vicon Nexus software (v2.11, Vicon, Oxford, UK).

Participants wore skin-tight shorts and short-sleeve shirts. Forty-five reflective markers with a diameter of 12 mm were placed according to the University of Ottawa Motion Analysis Model (UOMAM) using double-sided tape (Mantovani & Lamontagne, 2017). Sixteen sEMG electrodes were placed according to SENIAM guidelines with double-sided tape (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000) on muscles of the thigh and shank segments of both limbs.

3.3.3 Protocol

This thesis primarily incorporated the data from one large research program (Project GOLD - ClinicalTrials.gov identifier: NCT02589197), and all participants underwent the same motion capture protocol. The main exception was that some healthy participants in studies 1 to 3 were collected as part of other studies. Their data were obtained from the Human Movement Biomechanics Laboratory research database. Detailed descriptions are specified within each study (Chapters 4.1 to 4.8), and a brief explanation will follow in this section.

Participants arrived at the Human Movement Biomechanics Laboratory (University of Ottawa) for the motion capture protocol. Participants read and signed the informed consent forms, and the protocols were explained. They completed a KOOS questionnaire (Appendix – Chapter 8.2) (Roos et al., 1998) and then changed into spandex shorts and t-shirts to have a tight fit. They were asked to bring their own footwear (i.e., running shoes) to wear throughout the data collection. Participants performed a five-minute warm-up on a cycle ergometer, and then anthropometric measurements, including height, mass, leg length, knee width, and ankle width, were recorded.

sEMG electrodes were placed bilaterally on the following muscles: rectus femoris, vastus medialis, vastus lateralis, biceps femoris long head, semitendinosus, tibialis anterior, and both the medial and lateral heads of the gastrocnemius (Figure 3.3.2). All electrode placement locations were identified according to SENIAM guidelines (Hermens et al., 2000). The areas were shaved with a razor to remove hair, and the skin was cleaned with isopropyl alcohol. Probes were attached to the skin first with two-sided tape. The placement was verified by having the participants contract their muscles and view the signal on the computer. After signal/placement verification, probes were further secured, [1] with medical tape affixed to the skin; [2] wrapping the participants' thigh and shank segments with a polyurethane pre-taping foam under-wrap to protect the skin from tape; [3] rewrapping the thigh and shank segments with a 3" wide light elastic tape (Lightplast Pro, BSN, Luxembourg, LU); and [4] closing off the elastic tape with a 1.5" athletic tape to prevent it from coming undone during data collection.

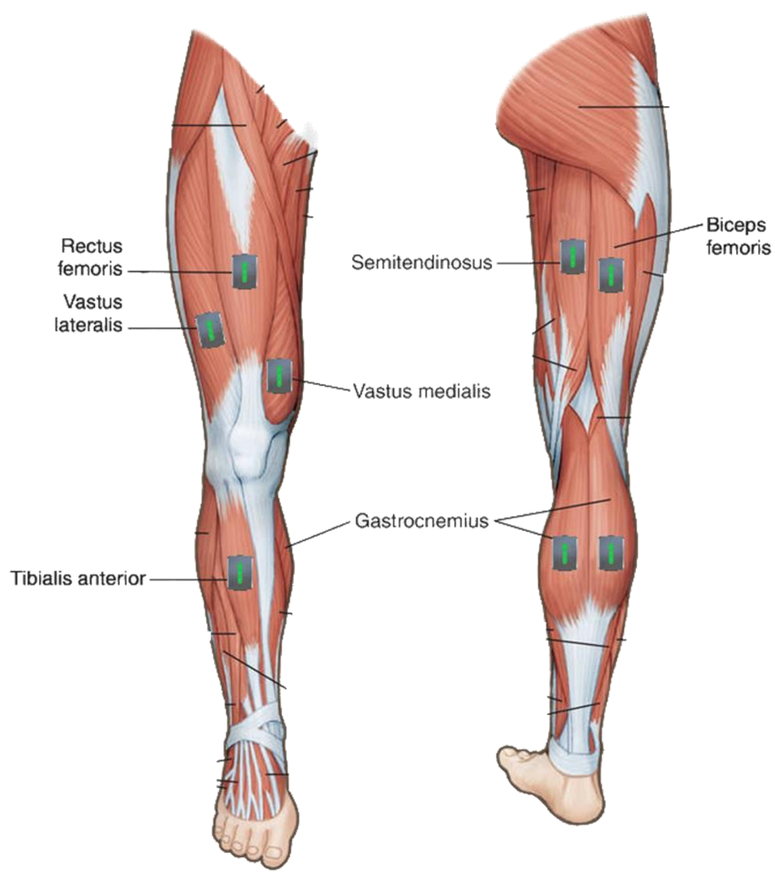


Figure 3.3.2: EMG electrode placement.

A protocol of maximum voluntary isometric contractions (MVICs) was performed to normalize the signals collected during the dynamic contractions and record the maximum isometric strength during knee flexion, extension, and ankle plantar and dorsiflexion. For the knee flexion and extension MVICs, participants sat in a modified knee flexion and extension exercise chair, instrumented with an S beam load cell (60001-300, Sensortronics, Covina, US), with their knee flexed at 60°. Ankle plantar flexion and dorsiflexion were performed with the participants seated in a Biodex isokinetic dynamometer with their knees at 60° and ankles at 90°. Each MVICs task lasted 5 seconds and was performed three times with a 30-second rest period between trials.

Once MVICs were completed, participants were outfitted with 45 reflective markers according to the University of Ottawa Motion Analysis Model (UOMAM) marker set (Mantovani & Lamontagne, 2017) – illustrated in the appendix (Chapter 8.3). Heel, toe, thigh, and medial knee and ankle markers were secured with additional pieces of tape.

The motion capture protocol first involved a static trial, where the participants had their feet parallel, pointing forward and one on each force platform, hip-width apart. Their arms and elbows were flexed, with palms facing down in a “motorcycle” pose. The static pose lasted five seconds and was further used for labelling and scaling. Following the static trial, participants performed several ADLs, including gait, ramp ascent/descent (Figure 3.3.3), stairs ascent/descent (Figure 3.3.4), sit-to-stand and stand-to-sit tasks. These tasks were performed in a random sequence, and for this thesis, the level walking, ramp descent, stair descent, and sit-to-stand functions were observed. Participants were given as many practice trials as necessary to perform the motions comfortably. Participants completed five trials of each task, which was performed at the participant’s preferred speed.

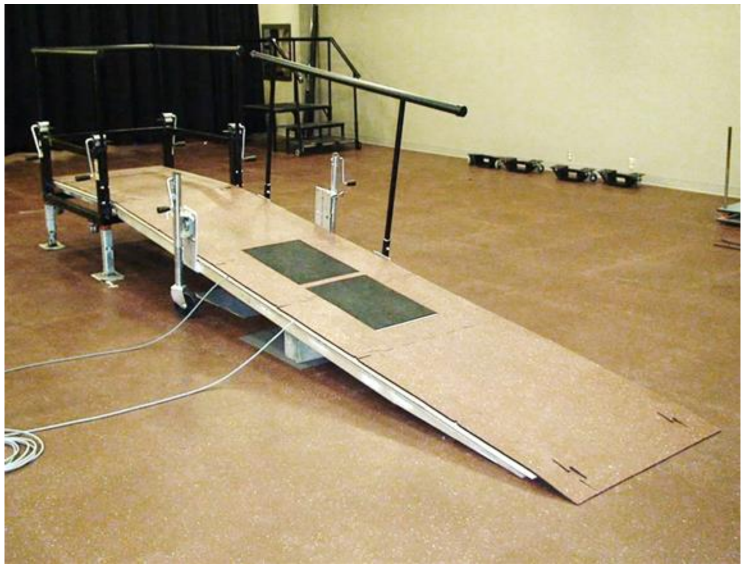


Figure 3.3.3: Instrumented ramp with slope set to 9 degrees.

Figure 3.3.4: Instrumented staircase.

3.3.4 Data processing

Motion capture data were first processed in Vicon Nexus software (version 2.11.0), where marker trajectories were labelled according to the UOMAM and filtered (Woltring, mean squared error = 15 mm^2) (Woltring, 1986). Ground reaction forces were also filtered (zero-lag, fourth-order Butterworth, cut-off 10 Hz). Gait events were identified with the aid of the force platforms using a threshold of 20 N. Following filtering and event detection, data were modelled within the Vicon Nexus software with the UOMAM (Mantovani & Lamontagne, 2017) to obtain joint angles, moments, and powers. For study 8 (Chapter 4.8), data were prepared for OpenSim format by exporting the .c3d data as .trc and .mot files format for further processing. Relevant data for each study were extracted using custom-written scripts within MATLAB (2019b, Mathworks, Natick, US), where kinematic data were time-normalized to 101 points (0-100%) to represent the entire gait cycle, and kinetic data were time normalized to the 63 points (0-62%) to represent the stance phase (Winter, 1984).

Electromyography (EMG) data were processed within MATLAB according to standards recommended by the International Society of Electrophysiology and Kinesiology (Merletti & Di Torino, 1999). This included band-pass filtering with a fourth-order recursive Butterworth filter, wave rectification, and applying a low-pass, fourth-order recursive Butterworth filter at 6 Hz to create a linear envelope. The linear envelopes of each muscle were then amplitude-normalized by their respective MVIC values, which were determined from rectified data using a moving-average window (100ms) (Rutherford, Hubley-Kozey, & Stanish, 2011). Study 7 (Chapter 4.7) explains the methods used for signal onset detection and extracting the variables of interest from the EMG signals.

All simulations were performed in OpenSim 3.3 (Delp et al., 2007) using the model described in study 8 (Chapter 4.8), which was based on the full-body MSK model designed by Uhlrich (Uhlrich, Jackson, Seth, Kolesar, & Delp, 2022). This was an updated version of the Rajagopal MSK model (Rajagopal et al., 2016). The Uhlrich model updated passive muscle force curves and improved hip abductor paths (Uhlrich et al., 2022). Adaptations of the Uhlrich model were made to compute medial and lateral tibiofemoral contact forces (Bedo et al., 2020; Lerner et al., 2015). The model included 37 degrees of freedom to define joint kinematics, Hill-type models of 80 muscle-tendon units actuating the lower limbs (Millard, Uchida, Seth, & Delp, 2013; Zajac, 1989), and 17 ideal torque actuators driving the upper body. Musculotendon parameters were developed from the anatomical measurements of cadaveric specimens (Ward, Eng, Smallwood, & Lieber, 2009) and magnetic resonance images of healthy young individuals (Handsfield, Meyer, Hart, Abel, & Blemker, 2014). All participants were scaled from the generic MSK model. The Batch OpenSIM Processing Scripts (BOPS) MATLAB toolbox was used to process inverse kinematics, inverse dynamics, muscle analysis, static optimization, and joint reaction analysis (Bedo et al., 2021). Static optimization was used to compute muscle forces, which minimized the sum of squared muscle activation (Buchanan & Shreeve, 1996; Todorov & Jordan, 2002). Joint reaction forces were computed using the resultant forces and moments acting on each articulating joint from all muscle forces and the internal and external loads applied to the model. Muscle forces and knee contact forces were normalized to body weight for each participant.

The variables of interest included: KOOS scores, isometric strength measures, knee joint angles, knee joint moments, knee joint powers, muscle onsets, muscle total time on, peak linear envelopes, total muscle activity, muscle forces, and KCF. The statistical analyses utilized are described in each study. activity, muscle forces, and KCF. The statistical analyses utilized are described in each study. Discrete variables were assessed for normal distribution using a Shapiro-Wilk's test ($p > .05$), and for homogeneity of variance using a Levene's test ($p < .05$). If data was not normally distributed, assumption of homogeneity of variance was violated, or if sample sizes were small, non-parametric statistics were used (Chapters 4.2 and 4.8). Otherwise, parametric statistic statistics were utilized.

Women and men were included in the sampled population for this thesis. Although some evidence suggests sex-differences exist after TKA such as men walking with greater stride lengths (McClelland, Feller, & Webster, 2018) and women experiencing less decline in quadriceps and hamstrings strengths, these differences did not persist after 6 months (Gustavson et al., 2016). Instead of using sex as a covariate, this thesis had similar number of males and females in each group. The current sample size used throughout the thesis could result in the misinterpretation or misrepresentation of any potential differences identified between the sexes.

4 Results

4.1 Side does not matter in healthy young and older individuals - Examining the importance of how we match limbs during gait studies

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Contents of this chapter published as an article in *Gait & Posture*

January 2019

Kowalski, E., Catelli, D. S., & Lamontagne, M. (2019). Side does not matter in healthy young and older individuals - Examining the importance of how we match limbs during gait studies. Gait & posture, 67, 133–136. https://doi.org/10.1016/j.gaitpost.2018.10.008

4.1.1 Abstract

Background: Various methods exist when comparing gait data between groups and include the analysis of a single limb, or taking an average of both limbs. Evidence exists suggesting that both limbs are not symmetrical, so statistical differences may exist in biomechanical variables when comparing gait with different limb-matching methods.

Research question: Does limb-matching method have an effect on statistical outcome when comparing biomechanical variables during a gait task?

Methods: This retrospective study compared forty participants separated into a younger and older group as they completed a gait task. Twenty-five commonly used biomechanical variables were compared between the two groups using four different limb-matching methods: (i) average of both limbs; (ii) dominant limb; (iii) non-dominant limb; (iv) random limb. A mixed linear model was used to compare all the biomechanical variables between the younger and older group using the different limb-matching methods.

Results: Limb-matching methods only had a significant effect for 1/25 variables examined. Group effects between the younger and older groups were more prevalent, with the most significant effects occurring at the ankle joint.

Significance: Limb-matching methods do not have a direct effect on biomechanical outcomes when comparing gait in healthy young and old groups. Gait is cyclical, so limb symmetry is often assumed. However, if the complexity of the task increases, or when comparing against groups with impaired gait, both limbs may behave differently, so limb-matching method may become more crucial.

4.1.2 Introduction

Gait is a basic requirement of daily life, and a major determinant of independence and quality of life (Schmid et al., 2007). However, gait is frequently impaired by a variety of musculoskeletal and neurological conditions, or surgical intervention (e.g. osteoarthritis, Parkinson's, stroke, total knee arthroplasty, etc.). In rehabilitation, gait receives a lot of attention due to its importance on restoring patients' independence (Latham et al., 2005). To understand how gait changes from impairment to rehabilitation, it is often necessary to use healthy individuals as a comparison.

Walking may seem like a simple task, but is a complex motor skill requiring several inter-linked pathways from the cortex to the muscles (Joffe, 1992). There is a large interaction involving the central nervous system and various muscles in order to maintain the body upright, while at the same time moving in a smooth, rhythmical motion. Since gait in healthy individuals is seen as smooth and rhythmical, gait symmetry is often assumed in the literature for the sake of data collection and analysis (Sadeghi, Allard, Prince, & Labelle, 2000). This simple assumption could be one of the primary reasons why many gait studies rely either on unilateral data collection (Cavanagh & Gregor, 1975; Crowinshield, Brand, & Johnston, 1978; Eng & Winter, 1995) or an average of the left and right limb (Hannah, Morrison, & Chapman, 1984; Ounpuu, Gage, & Davis, 1991; Winter, 1984). However, in these studies gait symmetry was not tested as it was assumed.

Before determining if gait is symmetrical or asymmetrical, it is essential to establish a definition of gait symmetry. Previously gait symmetry has been defined as a perfect agreement between the actions of the lower limbs (Herzog, Nigg, Read, & Olsson, 1989); the definition we prefer for gait symmetry is no statistical differences noted on parameters measured bilaterally

(Gundersen et al., 1989) . Gait studies which have examined gait symmetry have either confirmed gait symmetry (Hamill, Bates, & Knutzen, 1984; Vanderstraaten & Scholten, 1978), or have found differences in gait parameters between both limbs (Chatinier & Rozendal, 1970; Singh, 1970). With many studies in contradiction with each other, perhaps our reliance on unilateral or averaged left and right limbs during gait studies is incorrect.

When relying on unilateral data, such as using only the left or right limbs, the researcher is also assuming that both limbs are similar. However, as with limb dominance in the upper extremity (i.e., handedness), functional asymmetries between the lower-limbs have been documented in tasks other than walking (Gabbard & Hart, 1996; Gabbard & Iteya, 1996). The idea of limb dominance is that the two hemispheres of the human brain are functionally dissimilar (Gabbard & Hart, 1996; Sadeghi et al., 2000). Differences between the dominant and non-dominant limbs have been made based on the roles of the lower limbs. The foot used in activities is the dominant foot, whereas the foot providing the stability and postural support is the non-dominant foot (Peters, 1988). Other terms used in the literature to differentiate limbs have been the mobilization and stabilization, which represent the dominant and non-dominant limbs, respectively (Dargentpare, Deagostini, Mesbah, & Dellatolas, 1992; Gabbard, 1989; Peters, 1988). For continuity and simplicity, we will use the terms dominant and non-dominant limbs.

With certain tasks, there is a clear difference between the dominance of lower limbs (e.g., kicking a ball) (Gabbard & Hart, 1996; Gabbard & Iteya, 1996). With gait, it is not as clear, since it depends on which definition of symmetry is used and what variables are studied. However, researchers continue to use various limb-matching methods when comparing between subjects. These different methods may change the statistical outcome of the studies. When using gait data of healthy subjects to compare to impaired gait, such as those with osteoarthritis or a prosthetic joint in one limb, the method used to match the limb may be even more important to the statistical outcome. The purpose of this study was to explore various limb matching methods and its effect on the statistical outcome when comparing gait biomechanics between a younger and older group.

4.1.3 Methods

Data were collected from an ongoing standardized database that includes gait data of healthy individuals who were collected since 2009. The original study that the participants were volunteers was approved by the Research Ethics Board at the University and they provided written informed consent prior to participation in the study.

Participants were selected from the database if they were free from lower-limb injury and had no musculoskeletal or neurological disorder which would negatively impact gait. A total of 76 participants were eligible for inclusion in the study. From this cohort, participants were selected if they had a preferred walking speed (PWS) between 1.30 to 1.50 m/s. A total of 40 participants were included and were separated into two groups based on their age: a younger group (ages 23-44) and an older group (ages 54-81) (Table 4.1.1). For each participant, the dominant limb was identified as the preferred limb used to kick a ball (Chapman, Chapman, & Allen, 1987).

Table 4.1.1: Group mean demographic values. The group mean (SD) for each age group is provided for all of the recorded demographic variables.

	Younger	Older
Range (years)	23-44	54-81
Number (n)	20	20
Age (years)	30.5(7.2)	63.4(6.4)
Sex (male/female)	8/12	9/11
Body Mass Index (kg/m^2)	24.3(2.9)	25.4(3.3)
Dominant Limb (right/left)	19/1	18/2

Spatiotemporal and kinematic parameters of gait were measured using a 10-camera Vicon System (MX-13, Oxford Metrics, Oxford, UK) sampled at 200 Hz, and two Bertec force platforms (FP4060, Bertec Corporation, Columbus, USA) sampled at 1000 Hz. Participants were outfitted with 45 passive-reflective markers according to the University of Ottawa Motion Analysis Model (UOMAM) (Mantovani & Lamontagne, 2017). Following a static trial, all participants performed a standardized walking task: three trials of walking at their preferred speed, across a level 10 m walkway which included the two force plates in the middle.

Gait data were processed using Vicon Nexus 2.6 software (Oxford Metrics, Oxford, UK). Trajectories were filtered using a Woltring filter with a mean standard error of 15 and force platform data were filtered using a 4th order (zero lag) Butterworth filter with a cut-off frequency of 10 Hz. Gait events were identified with the help of the force plates and the walking trials were modeled with the UOMAM. Data were exported to Matlab R2016a (MathWorks, Natick, USA) to extrapolate spatiotemporal, kinematic and kinetic variables of interest.

A total of 25 commonly used variables in biomechanics were selected for analysis (Table 4.1.2). Group means for all these variables were compared between the younger and older groups with the following limb matching methods: (i) average of both limbs; (ii) dominant limb; (iii) non-dominant limb, and (iv) limb selected at random.

All statistical analyses were done using SPSS v.20 software (IBM Corporation, Armonk, USA) to perform a mixed linear model (MLM) of the relationship between age group and limb matching method on the various biomechanical variables. As fixed effects, we entered age group (young/old) and the four limb-matching methods into the model. Participants were entered as a random effect, and alpha was set at 0.05 for all tests. For any significant differences between limb-matching methods, a Bonferroni correction was used to identify where the significant difference occurred. Using the MLM, we were able to determine if there was a statistical effect of group, limb-matching method, as well as the interaction between them.

Table 4.1.2: Spatiotemporal, kinematic, and kinetic variables of interest.

Variable	Unit	Abbreviation
Spatiotemporal		
Walking speed normalized to leg length		
Stride time	(s)	
Step time	(s)	
Stride length normalized to leg length		
Step length normalized to leg length		
Kinematics		
Peak hip flexion angle during stance	(°)	HA1
Peak hip extension angle	(°)	HA2
Peak hip flexion angle during swing	(°)	HA3
Peak knee flexion angle	(°)	KA1
Peak ankle dorsiflexion angle	(°)	AA1
Peak ankle plantar flexion angle	(°)	AA2
Kinetics		
Support moment at early stance	(Nm/kg)	SM1
Support moment at late stance	(Nm/kg)	SM2
Peak hip flexion moment	(Nm/kg)	HM1
Peak hip extension moment	(Nm/kg)	HM2
Peak knee flexion moment	(Nm/kg)	KM1
Peak knee extension moment	(Nm/kg)	KM2
Peak ankle dorsiflexion moment	(Nm/kg)	AM1
Peak ankle plantar flexion moment	(Nm/kg)	AM2
Peak hip power absorption	(W/kg)	HP1
Peak hip power generation	(W/kg)	HP2
Peak knee power absorption	(W/kg)	KP1
Peak knee power generation	(W/kg)	KP2
Peak ankle power absorption	(W/kg)	AP1
Peak ankle power generation	(W/kg)	AP2

4.1.4 Results

The complete results comparing the different limb matching methods for all the variables of interest are located in the appendix (Tables 4.1.4 to 4.1.7). In general, most variables did not have a significant limb matching method effect. The only variable which had a significant limb matching method effect was peak ankle plantar flexion moment, which had a significant difference between the dominant and non-dominant limb matching methods ($p = 0.014$) (Figure 4.1.1). Several variables, primarily at the ankle joint, had a significant group effect between the young and old groups (Table 4.1.3). None of the other biomechanical variables examined had a significant group-limb matching method interaction.

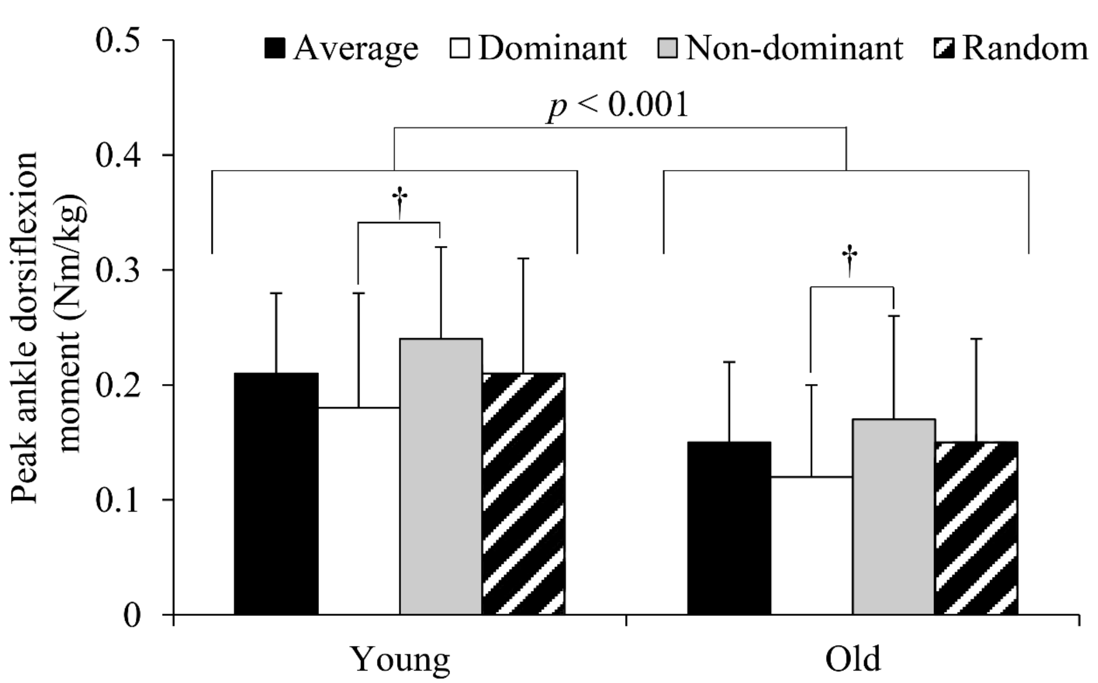

Figure 4.1.1: Peak ankle dorsiflexion moment (Nm/kg) of younger and older participants during a gait trial compared using various limb-matching methods. Mixed linear model group effect is indicated in the figure, and a significant limb-matching method is represented by a †.

4.1.5 Discussion

Researchers in biomechanics use a variety of limb-matching methods to compare data amongst groups. The aim was to identify if limb-matching method has a statistical effect on the biomechanical variables. Many gait studies rely either on unilateral data collection (Cavanagh &

Table 4.1.3: Group mean ankle joint kinematic and kinetic values during gait in a young and older group. The group mean (SD) for each of the four limb matching methods is provided for the spatiotemporal variables. The *p*-values are calculated from the linear mixed model where group and limb matching methods were fixed variables, and participants were set as random variables. Where applicable, a significant limb matching method effect was adjusted with a *Bonferroni* correction to identify where the significant difference occurred (†). Interaction effect is the Group*Limb Matching Method effect.

Variable	Limb Matching Methods	Group		Group Effect (*p* – value)	Limb Matching Method Effect (*p* – value)	Interaction Effect (*p* – value)
		Young Mean (SD)	Old Mean (SD)			
Peak ankle dorsiflexion (°)	Average	10.8 (3.3)	10.7 (3.0)	0.961	0.534	0.995
	Dominant	10.1 (3.3)	10.2 (3.8)			
	Non-dominant	11.5 (4.3)	11.2 (3.6)			
	Random	10.5 (3.5)	10.5 (3.8)			
Peak ankle plantar flexion (°)	Average	22.3 (6.0)	20.0 (4.3)	0.014	0.639	0.498
	Dominant	24.2 (5.8)	19.9 (4.4)			
	Non-dominant	20.5 (9.6)	20.0 (5.1)			
	Random	22.7 (6.7)	20.2 (4.2)			
Peak ankle dorsiflexion moment (Nm/kg)	Average	0.21 (0.07)	0.15 (0.07)	<0.001	0.025	0.999
	Dominant	0.18 (0.10)	0.12 (0.08)			
	Non-dominant	0.24 (0.08)	0.17 (0.09)			
	Random	0.21 (0.10)	0.15 (0.09)			
Peak ankle plantar flexion moment (Nm/kg)	Average	1.13 (0.14)	0.86 (0.14)	<0.001	0.596	0.251
	Dominant	1.07 (0.17)	0.89 (0.15)			
	Non-dominant	1.19 (0.17)	0.85 (0.19)			
	Random	1.10 (0.15)	0.83 (0.26)			
Peak ankle power absorption (W/kg)	Average	0.71 (0.37)	0.91 (0.22)	0.001	0.947	0.909
	Dominant	0.71 (0.39)	0.87 (0.30)			
	Non-dominant	0.72 (0.42)	0.95 (0.21)			
	Random	0.74 (0.38)	0.87 (0.27)			
Peak ankle power generation (W/kg)	Average	3.45 (0.77)	2.84 (0.50)	<0.001	0.671	0.811
	Dominant	3.30 (0.82)	2.82 (0.59)			
	Non-dominant	3.61 (0.90)	2.84 (0.55)			
	Random	3.34 (0.84)	2.77 (0.51)			

Gregor, 1975; Crowinshield et al., 1978; Eng & Winter, 1995) or an average of the left and right limb (Hannah et al., 1984; Ounpuu et al., 1991; Winter, 1984) when making comparisons. These methods are so often used because we think of gait a cyclical motion, therefore, symmetrical motion (Sadeghi et al., 2000). Gait symmetry is when no statistical differences exist on parameters measured bilaterally (Gundersen et al., 1989). Although the consensus on gait symmetry is still ongoing, it is important to understand the statistical impact of matching limbs has on the outcomes during gait studies.

We used a MLM to compare a total of 25 commonly used biomechanical variables in a young and old group completing a walking task. The benefit of the MLM was that it allowed for identification of significant differences between the young and old group, and between the four limb matching methods. Although the means varied between the different limb matching methods, the MLM confirmed that the various limb-matching methods were statistically the same ($p > 0.05$). Only a single variable, peak ankle dorsi-flexion moment, had a significant difference between the dominant and non-dominant limb matching methods (Figure 1). Therefore, when comparing gait in healthy groups, limb-matching the groups is not necessary as they will produce similar statistical outcomes.

Although there are studies which confirm that gait is symmetrical (Hamill et al., 1984; Vanderstraaten & Scholten, 1978) or asymmetrical (Chatinier & Rozendal, 1970; Singh, 1970), our findings would confirm that gait is symmetrical except for peak ankle dorsi-flexion moment. We asked our participants to walk in a straight line at a self-selected pace. Although gait is a complex motor skill (Joffe, 1992), when moving in a straight line, without any obstacles, it is as smooth and rhythmical as a task can be. When performing more complex movements or sport-specific skills which involve more motor skills such as kicking a ball, the lower limbs may perform differently, so selecting appropriate limb-matching methods during these tasks may be more crucial.

Our cohort included young and old healthy adult participants. Gait studies often include analyses on participants whose gait is impaired by musculoskeletal and neurological conditions (e.g. Parkinson's, stroke, osteoarthritis), or surgical interventions (total hip arthroplasty, total knee arthroscopy). For these participants, gait symmetry should not be assumed, so when comparing them against healthy controls, limbs should be matched more carefully.

Gait analysis is more commonly being used before and after orthopedic intervention to provide objective evidence on the joint biomechanical function (Lamontagne, Beaulieu, Varin, & Beaulé, 2009). Researchers are encouraged to use a control group to make comparisons with, instead of using the contralateral limb, as this limb's function may be altered because of compensating for the affected limb (Lamontagne et al., 2009). Our findings suggest that limb-matching methods in healthy adult participants do not affect statistical outcome, so regardless of which limb-matching method researchers select, their comparison will be valid. Many studies have compared young adults' gait with older adults' gait. In general these studies have identified that older adults have a slower walking speed, shorter step length, reduced range of motion in the lower extremity joints, and a distal to proximal shift in joint torques (DeVita & Hortobagyi, 2000; Gabell & Nayak, 1984; Judge, Davis, & Ounpuu, 1996; Kerrigan, Todd, Della Croce, Lipsitz, & Collins, 1998). Although limb-matching methods were not specified in all instances, it is still evident that gait of older adults is different from younger adults. Our findings were generally in agreement, as the MLM identified significant age group effect for many variables. Ankle joint kinetics were significantly lower in older adults than younger adults (Judge et al., 1996; Kerrigan et al., 1998). Reductions in ankle joint mobility and strength may be the cause of the age-related differences observed between young and old adults; therefore, future studies could investigate this further.

This study has certain limitations which must be addressed. First, more than 90% of our participants identified as right limb dominant when asked for their preferred limb when kicking a ball (Chapman et al., 1987). One should not assume that using only the right limb will always provide the same statistical outcome as the dominant limb, because if a cohort has primarily left-limb dominant participants, this may not be the case. Second, we compared healthy young and old groups, so differences in biomechanical variables using different matching-limbs may exist in participants whose gait is impaired. The preferred walking speed which was homogeneous among the participants could also attenuate the limb asymmetry.

4.1.6 Conclusion

In conclusion, our findings have shown that the limb-matching methods do not have a direct influence on the statistical outcome when comparing gait of younger and older healthy individuals. The dominant and non-dominant limb matching methods had only a significant effect on the peak ankle dorsi-flexion moment. This study showed that there were mainly no significant limb-matching method effects; confirming that both the dominant and non-dominant sides were symmetrical for the healthy participants. If the cohort changes to include participants with impaired gait (e.g. stroke, Parkinson’s, osteoarthritis, or joint arthroplasty), both limbs may not perform identically. However, if using a control group for comparison with an impaired gait group, the way limbs are matched between groups is not critical. Therefore, future studies are necessary to investigate participants with impaired gait.

4.1.7 References

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4.1.8 Appendix

Table 4.1.4: Group mean spatiotemporal values during gait in the younger and older group. The group mean (SD) for each of the four limb matching methods is provided for the spatiotemporal variables. The *p*-values are calculated from the linear mixed model where group and limb matching methods were fixed variables, and participants were set as random variables. Where applicable, a significant limb matching method effect was adjusted with a *Bonferroni* correction to identify where the significant difference occurred (†). Interaction effect is the Group*Limb Matching Method effect.

Variable	Limb Matching Methods	Group Young Mean (SD)	 Old Mean (SD)	Group Effect (*p*–value)	Limb Matching Method Effect (*p*–value)	Interaction Effect (*p*–value)
Walking Speed normalized to leg length	Average	1.60 (0.10)	1.56 (0.10)	0.011	0.935	0.961
	Dominant	1.60 (0.09)	1.57 (0.10)			
	Non-dominant	1.60 (0.11)	1.55 (0.12)			
	Random	1.60 (0.10)	1.56 (0.12)			
Stride Time (s)	Average	1.05 (0.05)	1.10 (0.07)	<0.001	0.952	0.960
	Dominant	1.05 (0.05)	1.09 (0.07)			
	Non-dominant	1.05 (0.05)	1.10 (0.08)			
	Random	1.05 (0.05)	1.10 (0.08)			
Step Time (s)	Average	0.53 (0.03)	0.55 (0.04)	<0.001	0.931	0.946
	Dominant	0.52 (0.03)	0.55 (0.04)			
	Non-dominant	0.53 (0.03)	0.56 (0.05)			
	Random	0.53 (0.02)	0.56 (0.05)			
Stride Length normalized to leg length	Average	1.68 (0.08)	1.71 (0.12)	0.125	0.996	1.000
	Dominant	1.68 (0.08)	1.71 (0.11)			
	Non-dominant	1.68 (0.08)	1.71 (0.14)			
	Random	1.68 (0.09)	1.71 (0.13)			
Step Length normalized to leg length	Average	0.84 (0.04)	0.85 (0.06)	0.257	0.736	0.705
	Dominant	0.85 (0.06)	0.85 (0.06)			
	Non-dominant	0.83 (0.04)	0.85 (0.07)			
	Random	0.85 (0.04)	0.85 (0.06)			

Table 4.1.5: Group mean kinematic values during gait in a younger and older group. The group mean (SD) for each of the four limb matching methods is provided for the spatiotemporal variables. The *p*-values are calculated from the linear mixed model where group and limb matching methods were fixed variables, and participants were set as random variables. Where applicable, a significant limb matching method effect was adjusted with a *Bonferroni* correction to identify where the significant difference occurred (†). Interaction effect is the Group*Limb Matching Method effect.

Variable	Limb Matching Methods	Group Young Mean (SD)	 Old Mean (SD)	Group Effect (*p* – value)	Limb Matching Method Effect (*p* – value)	Interaction Effect (*p* – value)
Peak hip flexion during stance (°)	Average	31.4 (3.8)	33.5 (4.9)	0.003	0.993	0.995
	Dominant	31.4 (3.6)	33.3 (5.0)			
	Non-dominant	31.4 (4.3)	33.8 (5.7)			
	Random	31.3 (3.6)	33.4 (4.7)			
Peak hip extension (°)	Average	16.6 (4.1)	16.4 (3.4)	0.979	0.927	0.923
	Dominant	16.5 (4.0)	16.7 (3.6)			
	Non-dominant	16.8 (4.7)	16.1 (3.9)			
	Random	15.7 (4.3)	16.3 (4.0)			
Peak hip flexion during swing (°)	Average	33.7 (4.3)	36.0 (4.5)	0.001	0.981	0.855
	Dominant	33.4 (4.3)	36.5 (4.9)			
	Non-dominant	34.1 (4.9)	35.6 (4.6)			
	Random	33.7 (4.6)	36.3 (4.7)			
Peak knee flexion (°)	Average	64.9 (4.5)	65.3 (3.4)	0.760	0.882	0.583
	Dominant	63.8 (5.5)	65.6 (3.6)			
	Non-dominant	66.0 (4.6)	65.0 (5.0)			
	Random	65.2 (5.4)	64.8 (4.8)			
Peak ankle dorsiflexion (°)	Average	10.8 (3.3)	10.7 (3.0)	0.961	0.534	0.995
	Dominant	10.1 (3.3)	10.2 (3.8)			
	Non-dominant	11.5 (4.3)	11.2 (3.6)			
	Random	10.5 (3.5)	10.5 (3.8)			
Peak ankle plantar flexion (°)	Average	22.3 (6.0)	20.0 (4.3)	0.014	0.639	0.498
	Dominant	24.2 (5.8)	19.9 (4.4)			
	Non-dominant	20.5 (9.6)	20.0 (5.1)			
	Random	22.7 (6.7)	20.2 (4.2)			

Table 4.1.6: Group mean joint moment values during gait in a younger and older group. The group mean (SD) for each of the four limb matching methods is provided for the spatiotemporal variables. The *p*-values are calculated from the linear mixed model where group and limb matching methods were fixed variables, and participants were set as random variables. Where applicable, a significant limb matching method effect was adjusted with a *Bonferroni* correction to identify where the significant difference occurred (†). Interaction effect is the Group*Limb Matching Method effect.

Variable	Limb Matching Methods	Group Young Mean (SD)	 Old Mean (SD)	Group Effect (*p* – value)	Limb Matching Method Effect (*p* – value)	Interaction Effect (*p* – value)
Support moment during early stance (Nm/kg)	Average	0.64 (0.38)	0.94 (0.42)	<0.001	0.939	0.816
	Dominant	0.63 (0.43)	1.01 (0.47)			
	Non-dominant	0.66 (0.39)	0.88 (0.42)			
	Random	0.57 (0.42)	0.94 (0.46)			
Support moment during late stance (Nm/kg)	Average	0.50 (0.18)	0.49 (0.21)	0.677	0.990	0.641
	Dominant	0.47 (0.22)	0.53 (0.22)			
	Non-dominant	0.52 (0.20)	0.46 (0.24)			
	Random	0.45 (0.22)	0.52 (0.23)			
Peak hip extension moment (Nm/kg)	Average	0.74 (0.21)	0.70 (0.20)	0.123	0.918	0.920
	Dominant	0.75 (0.24)	0.73 (0.26)			
	Non-dominant	0.74 (0.23)	0.66 (0.22)			
	Random	0.75 (0.23)	0.68 (0.24)			
Peak hip flexion moment (Nm/kg)	Average	1.10 (0.19)	1.07 (0.12)	0.347	0.979	0.851
	Dominant	1.13 (0.22)	1.06 (0.19)			
	Non-dominant	1.07 (0.22)	1.09 (0.19)			
	Random	1.12 (0.25)	1.08 (0.22)			
Peak knee extension moment (Nm/kg)	Average	0.70 (0.30)	0.55 (0.25)	0.004	0.995	0.664
	Dominant	0.65 (0.34)	0.58 (0.28)			
	Non-dominant	0.74 (0.31)	0.52 (0.28)			
	Random	0.67 (0.35)	0.57 (0.24)			
Peak knee extension moment (Nm/kg)	Average	0.15 (0.16)	0.50 (0.17)	0.090	0.789	1.000
	Dominant	0.13 (0.20)	0.19 (0.21)			
	Non-dominant	0.18 (0.16)	0.22 (0.23)			
	Random	0.14 (0.19)	0.19 (0.22)			
Peak ankle dorsiflexion moment (Nm/kg)	Average	0.21 (0.07)	0.15 (0.07)	<0.001	0.025	0.999
	Dominant†	0.18 (0.10)	0.12 (0.08)			
	Non-dominant†	0.24 (0.08)	0.17 (0.09)			
	Random	0.21 (0.10)	0.15 (0.09)			
Peak ankle plantar flexion moment (Nm/kg)	Average	1.13 (0.14)	0.86 (0.14)	<0.001	0.596	0.251
	Dominant	1.07 (0.17)	0.89 (0.15)			
	Non-dominant	1.19 (0.17)	0.85 (0.19)			
	Random	1.10 (0.15)	0.83 (0.26)			

Table 4.1.7: Group mean joint power values during gait in a younger and older group. The group mean (SD) for each of the four limb matching methods is provided for the spatiotemporal variables. The *p*-values are calculated from the linear mixed model where group and limb matching methods were fixed variables, and participants were set as random variables. Where applicable, a significant limb matching method effect was adjusted with a *Bonferroni* correction to identify where the significant difference occurred (†). Interaction effect is the Group*Limb Matching Method effect.

Variable	Limb Matching Methods	Group: Young Mean (SD)	Group: Old Mean (SD)	Group Effect (*p* – value)	Limb Matching Method Effect (*p* – value)	Interaction Effect (*p* – value)
Support moment during early stance (Nm/kg)	Average	0.64 (0.38)	0.94 (0.42)	<0.001	0.939	0.816
	Dominant	0.63 (0.43)	1.01 (0.47)			
	Non-dominant	0.66 (0.39)	0.88 (0.42)			
	Random	0.57 (0.42)	0.94 (0.46)			
Support moment during late stance (Nm/kg)	Average	0.50 (0.18)	0.49 (0.21)	0.677	0.990	0.641
	Dominant	0.47 (0.22)	0.53 (0.22)			
	Non-dominant	0.52 (0.20)	0.46 (0.24)			
	Random	0.45 (0.22)	0.52 (0.23)			
Peak hip extension moment (Nm/kg)	Average	0.74 (0.21)	0.70 (0.20)	0.123	0.918	0.920
	Dominant	0.75 (0.24)	0.73 (0.26)			
	Non-dominant	0.74 (0.23)	0.66 (0.22)			
	Random	0.75 (0.23)	0.68 (0.24)			
Peak hip flexion moment (Nm/kg)	Average	1.10 (0.19)	1.07 (0.12)	0.347	0.979	0.851
	Dominant	1.13 (0.22)	1.06 (0.19)			
	Non-dominant	1.07 (0.22)	1.09 (0.19)			
	Random	1.12 (0.25)	1.08 (0.22)			
Peak knee extension moment (Nm/kg)	Average	0.70 (0.30)	0.55 (0.25)	0.004	0.995	0.664
	Dominant	0.65 (0.34)	0.58 (0.28)			
	Non-dominant	0.74 (0.31)	0.52 (0.28)			
	Random	0.67 (0.35)	0.57 (0.24)			
Peak knee extension moment (Nm/kg)	Average	0.15 (0.16)	0.50 (0.17)	0.090	0.789	1.000
	Dominant	0.13 (0.20)	0.19 (0.21)			
	Non-dominant	0.18 (0.16)	0.22 (0.23)			
	Random	0.14 (0.19)	0.19 (0.22)			
Peak ankle dorsiflexion moment (Nm/kg)	Average	0.21 (0.07)	0.15 (0.07)	<0.001	0.025	0.999
	Dominant†	0.18 (0.10)	0.12 (0.08)			
	Non-dominant†	0.24 (0.08)	0.17 (0.09)			
	Random	0.21 (0.10)	0.15 (0.09)			
Peak ankle plantar flexion moment (Nm/kg)	Average	1.13 (0.14)	0.86 (0.14)	<0.001	0.596	0.251
	Dominant	1.07 (0.17)	0.89 (0.15)			
	Non-dominant	1.19 (0.17)	0.85 (0.19)			
	Random	1.10 (0.15)	0.83 (0.26)			

4.2 Knee biomechanics before and after a total knee arthroplasty with either a medial ball-and-socket or posterior stabilized implant

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Contents of this chapter prepared as an article for submission in *Gait & Posture*

October 2022

Kowalski, E., Catelli, D. S., Dervin, G., & Lamontagne, M. Knee biomechanics before and after a total knee arthroplasty with either a medial ball-and-socket or posterior stabilized implant. 2022. Submitted.

4.2.1 Abstract

Background: Gait biomechanics comparing various knee implant designs have reported mixed findings. However, many studies did not include a pre-operative assessment and rely on discrete measures. Gait is a continuous process, so statistical analyses which evaluate the entire movement cycle may be more appropriate for detecting biomechanical changes after a total knee arthroplasty (TKA).

Research question: Can the implant design of a TKA affect gait parameters, knee biomechanics, and patient reported outcome measures (PROMs), compared to a gait pattern of similar age and body mass index (BMI) healthy adults?

Methods: Twenty-eight patients randomly received either a medial ball-and-socket (MBS, n=14, age=62.7(5.8) years, BMI=27.0(3.8)kg/m^2, females=6) or posterior stabilized (PS, n=14, age=64.5(8.1) years, BMI=29.8(3.4)kg/m^2, females=6) implant and completed a gait analysis before and 12 months following TKA and were compared to 14 healthy controls(age=64.4(5.6) years, BMI=24.9(2.1) kg/m^2, females=7). Temporospatial, knee biomechanics, and PROMs were measured during five gait trials. Knee biomechanical measures were evaluated across the entire gait cycle using statistical parametric mapping.

Results: Preoperatively, the MBS and PS groups were similar, with no differences in age, BMI, PROMs, or knee biomechanics ($P>.05$). Pre-operatively, compared to healthy controls, both MBS and PS had different movement patterns, primarily in stance phase knee flexion angles ($P<.05$). No postoperative differences in knee joint angles or moments existed between the MBS and control groups ($P>.05$), whereas the PS group walked with less knee flexion angle and less

knee extension moment than the control group (P<.05). Neither surgical group achieved the same level of knee power absorption as the CTRL group (P>.05).

Significance: The MBS group achieved a gait pattern closer to the control group with no differences in knee joint angles or moments. However, both TKA groups had less knee power absorption prior to toe-off, which can be an indicative of a stiff knee gait.

4.2.2 Introduction

Total knee arthroplasty (TKA) is a cost-effective operation (Healy & Iorio, 2007) that reduces pain and improves quality of life in individuals with severe knee osteoarthritis (Roos & Toksvig-Larsen, 2003). Despite the overall success of TKA, approximately 20% of patients are not satisfied after surgery (Gunaratne et al., 2017; Noble, Conditt, Cook, & Mathis, 2006). Many knee implant designs exist (Dall'Oca et al., 2017), however, the TKA literature is inconclusive for which implant provides best outcomes with fewest complications (Kahlenberg, Lyman, Joseph, Chiu, & Padgett, 2019; Wright & Chitnavis, 2011).

Within fixed-bearing implants, three primary designs exist, including posterior stabilized (PS), cruciate-retaining (CR), and medial ball-and-socket (MBS) (Sabatini et al., 2018). Both CR and PS implants cause paradoxical tibiofemoral motion which is an irregular anterior sliding of the femoral component on the tibial plateau (Sabatini et al., 2018). The MBS implant was designed to reduce paradoxical motion by limiting anterior and posterior motion to the medial compartment (Atzori, Salama, Sabatini, Mousa, & Khalefa, 2016). Early research has shown good clinical results at one year follow-up with MBS implants (Sabatini et al., 2018), however, much of the growing literature comparing implants is based on patient reported outcome measures (PROMs) (Lan, Bell, Samuel, & Kamath, 2020).

A review of over 4000 patients comparing different implant designs found no clinically significant differences in the Knee and Osteoarthritis Outcome Scores (KOOS) or satisfaction rates two years after surgery (Kahlenberg et al., 2019). One of the limitations of PROMs is that only captures a subjective assessment of pain and functional deficits that persist following TKA (Stevens-Lapsley, Schenkman, & Dayton, 2011). To accurately evaluate changes in knee functions following TKA, biomechanical analysis remains an objective tool to quantify joint motion and moments during activities of daily living (Lamontagne, Beaulieu, Varin, & Beaule, 2009).

Gait biomechanics studies comparing various implant designs reported mixed findings (Esposito, Freddolini, Marcucci, Latella, & Corvi, 2020; Gray et al., 2020; Kulshrestha et al., 2020) such as: PS implants having greater gain in knee flexion (Kulshrestha et al., 2020), or MBS implants having more similar kinematic profiles to healthy controls than PS implants (Gray et al., 2020). Not all studies included a pre- and post-operative assessment, and without a pre-operative assessment, it is impossible to ascertain if reported differences were due to the TKA or were present prior to surgery. Many studies relied on discrete measures, such as peak knee flexion during gait. Gait is a continuous process, therefore, methods which examine the entire movement cycle rather than discrete parameters are more appropriate for detecting biomechanical differences.

Statistical methods such as statistical parametric mapping (SPM) allow for a continuous analysis through the entire gait cycle (Pataky, Robinson, & Vanrenterghem, 2016). This type of analysis compares the entire movement cycle, rather than just a single point in time. It also provides temporal information on where in the gait cycle differences occur. The location within the gait cycle of these differences may be of clinical significance. To our knowledge, no studies have utilized SPM to compare gait in patients before and after TKA.

The primary purpose of this prospective case-control study was to compare knee biomechanics and gait parameters during gait in patients who underwent TKA with either an MBS or PS implant designs, compared to healthy controls (CTRL) using SPM. The secondary aim was to compare KOOS between TKA with either an MBS and PS implants and compared to CTRLs.

4.2.3 Methods

Inclusion criteria and control group

This study initially screened 86 individuals with end-stage knee OA (Kellgren and Lawrence, grade 4 (Kellgren & Lawrence, 1957)) who were scheduled to undergo TKA by a single senior orthopedic surgeon (GD). Eligible participants were between the ages of 45 and 75 at the time of enrollment and needed to be willing complete the required study visits. Participants were excluded if they had a body mass index (BMI) and waist circumference measurement > 35 kg/m2 and 102 cm respectively for men, and > 35 kg/m2 and 88 cm respectively for women; any past or present condition, which in the opinion of the investigators may impact gait; or previous joint replacement of the enrolled knee or other lower limb joint replacement. TKA participants were excluded if they had degenerative condition (other than osteoarthritis in the enrolled knee) impacting joints of the lower extremities. TKA participants were recruited from The Ottawa Hospital. The study protocol was approved by the university and hospital ethics committees. All participants included in this study provided written consent. It was conducted in accordance with the principles of good clinical practice and the declaration of Helsinki. The study is referenced in the clinical trials website: NCT02589197.

Despite the pandemic COVID-19, thirty-two participants were eligible and underwent randomization to receive an MBS (MicroPort EVOLUTION® Medial Pivot System with Cruciate Sacrificing (CS) tibial inserts) or PS (Zimmer Biomet® NexGen® PS TKA system with PS inserts) TKA. Twenty-eight patients completed both pre- and 12-month post-operative gait lab visits and were included in the final analysis (Fig 4.2.1). Fourteen similarly aged healthy participants were recruited from the community to form the CTRL group. The CTRL group had the same inclusion criteria and were excluded if they had any degenerative condition impacting joints of the lower extremities. They completed a single gait lab visit.

Surgery

All surgical procedures were performed at a tertiary hospital by a senior arthroplasty surgeon (GD). A midline incision and subvastus approach was performed for all patients (Hofmann, Plaster, & Murdock, 1991). Manual instruments were used with the goal of mechanical neutral alignment with the femur first technique and the goal of the tibial component at coronal neutral angle. The protocol required for resurfacing of the patella and the posterior cruciate ligament was released in all patients. All components were cemented, and tourniquet use was restricted only to the time of cementation and then deflated prior to closure. No patients required additional soft tissue release, and there were no complications or revisions with the surgical cohort.

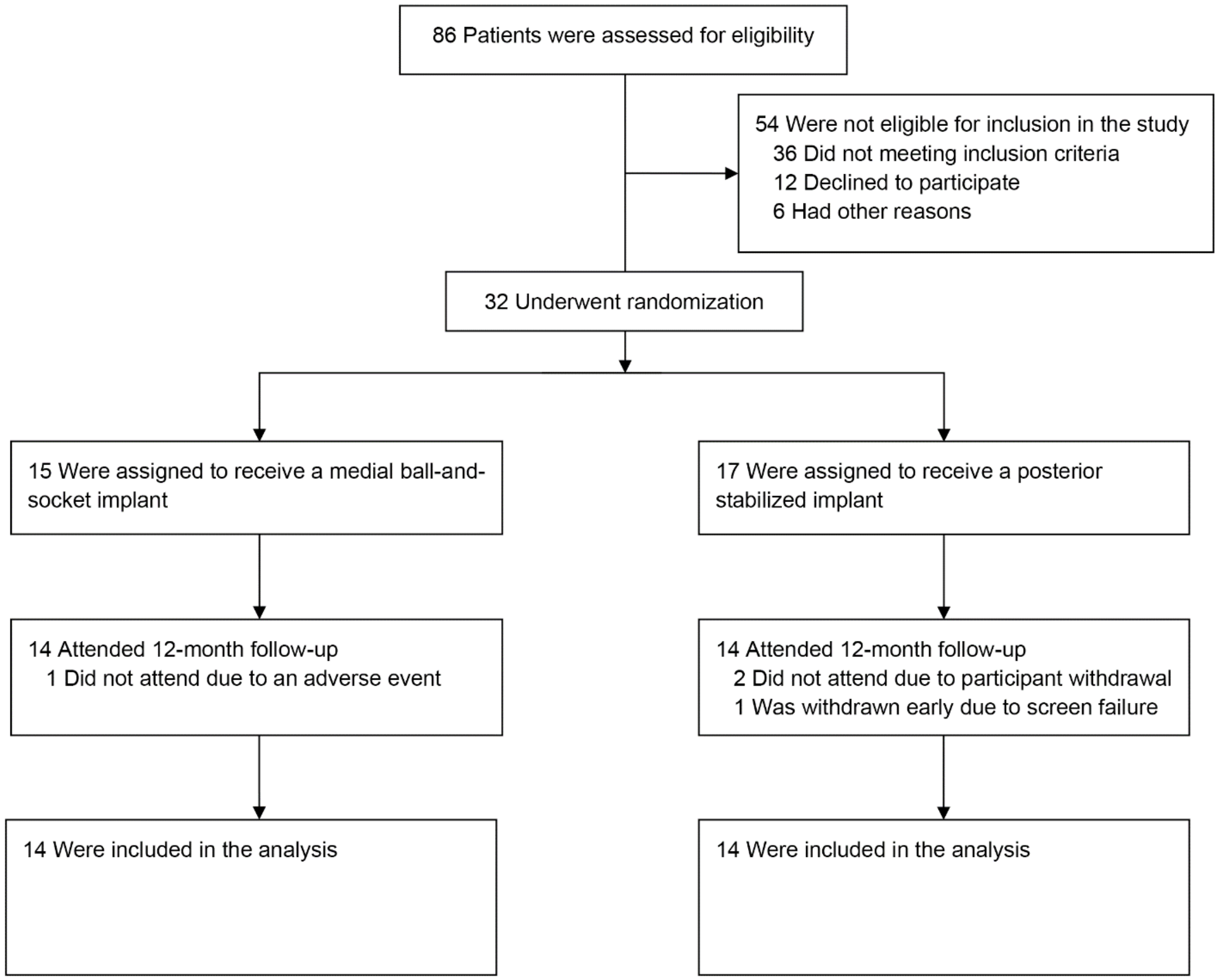

Figure 4.2.1: Consolidated Standards of Reporting Trials (CONSORT) flow diagram for enrolled patients.

Motion Capture

Level walking was captured using a 10-camera Vicon System (Oxford Metrics, Oxford, UK) sampled at 200 Hz. A 10m-long walkway had four force plates (two model 9268BA, Kistler, Winterhur, Switzerland and two model FP4060, Bertec Corporation, Columbus, OH) embedded in the middle of the walkway sampling at 1000 Hz. Participants were outfitted with 45 passive-reflective markers (Mantovani & Lamontagne, 2017). Participants completed five walking trials at their preferred walking speed. The TKA participants completed their first visit within one month of surgery and 12 months (+/- one month) after surgery, the CTRL group completed one visit. At each visit, participants completed a KOOS questionnaire.

Motion capture data were filtered using a Woltring filter with a mean standard error of 15 mm and force platform data were filtered using a 4th order (zero lag) Butterworth filter with a cut-off frequency of 10 Hz. Gait trials were modeled (Mantovani & Lamontagne, 2017), and relevant data were extracted using a custom-written Matlab script (2019b, MathWorks, Natick, USA).

Stature of participants can affect gait parameters, i.e., smaller participants tending to walk with smaller steps, so several temporospatial variables were normalized to leg length and included: walking speed, stride length, step length, and step width (Hof, 1996). Knee variables of interest included sagittal and frontal angles and moments, as well as joint power. Angles were normalized to 100% gait cycle, whereas moments and powers were normalized from heel strike to foot-off (0-62% stance phase) (Winter, 1984).

Statistical Analysis

Statistical analyses for demographic, temporospatial, and KOOS variables were done using SPSS v.27 software (IBM Corporation, Armonk, USA). Pre- vs post-operative visits were completed with paired t-test ($p<.05$). Between-group comparisons were done using a One-Way Analysis of Variance with a Bonferroni post hoc correction.

Post-operative improvement was evaluated for PROMs, temporospatial and sagittal knee range of motion (ROM) using percentage change (Eq 4.2.1).

$$Percentage\ Change = \frac{\left(M_{post} - M_{pre}\right)}{M_{pre}} \times 100\% \tag{4.2.1}$$

Knee joint angles, moments and powers were compared at each point of the gait cycle using statistical parametric mapping (SPM) (Pataky, Robinson, & Vanrenterghem, 2013). Pre- and post-operative visits were done using a paired SPM (p<.05), whereas between-group comparisons were completed using analysis of variance SPM (p <.05).

Walking speed could influence gait biomechanics (Fukuchi, Fukuchi, & Duarte, 2019), therefore if significant differences existed, it would be used as covariate.

4.2.4 Results

MBS and PS groups were similar pre-operatively, with no differences in age, BMI or KOOS scores (Table 4.2.1). Pre-operatively, the PS group walked slower than the MBS group (p=.027) and CTRL group (p=.003) (Fig 4.2.2), therefore, an ANCOVA with walking speed as a covariate was used to compare their group differences (Fukuchi et al., 2019). No other temporospatial differences existed between the MBS and PS groups pre-operatively (Fig 4.2.2). No pre-operative differences in knee biomechanics existed between the MBS and PS groups (Fig 4.2.3).

Mean KOOS for all groups are reported in Table 4.2.1. Post-operative scores were significantly higher amongst all KOOS five subscales for the MBS and PS groups compared with their pre-operative scores. Symptoms, sports and recreation, and quality of life subscales remained significantly lower post-operatively compared to the CTRL group for both MBS and PS groups. No differences in the change from pre-op to post-op (Δ Post-Pre) existed between the MBS and PS groups.

Step width was significantly greater in the PS group post-operatively compared to the MBS (p=.012) and CTRL (p<.001) groups. No other temporospatial variables were statistically significant.

Table 4.2.1: Group mean (SD) demographic and Knee Injury and Osteoarthritis Outcome Score (KOOS) values.

	MBS		**PS**		**CTRL**
	Pre	Post	Pre	Post	
Number participants (n)	14	14	14	14	14
Sex (female/male)	6/8	6/8	6/8	6/8	7/7
Age (years)	62.7 (5.8)	63.7 (5.7)	64.5 (8.1)	65.6 (8.1)	64.4 (5.6)
Height (m)	1.72 (.09)	1.72 (.09)	1.68 (.11)	1.68 (.11)	1.67 (.08)
Weight (kg)	82.7 (15.6)	81.7 (14.5)	84.6 (13.8)	85.8 (13.5)	71.5 (13.2)
Leg Length (m)	.91 (.05)	.91 (.06)	.88 (.08)	.86 (.08)	.87 (.08)
Body Mass Index (kg/m^2)	27.9 (3.8)	27.4 (3.5)	29.8 (3.4)§	30.3 (3.9)§	24.9 (2.1)
KOOS					
Symptoms	45.2 (15.4)*§	75.8 (20.4) *§	39.8 (14.9)*§	74.2 (21.0)*§	98.7 (2.6)
Pain	54.0 (12.6)*§	86.1 (12.3)*	45.0 (17.6)*§	85.7 (10.7)*	98.6 (3.2)
Function in daily living	63.9 (20.6) *§	91.3 (10.2)*	54.1 (17.3)*§	93.4 (7.1)*	100.0 (0.0)
Function in sport and recreation	23.6 (12.0)*§	69.6 (24.0)*§	27.5 (26.4)*§	60.4 (19.8)§	100.0 (0.0)
Quality of Life	26.8 (16.1)*§	67.4 (20.1)*§	17.4 (12.8)*§	71.0 (17.8)*§	100.0 (0.0)

* represents significant (p < .05) within-group pre- vs post-op difference;
† represents significant (p < .05) difference between MBS Pre-Op and PS Pre-Op visits;
‡ represents significant (p < .05) difference between MBS Post-Op and PS Post-Op visits;
§ represents significant (p < .05) difference from CTRL.

Post-operative improvements measured as percentage change for PROMs, temporospatial, and knee ROM are reported in Table 4.2.2. No differences in improvement existed between the MBS and PS groups.

Pre- versus post-operative knee joint angles, moments and powers for the MBS and PS groups are displayed in Fig 4.2.3. The MBS group had a larger post-operative knee adduction angle during swing phase (78-88% gait cycle (%$_{GC}$)) compared to their pre-operative measurements. Knee flexion moment increased post-operatively in the MBS group at initial contact (0-3%$_{GC}$), whereas knee extension moment increased post-operatively in the PS group through loading response and midstance (7-14%$_{GC}$).

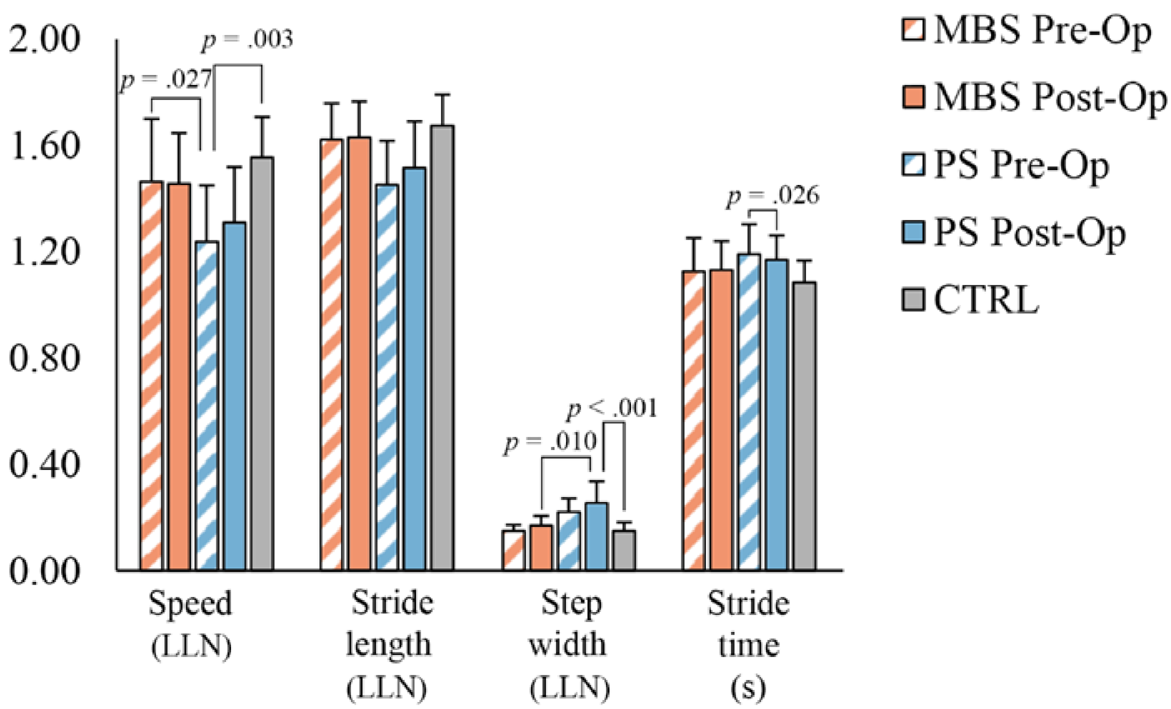

Figure 4.2.2: Group mean and standard deviations for temporospatial outcomes. Walking speed, stride length and step width were all leg length normalized (LLN). All variables were compared between the groups using walking speed as a covariate.

Post-operative differences between the MBS, PS, and CTRL groups for knee joint angles, moments and powers are displayed in Fig 4.2.4. The PS group had less knee flexion during terminal stance (29-37%$_{GC}$) and differences in knee adduction angle existed throughout swing phase (70-92%$_{GC}$) compared to the MBS group. Compared with the CTRL group, the PS group had less knee flexion angle during midstance (14-22%$_{GC}$) and less knee extension moment during pre-swing (49-57%$_{GC}$). Both the MBS (50-53%$_{GC}$) and PS (49-57%$_{GC}$) groups had less knee power absorption during pre-swing compared to the CTRL group.

Table 4.2.2: Group mean (SD) improvement for Knee Injury and Osteoarthritis Outcome Score, temporospatial and knee range of motion values represented as a percentage change from pre- to post-operative visits. Positive values indicate a post-operative increase.

	MBS (% change)	**PS** (% change)	***p*-value**
KOOS			
Symptoms	67.4 (53.8)	107.8 (78.0)	.155
Pain	60.8 (38.5)	118.8 (119.0)	.122
Function in daily living	48.0 (55.3)	78.2 (53.6)	.188
Function in sport and recreation	319.0 (479.9)	273.0 (363.2)	.793
Quality of Life	192.1 (167.9)	398.0 (364.1)	.089
Temporospatial			
Walking speed	2.0 (14.1)	7.7 (16.5)	.339
Step length	1.1 (9.6)	2.4 (15.9)	.792
Stride length	2.8 (13.7)	1.1 (20.5)	.801
Step width	11.0 (31.2)	14.4 (34.2)	.189
Stride time	-0.3 (5.9)	-4.8 (6.7)	.071
Step Time	-1.0 (8.0)	-6.4 (8.3)	.093
Knee range of motion	6.0 (5.9)	8.4 (12.2)	.512

4.2.5 Discussion

The primary aim of this study compared gait parameters and knee biomechanics in patients who underwent TKA with either an MBS or PS implant and compared them to a group of similar aged CTRLs. The secondary aim was to use PROMs to determine if the MBS or PS group improved their KOOS to the level of the control group following TKA.

Performance-based tests are necessary to fully characterize the change in physical function of patients after TKA, as they provide objective information of how the patients function that are not captured by PROMs (Mizner et al., 2011). Both the PS and MBS groups had significantly better scores for all KOOS sub-scores post-operatively when compared to their pre-operative measures (Table 4.2.1). Neither group achieved a better improvement in KOOS, temporospatial parameters, or sagittal knee ROM (Table 4.2.2). This would suggest that both implants provide similar outcomes but relying just on PROMs or discrete gait measures cannot capture how the patient is moving throughout the entire gait cycle. This can only be achieved with a gait analysis that evaluates the entire movement cycle using continuous statistical methods such as SPM.

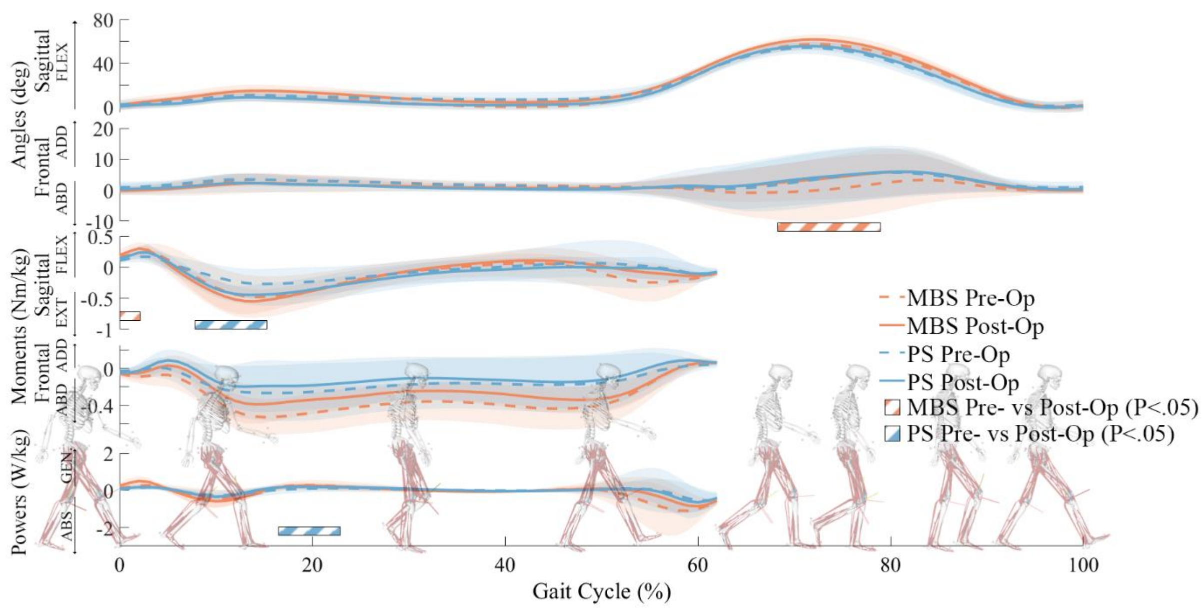

Figure 4.2.3: Group mean, standard deviations and SPM{t} results for knee angles, moments, and joint powers in MBS and PS groups as they completed a level walking task at a self-selected speed. The filled areas in the bottom bar graph below each waveform represents where in the cycle significant differences were identified using the statistical parametric mapping (SPM). The blue-white area indicates significant pre- and post-operative differences between the PS group, whereas the orange-white area indicates significant pre- and post-operative differences between the MBS group. The joint angles (degrees) were normalized to 100 % GC, whereas the joint moments (Nm/kg) and joint powers (W/kg) were normalized to 62 % stance phase. All variables were compared between the groups using walking speed as a covariate.

A limitation of previous studies was they only evaluated participants after surgery (Esposito et al., 2020; Kramers-de Quervain et al., 1997). Therefore, previously identified differences in gait biomechanics between TKA and CTRL groups may have already been present prior to surgery, and not as a direct result of the surgery or selected implant. This study addressed that limitation as it evaluated patients pre-operatively and 12 months post-operatively after undergoing TKA with either a PS or MBS implant. This allowed for the comparison of participants pre-operatively to ensure they were similar and to compare pre- and post-operative differences to evaluate how patients adapted their gait patterns following surgery.

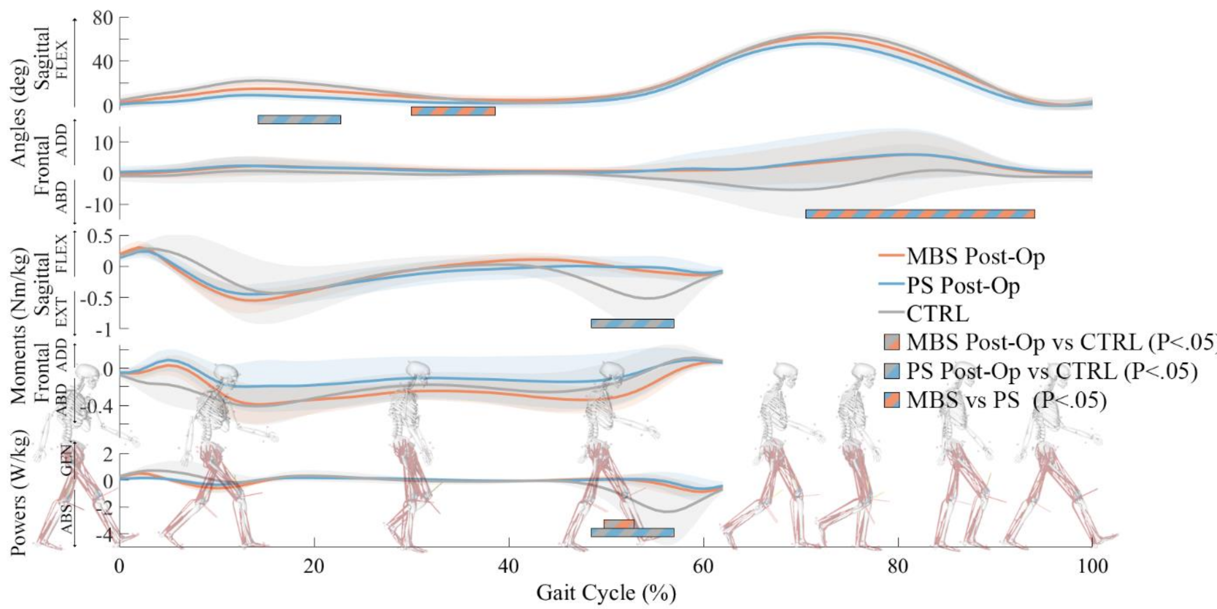

Figure 4.2.4: Group mean, standard deviations and SPM{t}results for knee angles, moments, and joint powers in post-operative MBS and PS groups, and healthy CTRLs as they completed a level walking task at a self-selected speed. The filled areas in the bottom bar graph below each waveform represents where in the cycle significant differences were identified using the statistical parametric mapping (SPM). The orange-blue area indicates significant difference between the MBS and PS groups, the orange-grey area indicates significant difference between the MBS and CTRL groups, whereas the blue-grey area indicates significant difference between the PS and CTRL groups. The joint angles (degrees) were normalized to 100 % GC, whereas the

joint moments (Nm/kg) and joint powers (W/kg) were normalized to 62 % stance phase. All variables were compared between the groups using walking speed as a covariate.

Postoperatively, PS group walked with wider steps compared to the MBS and CTRL groups (Fig 4.2.2). However, the pre-operative assessment did not show a difference, therefore, this adaptation was likely to increase the walking stability after TKA (Sample, Thorsen, Weinhandl, Strohacker, & Songning, 2020).

Walking with a wider step width can decrease knee abduction moment, which is indicative of loading the medial compartment of the knee (Sample et al., 2020). Even though the PS group walked with wider steps than the MBS and CTRL groups, it did not have a measurable effect on frontal plane knee joint moments (Fig 4.2.4). Wider step width in the PS group may have been developed to increase stability as it widened their base of support.

Surgical planning is focused on frontal plane knee joint alignment, so it is unsurprising that the patients in this study had similar front plane moments after surgery (Fig 4.2.4), as all patients had the same surgeon and surgical approach. Ro et al. (2020) identified that following surgery with a mechanical alignment, frontal knee moments were like healthy controls (Ro et al., 2020). Other studies also identified improvements in frontal plane angles and moments after TKA (Hatfield, Hubley-Kozey, Astephen Wilson, & Dunbar, 2011; Outerleys, Dunbar, Richardson, Hubley-Kozey, & Astephen Wilson, 2021).

Lower knee extension moments can be indicative of a pain avoidance strategy to minimize knee joint loading (Briem & Snyder-Mackler, 2009). Post-operatively, the PS group had lower second-peak knee extension moments compared to the CTRL group (Fig 4.2.4). This second-peak knee extension moment occurred as the body transitioned from single-limb support on the affected knee to double-limb support. This knee avoidance strategy was also evident in knee joint power, as both groups had significantly reduced power transfer through their knee joint during this period compared to the CTRL group (Fig 4.2.4). The CTRL group absorbed energy through their knee during this period, whereas this did not occur in either TKA group. Insufficient energy absorption during this period suggests that both implant groups utilized a stiff knee gait pattern to avoid loading their knee (Dorr, Ochsner, Gronley, & Perry, 1988).

The study findings disagreed with a recent study which compared gait biomechanics during level walking in patients following a TKA with either an MBS or PS implant (Esposito et al., 2020). Their MBS group had less peak knee flexion angles compared to their PS and control groups, and less knee extension moments than their controls. However, that study was limited to only a post-operative assessment and relied on a young adult population for their control group (Esposito et al., 2020). The current study found that the MBS group had similar sagittal joint angles and moments as the CTRL group, whereas the PS group could not achieve the same peak knee flexions during stance phase (Fig 4.2.4). The smaller period (MBS: Δ3%GC vs. PS: Δ8%GC) of significant differences in knee joint power supports that the MBS group provided a gait pattern closer to healthy CTRLs compared to the PS group.

PROMs do indicate that TKA is generally successful at relieving pain and showing an overall functional improvement (Gunaratne et al., 2017). The KOOS indicated that TKA improved reported pain scores for both MBS and PS groups exceeding the minimum clinical important change and substantial clinical benefit change (Lyman, Lee, McLawhorn, Islam, & MacLean, 2018). However, both groups were still significantly lower compared with the CTRL group, and neither group provided better pain relief (Table 4.2.1). Although PROMs are reliable, valid, and easy to administer, they are subjective measurements and suffer from ceiling effects which cannot detect differences between well-performing groups (Eckhard et al., 2020). Reliance on PROMs within the TKA literature continues to increase with time (Lan et al., 2020). However, it is necessary to move beyond qualitative tools to identify performance differences with different implant designs, which only pre- and post-operative biomechanical assessments can provide.

The TKA patients in this study were provided with eight publicly funded post-surgery physiotherapy sessions. However, some patients may have continued beyond the publicly funded sessions. This was not controlled in this study, so future studies could compare if additional physiotherapy sessions improve post-operative biomechanics and PROMs. Furthermore, post-surgery rehabilitation can help restore joint mobility and muscle strength, so future studies should compare muscle activity between the implant groups, especially during tasks that are more demanding than level walking. Although this study compared both females and males, the current sample size could result in the misinterpretation or misrepresentation of any potential differences identified between sexes.

Limitations

The relatively small sample size of 14 patients in each group may have been underpowered to identify differences between our groups, potentially leading to a type II error. It also compared patients at 12±1 months after surgery, so patients may still have been recovering as studies which compared patients at 24 months found fewer differences between groups (Kulshrestha et al., 2020; Stolarczyk et al., 2022). The decision was made to exclude obese (BMI>35kg/m2) individuals. While obese individuals are represented within the general patient population, excessive adiposity can affect the accuracy of gait analysis (da Silva-Hamu et al., 2013). It also excluded individuals with a previous lower limb joint replacement, so findings cannot be generalized to those cohorts. Many different types of TKA implants exist, and this study only evaluated a single type of MBS and PS implant, so the findings cannot be generalized to all implants. Additional group differences may be uncovered during more demanding activities of daily living, such as stair or ramp climbing, so future studies should incorporate additional tasks. This study did not control for the post-operative rehabilitation of the TKA patients. All patients were provided with eight publicly funded physiotherapy sessions. Some patients may have continued beyond these sessions and that information was not recorded. Finally, clinical and radiographic parameters were not evaluated as part of this study.

4.2.6 Conclusion

Pre-operative comparisons are necessary to determine if post-operative observations are due to the surgical procedure or are persistent adaptations acquired while awaiting TKA. The MBS group achieved a gait pattern that resembled the CTRLs, as there were no significant differences in knee joint angles or moments, whereas the PS group could not achieve the same amount of knee flexion angle and knee extension moment. However, both TKA groups could not reach the same level of knee energy absorption as the control group, which may represent a stiff knee gait to avoid loading their operated knee. Both groups improved post-operatively on all metrics measured by the KOOS, but the KOOS was not able to identify differences between the MBS and PS groups.

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4.3 A waveform test for variance inequality, with a comparison of ground reaction force variance during walking in younger vs. older adults

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Contents of this chapter published as an article in *Journal of Biomechanics*

October 2021

Kowalski, E., Catelli, D. S., & Lamontagne, M. (2021). A waveform test for variance inequality, with a comparison of ground reaction force during walking in younger vs. older adults. In Journal of Biomechanics (Vol. 127, p. 110657). Elsevier BV. https://doi.org/10.1016/j.jbiomech.2021.110657

4.3.1 Abstract

Various methods have been suggested for estimating the variability in biomechanical variables during gait. However, all current measures of variability are performed on discrete measurements extracted from the kinematic or kinetic waveforms, which provide no temporal information on where differences in variability occur. This study used a variance equality test to compare temporal differences in group variance along the entire ground reaction force waveform. The variance equality test used an F-statistic whose critical value was determined using the random field theory function within the one-dimensional statistical parametric mapping package. Twenty healthy younger and twenty older adults were included in the study and completed gait analysis as they walked along a level walkway at a self-selected pace. Variance for each group was calculated and compared at each interval along the waveform to produce the F-value. The F-value was compared against a calculated F-critical value to determine where in the waveform significant differences in ground reaction force variance occurred. Results suggest that younger individuals may exhibit greater ground reaction force variance during heel contact in the vertical and posterior directions, and that older individuals may exhibit greater variability in the mediolateral direction at toe-off. This study was able to identify differences in ground reaction force variance within the gait cycle between younger and older adults. The findings of this study warrant the use of the function as a suitable method to compare variance along the entire waveform between two groups.

4.3.2 Introduction

Gait variability is the fluctuation in the value of kinematic (e.g., joint angles), kinetics (e.g., joint moments), temporospatial (e.g., stride time) or electromyography measurements. This fluctuation may be observed in repeated measurements over time, across or within participants, or between different measurement, intervention or health conditions (Chau, Young, & Redekop, 2005).

Various methods have been suggested for estimating the amount of gait variability in biomechanical variables. Examples include standard deviation (Owings & Grabiner, 2004), coefficient of variation (Winter, 1984), and coefficient of multiple correlation (Growney, Meglan, Johnson, Cahalan, & An, 1997). The most prevalent measure of gait variability among clinical studies was stride time variability, expressed by the coefficient of variation (König, Singh, Von Beckerath, Janke, & Taylor, 2014). The coefficient of variation is applied on kinematic and kinetic parameters (Smith, 1993; Winter, 1987), giving an overall measure of the variation across the waveform. Although you can compare the coefficient of variation between groups, this discrete measurement does not provide temporal information on where the variability differences occurred. Current computing power allows researchers to move away from discrete measurements and make comparisons across entire waveforms.

Gait is an inherently variable task, and variability increases with normal aging (Oberg, Karsznia, & Oberg, 1993, 1994; Owings & Grabiner, 2004). Although both internal (e.g. natural variation, aging effects, pathological mechanisms) and external (e.g. methodological, environment) sources of variability exist (Chau et al., 2005), the internal sources are most pertinent. An increase in gait variability can be a strong predictor of falling in the elderly population (Gabell & Nayak, 1984; Maki, 1997). Thus gait variability is the result of impaired motor control, which may reflect the lack of ability to control the destabilizing motion of the centre of mass of the body ultimately leading to an increased likelihood of trips or missteps (Maki, 1997).

The primary aim of this study is to propose a waveform-level variance equality test that has not previously been used in the literature, and to demonstrate the utility of the proposed test by comparing ground reaction force (GRF) variability in younger vs. older adults. The secondary aim of this study is to compare the results of the variance equality test for continuous data to the results of a simple scalar variance comparison (i.e., the coefficient of variation). It is hypothesized that older adults have larger GRF variability measured by both the variance equality test and the scalar variance comparison.

4.3.3 Methods

Participant Selection

This retrospective study approved by the Research Ethics Board of the University included a subset of 107 healthy adult participants between 2009 to 2020. These volunteers provided written informed consent prior to participation.

To be eligible for inclusion, participants were free from lower-limb injury, had no musculoskeletal or neurological disorder, had bi-lateral gait analysis, included a minimum of five recorded trials, and were between the ages of 19-44 for the younger, or 55-79 for the older groups. Thirty younger and twenty older adults were eligible for this study. The eligible participants were randomly selected to include 10 females and 10 males in each group (Table 4.3.1). The comparisons between the younger and older groups were done only on the dominant limb (Kowalski, Catelli, & Lamontagne, 2019), which was defined by the preferred limb used to kick a ball (Chapman, Chapman, & Allen, 1987).

Data Collection

Gait trials were collected as participants walked over four force platforms (two FP4060, Bertec Corporation, USA, and two 9268BA, Kistler, Switzerland) sampled at 1000 Hz. Participants completed five walking trials at their preferred walking speed along a 10m level pathway which had the four embedded force platforms. Force platform data were filtered using a 10Hz low-pass 4th order (zero-lag) Butterworth filter. Gait event identification was assisted using the ground reaction forces (GRF). GRF in all three planes were extracted with a custom-written script with Matlab 2019b (MathWorks, USA) and were normalized to the stance phase (Winter, 1984).

Variance Equality Test

The comparison of variance between two groups along an entire waveform was obtained by a variance equality test, while using a new Matlab function called 'gwv1d'. This function evaluates one-dimensional group waveform variance (GWV). The 'gwv1d' function uses the validated, free and open-sourced 'SPM1D' software package which is available for use with both Matlab and Python (Python Software Foundation) (Friston et al., 1994; Pataky, 2012, 2016; Pataky, Robinson, & Vanrenterghem, 2013). The 'gwv1d' function enables the use of an F-test to compare group variances across an entire waveform.

F-statistic is calculated with the typical method, by dividing the group with the larger variance by the group with the smaller variance to force the F-test into a right-tailed test (Eq. 4.3.1). This is repeated at each time-point interval along the waveform.

$$Fstatistic = \frac{Larger\ Sample\ Variance}{Smaller\ Sample\ Variance} = \frac{{\sigma_1}^2}{{\sigma_2}^2} \quad (4.3.1)$$

Common calculations of the critical F value ("F-critical") depend on just the degrees of freedom and alpha, but this simple approach cannot be utilized for waveform data because multiple F values are calculated across time. To calculate F-critical, this function used the one-dimensional random field theory (Pataky, 2016) available within the SPM1D software package (Pataky, 2010). The F-critical is the inverse survival function of the F distribution. This function required several inputs, including alpha (α= .05), degrees of freedom, the number of discrete field nodes and the field smoothness. Field smoothness was estimated using the full-width-half-maximum function within SPM.

Coefficient of variance (CV) analysis for simple scalars

Scalar variance comparison was completed using the coefficient of variation (CV), which was calculated for the GRF parameters to compare to the GWV results (Eq. 4.3.2) (Winter, 1987).

$$CV = \frac{\sqrt{\frac{1}{N}\sum_{i=1}^{N}\sigma_i^2}}{\frac{1}{N}\sum_{i=1}^{N}|X|_i} \times 100 \quad (4.3.2)$$

N is the number of intervals

X is the amplitude of the variable of interest at the *i*th interval

σ_i^2 is the standard deviation of *X* ath the *i*th interval

Statistical analyses for demographics, walking speed, and CV variables were done using SPSS v.24 software (IBM Corporation, USA) to perform a nonparametric Mann Whitney U-test between the younger and older groups.

4.3.4 Results

No significant difference in body mass index existed between the groups (p = .150). There was no significant difference in walking speeds between the groups (p = .626) (Table 4.3.1).

Table 4.3.1: Group mean demographic and walking speed values. The group mean (SD) for each age group is provided for all of the recorded demographic variables.

	Younger	**Older**	***p*-value**
Number of participants (n)	20	20	
Sex (female/male)	10/10	10/10	
Range (years)	19-44	55-79	
Age (years)	29.9 (7.0)	63.6 (5.5)	**<0.001**
Body Mass Index (kg/m^2)	24.6 (3.2)	25.9 (2.7)	0.150
Dominant Limb (right/left)	20/0	20/0	
Walking Speed (Normalized to leg length)	1.62 (0.12)	1.60 (0.11)	0.626

Group waveform variance of the GRFs was reported in Figure 4.3.1. The older group had significantly greater ground reaction force variability in the mediolateral (5% and 60-62%) direction. Anteroposterior (1-4%) and vertical (0-3%) ground reaction forces were significantly greater in the younger group at initial contact.

Table 4.3.2: Group mean coefficient of variation measures for the ground reaction force variables. The group mean (SD) for each age group is provided for all of the recorded variables.

Variable	**Younger**	**Older**	***p*-value**
GRFx (medio-lateral)	4.0 (2.5)	5.6 (4.6)	.176
GRFy (antero-posterior)	1.7 (0.9)	3.4 (1.9)	**.006**
GRFz (vertical)	0.4 (0.1)	0.4 (0.1)	.766

Coefficient of variation (CV) comparisons of the GRFs were reported in Table 2. The anteroposterior axis of the GRF was the only variable that identified a significant difference in CV between the younger and older groups, with the older group having greater variability (*p* = .006).

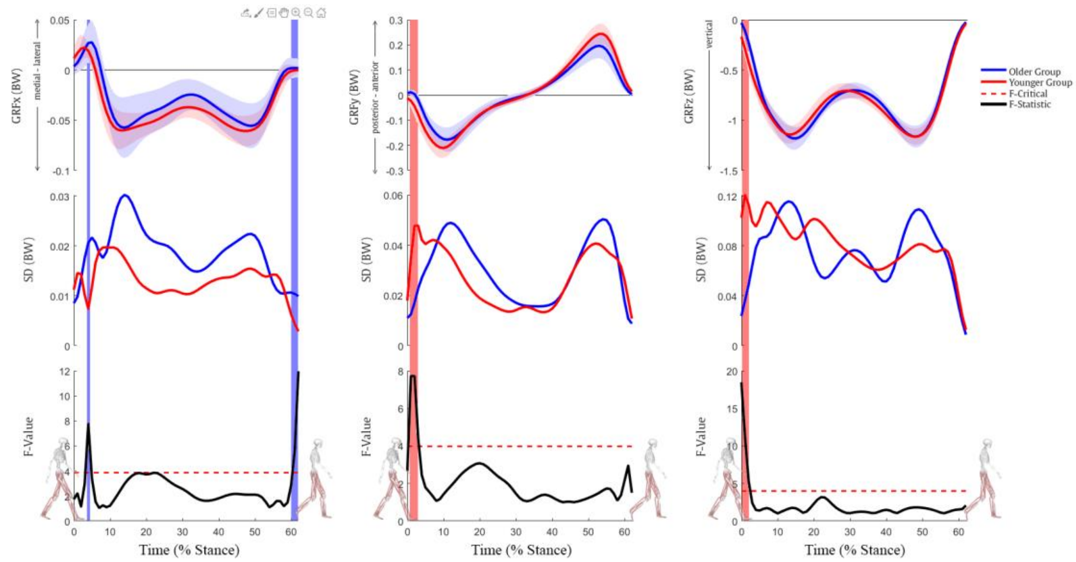

Figure 4.3.1: Top panels: group means and standard deviations (SD) for ground reaction forces (GRF) during overground, level walking. Middle panels: SD values. Bottom panels: equal variance tests; vertical shaded bars highlight temporal regions where F values reached significance. Shaded blue and red vertical bars represents the area where variance was significantly greater for the older or younger groups, respectively.

4.3.5 Discussion

The primary goal of this retrospective study was to compare variance in ground reaction force during gait across the entire waveform between the healthy younger and older adult groups using a waveform-level variance equality test. The secondary aim compared the variance equality test outcomes to a scalar variance comparison (i.e., the coefficient of variation). In general, our hypotheses were partially correct, as the older group had periods of significantly greater variability for the mediolateral GRF variable (Figure 4.3.1), whereas the CV identified the older group as having greater variability for only a single GRF variable (Table 4.3.2). The younger group exhibited larger variance for vertical and posterior GRF near initial contact, which was contrary to our hypothesized expectation. This unexpected outcome emphasizes how the proposed waveform-level variance test can be a valuable addition to gait analysis research.

Previous studies have shown that older adults typically walk slower than younger individuals (Prince, Corriveau, Hébert, & Winter, 1997; Winter, Patla, Frank, & Walt, 1990), which have also been associated with greater gait variability (Oberg et al., 1993, 1994; Owings & Grabiner, 2004; Winter, 1983). However, when they are asked to walk at speeds slower or faster than their preferred pace, older adults had increased gait variability compared with younger adults for several temporospatial variables (Kang & Dingwell, 2008). The participants in our study were asked to walk at their preferred walking speed, and there were no differences between the groups in their preferred walking speed (p=0.626). Therefore, the observed differences in the GRFs would be due to age-related differences, as opposed to differences in walking speed.

Numerous sources of variability in gait measurements have been reported, and are either internal or external to the individual (Schwartz, Trost, & Wervey, 2004). Internal variability relates to the neurological, metabolic, and musculoskeletal health of the individual. Internal sources of variability can be further divided into natural and pathological fluctuations due to aging. Walking requires the complex neurological integration of many different inputs, including visual and auditory stimuli, vestibular, proprioceptive and kinesthetic inputs (Chau et al., 2005; Hausdorff et al., 1996). Disturbances to any of these integrated inputs, either due to aging or pathology, could alter the internal variability in gait measurements (Chau et al., 2005).

External sources of variability relate to the physical environment, including the walking surface and lighting, which could alter cadence and step width (Menz, Lord, & Fitzpatrick, 2003). However, other external sources of variability can be due to the methodology, including instrumentation or experimenter error (Chau et al., 2005). Researchers employ sound methodological practices to minimize external sources of variability, therefore, observed variability would be primarily due to internal sources of variability.

The comparisons of the GWV to the CV were not in total agreement. The GWV identified that all three GRF axes had periods of significant variability differences (Figure 4.3.1), whereas the CV identified only the anteroposterior GRF were significantly different (Table 4.3.2). For the anteroposterior axis, both the GWV identified that the younger group had greater variability whereas the CV measurements identified that the older group had greater variability. This is because the GWV and CV are measuring two different phenomena. The CV is a measure of global variability (Winter, 1984), which integrates variance across the time domain to yield a single scalar value. Whereas GWV is an instantaneous measurement of variability, which considers variability as a temporally continuous variable. Therefore, small portions of the gait cycle could have significant differences identified by the GWV, but the entire waveform was not significantly different when measured using the CV, as a single scalar value clearly cannot embody temporal patterns in variance changes.

This is likely what occurred as variables such as mediolateral and vertical GRFs (Figure 4.3.1), that had significant differences in variability identified by the GWV, but the CV was not sensitive enough to detect this difference (Table 4.3.2).

Whether excessive variability is good or not depends on the type of variable (i.e., kinematic or temporospatial) (Hausdorff, 2007). Low variability in kinematic measurements is associated with the movement being achieved within specific margins of error. Kinetic measurements have greater variability due to the greater number of degrees of freedom caused by the number of muscles that cross each lower limb joint. The greater variability caused by this redundancy indicates adaptability needed to negotiate variation in environmental conditions which may occur during gait (Smith, 1993).

However, where this variability is occurring during the gait cycle may be of most importance. If excessive variability occurs during the transition from double- to single-limb support, it may lead to an increased risk of fall. For example, excessive variability during the transition from double-limb to single-limb support may lead to increased fall risk in patients with Parkinsons or a pain-avoidance adaptation in people with knee osteoarthritis.

This study demonstrated that a variance equality test is a valid way to identify temporal differences in continuous variables. Future studies should expand the use of the GWV to other biomechanical variables, populations, and tasks.

4.3.6 Conclusions

This study introduced the variance equality test under the novel 'gwv1d' function which was able to compare group variance of ground reaction forces along the entire waveform between healthy younger and older adults. This method was able to identify where in the gait cycle significant differences in variance occurred. Overall, the findings in this study warrant the use of variance equality test to compare temporal differences in variance for other biomechanical variables, populations and tasks.

4.3.7 References

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4.4 Gait variability between younger and older adults: an equality of variance analysis

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Contents of this chapter published as an article in *Gait & Posture*

June 2022

Kowalski, E., Catelli, D. S., & Lamontagne, M. (2022). Gait variability between younger and older adults: An equality of variance analysis. Gait & posture, 95, 176–182. https://doi.org/10.1016/j.gaitpost.2022.04.022

4.4.1 Abstract

Background: To estimate gait variability, several methods have been routinely used which provide a measure of global variability. A recent study introduced a group waveform variability method which provides a point-by-point measurement of data variance equality. This can identify where in the gait cycle the significant differences in variability exist.

Research question: Do waveform differences exist in equality of variance and group means in lower limb biomechanical variables between healthy younger and older adults during a gait task?

Methods: Twenty healthy younger (19-44 years old, age = 29.9 (7.0) years, body mass index = 24.6 (3.2) kg/m^2, females = 10) and 20 healthy older (55-79 years old, age = 63.6 (5.5) years, body mass index = 25.9 (2.7) kg/m^2, females = 10) adults who were free from lower limb injuries and had no musculoskeletal or neurological disorders. Temporospatial outcomes, sagittal and frontal lower limb joint angles and moments, along with joint powers were examined as participants walked at a self-selected pace. Waveform patterns were normalized to the gait cycle and compared using equality of variance and statistical parametric mapping techniques.

Results: No difference in walking speed existed between the younger or older groups ($P>.05$). The older group had greater variability ($P<.05$) in sagittal hip angles, as well as greater frontal ankle angle and moment variability. The younger group had significantly greater mean ($P<.05$) ankle power generation prior to toe-off.

Significance: This study provided a baseline of temporal differences in variance between healthy younger and older individuals. Its findings warrant the use of the equality of variance test to compare temporal differences for a variety of populations and tasks. Older adults generally had more variability than the younger adults, with many differences occurring near the transition

from double- to single-limb support. The statistical parametric mapping analysis showed that the older adults could not generate as much ankle power as the younger adults prior to toe-off.

4.4.2 Introduction

Measuring gait variability allows researchers to compare the variation in kinematic, kinetic, spatiotemporal, or electromyography measurements. These variations can be compared across or within participants, or among different measurements, interventions or health conditions (Chau, Young, & Redekop, 2005). Several methods to estimate the amount of gait variability have been routinely used and include standard deviation (Owings & Grabiner, 2004), coefficient of variation (CV) (Winter, 1984), coefficient of multiple correlation (Growney, Meglan, Johnson, Cahalan, & An, 1997), interquartile range, median absolute deviation (Chau et al., 2005), standard error of measurement, and minimum detectable difference (Weir, 2005). These methods provide an overall measure of global variability, which integrates the variability across the time domain to yield a single scalar value (Kowalski, Catelli, & Lamontagne, 2021). None of these methods can provide temporal information to identify where significant differences in variability occur in the movement cycle.

Considering gait as a continuous process, methods which examine the entire gait cycle rather than discrete parameters seem to be more appropriate for detecting biomechanical differences, because in certain situations, discrete analysis may compromise the spatiotemporal integrity of the original fields (Pataky, 2010). Many studies are now moving away from only comparing discrete parameters, such as peak knee joint flexion, and instead are applying point-by-point methods such as statistical parametric mapping (SPM) to examine the continuous dynamic information of gait data (Ismailidis et al., 2020; Ismailidis et al., 2021). SPM has been successfully applied to analyze gait data for various populations and outcome measurements including patients with knee (Ismailidis et al., 2020) and hip osteoarthritis (Ismailidis et al., 2021). The inclusion of temporal information on where in the gait cycle significant differences occur provides a comprehensive evaluation of gait parameters which can instantaneously provide temporal information on where two groups differ in their gait variability.

A recent study introduced an equality of variance which provides a point-by-point measurement of data variance equality (Kowalski et al., 2021). It considers variability as a temporally continuous variable, which can identify where in the gait cycle the significant differences in variability exist. Therefore, specific portions of the gait cycle could have significant differences identified by the equality of variance test, but the entire waveform may not be significantly different when measured by a single scalar value, such as the CV (Kowalski et al., 2021). Another recent study compared waveform variability by taking the mean of standard deviation and comparing their groups using an ANOVA SPM (Sarvestan et al., 2021).

While both methods appear similar, they are testing different hypotheses. The equality of variance test considers the original data (e.g., joint angles) to be the dependent variable, whereas the other method regards the mean standard deviation as the dependent variable. Standard deviations are constrained to be positive, which may result in the data not being normally distributed, and thus the parametric results may not be valid. The only assumption for the equality of variance test is that the populations are normally distributed, and the test results in a more direct estimate of population variance. Identifying where in the gait cycle significant differences in data variance occur can help uncover important information when comparing different groups.

The goal of this study is to compare lower limb joint kinematics and kinetics of healthy younger and older participants during level walking. The first aim was to compare the equality of variance of the spatiotemporal parameters, as well as hip, knee, and ankle joint kinematics, moments, and powers during the whole cycle analyzed with equality of variance test. The second aim was to compare the mean differences in the continuous trajectories amongst the same variables and between same groups by using SPM.

4.4.3 Methods

Participant Selection

This retrospective study included a subset of 107 healthy adult participants who visited the gait laboratory from 2009 to 2020. The study was approved by the Research Ethics Board of the University and all volunteers provided written informed consent prior to participation.

All eligible participants were lower-limb injury-free, had no musculoskeletal or neurological disorder, had bi-lateral gait analysis, included a minimum of five recorded trials, and were between the ages of 19-44 for the younger, or 55-79 for the older groups. Although changes in gait parameters appear to periodically change after the age of 50 (Verlinden et al., 2013), the reality for many studies, including those that examine orthopedic conditions such as total hip or total knee replacements, often include cohorts within the age range of this study (Beaulieu, Lamontagne, & Beaule, 2010; Benedetti et al., 2003; Ismailidis et al., 2020; Ismailidis et al., 2021). In Canada, approximately 50% of all total hip replacements and 70% of total knee replacements are performed on individuals between the ages of 55-74 (Canadian Institute for Health Information. Canadian Joint Replacement Registry: 2019–2020 Full Annual Report, 2021), therefore, including this wide of an age range is applicable and pertinent.

Twenty younger (19-44 years old) and 20 healthy older (55-79 years old) adults volunteered for this study. The eligible participants were randomly selected to include 10 females and 10 males in each group (Younger group: age = 29.9 (7.0) years, body mass index = 24.6 (3.2) kg/m^2; Older group: age = 63.6 (5.5) years, body mass index = 25.9 (2.7) kg/m^2) (Table 4.4.1). All comparisons were completed on the dominant limb (Kowalski, Catelli, & Lamontagne, 2019), which was defined by their preferred limb used to kick a ball (Chapman, Chapman, & Allen, 1987).

Table 4.4.1: Group mean demographic values. The group mean (SD) for each age group is provided for all of the recorded demographic variables.

	Younger	**Older**	**Mean Difference (95% CI)**	***p*-value**	**Effect size (d)**
Number of participants (n)	20	20			
Sex (female/male)	10/10	10/10			
Range (years)	19-44	55-79			
Age (years)	29.9 (7.0)	63.6 (5.5)	-33.8 (-37.9, -29.7)	**<0.001**	5.335
Body Mass Index (kg/m^2)	24.6 (3.2)	25.9 (2.7)	-1.4 (-3.3,0.5)	0.150	0.465
Dominant Limb (right/left)	20/0	20/0			

Motion Capture

Motion capture of gait trials was collected using a 10-camera (MX-13, Vicon, UK between 2009-2016; and 2016-onwards two Vantage V5 and eight Vero v2.2, Vicon, UK) sampled at 200 Hz, four force platforms (two FP4060, Bertec Corporation, USA, and two 9268BA, Kistler, Switzerland) sampled at 1000Hz. Participants were outfitted with 45 passive-reflective markers according to the University of Ottawa Motion Analysis Model (UOMAM) (Mantovani & Lamontagne, 2017). Participants completed five walking trials at their preferred walking speed along a 10m level pathway which had the four embedded force platforms.

Motion data were processed using Vicon Nexus 2.8 (Vicon, UK). Trajectories were filtered using a Woltring filter with a mean standard error of 15mm and force platform data were filtered using a 10Hz low-pass 4th order (zero-lag) Butterworth filter. Gait event identification was assisted using the ground reaction forces (GRF) and the walking trials were modeled with the UOMAM (Mantovani & Lamontagne, 2017). To extract the variables of interest, data were exported to Matlab 2019b (MathWorks, USA) with a custom-written script.

Coefficient of variance (CV) analysis for simple scalars

Scalar variance comparison was completed using the coefficient of variation (CV), which was calculated for all parameters to compare to the equality of variance test results(Winter, 1987).

Statistical Analyses

Statistical analyses for the temporospatial and CV variables were processed using the SPSSv.24 software (IBM Corporation, USA) to perform independent samples t-test between the younger and older groups. The comparison of the continuous trajectories of the joint biomechanics were compared between the younger and older groups using a custom-written script in Matlab 2019b (MathWorks, USA). Variance equality was compared throughout the time-normalized gait cycle ($\%_{GC}$) waveform using the equality of variance test function within the spm1d package(Kowalski et al., 2021) ($P < .05$), while mean group waveforms were compared using statistical parametric mapping (SPM) (Pataky, 2010) ($P < .05$). A sample of the output of the equality of variance test calculation is presented in Figure 4.4.1.

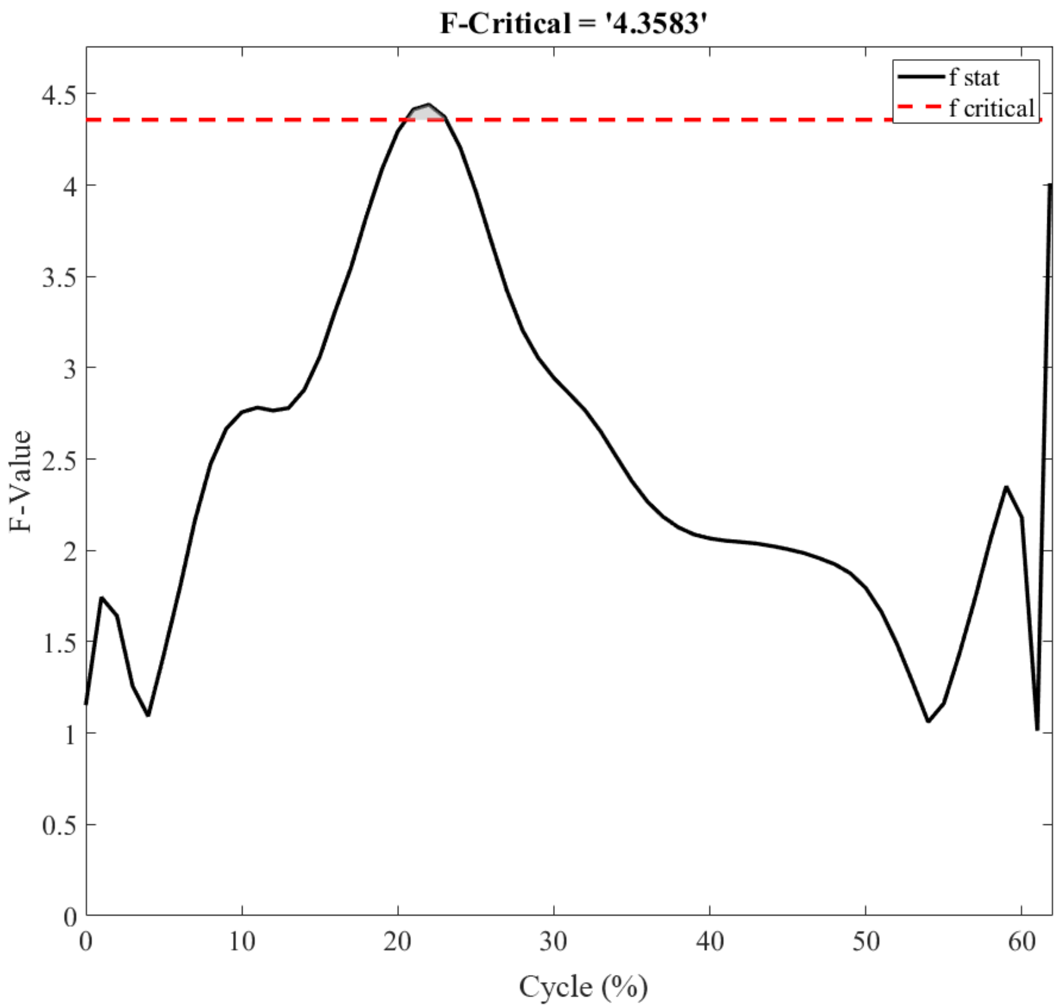

Figure 4.4.1: Equality of Variance Test Method. Calculated F-statistic for younger and older adults plotted over the F-Critical value (4.3583) for the frontal ankle moment during the gait cycle. Shaded area represents the significant areas where the F-statistic was greater than the F-critical value.

4.4.4 Results

No significant difference in BMI existed between the groups (Table 4.4.1). Walking speed was similar between the younger (1.62±0.12) and older (1.60±0.11) groups (*p*=.575). There was a statistically significant difference in step time and double support time between the younger and older groups. The younger group had a shorter step time than the older group, -.02 (95% CI, -.04 to -.002), $t(38) = -2.240$, $p = .031$. The younger group spent less time in double support phase than the older group, -.04 (95% CI, -.06 to -.02) $t(38) = -4.084$, $p < .001$ (Table 4.4.2).

Table 4.4.2: Group mean temporospatial variables. The group mean (SD) for each age group is provided for all of the recorded variables.

Variable	**Younger**	**Older**	**Mean Difference (95% CI)**	***p*-value**	**Effect size (d)**
Cadence	115.7 (5.7)	111.9 (7.1)	3.86 (-.24, 7.98)	0.065	.602
Stride Time (s)	1.04 (0.05)	1.08 (0.07)	-.04 (-.08, .002)	0.065	.602
Step Time (s)	0.51 (0.02)	0.54 (0.04)	-.02 (-.04, .002)	**0.031**	.708
Single Support Time (s)	0.41 (0.02)	0.41 (0.03)	-.002 (-.02, .01)	0.789	.085
Double Support Time (s)	0.21 (0.02)	0.25 (0.04)	-.04 (-.06, -.02)	**< 0.001**	1.292
Step Width (LLN)	0.18 (0.04)	0.18 (0.02)	.01 (-.01, .02)	0.510	.210
Step Length (LLN)	0.85 (0.04)	0.87 (0.06)	-.02 (-.05, .01)	0.209	.405
Stride Length (LLN)	1.68 (0.08)	1.71 (0.10)	-.04 (-.09, .02)	0.174	.438
Walking Speed (LLN)	1.62 (0.12)	1.60 (0.11)	0.02 (-.05, .09)	0.575	.179

LLN – leg length normalized

Differences in Group Variance

Coefficient of variation (CV) comparisons of the temporospatial, joint kinematics, moments and powers were reported in Table 4.4.3. The CV identified 6/22 variables that had significant differences between the younger and older groups. For all significantly different variables, the older group had more variability than the younger group, except for frontal ankle angle. The younger group had more frontal angle CV than the older group , 11.89 (95% CI, 2.50 to 21.29), $t(38) = -2.210$, $p = .033$. The younger group had less sagittal hip moment CV, -0.67 (95% CI, -1.23 to -0.11), $t(38) = -2.442$, $p = .019$, less frontal hip moment CV, -1.08 (95% CI, -1.73 to -0.43), $t(38) = -3.344$, $p = .002$, less hip power CV, -6.98 (95% CI, -12.94 to -1.03), $t(38)$

= -2.375, *p* = .023, and less ankle power CV, -18.26 (95% CI, -35.34 to -1.19), *t*(38) = -2.116, *p* = .037, than the older group.

Table 4.4.3: Group mean coefficient of variation measures for the temporospatial, joint kinematic and kinetic variables. The group mean (SD) for each age group is provided for all of the recorded variables.

Variable	**Younger**	**Older**	**Mean Difference (95% CI)**	***p*-value**	**Effect size (d)**
Temporospatial variables					
Cadence	2.2 (1.4)	2.0 (1.5)	0.15(-0.76,1.07)	0.735	0.108
Stride Time (s)	2.7 (1.6)	2.6 (1.8)	0.14(-0.95,1.22)	0.799	0.081
Step Time (s)	1.3 (0.5)	1.2 (0.4)	0.10(-0.20,0.40)	0.496	0.217
Step Width (LLN)	9.1 (4.7)	10.5 (6.6)	-1.39(-5.04,2.26)	0.446	0.244
Step Length (LLN)	1.8 (0.8)	1.7 (1.2)	0.03(-0.60,0.66)	0.924	0.030
Stride Length (LLN)	2.7 (1.3)	2.2 (1.3)	0.49(-0.33,1.31)	0.235	0.382
Walking Speed (LLN)	3.0 (1.7)	2.6 (2.4)	0.36(-0.96,1.68)	0.583	0.175
Joint Angles (degrees)					
Sagittal Hip	1.8 (0.5)	2.1 (0.5)	-.33(-.68,-0.004)	0.053	0.632
Frontal Hip	7.0 (3.3)	7.2 (3.0)	-.20(-2.21,1.81)	0.840	0.064
Sagittal Knee	1.9 (0.4)	2.0 (0.6)	-.14(-.47,0.17)	0.363	0.291
Frontal Knee	14.6 (6.8)	13.8 (4.9)	.77(-3.01,4.55)	0.682	0.131
Sagittal Ankle	5.4 (1.5)	6.4 (1.1)	-.92(-1.76,-0.08)	**0.033**	0.699
Frontal Ankle	21.9 (19.9)	10.0 (5.8)	11.89(2.50,21.29)	**0.015**	0.810
*Joint Moments (N*m/kg)*					
Sagittal Hip	0.6 (0.2)	1.3 (1.2)	-0.67(-1.23, -0.11)	**0.019**	0.772
Frontal Hip	1.1 (0.7)	2.1 (1.3)	-1.08(-1.73, -0.43)	**0.002**	1.057
Sagittal Knee	1.3 (0.6)	1.7 (1.2)	-0.43(-1.01, 0.16)	0.149	0.466
Frontal Knee	1.6 (0.9)	2.0 (0.7)	-0.43(-0.95, 0.09)	0.105	0.525
Sagittal Ankle	0.5 (0.1)	1.0 (1.2)	-0.49(-1.03, 0.05)	0.072	0.584
Frontal Ankle	3.4 (1.5)	3.7 (3.1)	-0.36(-1.92, 1.19)	0.640	0.149
Joint Powers (W/kg)					
Hip	27.2 (5.8)	34.2 (11.8)	-6.98(-12.94, -1.03)	**0.023**	0.751
Knee	31.6 (15.5)	50.2 (40.1)	-18.58(-38.05, 0.88)	0.061	0.611
Ankle	19.3 (5.4)	37.6 (37.3)	-18.26(-35.34, -1.19)	**0.037**	0.685

LLN – leg length normalized

Differences in sagittal joint kinematics and moments, and joint powers variability, are presented in Figure 4.4.2. The older group had greater hip flexion variability through early stance phase (9-11% GC). Knee moment variability was greater in the younger group during the initial foot strike (1-4% GC). During initial contact, the older group had greater ankle moment variability (1-3%) younger group had greater ankle power variability (0-3% GC).

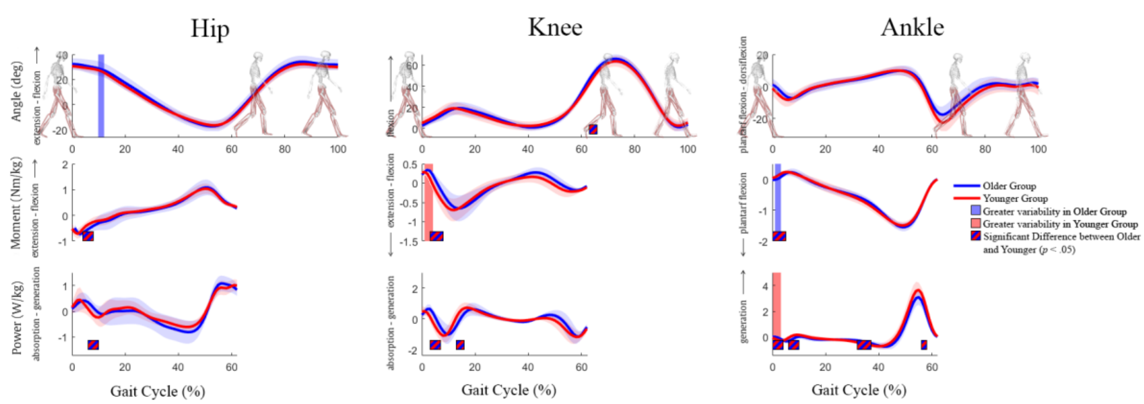

Figure 4.4.2: Group means and standard deviations (SD) for hip, knee, and ankle sagittal angles (degrees), moments (Nm/kg), and powers (W/kg). Vertical shaded bars highlight temporal regions where F values reached significance. Shaded blue and red vertical bars represents the area where variance was significantly greater for the older or younger groups, respectively. Alternating blue/red bars represent area where means were significantly different between the younger and older groups.

Differences in frontal joint kinematics and moments variability are presented in Figure 4.4.3. The older group had greater ankle moment variability during the midstance phase (21-24% GC).

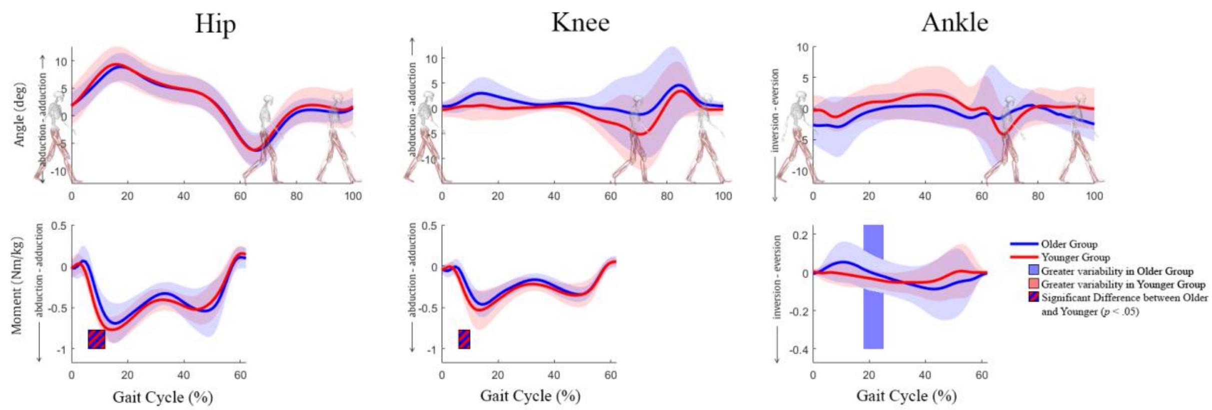

Figure 4.4.3: Group means and standard deviations (SD) for hip, knee, and ankle frontal angles (degrees) and moments (Nm/kg). Vertical shaded bars highlight temporal regions where F values reached significance. Shaded blue and red vertical bars represents the area where variance was significantly greater for the older or younger groups, respectively. Alternating blue/red bars represent area where means were significantly different between the younger and older groups.

Differences in Group Means

Differences in sagittal joint kinematics and moments, and joint powers, are presented in Figure 4.4.2. No differences in sagittal joint angles existed between the two groups. During the loading response phase (initial 10% GC), significant differences between the younger and older groups existed in hip, knee, and ankle joint moments. The younger group had significantly greater hip power absorption prior to opposite toe-off (7-10% GC) compared to the older group. Significant differences in knee and ankle powers existed between the two groups throughout the stance phase. The younger group had greater ankle power generation before toe-off (56-58%).

Differences in frontal joint kinematics and moments are presented in Figure 4.4.3. No differences in frontal joint angles existed between the two groups. The younger group had larger hip (6-12% GC) and knee (6-10% GC) abduction moments during the transition from double- to single-limb support compared to the older group.

4.4.5 Discussion

The main objective of this retrospective study compared the equality of variance in joint kinematics, moments, powers between healthy younger and older adults along the entire waveform during gait at a self-selected pace. The secondary objective was to compare the group mean differences in the joint angles, moments and powers along the entire waveform using SPM.

Differences in walking speed can account for observed differences in gait biomechanics between groups (Ko, Hausdorff, & Ferrucci, 2010; Tolea et al., 2010). Studies that compared younger and older adults have found that older adults typically walk slower (Prince, Corriveau, Hébert, & Winter, 1997; Winter, Patla, Frank, & Walt, 1990) and with greater gait variability (Oberg, Karsznia, & Oberg, 1993, 1994; Owings & Grabiner, 2004). When forced to walk at speeds other than their preferred pace, older adults had more gait variability than younger adults (Kang & Dingwell, 2008). This study had all participants walk at their preferred pace and found no significant difference in walking speed between the healthy younger and older groups (Table 4.4.2). Observed differences in this study are likely age-related differences as opposed to differences in walking speed which may have caused previously observed differences.

Differences in group variance

Previous knowledge on gait variability between older and younger adults has been primarily limited to scalar variance comparisons. Nonetheless, variables such as step width variability have been linked to increased risk of falling (Hausdorff, 2005), been able to predict falls (Hausdorff, Rios, & Edelberg, 2001), and able to differentiate fallers from non-fallers (Yang & Pai, 2014). This link between fall risk and increased stride-to-stride variability has been suggested to reflect an unsteady gait pattern which challenges the dynamic control of balance, leading to a greater predisposition to falls (Hausdorff, 2007). However, temporospatial variability cannot identify which joints may be responsible for differences in gait variability.

Kinematic or movement variability has been used to assess injury risk at various joints and discriminate between injured and uninjured populations (Baida, Gore, Franklyn-Miller, & Moran, 2018). However, contrasting results have found that reduced variability during movement may lead to repetitive loading on a structure, leading to eventual injury (Hamill, van Emmerik, Heiderscheit, & Li, 1999), or that greater variability is associated with injury, as the variations in movement represent abnormal neuromuscular control, which leads to excessive stress and injury (Schmidt, 2003). Understanding where movement cycle variability differences occur could provide valuable information in preventing falls, injuries, or aiding in rehabilitation.

When comparing the younger and older adults in this study, many of the variability differences measured with the equality of variance test occurred around the early portions of the stance phase, as the body is transitioning from double- to single-limb support (~10% GC). During this transition from double- to single-limb support, balance is disturbed along the anteroposterior and mediolateral directions because lifting the swing foot from the ground induces a gap between the center of mass and center of pressure. This gap generates body instability and a controlled fall towards the swing leg side. If these perturbations are not properly counterbalanced, they may lead to falls, particularly in the elderly or in populations with sensory or motor deficits (Tisserand, Robert, Chabaud, Bonnefoy, & Chèze, 2016). Future research on people with history of falls and non-fallers could confirm if differences between the groups occur during these transition periods within the gait cycle.

In general, this study identified that the older adults had more variability than the younger adults. Previous studies have identified that elderly adults have more gait variability in spatiotemporal variables, such as stride lengths (Guimaraes & Isaacs, 1980) and stride widths (Gabell & Nayak, 1984) compared to younger adults, when measured with a scalar variance comparison, such as the CV. However, this study did not find any differences in temporospatial variability between the younger and older adults (Table 4.4.3). When comparing the variability differences measured by the CV and the equality of variance test, they were not in total agreement.

The equality of variance test identified significantly different periods within sagittal knee moments (Figure 4.4.2) and frontal ankle moments (Figure 4.4.3), which were not identified as significantly different when measured with the CV (Table 4.4.3). The opposite was also true as CV identified that sagittal and frontal hip moments (Table 4.4.3) were significantly different between the younger and older groups, but they were not identified by the equality of variance test (Figures 4.4.2 & 4.4.3). It is not surprising as the equality of variance test and CV show differences as they are measuring two different phenomena. The CV integrates variance across the time domain to yield a single scalar value, providing a measure of global variability. Whereas the equality of variance test considers variability as a temporally continuous variable, providing a measurement of variability at a specific time-point (Kowalski et al., 2021). Therefore, with sagittal knee moments and frontal ankle moments, specific portions of the gait cycle were significantly different and identified by the equality of variance test, but the entire waveforms were not significantly different when measured using the CV. The opposite was also true with frontal hip moments as overall the entire waveform was significantly different when measured using the CV, but no section alone was significantly different when measured with the equality of variance test. The equality of variance test is recommended to identify where in the movement cycle differences in variance exist, as a single scalar value, such as the CV, cannot embody temporal patterns in variance changes (Kowalski et al., 2021).

It has been argued whether variability is a good factor or not depends on the type of variable (i.e., kinematic or temporospatial) (Hausdorff, 2007). For physiological outputs, high variability may reflect adaptability and flexibility, where more variability is better, or it can represent the inability of the physiological control system to regulate a given parameter, with more variability being worse in this situation (Hausdorff, 2007). In gait studies, large variability in temporospatial variables seems to be associated with worse outcomes (Hausdorff, 2007). However, greater variability in joint kinematics, is associated with a good outcome as it reflects adaptability, while lower variability for temporospatial variables is better as it represents automaticity and rhythmicity (Hausdorff, 2007). Kinetic measurements have greater variability due to the greater number of degrees of freedom caused by the number of muscles that cross each lower limb joint (Hausdorff, 2007). The greater variability caused by this redundancy indicates adaptability needed to negotiate variation in environmental conditions which may occur during gait (Smith, 1993).

Less or more data variability remains a topic of contention for falls (Hausdorff, 2007), sports performance (Preatoni et al., 2013), injury risk (Baida et al., 2018), and disease severity (Hall et al., 2020). However, the location of these variability differences may be of most importance. Decreased variability during the transitions from double- to single-limb support may be representative of increased fall risk in elderly populations or may indicate a pain avoidance strategy in patients with knee osteoarthritis, however, future studies would need to confirm this hypothesis. The equality of variance test method is adaptable for a variety of tasks and populations, so its future application by other researchers may provide valuable insight.

Differences in group means

Studies comparing gait of younger and older adults have consistently found that healthy, older individuals have lower ankle power generation (DeVita & Hortobagyi, 2000; Judge, Davis, & Ounpuu, 1996; Kerrigan, Todd, Della Croce, Lipsitz, & Collins, 1998; Riley, DellaCroce, & Kerrigan, 2001; Winter et al., 1990). This lower ankle plantar-flexor power is assumed to be an age-related impairment that limits the forward progression of the body and momentum of the swing leg, leading to shorter step lengths (McGibbon, 2003). This impairment is related to ankle plantarflexion strength, and the inability of older adults to generate sufficient ankle power (Thelen, Schultz, Alexander, & Ashton-Miller, 1996).

This lower ankle power generation for the older group was evident in Figure 2. The hip joint can limit ankle plantar flexion due to reduced hip extension caused by hip flexion contracture (Kerrigan et al., 1998; Riley et al., 2001). Neither hip extension nor ankle plantarflexion angle were significantly different between the younger and older groups in our study, therefore, the reduction in terminal stance ankle power generation is likely, not due to compensatory effects from the hip joint (Kerrigan et al., 1998; Riley et al., 2001).

Lower ankle kinetics may be the result of lower ankle mobility and strength with aging, as normative data has shown an approximate five degrees reduction in ankle plantarflexion range of motion in both females and males when comparing younger (20–44 years) to older (45–69 years) individuals (Soucie et al., 2011). Although neither joint strengths nor passive ranges of motion were measured in this study, age-related decreases in ankle strength (Thelen et al., 1996) and mobility (Soucie et al., 2011) could explain the differences in ankle kinematics and kinetics between the younger and older groups.

It has been suggested that aging causes a distal to proximal redistribution of joint torques and powers, which has older adults using primarily their hip extensors, while younger adults use their knee extensors and ankle plantar flexors while walking (DeVita & Hortobagyi, 2000). This is partially supported by our findings, as the older group did have greater hip extension moment, but lower ankle power generation (Figure 4.4.2). However, this shift in the neuromuscular pattern of gait with aging is not as straightforward for all individuals.

Studies have found conflicting results in the way the hip muscles compensate for distal muscle disfunction. Some studies found that older adults increase hip extension moments (DeVita & Hortobagyi, 2000), while other studies found reduced hip extension moments (Kerrigan et al., 1998), or no difference between the younger and older groups (Judge et al., 1996). These differences in hip kinetics compensatory roles exist for different reasons. If trunk stabilization is required, then hip extensors could play a primary compensatory role, but a hip flexor compensatory role would be to assist with leg advancement. Hip power generation during early stance could also assist the pelvis in pulling the contralateral leg into the swing (McGibbon, 2003).

Limitations

This study normalized the biomechanical data by converting the trajectories to a percentage of the gait cycle (Winter, 2009). Regardless of which method is used to temporally align gait data, whether converting to percentage of the gait cycle (Winter, 2009) or aligning corresponding subphases (Sadeghi et al., 2000), can impact point-by-point comparisons (such as SPM or equality of variance test). Variation at any one time point can be inflated, peaks and valleys in the averaged gait cycles will be broader and flatter than for any single trial (Helwig, Hong, Hsiao-Wecksler, & Polk, 2011; Sadeghi et al., 2000). Therefore, some differences identified in this study with either the equality of variance test or SPM may be due to temporal shift as these methods do not correct for this. Future studies could compare various temporal alignment techniques to determine if one method can provide more meaningful and accurate point-by-point comparisons for variability and group means using both the equality of variance test and SPM methods.

This study had a limited sample size, with only 20 individuals in each group. Although this study was able to identify mean and variance differences between the age groups, it would be interesting if future studies had a larger sample size where they could compare the equality of variance across the different decades. This would help determine at what age variance differences become most apparent.

Although the data analysis of this study combined both females and males, the current sample size could result in misrepresentation or misinterpretation of any potential differences identified between sexes. Future studies should be designed to determine if sex differences exist in either gait variance or gait biomechanics and to examine groups of individuals at various decades of life especially at the older ages, to determine at what age do these joint torque redistributions occur, and if that redistribution causes any risks to patients such as fall risk, development of hip or knee osteoarthritis.

4.4.6 Conclusions

The findings of this study provide a baseline of temporal differences in variance between healthy younger and older individuals and warrant the use of the equality of variance test to compare temporal differences for a variety of populations and tasks. Several differences were identified, and the older group generally had more variability than the younger group, with many differences occurring near the transition from double- to single-limb support. Group mean comparisons found that the older group could not generate as much ankle power as the younger group before toe-off. The equality of variance test identified significant differences in kinematic and kinetic variables within gait cycle that were not identified as significantly different when measured with the CV.

4.4.7 References

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4.5 Knee biomechanics variability before and after total knee arthroplasty: a prospective study comparing patients with different implant designs

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Contents of this chapter prepared as an article for submission in *The Knee*

Kowalski, E., Catelli, D. S., Dervin, G., & Lamontagne, M. Knee biomechanics variability before and after total knee arthroplasty: a prospective study comparing patients with different implant designs. 2023. Submitted.

4.5.1 Abstract

This study evaluated gait variability in patients before and after TKA with either a posterior stabilized (PS) or medial ball-and-socket (MBS) implant using the equality of variance method to determine where variability differences occur in the movement cycle. Twenty-eight patients underwent TKA with either cruciate-sacrificed MBS or PS implants. Patients underwent motion analysis which measured knee biomechanics as they walked overground at their preferred pace before and 12 months after TKA. Results were compared with 14 healthy controls of similar age. Before surgery, patients with knee OA had reduced knee moment variability throughout the stance phase compared to controls. Knee moment and power variability further decreased following TKA, with neither the MBS nor PS implant providing movement variability like the controls. A reduction in knee moment and power variability in the PS and MBS groups after TKA may represent that patients are less stable and capable of adapting to unanticipated situations while walking, which may ultimately lead to a fall. This study could not determine the reason for the reduction in variability. Still, it may be due to the cruciate sacrificing design of the MBS and PS implants, an unresolved adaptation developed while awaiting TKA, or other unknown reasons.

4.5.2 Introduction

Pain and impaired mobility are a daily reality for those living with symptomatic knee osteoarthritis (OA) (Guccione et al., 1994). As knee OA progresses, greater pain and stiffness occur at the knee joint, which ultimately requires surgical intervention with a total knee arthroplasty (TKA). Despite TKA being a successful surgery, 20% of patients remain dissatisfied after their TKA and often report loss of stability, decreased functional outcomes, reduced knee range of motion, greater difficulty performing daily activities compared to their age-matched peers (Varadarajan et al., 2015), and their gait becomes less variable (Smith, Christensen, Marcus, & LaStayo, 2014). The question remains why patients are still unsatisfied, which could be related to knee instability after TKA.

Orthopaedic surgeons evaluate knee joint instability by assessing knee joint laxity; however, self-reported instability is unrelated to knee joint laxity (Schmitt, Fitzgerald, Reisman, & Rudolph, 2008). Greater knee joint instability can cause a loss of balance and lead to an eventual fall (Nevitt et al., 2016). The measure of knee joint laxity assesses static stability but falls generally occur during movements, implying deficits in dynamic stability (Reeves, Narendra, & Cholewicki, 2007; Yakhdani et al., 2010). Researchers have assessed the dynamic stability of the knee by evaluating the variability of temporospatial and knee biomechanical measures during gait (Lewek, Scholz, Rudolph, & Snyder-Mackler, 2006; Yakhdani et al., 2010).

Gait variability is the amplitude of the fluctuations in the time series with respect to the mean of kinematic (e.g., joint angles) or kinetic (e.g., joint moments) measurements (Chau, Young, & Redekop, 2005). Several methods to estimate the amount of gait variability are routinely used and include standard deviation (Owings & Grabiner, 2004), Lyapunov exponent (Yakhdani et al., 2010), or the coefficient of variation (CV) (Ferreira, 2021; Winter, 1984). These methods provide a measure of global variability, which integrate the variability across the time domain to yield a single scalar value. A review on gait variability in patients with knee OA highlighted that variability remains lower than healthy controls before and after a TKA (Smith et al., 2014). However, this was determined using single scalar values of gait variability, so it is unknown where in the gait cycle these variability differences occur. Understanding where significant differences in variance occur in a movement cycle may identify certain motor impairments. This could help influence rehabilitation interventions in individuals with knee OA before or after undergoing a TKA.

Recent studies introduced the equality of variance test, which provides a time-point measurement of data variance inequality for kinetic and kinematic variables (Kowalski, Catelli, & Lamontagne, 2021, 2022). Using this method, it can identify where in the movement cycle variability differences occurred. This may be more sensitive and uncover additional variability differences, which typical global variability measurement may miss.

To the researchers' knowledge, no study has evaluated gait variability in patients before or after TKA using the equality of variance test or examined the effect of implant design on gait variability. Therefore, this study aimed to evaluate overground walking in knee OA patients before and after a TKA with either a PS or MBS implant and assessed the variability of knee joint angles, moments and powers using the equality of variance method. It also compared patients to a group of healthy participants of similar age. A recent study comparing knee biomechanics during gait in patients before and after a TKA with either an MBS or PS implant identified differences in knee extension moments and knee joint powers, particularly during the single-support phase. During this phase in the gait cycle, all the body's weight is supported on a single limb so that it requires the greatest amount of stability. For this reason, it is hypothesized that variability will be greater in the TKA groups during single-limb support.

4.5.3 Methods

Participants

The university and hospital review boards approved this randomized control study by the U.S. Food and Drug Administration, and it has been registered in the ClinicalTrials.gov database. To be eligible for participation, patients had to meet the inclusion and exclusion criteria fully.

For inclusion, participants were between the ages of 45 and 75 at the time of enrollment and were willing and able to complete required study visits and assessments. Exclusion criteria for all participants included having a body mass index (BMI) and waist circumference measurements > 35 kg/m2 and 102 cm respectively for men, and > 35 kg/m2 and 88 cm respectively for women; any past or present condition, which in the opinion of the investigators may impact gait; and had a previous joint replacement of the enrolled knee or other lower limb joint replacement. TKA participants could not have a degenerative condition (other than OA in the enrolled knee) impacting joints of the lower extremities. Healthy controls could not have a degenerative condition affecting the lower extremity joints.

Eighty-six symptomatic patients with severe knee OA (Kellgren and Lawrence grade 4 (Kellgren & Lawrence, 1957)) scheduled for TKA were screened. Fifty-four cases were not eligible for participation; their reasons are outlined in Figure 4.5.1. Thirty-two patients fully met the inclusion criteria and were randomly assigned to the MBS (MicroPort EVOLUTION® Medial Pivot System with Cruciate Sacrificing tibial inserts) and PS (Zimmer Biomet NexGen® PS TKA System with PS inserts) groups (Figure 1). A total of 28 patients completed the 12-month follow-up and were included in the final analysis: MBS (14 patients) and PS (14 patients). Fourteen healthy, similarly aged controls were recruited from the community and included in the study (Table 1). TKA and CTRL groups completed the Knee Injury and Osteoarthritis Outcome Score (KOOS) questionnaire at each visit (Roos & Lohmander, 2003).

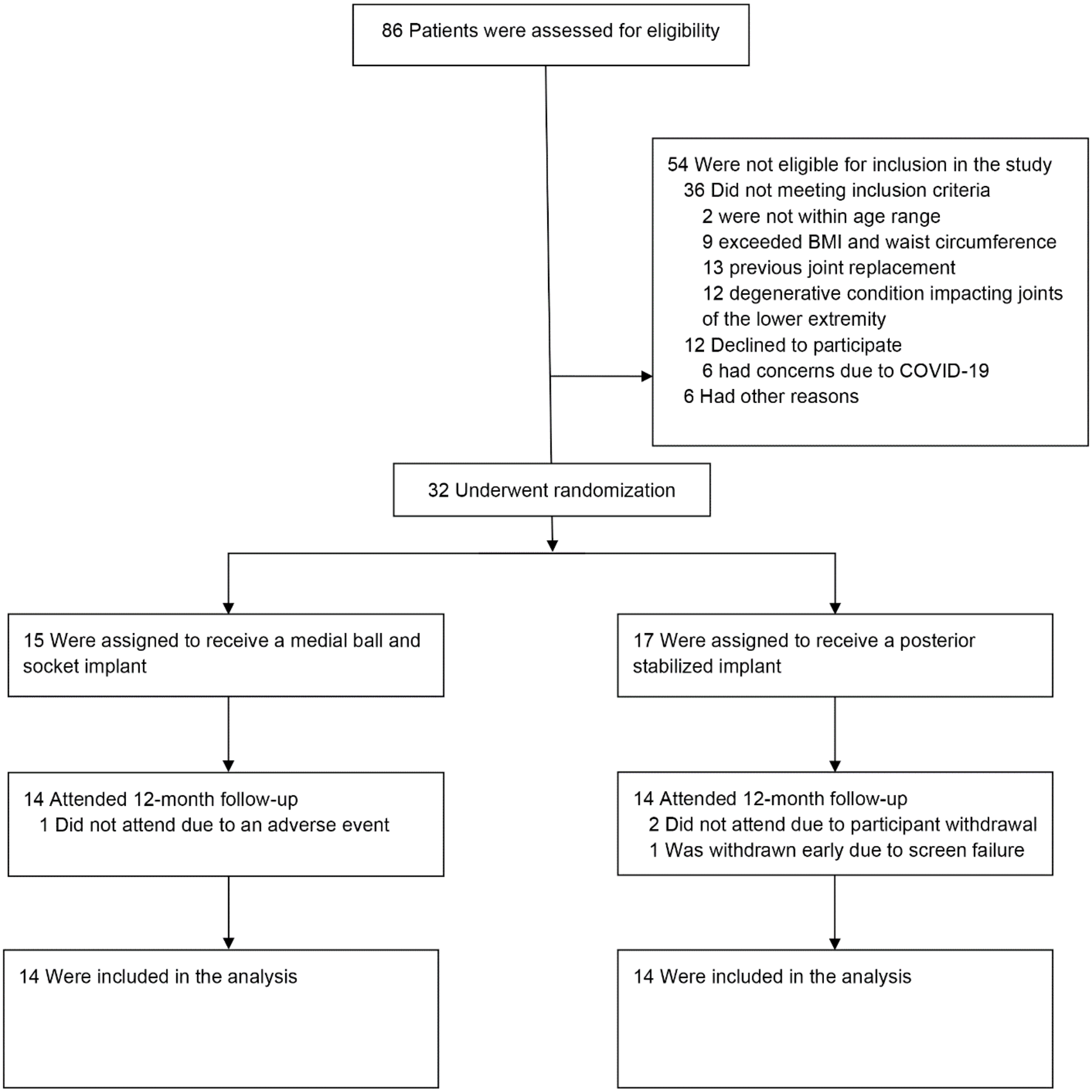

Figure 4.5.1: Consolidated Standards of Reporting Trials (CONSORT) flow diagram for enrolled patients.

Surgery

A senior arthroplasty surgeon (GFD) performed all surgical procedures, and all TKA patients followed the same protocol. After a suitable anesthetic, patients were given an appropriate weight dose of cefazolin and 15 mg/kg of tranexamic acid intravenously. A midline incision and subvastus approach was performed for all patients (Hofmann, Plaster, & Murdock, 1991). Manual instruments were used with the goal of mechanical neutral alignment with the femur first technique and the goal of the tibial component at a coronal neutral angle. The protocol called for resurfacing all patellae and the PCL was released for all patients. All components were cemented, and tourniquet use was restricted only to the time of cementation and then deflated before closure. Many patients were discharged the same day and continued with eight publicly funded outpatient physical therapy sessions. No patients required additional soft tissue release, and there were no complications or revisions with the surgical cohort.

Motion Analysis

All motion capture was completed using a 10-camera Vicon System (two Vantage V5, eight Vero 2.2, Oxford Metrics, Oxford, UK) sampled at 200 Hz. The capture volume included four force platforms sampled at 1000 Hz: two Bertec force platforms (model FP4060, Bertec Corporation, Columbus, USA) and two Kistler force platforms (model 9286BA, Kistler, Winterthur, Switzerland).

Participants were outfitted with 45 passive-reflective markers placed on anatomical locations using the University of Ottawa Motion Analysis Model (UOMAM) (Mantovani & Lamontagne, 2017). Participants completed a static trial and five trials of level walking at their preferred, self-selected walking speed. The starting spot for each participant was selected so their first step onto the force platform was always done with the affected limb. The TKA groups completed their first visit within one month of surgery, and 12 months (+/- one month) after surgery, the CTRL group completed one visit.

Motion capture data were processed using Vicon Nexus 2.9.2 software (Oxford Metrics, Oxford, UK). Trajectories were filtered using a Woltring filter with a mean, standard error of 15 mm and force platform data using a 4th-order (zero lag) Butterworth filter with a cut-off frequency of 10 Hz. Gait event detection was done with the assistance of the ground reaction forces. Trials were modelled with the UOMAM (Mantovani & Lamontagne, 2017), and relevant data were extracted with a custom-written Matlab script (2019b, MathWorks, Natick, USA). Walking speed was extracted and normalized to leg length. Knee variables of interest included sagittal and frontal angles and moments and knee joint power. Knee angles were normalized to 100% gait cycle, whereas knee moments and powers were normalized to 62% stance phase (Winter, 1984). Data were extracted from the affected limb in the MBS and PS groups and the dominant limb in the CTRL group (Kowalski et al., 2019), defined as the participants' preferred leg to kick a ball (Chapman, Chapman, & Allen, 1987). All five trials were included in the final analyses and were not averaged together.

Statistical Analyses

Scalar variance comparison was completed using the coefficient of variation (CV), which was calculated and compared to the equality of variance results (Eq. 4.5.1) (Winter, 1984).

$$CV = \frac{\sqrt{\frac{1}{N}\sum_{i=1}^{N}\sigma_i^2}}{\frac{1}{N}\sum_{i=1}^{N}|X|_i} \times 100 \tag{4.5.1}$$

N is the number of intervals

X is the amplitude of the variable of interest at the i^{th} interval

σ_i^2 is the standard deviation of X at the i^{th} interval

Percentage change $(X_{post} – X_{pre}) / X_{pre} * 100\%$ was determined for CV and KOOS variables which were compared between the MBS and PS groups.

Statistical analyses for the KOOS, CV, improvement, and walking speed variables were processed using the SPSS v.27 software (IBM Corporation, Armonk, USA). A One-Way Analysis of Variance with a Bonferroni post-hoc correction was used for the between-group comparisons. In contrast, an independent samples t-test was used to compare the percentage change between MBS and PS groups, and significance was set to $p < .05$ for all comparisons. Effect size is reported as ω^2 for ANOVA and Cohen's d for the independent samples t-test.

The comparisons of continuous trajectories of the joint kinematics, moments, and powers were compared between the groups. Equality of variance was compared between the groups using the 'gwv1d' function (Kowalski et al., 2021) throughout the waveform in Matlab.

4.5.4 Results

Demographics, walking speed and KOOS

No significant differences in age existed between the groups $F(4,65) = 0.362$, $p = .835$, $\omega^2 = .038$ (Table 4.5.1). Body mass index (BMI) was statistically different between the groups $F(4,65) = 5.647$, $p < .001$, $\omega^2 = .210$. The PS group had greater BMI compared to the CTRL group pre-operatively (4.9, 95% CI (1.2 to 8.6), $p = .003$) and post-operatively (5.4, 95% CI (1.7 to 9.1), $p = .003$).

Walking speed was statistically different between the groups $F(4,65) = 5.448$, $p < .001$, $\omega^2 = .203$ (Table 4.5.1). Pre-operatively the PS (1.21 ± 0.20) group walked slower than the CTRL (1.51 ± 0.18) (0.30, 95% CI (.08 to .53), $p = .003$) and the pre-operative MBS (1.45 ± 0.25) (.25, 95% CI (.02 to .48), $p = .027$) groups. Walking speed was used as a covariate to compare these groups.

All KOOS subscale scores significantly ($P<.05$) increased from pre-operative to 12-month post-operative visits for both the MBS and PS groups (Table 4.5.1). However, post-operatively both groups remained with lower symptoms, function in sports and recreation, and quality of life subscales scores compared to the CTRL group.

Coefficient of variance (CV) analysis for simple scalars

No significant differences in CV for any temporospatial variables existed between the groups ($P > .05$) (Table 4.5.2). Sagittal knee angle CV was statistically different between the groups $F(4,65) = 7.660$, $p < .001$, $\omega^2 = .276$. Pre-operatively the PS group had greater knee angle CV compared to the CTRL group (0.5, 95% CI (0.1 to 1.0), $p = .005$), as well as post-operatively, the PS had greater variability compared to the CTRL (0.7, 95% CI (.2 to 1.1), $p < .001$) and MBS (0.2, 95% CI (.1 to 1.0), $p = .007$) groups.

Knee power CV was statistically different between the groups $F(4,65) = 6.531$, $p < .001$, $\omega^2 = .240$. Compared to the CTRL group, the PS group had greater knee power CV both pre-operatively (41.7, 95% CI (15.8 to 67.6), $p < .001$) and post-operatively (32.7, 95% CI (1.6 to 63.9), $p = .005$).

Table 4.5.1: Group mean (SD) demographic, walking speed, and knee injury and osteoarthritis score (KOOS) values. * represents significant ($p < .05$) within-group difference between pre- and post-operative visits. † represents significant ($p < .05$) difference between MBS Pre-Op and PS Pre-Op visits; ‡ represents significant ($p < .05$) difference between MBS Post-Op and PS Post-Op visits; § represents significant ($p < .05$) difference from CTRL.

	MBS		PS		CTRL
	Pre	Post	Pre	Post	
Number of participants (n)	14	14	14	14	14
Sex (female/male)	6/8	6/8	6/8	6/8	7/7
Age (years)	62.7 (5.8)	63.7 (5.7)	64.5 (8.1)	65.6 (8.1)	64.4 (5.6)
Body Mass Index (kg/m^2)	27.9 (3.8)	27.4 (3.5)	29.8 (3.4) §	30.3 (3.9) §	24.9 (2.1)
Walking speed (leg-length normalized)	1.45 (0.25) †	1.46 (0.20)	1.21 (0.20) †§	1.29 (0.21)	1.51 (0.18)
KOOS					
Symptoms	45.2 (15.4) *§	75.8 (20.4) *§	39.8 (14.9) *§	74.2 (21.0) *§	98.7 (2.6)
Pain	54.0 (12.6) *§	86.1 (12.3) *	45.0 (17.6) *§	85.7 (10.7) *	98.6 (3.2)
Function in daily living	63.9 (20.6) *§	91.3 (10.2) *	54.1 (17.3) *§	93.4 (7.1) *	100.0 (0.0)
Function in sport and recreation	23.6 (12.0) *§	69.6 (24.0) *§	27.5 (26.4) *§	60.4 (19.8) *§	100.0 (0.0)
Quality of Life	26.8 (16.1) *§	67.4 (20.1) *§	17.4 (12.8) *§	71.0 (17.8) *§	100.0 (0.0)

Table 4.5.2: Group mean (SD) coefficient of variation values. * represents significant ($p < .05$) within-group difference between pre- and post-operative visits. † represents significant ($p < .05$) difference between MBS Pre-Op and PS Pre-Op visits; ‡ represents significant ($p < .05$) difference between MBS Post-Op and PS Post-Op visits; § represents significant ($p < .05$) difference from CTRL.

	MBS		**PS**		**CTRL**
	Pre	Post	Pre	Post	
Temporospatial					
Speed	3.5 (2.6)	3.7 (1.9)	3.5 (2.7)	4.7 (3.2)	3.8 (4.0)
Step length	2.1 (1.5)	2.5 (1.7)	2.9 (2.2)	2.4 (1.2)	2.1 (1.6)
Stride length	2.7 (1.7)	3.4 (2.1)	4.3 (3.3)	3.8 (1.7)	3.3 (2.7)
Step width	14.2 (11.1)	14.0 (11.6)	12.2 (4.9)	11.8 (7.1)	13.6 (8.1)
Stride time	2.2 (1.5)	2.4 (1.3)	2.1 (1.2)	4.2 (3.3)	2.8 (3.0)
Step time	2.7 (2.1)	4.0 (3.6)	3.3 (1.9)	7.7 (7.1)	4.4 (5.7)
Knee Variables					
Sagittal angle	2.0 (0.4)	2.0 (0.4)‡	2.5 (0.6) §	2.6 (0.4))‡§	1.9 (0.3)
Frontal angle	12.9 (8.7)	12.5 (7.0)	12.6 (4.7)	17.1 (13.6)	13.6 (7.4)
Sagittal moment	33.8 (21.3) †	45.2 (11.2)	69.6 (30.4) †§	31.5 (16.2)	30.5 (13.1)
Frontal moment	32.6 (17.2) †	40.0 (12.1)	77.9 (68.0) *†	68.1 (48.3) *	40.7 (24.6)
Power	44.9 (24.1)	46.6 (14.9)	70.3 (34.5) *§	61.3 (23.9) *§	28.6 (14.6)

Equality of variance analysis

Pre-operative knee biomechanics variability compared to the CTRL group is presented in Figure 4.5.2. The PS group had greater knee flexion angle variability compared with the MBS (28-63%, 79-89%, and 97-99% gait cycle (%) and CTRL (28-55 and 73-89%). The MBS group had periods of greater knee power variability than the PS group (0-3, 21-28, 38-40, and 45-50%). However, compared to the CTRL, the MBS (2-13, 16-46, and 48-61%) and PS (0-14, 17-61%) groups remained with lower knee power variability throughout most of the stance phase.

Pre- versus post-operative knee biomechanics variability is presented in Figure 4.5.3. In general, variability decreased post-operatively for both the PS and MBS groups. The PS group reduced knee angle variability post-operatively (12-51, 69-76, and 97-99%). The MBS group had lower knee moment variability throughout most of the stance phase (4-61%), whereas the PS group had lower post-operative knee moment variability at opposite toe off (10-19%) and throughout terminal stance (41-57%). The PS group had similar periods of decreased knee power variability post-TKA (10-19 and 48-56%), whereas the MBS group had periods of reduced knee power variability throughout the stance phase (12-14, 20-30, 38-40, and 47-62%).

Post-operative knee biomechanics variability compared to the CTRL group is presented in Figure 4.5.4. The MBS group had less knee flexion angle variability than the PS group earlier in the gait cycle (2-26%). The PS group had greater variability than the MBS group later in the gait cycle (50-64 and 90-93%). The PS group achieved greater variability than the MBS group throughout the stance phase (4-13 and 24-62%). However, compared to the CTRL, the MBS (4-40 and 42-62%) and PS (5-23 and 47-59%) groups remained with less knee moment variability throughout the stance phase. The MBS achieved periods of greater knee power variability than the PS group (1-4, 8-12. 15-17, and 46-51%). However, both the MBS (3-62%) and PS (2-60%)

groups remained with less variability compared to the CTRL group throughout most of the stance phase.

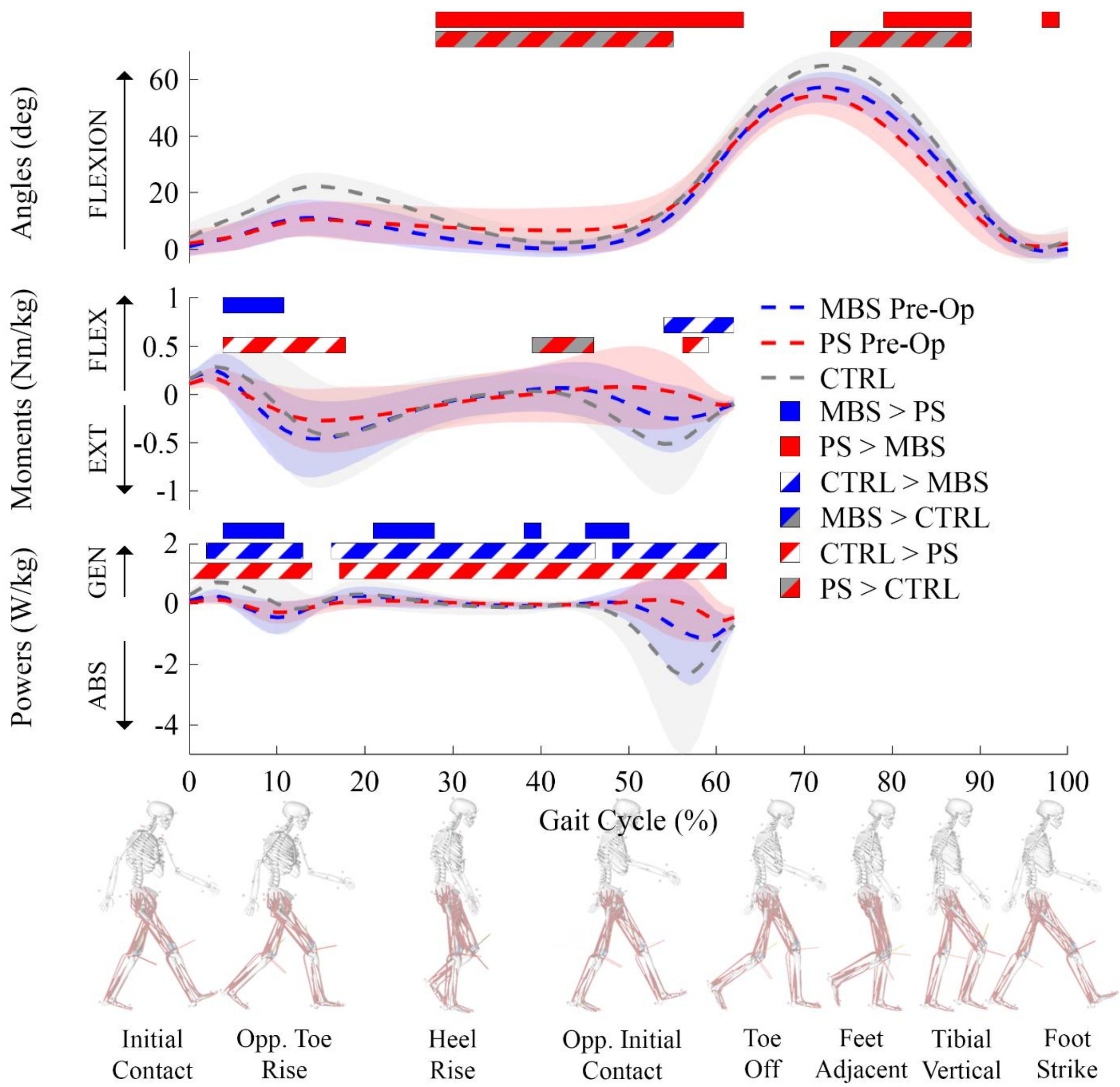

Figure 4.5.2: Group means and standard deviations (SD) for sagittal knee angles (degrees), moments (Nm/kg), and powers (W/kg) for the pre-operative medial ball-and-socket, pre-operative posterior stabilized, and control groups. Horizontal bars represent where in the movement cycle differences in variability occur and identify which group had greater variability.

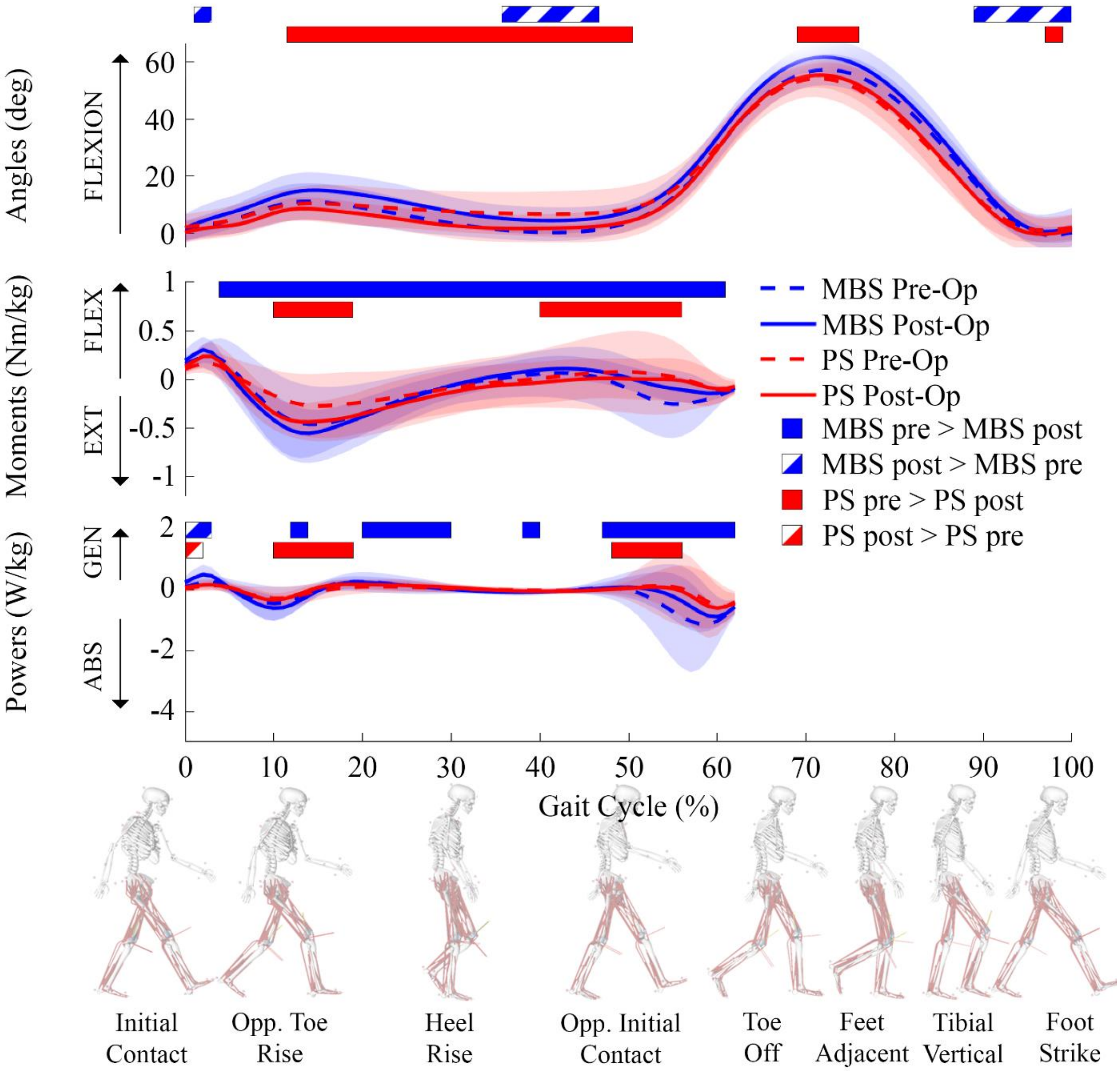

Figure 4.5.3: Group means and standard deviations (SD) for sagittal knee angles (degrees), moments (Nm/kg), and powers (W/kg) for pre-and post-operative medial ball-and-socket and posterior stabilized groups. Horizontal bars represent where in the movement cycle differences in variability occur and identify which group had greater variability.

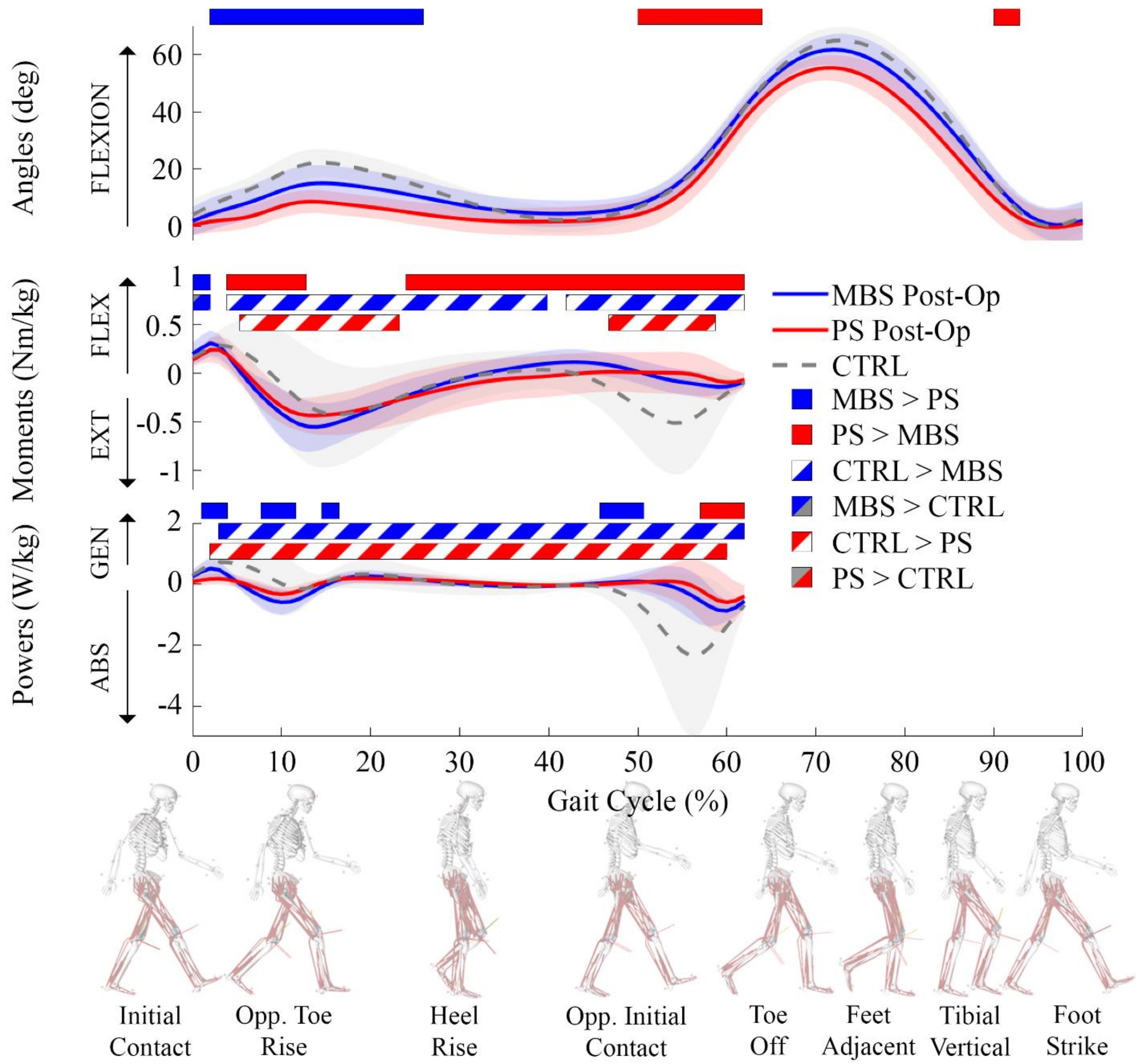

Figure 4.5.4: Group means and standard deviations (SD) for sagittal knee angles (degrees), moments (Nm/kg), and powers (W/kg) for the post-operative medial ball-and-socket, post-operative posterior stabilized, and control groups. Horizontal bars represent where in the movement cycle differences in variability occur and identify which group had greater variability.

No significant differences in the pre-to-postoperative percentage change existed for any KOOS subscales, temporospatial, or knee CV variables between the groups (*P* > .05) (Table 4.5.3).

Table 4.5.3: Group mean (SD) percentage change for Knee Injury and Osteoarthritis Outcome Score, temporospatial coefficient of variability, and knee biomechanics coefficient of variation represented the difference from pre- to post-operative visits. Positive values indicate a post-operative increase.

	MBS	**PS**	**Mean Difference (95% confidence interval)**	***p*-value**	**Effect Size (*d*)**
KOOS					
Symptoms	67.4 (53.8)	107.8 (78.0)	-40.3 (-97.0 to 16.4)	.155	.304
Pain	60.8 (38.5)	118.8 (119.0)	-58.1 (-133.0 to 16.8)	.122	.509
Function in daily living	52.4 (55.8)	78.2 (53.6)	-25.8 (-73.2 to 21.6)	.271	.649
Function in sport and recreation	202.9 (107.0)	273.0 (363.2)	-70.1 (-319.1 to 178.8)	.563	.483
Quality of Life	168.0 (263.3)	341.0 (320.7)	-148.8 (-368.0 to 70.3)	.172	.628
Temporospatial					
Speed	103.5 (235.3)	47.4 (104.3)	-56.0 (-197.4 to 85.3)	.423	-.308
Step length	18.3 (100.5)	56.3 (104.7)	38.0 (-41.7 to 117.7)	.336	.370
Stride length	26.8 (89.7)	62.0 (115.3)	35.2 (-45.0 to 115.4)	.376	.341
Step width	3.8 (58.9)	39.2 (93.2)	35.5 (-25.1 to 96.0)	.239	.455
Stride time	144.7 (194.0)	40.3 (108.8)	-104.4 (-226.6 to 17.8)	.091	-.664
Step time	162.6 (217.8)	113.7 (260.8)	-48.9 (-235.5 to 137.8)	.595	-.203
Knee Variables					
Sagittal angle	4.8 (33.6)	8.2 (21.4)	-3.3 (-25.2 to 18.6)	.757	-.118
Frontal angle	17.1 (74.0)	63.6 (178.2)	-46.5 (-152.5 to 59.5)	.375	-.341
Sagittal moment	12.5 (37.7)	-16.2 (38.9)	28.8 (-1.0 to 58.5)	.057	.751
Frontal moment	45.6 (58.1)	14.1 (59.0)	31.5 (-14.0 to 77.0)	.167	.537
Power	24.6 (71.0)	6.2 (55.9)	18.3 (-31.3 to 68.0)	.455	.287

4.5.5 Discussion

This study aimed to determine gait cycle differences in variability in patients with knee OA before and after undergoing a TKA with either an MBS or PS implant and compared to healthy, similarly aged adults. Whereas previous studies were limited to a simple scalar analysis of variability, which provided an estimate of global variability, this study implemented equality of variance comparison at each interval of the gait cycle (Kowalski et al., 2021, 2022), which provided the location where the gait cycle differences in variability occurred.

Pre-operatively, knee OA patients had less knee moment and power variability than healthy individuals (Figure 4.5.2). Knee moment and power variability further decreased post-TKA for both the MBS and PS groups (Figure 4.5.3) and remained lower than the CTRLs for most of the stance phase (Figure 4.5.4). These findings generally align with previous studies showing knee OA patients had lower variability than controls, which decreased after TKA (Yakhdani et al., 2012; Yakhdani et al., 2010; Kiss, Bejek, & Szendrői, 2012).

Evaluating variability can assess system stability. For example, high variability of physiological outcomes, like heart rate variability, is favourable as it reflects greater adaptability and a wider ability to respond. In other situations, high variability is unfavourable as it represents the inability of the physiological control system to regulate a given parameter (Hausdorff, 2007; Hausdorff, Rios, & Edelberg, 2001). Applied to gait, greater kinematic and kinetic variability is favourable as it reflects adaptability (Hausdorff, 2007). The further reduction in knee moment and power variability in the PS and MBS groups after TKA could mean patients are less stable and less capable of adapting to unanticipated situations while walking.

It is estimated that 7 to 38% of TKA recipients experience a fall within the first 12 months after TKA (Chan, Jehu, & Pang, 2018; Liu et al., 2020). Reduced movement adaptability can affect a TKA patient's ability to respond to walking disturbances, such as stepping over an obstacle or regaining balance after perturbation, which may lead to a fall (Smith, 1993). However, this variability reduction may result from movement strategies patients utilize to reduce pain and loading on the affected knee. Studies have shown that individuals with medial knee OA (Lewek et al., 2006) and after TKA (Benedetti et al., 2003) walk with a 'stiff knee gait pattern' characterized by prolonged muscle contractions during the stance phase. Patients could increase the co-contraction around the knee to increase stability but reduce their movement variability (Yakhdani et al., 2012).

Passive knee stabilization is achieved through knee ligaments, whereas muscles around the knee achieve active stability, and both systems work together to allow reliable knee joint function (Abulhasan & Grey, 2017). These structures provide proprioceptive feedback to allow for the perception of joint movements and the position of joint segments in space. In the knee, proprioception provides three fundamental functions, stabilization during the static posture, protection against excessive and possible injurious movements via reflex responses, and coordination of complex movements and precise knee joint motions (di Laura Frattura et al., 2019). Proprioceptive feedback is negatively affected in individuals with knee OA, but removing damaged structures during TKA has been reported to improve this proprioceptive feedback (Smith et al., 2014). However, this improvement in proprioception and joint stability after TKA is accompanied by a further reduction in variability. These changes need to be better understood with predicting joint function.

This study identified reduced variability in both TKA groups, regardless of the implant they received (Figures 4.5.3 & 4.5.4). The MBS implant was designed with a medial congruent tibial insert to provide greater stability from full extension through deep flexion while limiting anterior sliding (Sabatini et al., 2018). However, this increased inherent stability did not translate to improved gait variability over the PS implant. Although the MBS group achieved greater periods of knee power variability, the PS group achieved greater knee moment variability. Still, both implant groups remained with significantly less knee moment and power variability throughout most of the stance phase when compared to the CTRL group (Figure 4.5.4). KOOS measures were similar between both TKA groups (Table 4.5.1), and neither implant achieved greater improvement in CV or KOOS measurements (Table 4.5.3), so it is unclear if one implant provided greater outcomes over the other. More research is necessary to understand better the mechanisms used to reduce movement variability and evaluate how these different implant groups activate their muscles to provide joint stability.

This study was not designed to determine the source of the decreased movement variability following TKA. However, several possibilities are proposed, which will require further exploration. The MBS and PS implants in this study had cruciate sacrificing tibial inserts, meaning the anterior and posterior cruciate ligaments were removed during surgery. These cruciate ligaments play an essential role in the passive stabilization of the knee (Abulhasan & Grey, 2017), so the muscles may have to compensate for this reduced stability by creating more co-contraction, ultimately reducing gait variability (Fallah Yakhdani et al., 2010). Alternative implant designs preserve the posterior or both cruciate ligaments, improving proprioception more than the cruciate sacrificing TKA (Isaac et al., 2007). However, it is unknown if this would result in greater movement variability.

Although TKA successfully reduces pain and improves strength around the knee, patients still move with atypical movement patterns suggesting an adapted movement pattern to reduced pain while awaiting TKA (Farquhar, Reisman, & Snyder-Mackler, 2008). Several strategies to reduce variability have been identified, including stiffening the knee joint through co-contraction, walking slower, or paying more attention while walking (Yakhdani et al., 2010), which could have been used by the patients in this study. These strategies do not imply conscious cognitive involvement, as it is suspected that patients do not know how they are adapting or why they are doing so (Dijksterhuis & Aarts, 2010; Yakhdani et al., 2010). The knee OA patients in this study walked with less variability than the controls (Figure 4.5.2), so the further reduction in variability may have been due to these movement strategies they adopted due to the OA, which were further reduced due to either the implant design or the surgery itself.

Limitations

This study evaluated MBS and PS implants with cruciate sacrificing inserts, so the findings of this study cannot be generalized to all implant types. It also assessed patients 12 months after surgery. Patients' functions may continue to improve beyond this time (Cheng, Dashti, & McLeod, 2007), so gait variability may increase over time. Greater variability indicates system flexibility and wider functional responsiveness (Joffe, 1992), so variability analysis may prove to be a potential measure of recovery after TKA.

Like many biomechanical studies, this study had a relatively small sample size which may increase the risk of type II errors (Knudson, 2017). Although patients with a body mass index above 35 are included in the general patient population, the decision was taken to exclude them from participation to improve the accuracy of the motion capture measurements, as the excessive adiposity can affect the placement of reflective markers and create additional skin movement artifacts (da Silva-Hamu et al., 2013). Some evidence has shown that gait variability associated with knee OA is sex-dependent (Lewek et al., 2006). To overcome this shortcoming, we included the same number of female and male participants in the MBS and PS groups and had an equal number of females and males in the control group.

4.5.6 Conclusions

In conclusion, this study identified that patients with knee OA had reduced knee moment variability throughout the stance phase compared to healthy participants of similar age. Knee moment and power variability further decreased following TKA, with neither the MBS nor PS implant providing movement variability like the controls. Future research is necessary to determine the source for the reduction in variability and whether it is due to the cruciate sacrificing design of the MBS and PS implants, an unresolved adaptation developed while awaiting TKA, or other unknown reasons.

4.5.7 References

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4.6 Total knee arthroplasty recipients descend a ramp with a cautious gait pattern twelve months after surgery independent of implant design

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Contents of this chapter prepared as an article for submission in *Journal of Orthopedic Research*

Kowalski, E., Catelli, D. S., Dervin, G., & Lamontagne, M. Total knee arthroplasty recipients descend a ramp with a cautious gait pattern twelve months after surgery independent of implant design. 2023. Submitted.

4.6.1 Abstract

Poor knee function after surgery is one of the main factors of patient dissatisfaction after total knee arthroplasty (TKA). Descending tasks are more demanding on the knee than level walking, and patients report discomfort performing these tasks. This study investigated how implant designs can influence knee biomechanics during a ramp descent gait task. Twenty-eight patients underwent TKA with either cruciate-sacrificed medial congruent or posterior stabilized implants. Patients underwent motion analysis which measured knee biomechanics as they descended a nine-degree instrumented ramp before and 12 months after TKA. Results were compared with 14 healthy controls of similar age. Pre-operatively, different gait patterns were observed in the patient groups. However, post-operatively all TKA patients adopted the same movement pattern, characterized by a knee extension moment and increased power absorption throughout the stance phase. These differences existed in all TKA patients, regardless of implant type or how they walked pre-operatively. They adopted a cautious gait pattern to increase stability by widening their steps and stiffening their knee to absorb energy as their body moved into single-limb support. Patients after TKA lack joint stability, and their muscles work primarily to stabilize the knee and dissipate mechanical energy. In contrast, half the controls could dissipate and generate power at the knee during the ramp descent task, indicating a more coordinated movement pattern. In conclusion, after one year of surgery, all TKA patients descended a ramp with a cautious gait pattern, regardless of which implant they received or how they walked pre-operatively.

4.6.2 Introduction

Total knee arthroplasty (TKA) is successful at reducing pain for patients with end-stage knee osteoarthritis (OA) (Roos & Toksvig-Larsen, 2003). Although most patients are satisfied with the outcomes (Choi & Ra, 2016), some remain dissatisfied with their TKA, often citing limitations while performing various activities of daily living (Gunaratne et al., 2017). Descending tasks are more demanding on the knee joint than level walking, creating larger knee extension moments that represent knee joint load (Pickle, Grabowski, Auyang, & Silverman, 2016). The increased demands caused by descending tasks create more difficulties for patients, who report feeling uncomfortable performing these tasks (Collins, Misra, Felson, Crossley, & Roos, 2011). However, the patient-reported outcome measures cannot capture the objective functional deficits that may remain following TKA (Stevens-Lapsley, Schenkman, & Dayton, 2011). Therefore, a biomechanical analysis is imperative to understand the functional adaptations after TKA (Lamontagne, Beaulieu, Varin, & Beaule, 2009).

Several studies have identified altered biomechanics in patients after TKA while descending a ramp (Simon, Della Valle, & Wimmer, 2018; Wen, Cates, Weinhandl, Crouter, & Zhang, 2022; Wiik, Aqil, Tankard, Amis, & Cobb, 2015). Additionally, only a few studies had the purpose of comparing different implant designs during these tasks. Implant designs can influence knee joint function following TKA (Andriacchi et al., 1997). However, when different implant designs are pooled together in a study, it is difficult to determine the effect of the implant design. Many different knee implant designs exist (Dall'Oca et al., 2017), and the debate is still ongoing on if one implant design provides better functional outcomes with fewer complications (Kahlenberg, Lyman, Joseph, Chiu, & Padgett, 2019). These different implant designs have features which claim to provide increased knee joint stability for patients. Medial ball and socket (MBS) implants have a highly congruent medial compartment and less conforming lateral compartment, which were designed to reduce anterior sliding of the medial femoral component on the tibial component that is experienced in patients with posterior stabilized (PS) implants

(Atzori, Salama, Sabatini, Mousa, & Khalefa, 2016). The more congruent design of an MBS implant may provide additional stability during functional tasks. However, during level walking, findings were inconsistent with either PS (Kulshrestha et al., 2020) or MBS (Gray et al., 2020) implants having similar knee kinematics to healthy controls, so it is unknown how these different implant designs would perform during descending tasks.

Often gait studies comparing TKA implants focus on discrete biomechanical outcomes, such as peak extension moment or knee range of motion. However, gait is a continuous process. Therefore, methods that examine the entire movement cycle may uncover additional information which may have been missed with discrete analyses. Statistical parametric mapping (SPM) analyzes the whole gait cycle and can identify multiple areas within the movement cycles where significant differences exist (Pataky, Robinson, & Vanrenterghem, 2016). Analyzing the entire movement cycle with SPM may be more sensitive and provides additional differences between the MBS and PS groups which traditional discrete statistics may have missed. A previous study which compared gait in patients who underwent total hip arthroplasty with different surgical approaches did not find any differences between the approaches using discrete statistics but was able to identify differences when compared using SPM (Pincheira, De La Maza, Silvestre, Guzmán-Venegas, & Becerra, 2019).

To our knowledge, no study exists in the TKA literature which compared knee biomechanics throughout the entire movement cycle during a ramp descent (RD) task. Therefore, this study aimed to investigate how implant designs can influence knee biomechanics during a ramp descent task one year after TKA. It was hypothesized that the congruent design of the MBS implant would provide additional knee joint stability and allow patients to achieve a movement pattern closer to healthy controls than those with a PS implant.

4.6.3 Methods

Subjects

This randomized control trial initially screened 86 individuals at end-stage knee OA (Kellgren and Lawrence, grade 4 (KL4) [19]) who were scheduled to undergo TKA by a single senior orthopedic surgeon (GD). Eligible participants were between the ages of 45 and 75 at the time of enrollment and needed to be willing complete the required study visits. Participants were excluded if they had a body mass index (BMI) and waist circumference measurement > 35 kg/m2 and 102 cm respectively for men and > 35 kg/m2 and 88 cm respectively for women; any past or present condition, which in the opinion of the investigators may impact gait; or previous joint replacement of the enrolled knee or other lower limb joint replacement. TKA patients were excluded if they had a degenerative condition (other than osteoarthritis in the enrolled knee) impacting joints of the lower extremities. Ethics approval was approved by the university and hospital research ethics boards following approval by the U.S. Food and Drug Administration. The trial was registered in the ClinicalTrials.gov database (NCT02589197). All participants were informed about the study and provided signed consent.

Thirty-two participants were eligible and underwent randomization to receive either an MBS (MicroPort EVOLUTION® Medial Pivot System with Cruciate Sacrificing tibial inserts) or PS (Zimmer Biomet NexGen® PS TKA system with PS inserts) implants (Figure 4.6.1). Participants were required to attend the gait laboratory within one month before undergoing TKA and 12 months post-surgery. At each visit, participants completed the KOOS questionnaire (Roos, Roos, Lohmander, Ekdahl, & Beynnon, 1998). Due to technical difficulties during the data collection, pre-operative motion capture data was unavailable for three MBS participants. These three participants were also excluded from the post-operative analysis. Twenty-five TKA participants were included in the final analysis (Figure 4.6.1). A sample of 14 similarly aged healthy participants was recruited from the community to form a control group (CTRL). The CTRL participants had the same inclusion criteria and were excluded if they had degenerative conditions impacting the lower extremities' joints. The CTRL group completed a single gait lab visit.

Surgery

All surgical procedures were performed with a midline incision and subvastus approach at a tertiary hospital by a single senior arthroplasty surgeon (GD). The protocol required for resurfacing all patellae and PCL was released. All components were cemented, and tourniquet use was restricted to the time of cementation and then deflated before closure (Hofmann, Plaster, & Murdock, 1991). No patients required additional soft tissue release.

Motion Analysis

Motion analysis was completed using a three-dimensional 10-camera motion capture system (200Hz, VICON, Oxford Metrics, Oxford, England) synchronized with two force platforms (1000Hz, Kistler, Winterthur, Switzerland). An instrumented 3x1m ramp with a 1x1m platform at the top was inclined to 9° and had two force platforms embedded in the middle of the ramp to record a step from each foot as participants walked down the ramp. Although some provincial building codes limit ramps to a slope gradient of 1:10 (5.7°), larger slopes are encountered within the natural environment, and studies which have included slopes outside of this building code range and shown that knee extension moments increase as the slope increased to 9° (Pickle et al., 2016; Wen et al., 2022).

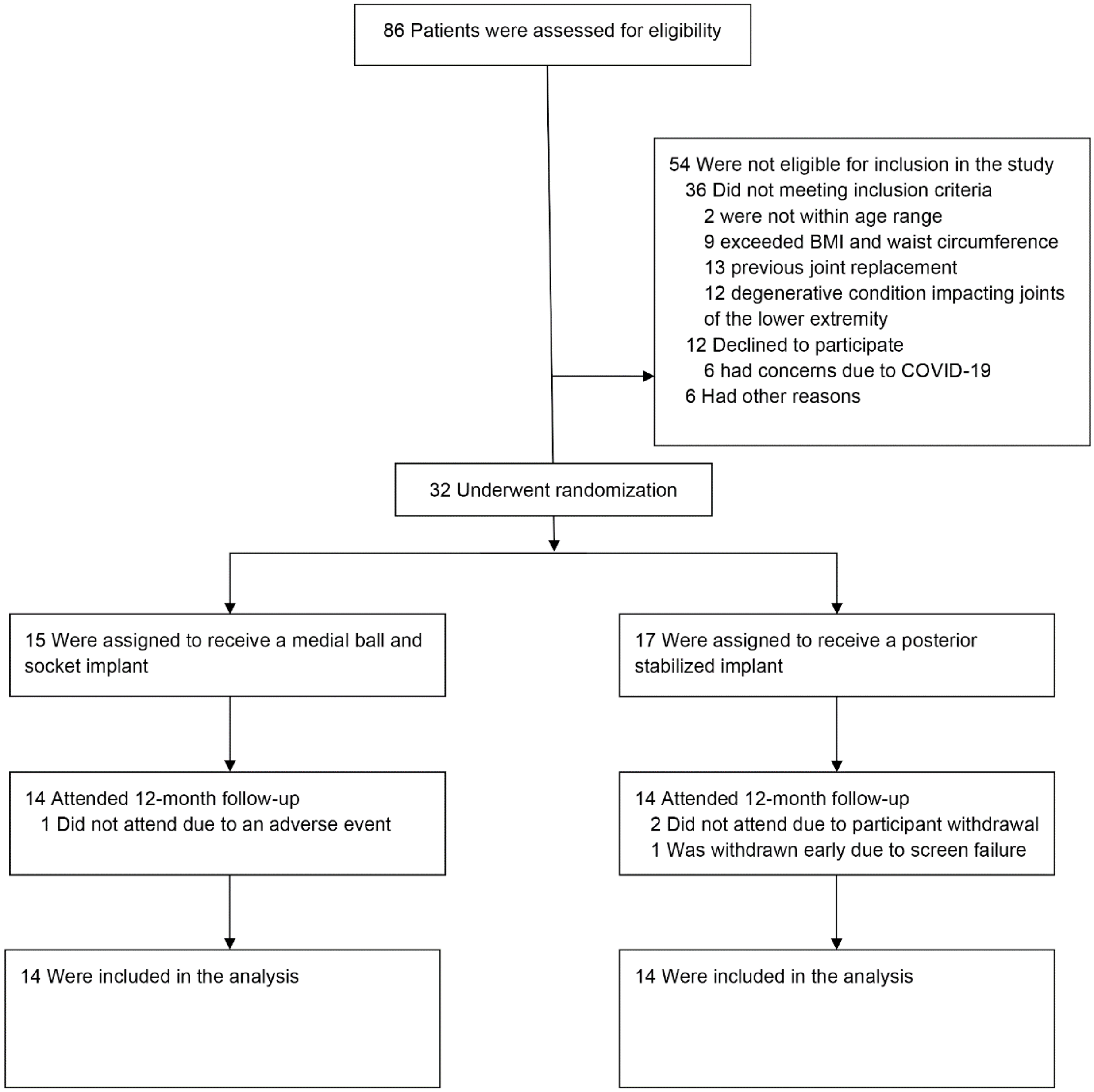

Figure 4.6.1: Consolidated Standards of Reporting Trials (CONSORT) flow diagram for enrolled patients.

Forty-five spherical passive-reflective markers (15mm) were placed bilaterally on all participants according to the University of Ottawa Motion Analysis Model (UOMAM) (Mantovani & Lamontagne, 2017). Participants completed a static trial and five trials descending the ramp in their own footwear, at their preferred pace, and without holding onto the handrail.

Motion capture data were processed using Vicon Nexus 2.9.2 software (Oxford Metrics, Oxford, UK). Trajectories were filtered using a Woltring filter with a mean, standard error of 15mm, and force platform data using a 4th order (zero lag) Butterworth filter with a cut-off frequency of 10Hz. Gait event detection was done with the assistance of the ground reaction forces (GRF), and the trials were modelled with the UOMAM (Mantovani & Lamontagne, 2017) to compute the variables of interest. A custom-written Matlab script (2019b, MathWorks, Natick, USA) was used to extract relevant data.

Outcome Variables

Comparisons were made on the affected side undergoing TKA for the MBS and PS groups and the dominant limb in the CTRL group (Kowalski, Catelli, & Lamontagne, 2019), defined as the limb they would use to kick a ball (Chapman, Chapman, & Allen, 1987). The variables of interest were sagittal knee angles, moments, and joint powers. Knee moments were expressed as external moments normalized to body mass (Nm/kg), and knee power was normalized to body mass (W/kg). Knee biomechanical variables were normalized to 100% gait cycle (GC), which was determined with the assistance of the GRF. The stance phase was represented as 0 to 62% GC, and the swing phase from 62 to 100% GC (Winter, 1984). The entire normalized waveforms of the knee biomechanical variables were evaluated between the participants. All five trials were included and were not averaged together for each participant.

Several temporospatial variables were extracted. Since the stature of participants can affect these parameters, i.e., taller individuals tend to walk with longer steps. Several temporospatial variables were normalized to leg length, including walking speed, step length, and step width (Hof, 1996). All five subscales of the KOOS were compared between the groups.

Data Analysis

All discrete statistical analyses of demographic, temporospatial and KOOS variables were done using SPSS v.27 software (IBM Corporation, Armonk, USA). A paired t-test ($P<.05$) assessed changes between pre-operative and post-operative time points. A One-Way ANOVA with a Bonferroni post hoc correction to measure differences between the groups.

A point-by-point analysis of knee biomechanics along the entire movement cycle was completed using statistical parametric mapping (SPM) (Pataky et al., 2016). To assess differences between pre-operative and post-operative time points, a paired SPM ($P<.05$) was used, whereas between-group comparisons were completed using analysis of variance SPM ($P<.05$).

Several biomechanical outcomes may be influenced by walking speed (Fukuchi, Fukuchi, & Duarte, 2019). Therefore, if walking speed were identified to be significantly different, it would be used as a covariate for all knee biomechanics (angles, moments, and power) waveforms, step length, and step width.

Initial inspection of the data revealed different movement patterns pre-operatively in the knee OA participants (Appendix - Figure 4.6.6) and in the CTRLs (Appendix - Figure 4.6.7). These movement patterns were characterized by the sagittal knee moment curve pattern, with distinct peaks occurring at the start and end of the single support phase (occurs between ~10-50% ramp descent cycle ($\%_{RDC}$)). Most individuals with OA (n=18) walked with an extension moment throughout ramp descent (OA_{Ext}), whereas several (n=7) initially had a knee flexion moment at the start of the single support phase, followed by a knee extension moment ($OA_{ExtFlex}$). Patients were randomized to either the MBS or PS group before their pre-operative gait lab visit, so the distribution of walking patterns between the two implant groups is unequal (Appendix – Table 4.6.3). Seven of the CTRLs walked with a knee extension moment throughout the stance phase ($CTRL_{Ext}$), whereas the other seven walked with a knee flexion and extension moment pattern ($CTRL_{FlexExt}$).

4.6.4 Results

Regardless of how patients walked pre-operatively or which implant they received, they all adopted the knee extension moment gait pattern one year following TKA (Figure 4.6.2). Some post-operative differences arose based on their pre-operative gait pattern. The $OA_{FlexExt}$ group had more knee flexion near the return to double-limb support compared to the OA_{Ext} group (40-51$\%_{GC}$).

After controlling for speed, the MBS group walked with more knee flexion during the early stance (0-12$\%_{GC}$) and swing (68-82$\%_{GC}$) phase (Figure 4.6.3). The MBS group had a larger peak knee extension moment (15-17$\%_{GC}$) and greater power absorption (11-19$\%_{GC}$) during single limb support compared to the PS group. In general, both PS and MBS groups walked with similar knee joint biomechanics as the $CTRL_{Ext}$ group. Some differences did occur during midstance and not at the transitioning periods between single- and double-limb support.

Compared to the $CTRL_{FlexExt}$ (Figure 4.6.4), both MBS and PS groups had altered knee biomechanics, with the knee in an extension moment throughout single limb support, whereas the $CTRL_{FlexExt}$ had a flexion moment. Before toe-off, both groups had less knee extension moment than the $CTRL_{FlexExt}$ (MBS: 50-58$\%_{GC}$, PS: 51-59$\%_{GC}$). At the start of single limb support, the $CTRL_{FlexExt}$ generated power at the knee, whereas the MBS (3-14$\%_{GC}$) and PS (2-

16%$_{GC}$) absorbed power. All groups absorbed power at the knee before toe-off. However, the CTRL$_{FlexExt}$ absorbed more than the MBS (50-59%$_{GC}$) and PS (47-60%$_{GC}$) groups.

Patient demographics comparing the PS, MBS and CTRL groups are reported in Table 4.6.1. No differences in age existed, but BMI was significantly different between the groups, $F(3,35)=4.545$, $p=.009$, with the PS having a larger BMI than CTRL$_{Ext}$ (5.42, 95%CI (-5.2 to 12.0, $p = .014$). MBS and PS groups improved in all subscales of the KOOS compared to their pre-operative measures. However, no differences in KOOS existed between the PS and MBS groups for any of the subscales ($p>.05$). Both MBS and PS groups remained with significantly lower KOOS Function in Sport and Recreation and Quality of Life subscale scores ($p < .01$) when compared to the control groups.

Temporospatial outcomes are presented in Figure 4.6.5. The MBS group walked significantly slower post-operatively compared to pre-op (0.08, 95% CI (0.03 to 0.13, $p < .001$). Although the PS group increased their walking speed post-TKA (0.11, 95% CI (0.06 to 0.16, $p < .001$), they still walked slower than the MBS (0.16, 95% CI (0.03 to 0.283, $p = .003$) and CTRL$_{Flex}$ groups (0.27, 95% CI (0.13 to 0.40, $p < .001$). The CTRL$_{FlexExt}$ walked faster than the CTRL$_{Ext}$ group (0.16, 95% CI (0.01 to 0.33, $p = .04$). Step widths increased post-operatively compared to pre-operative measures in both MBS (0.01, 95% CI (0.01 to 0.02, $p = .002$) and PS groups (0.01, 95% CI (0.01 to 0.02, $p = .02$). The PS group walked with a longer step time (0.03, 95%CI (0.01 to 0.05, $p = .044$) and wider step width (0.03, 95%CI (0.01 to 0.04, $p < .001$) compared to MBS. However, both MBS and PS had wider step widths than CTRL$_{FlexExt}$ ($p < .05$), and the PS group also had wider steps than CTRL$_{Ext}$.

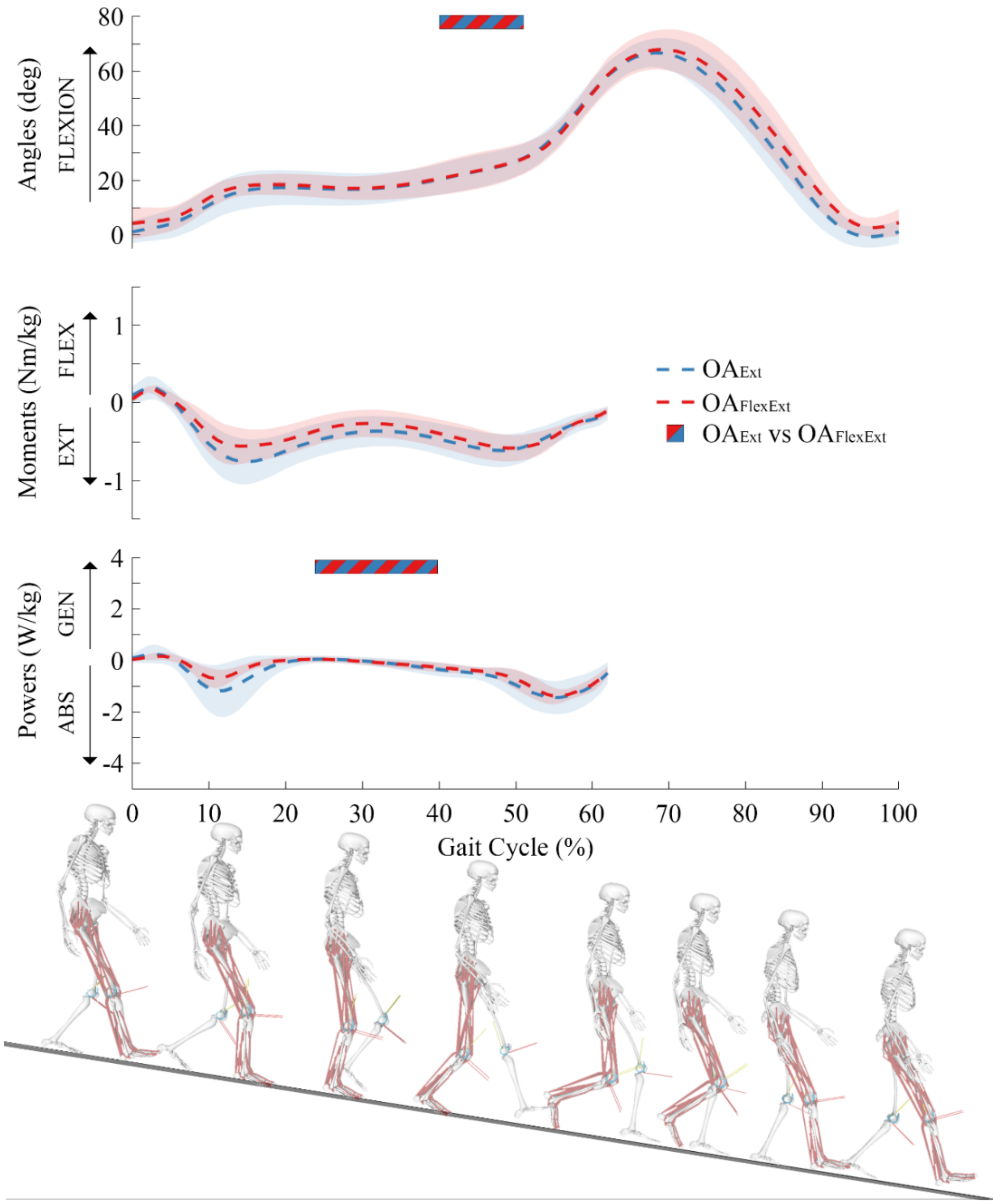

Figure 4.6.2: Post-operative group mean, standard deviations and statistical parametric mapping (SPM), results for knee angles, moments, and joint powers grouped based on pre-operative gait pattern. The filled areas in the horizontal bar graphs represent where significant differences were identified in the cycle using SPM. The joint angles (degrees) were normalized to 100% GC, whereas the joint moments (Nm/kg) and joint powers (W/kg) were normalized to 62% stance phase.

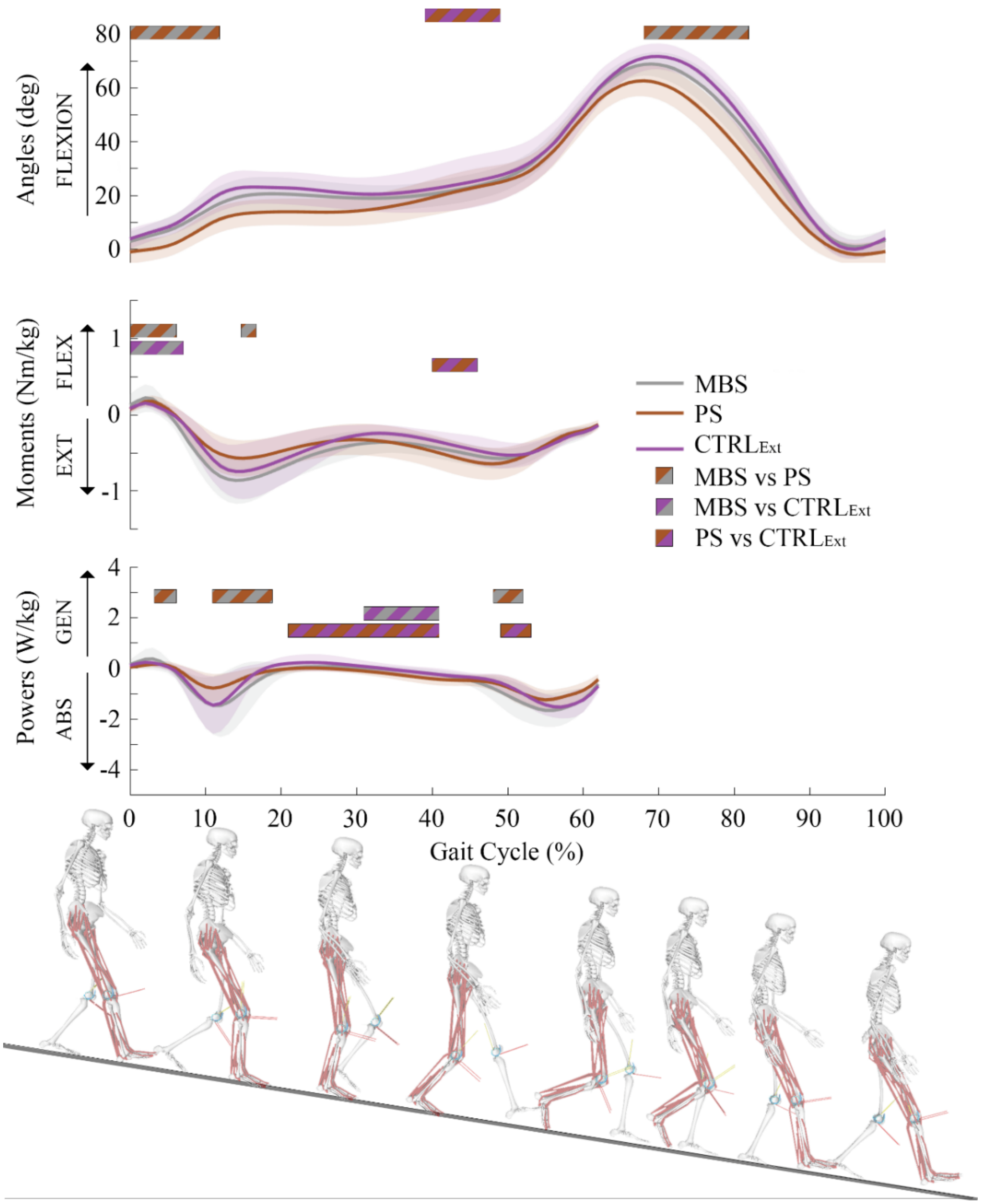

Figure 4.6.3: : Post-operative group mean, standard deviations and statistical parametric mapping (SPM) results for knee angles, moments, and joint powers comparing the MBS, PS, and CTRLExt groups. The filled areas in the horizontal bar graphs represent where significant differences were identified in the cycle using SPM. The joint angles (degrees) were normalized to 100% GC, whereas the joint moments (Nm/kg) and joint powers (W/kg) were normalized to 62% stance phase.

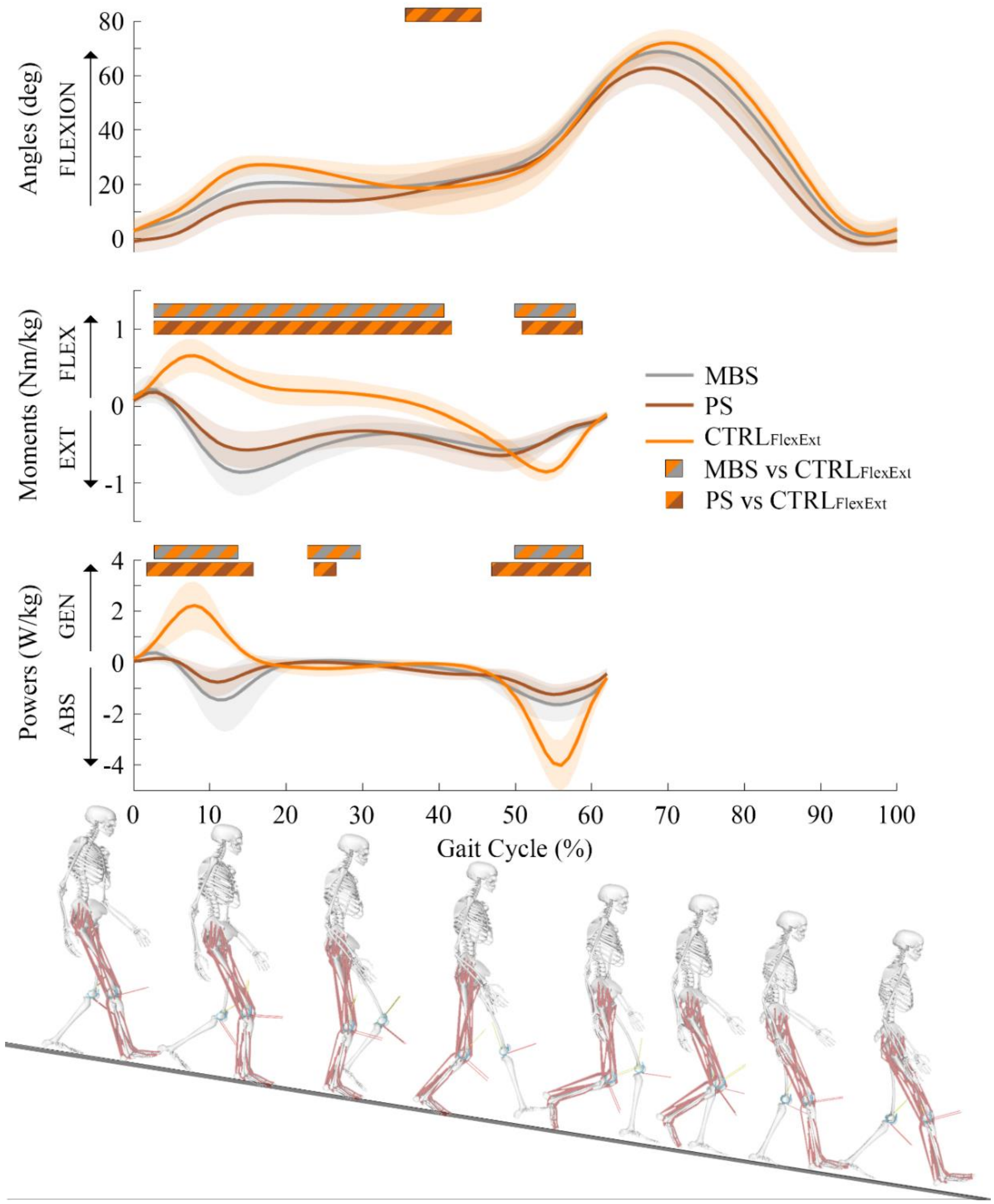

Figure 4.6.4: Post-operative group mean, standard deviations and statistical parametric mapping (SPM) results for knee angles, moments, and joint powers comparing the MBS, PS, and CTRLFlexExt groups. The filled areas in the horizontal bar graphs represent where significant differences were identified in the cycle using SPM. The joint angles (degrees) were normalized to 100% GC, whereas the joint moments (Nm/kg) and joint powers (W/kg) were normalized to 62% stance phase.

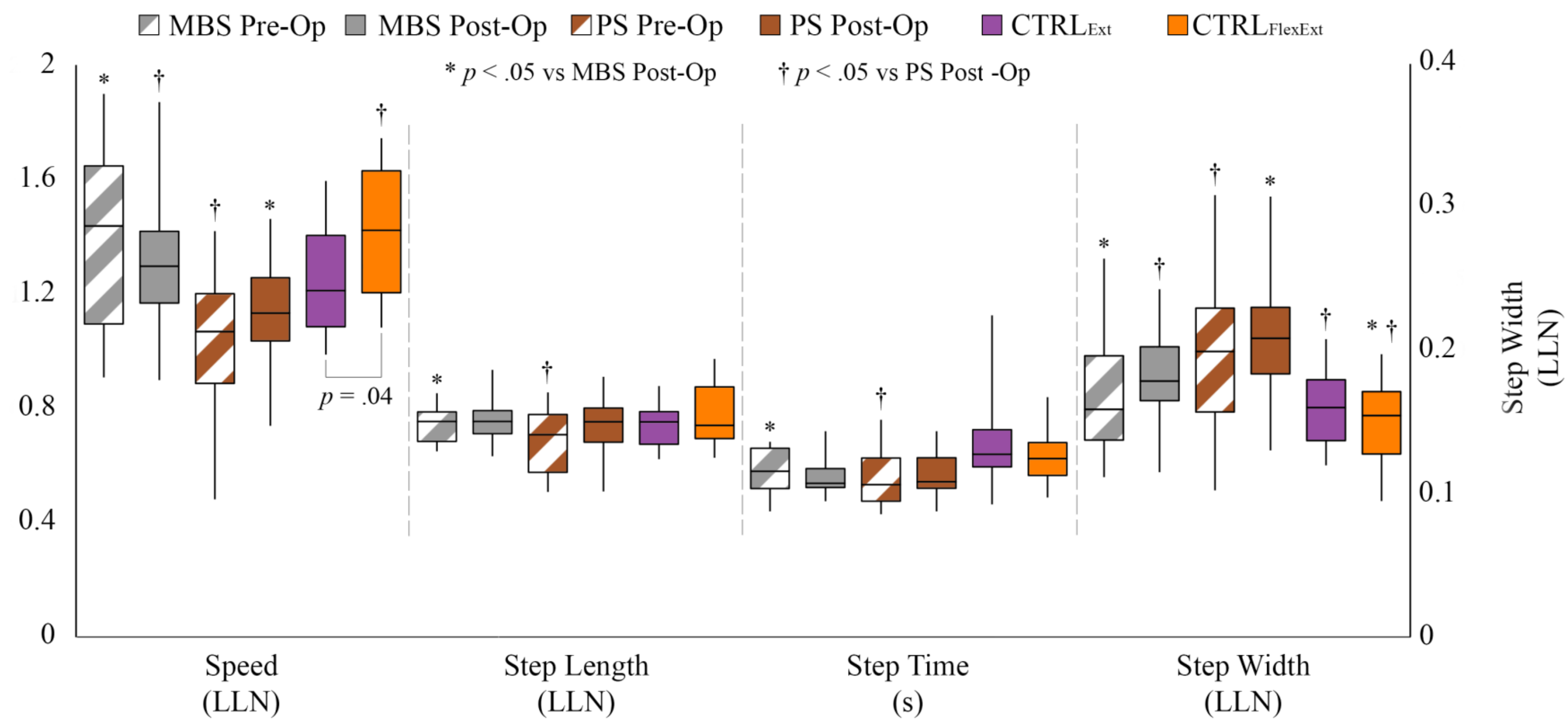

Figure 4.6.5: Group mean and standard deviations for temporospatial outcomes between the MBS, PS, CTRLExt, and CTRLFlexExt groups. The primary axis describes walking speed, step length, and step time, whereas the secondary axis describes step width. Walking speed, step length, and step width were all leg length normalized (LLN). * represents significant ($p < .05$) difference from MBS Post-operative visit; † represents significant ($p < .05$) difference from PS Post-operative visit.

Table 4.6.1: Post-operative group mean (SD) demographic and KOOS subscale values.

	MBS Pre-Op	Post-Op	PS Pre-Op	Post-Op	$CTRL_{Ext}$	$CTRL_{FlexExt}$
Number of participants (n)	11	11	14	14	7	7
Sex (female/male)	5/6	5/6	8/6	8/6	4/3	2/5
Age (years)	62.1 (5.7)	63.2 (5.7)	64.5 (8.1)	65.6 (8.1)	62.2 (5.6)	66.5 (5.1)
Body Mass Index (kg/m^2)	28.1 (4.2)	27.6 (3.9)	29.8 (3.4)	30.3 (3.9)	23.9 (2.1)†	25.9 (1.7)
KOOS						
Symptoms	43.2 (14.9)*	77.9 (18.9)	39.8 (14.9)†	74.2 (21.0)	98.5 (2.8)†	99.0 (2.7)†
Pain	54.8 (12.9)*	87.4 (10.3)	45.0 (17.6)†	85.7 (10.7)	99.2 (2.1)	98.0 (4.2)
Function in daily living	66.2 (21.9)*	91.8 (10.8)	54.1 (17.3)†	93.4 (7.1)	100.0 (0.0)	100.0 (0.0)
Function in sport and recreation	21.8 (12.9)*	70.0 (22.7)	27.5 (26.4)†	60.4 (19.8)	100.0 (0.0)*†	100.0 (0.0)*†
Quality of Life	24.4 (13.5)*	67.6 (19.1)	17.4 (12.8)†	71.0 (17.8)	100.0 (0.0)*†	100.0 (0.0)*†

* significant ($p < .05$) difference from MBS Post-Op

† significant ($p < .05$) difference from PS Post-Op

4.6.5 Discussion

This study compared knee biomechanics across the entire gait cycle during an RD task in patients with knee OA before and 12 months after undergoing a TKA and compared them to a healthy control group of similar age. To the researchers' surprise, different gait patterns were observed pre-operatively and within the control groups, characterized by different knee moment patterns during the stance phase. However, regardless of how patients walked pre-operatively or which implant they received, they all walked with a considered 'cautious' gait pattern after TKA. This cautious gait pattern was characterized by a knee extension moment and knee power absorption throughout the stance phase. In contrast, half the controls and several patients pre-operatively walked with a gait pattern with a knee flexion moment at the start of the stance phase, followed by a knee extension moment before toe-off at the end of the stance phase.

Human locomotion involves the generation and dissipation of mechanical energy (i.e., positive and negative work, respectively) throughout the gait cycle (Devita, Helseth, & Hortobagyi, 2007). This positive and negative work has been directly attributed to the work generated or dissipated by skeletal muscles contracting concentrically (shortening) or eccentrically (lengthening), respectively (Devita et al., 2007; Elftman, 1939; Winter, 1983). Humans generally adopt a gait pattern that minimizes energetic cost and maintains stability to avoid falling over. When walking downhill, gravitational energy can assist with propulsion, but at the expense of stability. However, studies have shown that individuals do not take advantage of the propulsion provided by gravity to decrease the energetic cost but instead opt for a more stable and costly gait pattern (Hunter, Hendrix, & Dean, 2010; Monsch, Franz, & Dean, 2012).

Descending a ramp presents a falling hazard due to the potential of slipping and loss of balance (Redfern & DiPasquale, 1997). As the ramp angle increases, individuals generate larger shear (i.e., braking) forces to increase stability (McVay & Redfern, 1994). These larger shear forces increase knee extension moments and power absorption throughout the stance phase (Redfern & DiPasquale, 1997), observed in this study. However, not all patients adopted this gait pattern preoperatively, as many individuals descended the ramp with a knee flexion moment during single limb support (Figure 4.6.2). This type of gait pattern has smaller braking forces and larger propulsive anteroposterior GRF (Redfern & DiPasquale, 1997). This resulted in a faster walking speed (Figure 4.6.5) and suggested a more efficient gait pattern. Much of the metabolic cost of walking occurs during the transition from one step to the other (Kuo & Donelan, 2010). The $OA_{FlexExt}$ and $CTRL_{FlexExt}$ groups generated power at the knee at the start of single limb support (~$10\%_{GC}$) when their opposite foot was leaving the ground, efficiently transferring power from one limb to the other. In contrast, the groups with a knee extension moment at this point in the gait cycle constantly absorbed power at the knee joint. Effectively, this extension moment pattern resulted in a 'controlled fall' from one foot to the other as they descended the ramp, which increased stability, but decreased efficiency (Hunter et al., 2010; Monsch et al., 2012).

Three previous studies (Simon et al., 2018; Wen et al., 2022; Wiik et al., 2015) have evaluated knee biomechanics during a ramp descent task in TKA patients. Wen et al. (Wen et al., 2022) compared operated and non-operated knee biomechanics in patients who descended ramps sloped at 5, 10, and 15°. The operated knee of their TKA patients had less peak knee extension moment compared to their non-operated knee and controls, suggesting that patients are still favouring their non-operated limb (Wen et al., 2022). The longer follow-up range of 7-46 months post-TKA (Wen et al., 2022) suggests that patients still utilize this cautious gait pattern even after longer recovery times. Other studies compared how different implant designs impact ramp descent gait, including cruciate retaining (CR) vs bicruciate retaining (BiCR) (Simon et al., 2018) and CR vs unicompartmental knee arthroplasty (UKA) (Wiik et al., 2015). Simon et al. (2018) found no differences in knee joint kinetics between the CR and BiCR groups. Still, they did find that the BiCR had less quadriceps muscle activity, suggesting the BiCR may provide some neuromuscular benefits for stabilizing the knee (Simon et al., 2018). Temporospatial and GRF were compared in CR and UKA patients as they continuously walked on a declined treadmill. No differences in preferred walking speed were identified, but UKA and TKA groups walked with wider steps than their control group (Wiik et al., 2015). However, no studies evaluated patients pre-operatively or compared cruciate sacrificing TKAs.

Both the PS and MBS implant groups in this study descended the ramp with a cautious gait pattern, regardless of how they walked pre-operatively (Figure 4.6.2). This adaptation likely increased stability as the MBS group reduced their walking speed post-TKA, and both MBS and PS groups walked with wider steps than pre-op (Figure 4.6.5). Walking slower and with wider steps are adaptations individuals make to increase their stability (Maki, 1997). While differences in gait biomechanics existed between the MBS and PS groups, the KOOS could not identify them (Table 4.6.1), highlighting the importance of a postoperative biomechanical assessment during the patient's follow-up.

The MBS group achieved greater knee flexion angles than the PS group and had larger knee extension moments and power absorptions at the start of single limb support (Figure 4.6.3). These findings suggest that the PS group used greater knee joint stiffness than the MBS group to reduce loading at the knee joint (Briem & Snyder-Mackler, 2009; McGinnis, Snyder-Mackler, Flowers, & Zeni, 2013). The differences in implant design may explain these observations, as the MBS implants have a highly congruent medial compartment and a less conforming lateral compartment. These were designed to improve stability and reduce anterior sliding for the femoral component experienced in patients with PS implants (Atzori et al., 2016). The greater knee joint stiffness within the PS group is likely due to increased co-contraction between the quadriceps and hamstrings (McGinnis et al., 2013). Simon et al. (2018) identified differences in muscle activity when comparing different implant types during a ramp descent task (Simon et al., 2018), so a future study which evaluates muscle activity may uncover neuromuscular differences between the MBS and PS groups, especially during more eccentric tasks where the knee has to maintain stability while lowering the body's centre of mass.

Our hypothesis was not proven, as few differences were observed when the MBS and PS groups were compared with the $CTRL_{Ext}$ group (Figure 4.6.3), so neither implant group achieved a more similar gait pattern to them. Often the $CTRL_{Ext}$ had larger mean values in their knee biomechanics than both the MBS and PS groups. Yet, only the comparisons between the MBS and PS groups were statistically significant. This was due to increased variability within the $CTRL_{Ext}$ group. Decreased variability generally indicates more stability (Stergiou & Decker, 2011). Still, too little variability also means the MBS and PS groups were more rigid in their movements and may be less able to adapt their movement patterns during the task (Hausdorff, 2007).

The current study cannot determine if having more knee extension or flexion moment at single limb support is more favourable, as they are simply different movement strategies to accomplish the same task. Moment generations are related to the strength and control requirements of walking down the ramp (Redfern & DiPasquale, 1997). Individuals adopted different strategies to be more efficient or stable (Hunter et al., 2010; Monsch et al., 2012). This could have been part of a general strategy to be more cautious or lessen the impact, thereby reducing pain (Yakhdani et al., 2010). These strategies do not necessarily imply conscious cognitive involvement, as many individuals do not know how they are adapting or why they are doing so (Dijksterhuis & Aarts, 2010). The larger slope of the ramp or not being allowed to hold onto the handrail during the task may have influenced their movement pattern during ramp descent.

This study was not without limitations. Initially, this study aimed to include 14 patients in both the MBS and PS groups. However, three individuals were excluded from the MBS group due to technical issues during their pre-operative motion capture session. The smaller sample size may have been underpowered, potentially leading to a type II error. However, the included sample size was similar to the previous TKA biomechanical studies on ramp descent tasks (Simon et al., 2018; Wen et al., 2022; Wiik et al., 2015). The smaller sample size also meant that the interaction between pre-operative walking pattern and implant type could not be evaluated, so there is potential that a particular implant may be suited for patients who walk with a specific gait pattern pre-operatively. The ramp inclination used was larger than what is within the building code limit. Differences do exist at smaller slopes (Wen et al., 2022), so there could be fewer differences between MBS and PS implants at smaller inclinations. Finally, this study only evaluated MBS and PS implants. Although similar findings were found in other studies comparing different implant types (Simon et al., 2018; Wiik et al., 2015), the results may not apply to all implant types.

4.6.6 Conclusions

In conclusion, this study demonstrated that after one-year post-surgery, all TKA patients descended a ramp with a cautious gait pattern, regardless of which implant they received or how they walked pre-operatively. Patients adopted this strategy to gain stability by increasing their step widths and having a knee extension moment throughout single limb support. The PS group had more knee joint stiffness, likely due to increased co-contraction, which resulted in less knee flexion angles and lower knee extension moments and powers than the MBS group.

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4.6.8 Appendix

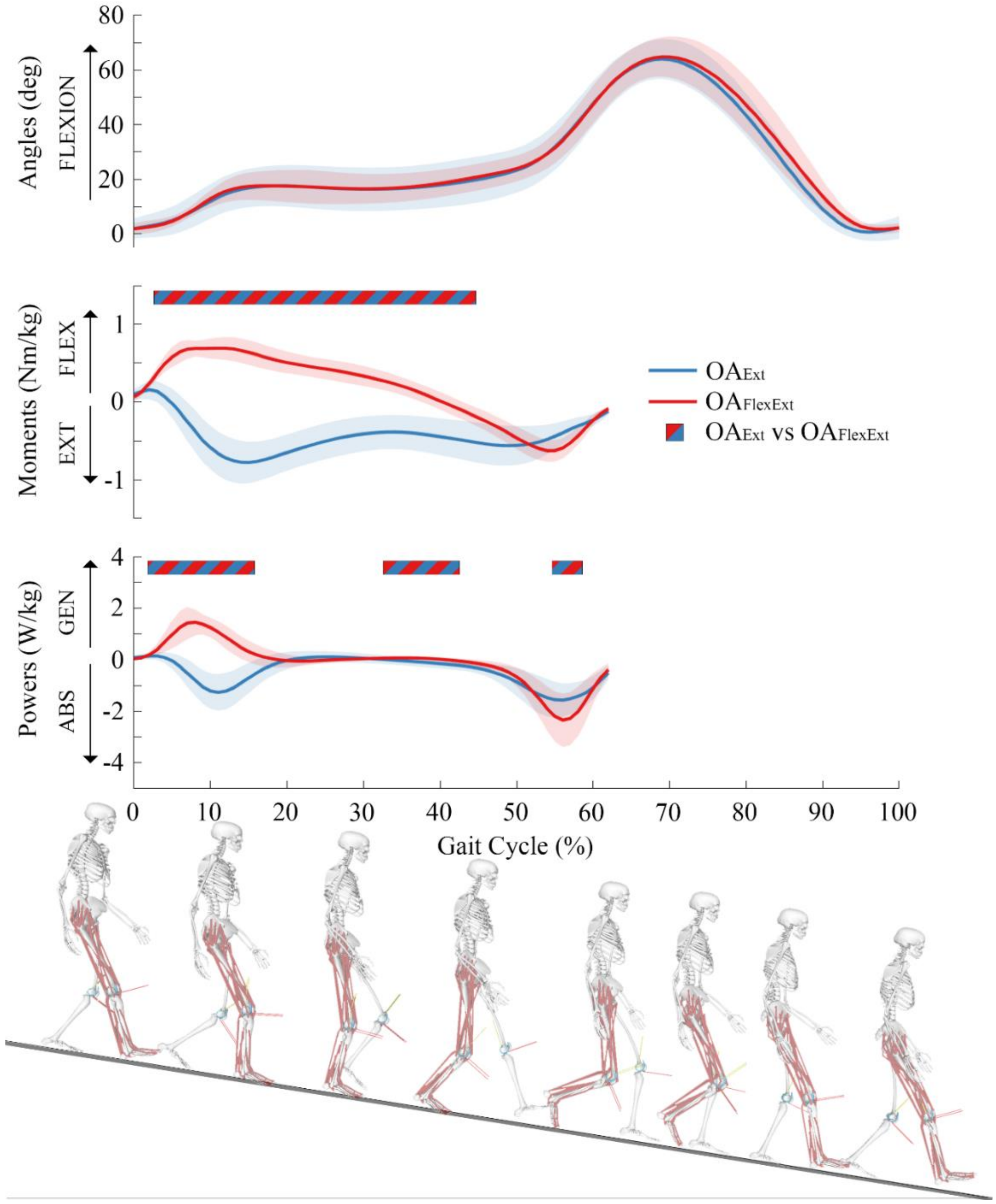

Figure 4.6.6: Group mean, standard deviations and statistical parametric mapping (SPM) results for knee angles, moments, and joint powers in the three different ramp descent gait patterns which existed pre-operatively. The filled areas in the horizontal bar graphs represent where significant differences were identified in the cycle using SPM. The joint angles (degrees) were normalized to 100% GC, whereas the joint moments (Nm/kg) and joint powers (W/kg) were normalized to 62% stance phase.

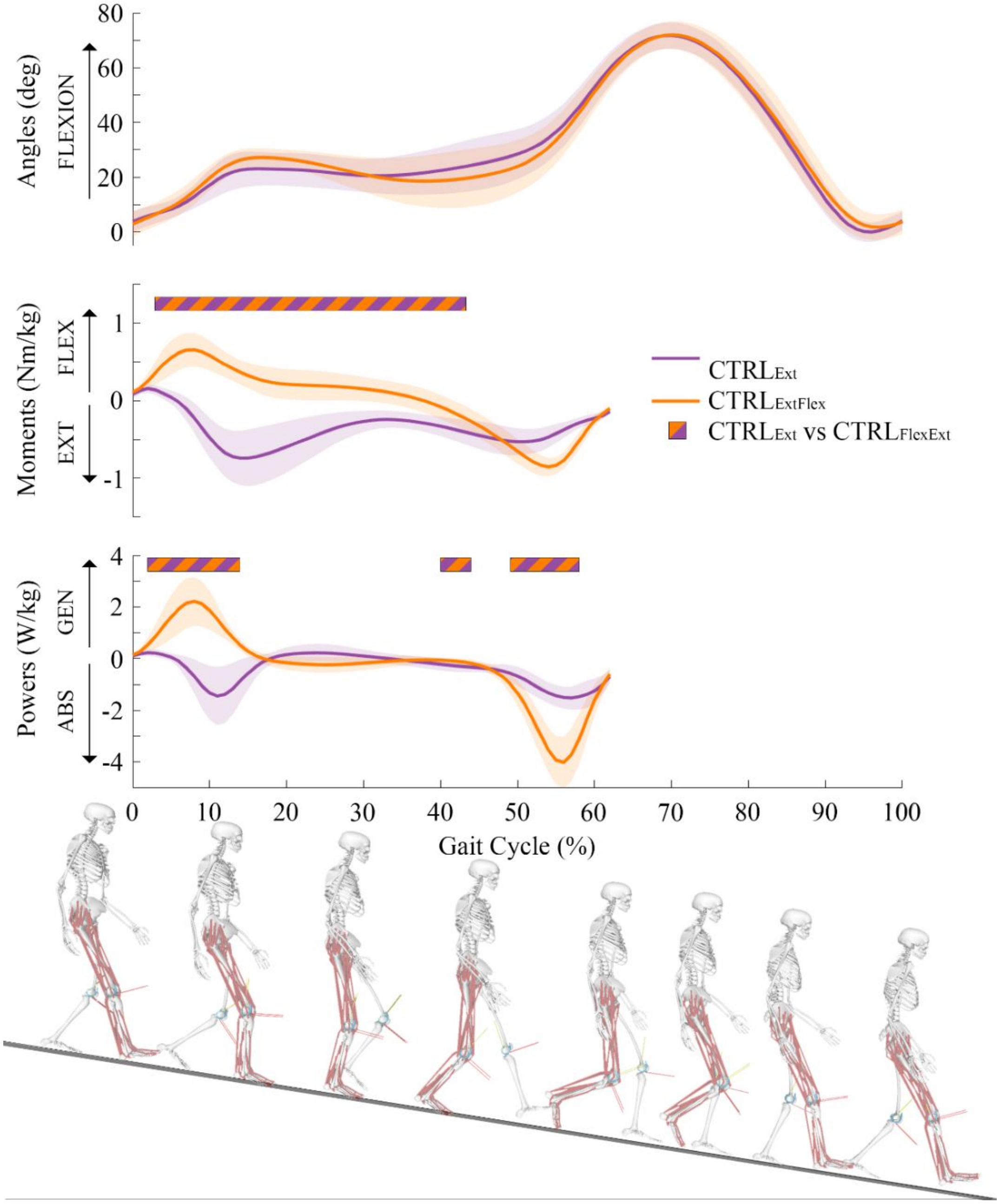

Figure 4.6.7: Group mean, standard deviations and statistical parametric mapping (SPM) results for knee angles, moments, and joint powers in the two different ramp descent gait patterns within the controls. The filled areas in the horizontal bar graphs represent where significant differences were identified in the cycle using SPM. The joint angles (degrees) were normalized to 100% GC, whereas the joint moments (Nm/kg) and joint powers (W/kg) were normalized to 62% stance phase.

Table 4.6.2: Pre-operative group mean (SD) demographic and KOOS subscale values.

	OA_{Ext}	$OA_{FlexExt}$	$CTRL_{Ext}$	$CTRL_{FlexExt}$
Number of participants (n)	18	7	7	7
Sex (female/male)	8/10	3/4	4/3	2/5
Received MBS implant (female/male)	4/4	1/2	N/A	N/A
Received PS implant (female/male)	4/6	2/2	N/A	N/A
Age (years)	62.7 (8.1)	65.4 (3.5)	62.2 (5.6)	66.5 (5.1)
Body Mass Index (kg/m^2)	28.3 (3.5)†	31.0 (4.0)*	23.9 (2.1)*†	25.9 (1.7)†
KOOS				
Symptoms	39.7 (14.8)	45.3 (14.7)	98.5 (2.8)*†	99.0 (2.7)*†
Pain	47.5 (17.4)	54.0 (12.3)	99.2 (2.1)*†	98.0 (4.2)*†
Function in daily living	60.1 (20.7)	57.8 (19.3)	100.0 (0.0)*†	100.0 (0.0)*†
Function in sport and recreation	25.8 (24.1)	22.9 (12.9)	100.0 (0.0)*†	100.0 (0.0)*†
Quality of Life	19.5 (13.9)	23.2 (12.4)	100.0 (0.0)*†	100.0 (0.0)*†

* significant ($p < .05$) difference from OA_{Ext}

† significant ($p < .05$) difference from $OA_{FlexExt}$

4.7 **Muscle activity and biomechanics while descending a staircase after total knee arthroplasty: A study comparing different posterior stabilized and medial congruent designs**

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Contents of this chapter prepared as an article for submission in *Journal of Arthroplasty*

Kowalski, E., Pelegrinelli, A.R.M., Ryan, N., Dervin, G., & Lamontagne, M. Muscle activity and biomechanics while descending a staircase after total knee arthroplasty: A study comparing different posterior stabilized and medial congruent designs. 2023. Submitted.

4.7.1 Abstract

Background

Many patients report more difficulty when descending stairs compared to level walking after total knee arthroplasty (TKA). Different implant designs can affect knee biomechanics and muscle activity during gait, but their effect during stair descent is unclear. The purpose of this study was to evaluate knee biomechanics and muscle activations of quadriceps, hamstrings, and gastrocnemius muscles during a stair descent task in patients who underwent TKA with either a posterior stabilized (PS) or medial ball-and-socket (MBS) implant and to compare them to a group of healthy controls.

Methods

Twenty-eight TKA patients were randomized to either an MBS (n=14) or PS (n=14) implant and were compared with 14 controls. Patients visited the biomechanics lab approximately 12 months after TKA, where knee biomechanics and muscle activity were measured as they descended a three-step staircase.

Results

Compared to the MBS and control groups, the PS group descended the stairs with reduced knee flexion angle and greater hamstring muscle activation throughout single limb support. Knee joint moments and power were similar between MBS and PS groups, but neither reached the level of the control group.

Conclusion

Lower knee flexion angles and increased hamstring muscle activity indicated that the PS group descended the stairs with a stiffer knee gait pattern than the MBS group. The MBS implant design may provide additional stability as patients required less muscle activity than the PS group. Increased coactivations may increase compressive forces and contribute to additional knee prosthesis wear.

4.7.2 Introduction

Some patients remain dissatisfied after total knee arthroplasty TKA and have functional limitations when performing various activities of daily living (Choi & Ra, 2016; Gunaratne et al., 2017). These difficulties often occur when descending a staircase (Collins, Misra, Felson, Crossley, & Roos, 2011). This task is more demanding than level walking as it generates larger forces at the knee joint (Lu & Lu, 2011; Pickle, Grabowski, Auyang, & Silverman, 2016) and requires patients to adjust their movement patterns to adapt to these larger loads (Catani et al., 2003; Standifird, Cates, & Zhang, 2014; Wilson et al., 1996). However, the neuromuscular adaptations after TKA during a stair descent task still need to be fully understood.

Several studies have evaluated stair descent biomechanics after TKA (Bolanos et al., 1998; Catani et al., 2003; Elkarif et al., 2021; Fenner, Behrend, & Kuster, 2017; Joglekar, Gioe, Yoon, & Schwartz, 2012; Kelman, Biden, Wyatt, Ritter, & Colwell Jr, 1989; Komaris et al., 2021; McClelland, Webster, & Feller, 2009; Saari, Tranberg, Zügner, Uvehammer, & Kärrholm, 2004; Standifird et al., 2014; Trinler et al., 2016; Wilson et al., 1996). Many of these studies have reported that TKA patients have less knee range of motion and lower knee extension moments than healthy controls (Catani et al., 2003; Standifird et al., 2014; Wilson et al., 1996).

However, the muscles are the major contributors to producing moments of force at the joint, and only a few studies have reported muscle activity (Bolanos et al., 1998; Catani et al., 2003; Elkarif et al., 2021; Kelman et al., 1989; Wilson et al., 1996) during stair descent. Muscle activation is typically measured using small sensors attached superficially to the skin to record the electrical activity produced by the underlying muscles (Stefano, Burridge, Yule, & Allen, 2004). Muscle onset and intensity are detected from the surface electromyography (sEMG) analysis (Lamontagne, Beaulieu, Varin, & Beaulé, 2009). Combining sEMG with biomechanical motion analysis identifies the main contributors to joint moment and can help understand whether a muscle is activating abnormally (Lamontagne et al., 2009).

Encountering a staircase is a daily reality for many individuals, yet our understanding of how TKA patients adapt to these tasks has yet to be thoroughly examined. Of the studies which did evaluate sEMG, several muscle adaptations have been identified and included prolonged muscle activation of the hamstrings (Catani et al., 2003; Kelman et al., 1989) and gastrocnemius muscles (Catani et al., 2003). Such an activation pattern suggests that TKA patients are descending stairs with increased co-contraction between the extensors (quadriceps) and flexors (hamstrings and gastrocnemii), which increases knee joint stiffness (McGinnis, Snyder-Mackler, Flowers, & Zeni, 2013). The primary limitations of previous studies were the number of muscles evaluated and which implant designs were used, making direct comparisons difficult. Several studies have compared different implant designs, and evidence suggests that different designs result in different muscle adaptations (Beach, Regazzola, Neri, Verheul, & Parker, 2019; Esposito, Freddolini, Marcucci, Latella, & Corvi, 2020; Simon, Della Valle, & Wimmer, 2018).

One aspect of implant design is the constraint, which refers to whether the posterior cruciate ligament (PCL) is sacrificed or preserved (Porteous & Curtis, 2021). Within cruciate sacrificing designs exist posterior stabilized (PS) and medial ball-and-socket (MBS) implants. These differ in design as the MBS implant has a highly congruent medial compartment and less conforming lateral compartment, designed to reduce the anterior sliding of the femoral component on the tibial component that patients with PS implants experience (Atzori, Salama, Sabatini, Mousa, & Khalefa, 2015). An MBS implant's more medial congruent design may provide additional stability during functional tasks and provide these patients with more efficient muscle activations. Previous studies which have compared MBS and PS implants have identified differences in biomechanics and muscle activity (Beach et al., 2019; Esposito et al., 2020; Simon et al., 2018). Still, to our knowledge, no study has compared these implant groups during a stair descent task. This study aimed to evaluate the magnitude and timing of muscle activations for quadriceps, hamstrings, and gastrocnemii muscles, as well as knee joint biomechanics during stair descent in patients one year after TKA with either an MBS or PS implant and compare them to a group of similarly aged healthy controls (CTRL).

4.7.3 Methods

Subjects

This prospective study recruited 86 individuals with severe knee OA(Kellgren & Lawrence, 1957) who were scheduled to undergo TKA by a single surgeon and were screened according to the inclusion/exclusion criteria outlined in Table 4.7.1. Thirty-two participants were eligible and underwent block randomization to receive either an MBS (MicroPort EVOLUTION® Medial Pivot System with Cruciate Sacrificing tibial inserts) or PS (Zimmer Biomet® NexGen® PS TKA system with PS inserts) implant. Twenty-eight patients who underwent TKA with either an MBS (female=6/male=8, age=63.7±5.7 years) or PS (female=6/male=8, age=65.6±8.1 years) were included in the final analysis and completed a pre-operative and a 12-month post-operative motion and muscle activity capture at the biomechanics lab (Figure 4.7.1). Although pre-operative assessments were completed, these results will focus on the post-operative outcomes. Fourteen similarly aged healthy participants were recruited from the community to form the CTRL group and completed the same motion and muscle activity protocol. The university and hospital ethics committees approved the study, all participants provided written informed consent, and the study was referenced on the clinical trials website: NCT02589197.

Table 4.7.1: Inclusion and exclusion criteria.

Inclusion criteria
Participants were between the ages of 45 and 75 at the time of enrollment
Participant was willing and able to complete the required study visits and assessments
Exclusion criteria
All Participants
Body mass index (BMI) and waist circumference measurements > 35 kg/m^2 and 102 cm, respectively, for men, and > 35 kg/m^2 and 88 cm, respectively, for women
Any past or present condition, which in the opinion of the investigators, may impact gait
Previous joint replacement of the enrolled knee or other lower limb joint replacement
Total Knee Arthroplasty Participants
Participant had a degenerative condition (other than osteoarthritis in the enrolled knee) impacting joints of the lower extremities
Healthy Controls
Participant had a degenerative condition impacting joints of the lower extremities

Surgery

All surgical procedures were performed by a single senior arthroplasty surgeon (GD) using a midline incision and subvastus approach. Manual instruments were used with the goals of mechanical neutral alignment with the femur first technique and the tibial component at a coronal neutral angle. The protocol required resurfacing of the patella, and the posterior cruciate ligament (PCL) was released in all patients. All components were cemented, and tourniquet use was restricted to the time of cementation and deflated before closure. No patients required additional soft tissue release, and there were no complications or revisions with the surgical cohort.

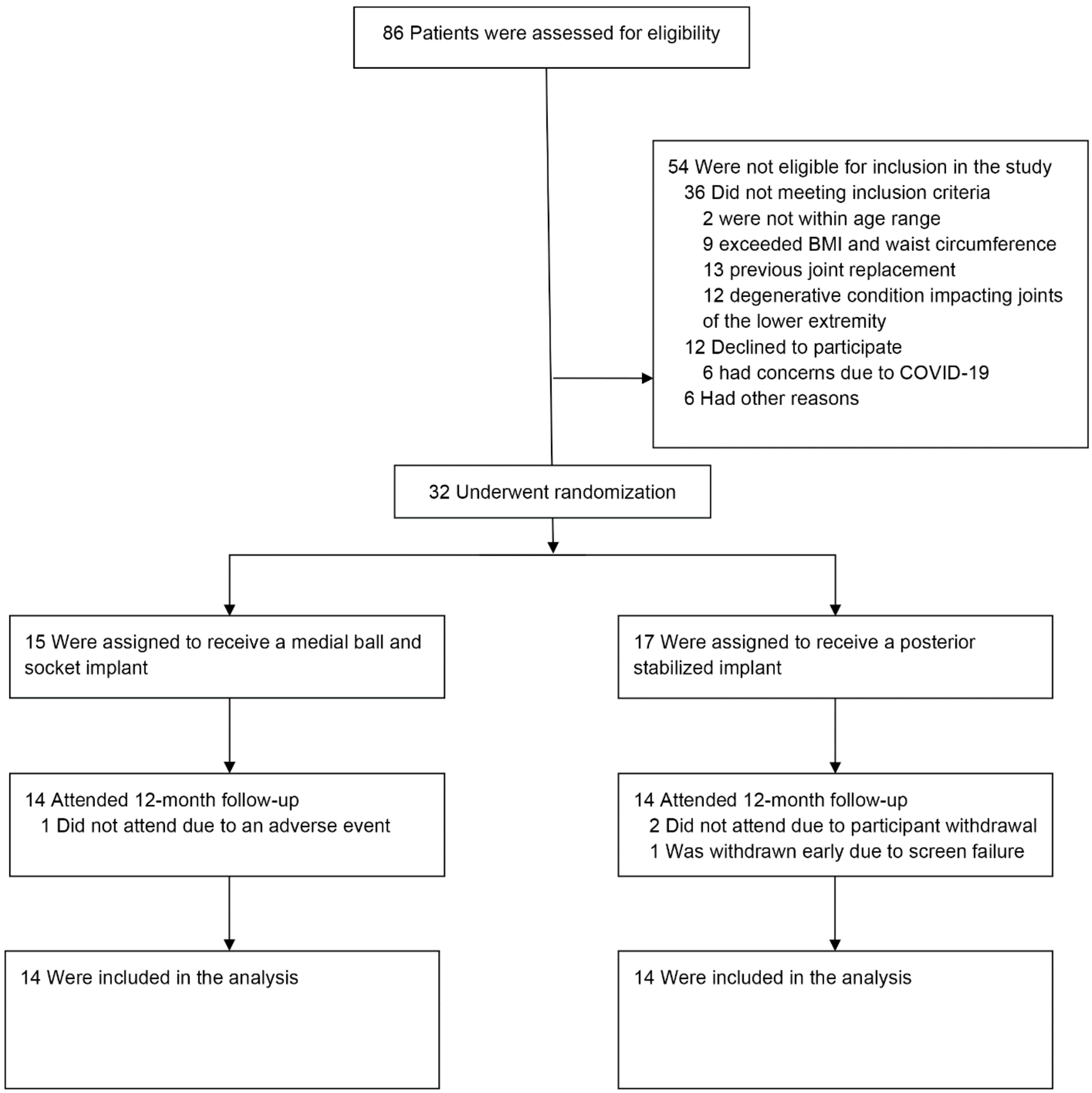

Figure 4.7.1: Consolidated Standards of Reporting Trials (CONSORT) flow diagram for enrolled patients.

Data collection

Sixteen sEMG electrodes (Trigno, Delsys, 2000 Hz, bandwidth 10-450 Hz, signal amplification to a gain of 1000) were placed bilaterally over the following muscles according to SENIAM guidelines (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000): vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), semitendinosus (ST), biceps femoris (BF), both medial (GM) and lateral (GL) heads of the gastrocnemius, and the tibialis anterior. Electrode placements were validated onscreen as participants contracted their muscles and were further secured with foam under-wrap and elastic athletic tape.

Participants completed a series of maximum voluntary isometric contractions (MVIC) to amplitude normalize the sEMG waveforms (Rutherford, Hubley-Kozey, & Stanish, 2011). Knee flexion and extension were conducted on an instrumented exercise chair, whereas ankle plantar flexion was performed on a Biodex in isometric mode according to the manufacturer's instructions. All MVIC tasks had the knee in 60° flexion. Three trials for each exercise were performed bilaterally. Each trial lasted 5s, and 30s rest was provided between each trial. Strength values for ankle plantar flexion and knee flexion and extension were determined by normalizing the force by the participant's body mass (Bazett-Jones, Cobb, Joshi, Cashin, & Earl, 2011).

Following MVICs, participants were outfitted with 45 passive reflective markers according to the University of Ottawa Motion Analysis Model (UOMAM) marker set (Mantovani & Lamontagne, 2017). Ten motion capture cameras (Vicon, Oxford Metric, 200 Hz) recorded patients as they completed a static trial and five trials descending a three-step staircase. The staircase (slope 22.6°, rise 18 cm, tread depths 43.2 cm, and a width of 91.5 cm) had force platforms placed on the first and second steps (model 9268BA, Kistler, 1000 Hz) and two embedded in the ground (FP4060, Bertec, 1000 Hz) to record ground reaction forces. Deeper tread depths were necessary to accommodate the width of the force platforms (60 x 40 cm).

Participants completed the five stair descent trials at their preferred pace, without holding onto the handrails and wearing their own comfortable footwear. The first step down was taken with the operated limb in the PS and MBS groups and the dominant limb in the control group (Kowalski, Catelli, & Lamontagne, 2019). The dominant limb was their preferred limb for kicking a ball (Chapman, Chapman, & Allen, 1987). All participants were provided enough practice trials to get comfortable with the protocol. The movement cycle was defined from foot placement onto the first step to the subsequent foot placement of the same limb (McFadyen & Winter, 1988).

Data processing

sEMG signals were processed according to ISEK guidelines("Standards for Reporting EMG Data," 2018). For each muscle, peak signals from the MVIC trials were used to amplitude normalize (%MVIC) the signals. Variables of interest were onset time, total time of activation, peak signal (peakEMG), total signal (iEMG), and co-contraction. Muscle onset and offset detections were completed using the procedure described by Roetenberg et al. (Roetenberg, Buurke, Veltink, Forner Cordero, & Hermens, 2003) which used a custom-written Matlab script. This method included the double threshold onset detection method with a Teager-Keiser Energy Operator, and onsets/offsets were verified with a visual inspection. The location of the muscle onset was normalized and referenced to its position in the movement cycle. Total activation duration was a percentage of the movement cycle and calculated based on the onset/offset times. PeakEMG was the maximum normalized signal (expressed as %MVIC), whereas iEMG (unitless value) represents the integrated sEMG signal. Both peakEMG and iEMG were calculated between the onset/offset time points when the muscles were activated. The muscle co-contraction index (CCI) was calculated using the equation developed by Rudolph et al. (Eq. 4.7.1) (Rudolph, Axe, Buchanan, Scholz, & Snyder-Mackler, 2001).

Eq. 4.7.1

$$CCI = \int \frac{EMG_S}{EMG_L}(EMG_S + EMG_L)$$

EMGs was the signal from the least active muscle, whereas EMGL was the signal from the most active muscle. The resulting curves were integrated to arrive at a single representative value of co-contraction between the two muscles and compared between the groups. The following CCIs muscle pairs were calculated about the medial and lateral sides of the knee joint: quadriceps-hamstrings (VM:ST and VL:BF) and the quadriceps-gastrocnemius (VM:GM and VL:GL) (Lewek, Rudolph, & Snyder-Mackler, 2004).

Motion capture data were filtered using a Woltring filter with a mean standard error of 15 mm, and force platform data were filtered using a 4th-order (zero lag) Butterworth filter with a cut-off frequency of 10 Hz. Gait trials were modelled (Mantovani & Lamontagne, 2017) and extracted data using a custom-written Matlab script (2019b, MathWorks). Knee variables of interest included sagittal joint angles, moments, and joint power. Knee angles and linear muscle envelopes of sEMG were normalized to a 100% movement cycle, whereas moments and powers were normalized to the stance phase (0 to 62% movement cycle) (Winter, 1984).

Statistical analysis

Discrete statistics were performed with SPSS (v27, IBM) using a One-Way ANOVA to determine differences between the MBS, PS, and CTRL groups. Knee biomechanics and linear muscle envelopes were compared across the entire stair descent movement cycle using statistical parametric mapping ANOVA (Pataky, Robinson, & Vanrenterghem, 2013). A Bonferroni correction was applied, and a p-value of 0.05 was used to evaluate group differences.

4.7.4 Results

Participant demographics, strength and KOOS subscale scores are presented in Table 4.7.2. The groups were similar in terms of age (p>.05), but the PS group (30.3±3.9 kg/m^2) had a larger (p<.001) BMI compared to the CTRL group(24.9±2.1 kg/m^2). After normalizing strength by body weight, the MBS group (.23±.07 BW) was stronger (p=.019) compared to the PS group (.15±.07 BW) during knee flexion. The KOOS could not identify differences between the MBS and PS groups for any subscale (p>.05), and both remained lower than the CTRL for most sub-scales.

Muscle onsets, total duration percentage on, and CCRs for the stair descent task are presented in (Table 4.7.3). The MBS group (71.6±19.9% stair descent cycle (%$_{SDC}$)) had earlier vastus lateralis onsets than the PS (84.8±20.6%$_{SDC}$, p=.011) and CTRL (89.5±3.8%$_{SDC}$, p<.001) groups. The MBS group (70.6±3.6%$_{SDC}$) also activated their vastus lateralis for a longer period than the PS (68.5±5.0%$_{SDC}$, p=.016) and CTRL (68.6±4.2%$_{SDC}$, p=.030) groups. The groups had no differences in onset times or total time on for the hamstring muscles. Medial quadriceps-hamstrings CCI was greater in the PS group (22.4±8.2) compared to the MBS (11.5±5.2, p<.001) and CTRL (12.1±5.5, p<.001). The PS group (26.1±19.2) had greater medial quadriceps-gastrocnemius CCI compared to the MBS (17.8±10.1, *p*=.019) and CTRL (18.2±5.7, *p*=.015) groups. Lateral quadriceps-hamstrings CCI was greater (*p*<.001) in the PS group (24.6±11.6) compared to the MBS group (17.8±8.8).

Table 4.7.2: Group mean (SD) demographics, strength and KOOS subscale values.

	MBS	**PS**	**CTRL**
Number of participants (n)	14	14	14
Sex (female/male)	6/8	6/8	6/8
Age (years)	63.7 (5.7)	65.6 (8.1)	64.4 (5.6)
Height (m)	1.72 (.09)	1.68 (.11)	1.67 (.08)
Mass (kg)	81.7 (14.5)	85.8 (13.5) †	71.5 (13.2)
Leg length (m)	.91 (.06)	.86 (.08)	.87 (.06)
Body mass index (kg/m^2)	27.4 (3.5)	30.3 (3.9) †	24.9 (2.1)
Months post-surgery	12.4 (0.5)	13.6 (2.2)	N/A
Strength			
Knee extension strength (BW)	.41 (.14)	.41 (.11)	.51 (.19)
Knee flexion strength (BW)	.23 (.07) *	.15 (.07) *	.17 (.06)
Ankle plantarflexion strength (BW)	.76 (.27)	.54 (.14)	.77 (.28)
Knee Osteoarthritis and Injury Score			
Symptoms	75.8 (20.4) †	74.2 (21.0) †	98.7 (2.6)
Pain	86.1 (12.3) †	85.7 (10.7) †	98.6 (3.2)
Activities of Daily Living	91.3 (10.2) †	93.4 (7.1)	100.0 (0.0)
Sport & Recreation	69.6 (24.0) †	60.4 (19.8) †	100.0 (0.0)
Quality of Life	67.4 (20.1) †	71.0 (17.8) †	100.0 (0.0)

BW – normalized to body mass; * significant MBS vs PS difference; † significant difference from CTRL

PeakEMG for all muscles during the stair descent task is presented in Figure 4.7.2. The MBS group (43.3±51.1%$_{MVIC}$) had greater peakEMG for the rectus femoris than the PS (28.2±12.7%$_{MVIC}$, *p*=.032) and CTRL (26.0±9.9%$_{MVIC}$, *p*=.012) groups. The PS group (47.7±21.7%$_{MVIC}$) had greater peakEMG for the biceps femoris than the MBS (31.5±14.4%$_{MVIC}$, *p*<.001) and CTRL (37.0±14.8%$_{MVIC}$), *p*=.010) groups. PeakEMG for the semitendinosus was also greater in the PS group (32.0±10.7%$_{MVIC}$) than the MBS (23.6±12.8%$_{MVIC}$, *p*=.002) and CTRL (19.6±9.0%$_{MVIC}$, *p*<.001) groups.

Table 4.7.3: Group means (SD) for muscle onsets and total time on during a stair descent task.

	MBS	**PS**	**CTRL**
Muscle Onsets (% location within the Stair Descent Cycle)			
Vastus Lateralis	76.1 (19.9) *†	84.8 (20.6) *	89.5 (3.8)
Vastus Medialis	83.5 (22.6)	85.4 (18.9)	86.3 (12.7)
Rectus Femoris	80.8 (23.1) †	87.3 (23.7)	90.5 (9.2)
Biceps Femoris	88.5 (18.0)	85.8 (28.0)	80.8 (15.0)
Semitendinosus	72.5 (16.4)	74.8 (27.1)	73.1 (13.0)
Lateral Gastrocnemius	74.8 (19.5)	81.9 (21.2)	77.4 (12.9)
Medial Gastrocnemius	74.3 (16.5) *	87.1 (18.6) *†	76.2 (10.5)
Total Time On (% of entire Stair Descent Cycle)			
Vastus Lateralis	70.6 (3.6) *†	68.5 (5.0) *	68.6 (4.2)
Vastus Medialis	67.4 (12.7)	66.3 (5.4)	68.9 (4.7)
Rectus Femoris	73.9 (9.8) *	68.3 (11.5) *†	73.1 (8.9)
Biceps Femoris	61.6 (21.2)	55.8 (17.6)	61.9 (17.7)
Semitendinosus	66.3 (18.9)	65.8 (17.8)	65.7 (17.9)
Lateral Gastrocnemius	67.0 (14.1) †	62.8 (17.7) †	76.0 (13.0)
Medial Gastrocnemius	52.0 (17.1)	49.5 (14.5) †	58.6 (16.4)
Co-contraction Ratio			
Quadriceps-Hamstrings			
Medial (VM:ST)	11.5 (5.2) *	22.4 (8.2) *†	12.1 (5.5)
Lateral (VL:BF)	17.8 (8.8) *	24.6 (11.6)	20.2 (6.2)
Quadriceps-Gastrocnemii			
Medial (VM:GM)	17.8 (10.9) *	26.1 (19.2) *†	18.2 (5.7)
Lateral (VL:GL)	21.8 (12.6)	25.4 (13.2)	23.7 (11.7)

* significant MBS vs PS difference; † significant difference from CTRL; VM – vastus mediali; VL – vastus lateralis; ST – semitendinosus; BF – biceps femoris; GM – medial head of gastrocnemius; GL – lateral head of gastrocnemius.

iEMG for all muscles during the stair descent task is presented in Figure 4.7.3. The PS group (438.8±321.4) had greater iEMG for the biceps femoris than the MBS (253.7±125.8, *p*<.001) and CTRL (295.9±133.6, *p*=.005) groups. iEMG for the semitendinosus was also greater in the PS group (320.6±185.3) than the MBS (174.8±93.7, *p*<.001) and CTRL (154.8±89.1,

p<.001) groups. Additionally, iEMG for the medial gastrocnemius was greater in the PS group (555.3±255.4) compared to the MBS (423.0± 275.7, p=.035) and CTRL (439.1±206.9, p=.049) groups.

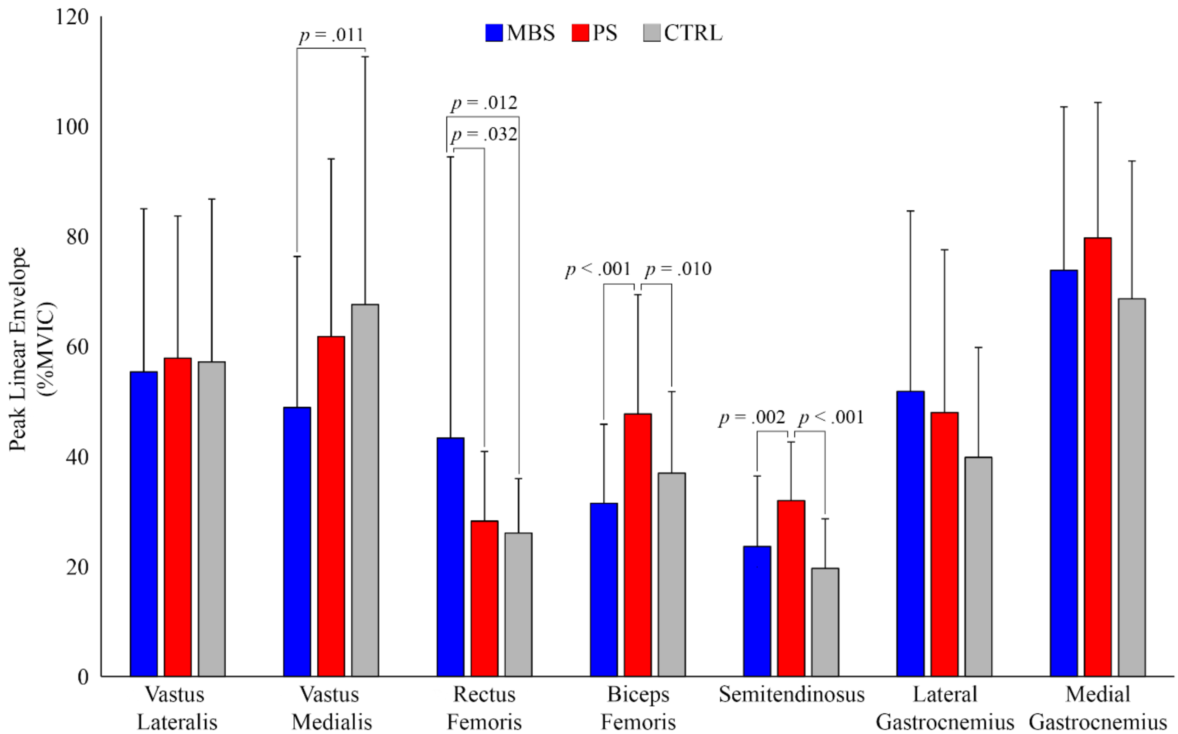

Figure 4.7.2: Group mean and standard deviations (SD) peak muscle activity (peakEMG) for each muscle during a stair descent cycle.

Average muscle activities for the quadriceps, hamstrings, and gastrocnemius muscle groups during the stair descent task are compared in Figure 4.7.4. For individual muscles, see (Appendix – Table 4.7.4). No differences in average quadriceps or gastrocnemius muscle activity existed between the groups. The PS group had increased average hamstrings (biceps femoris and semitendinosus) activation throughout forward continuance compared to the MBS (8 to 51% stair descent cycle ($\%_{SDC}$)) and CTRL group (13 to 44$\%_{SDC}$). The MBS group had less hamstrings activation throughout leg pull-through than the PS (58 to 84%$_{SDC}$) and CTRL (62 to 78%$_{SDC}$) groups.

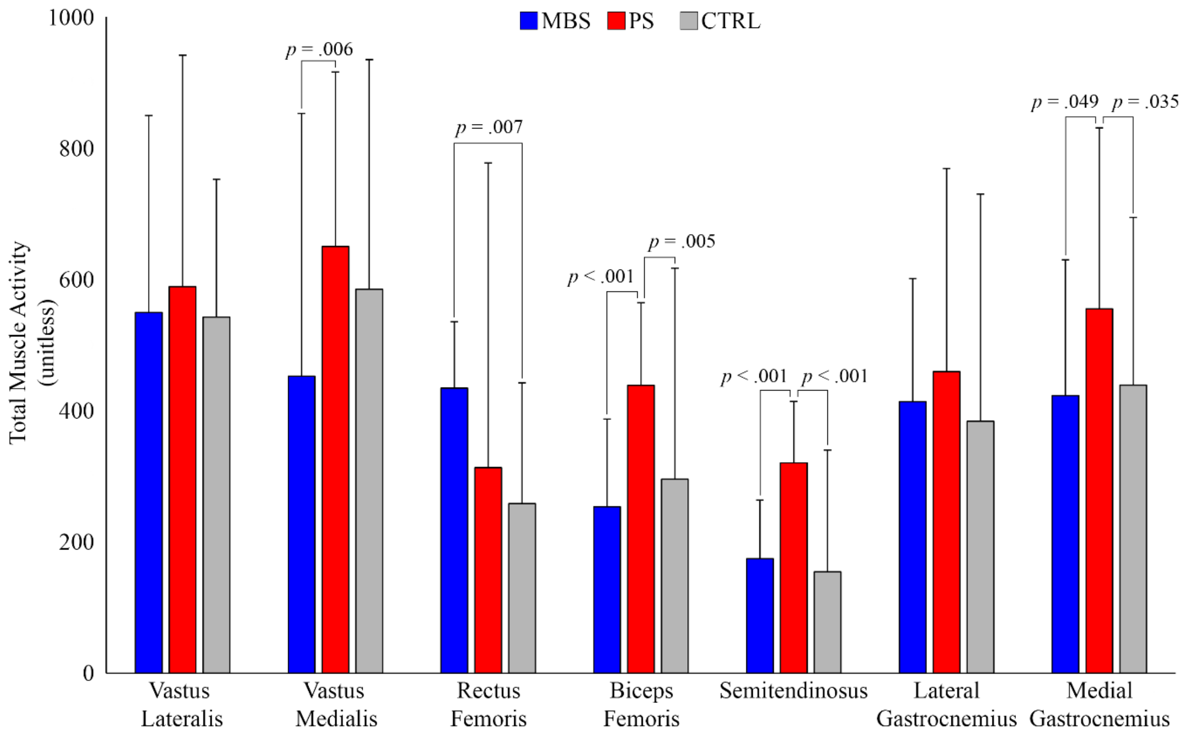

Figure 4.7.3: Group mean and standard deviations (SD) peak muscle activity (peakEMG) for each muscle during a stair descent cycle.

Knee joint angles, moments, and powers are also displayed in Figure 4.7.4. MBS and PS groups had less knee flexion angle than the CTRL group throughout the weight acceptance and forward continuance phases. However, the MBS group did achieve more knee flexion than the PS group during this period (0 to 31%$_{SDC}$). Knee extension moments were lower in the MBS (9 to 21%SDC) and PS (9 to 26%SDC) groups than in the CTRL group. Knee power absorption and generation did not reach the level of the CTRL group for either the MBS or PS groups throughout the weight acceptance and forward continuance phases.

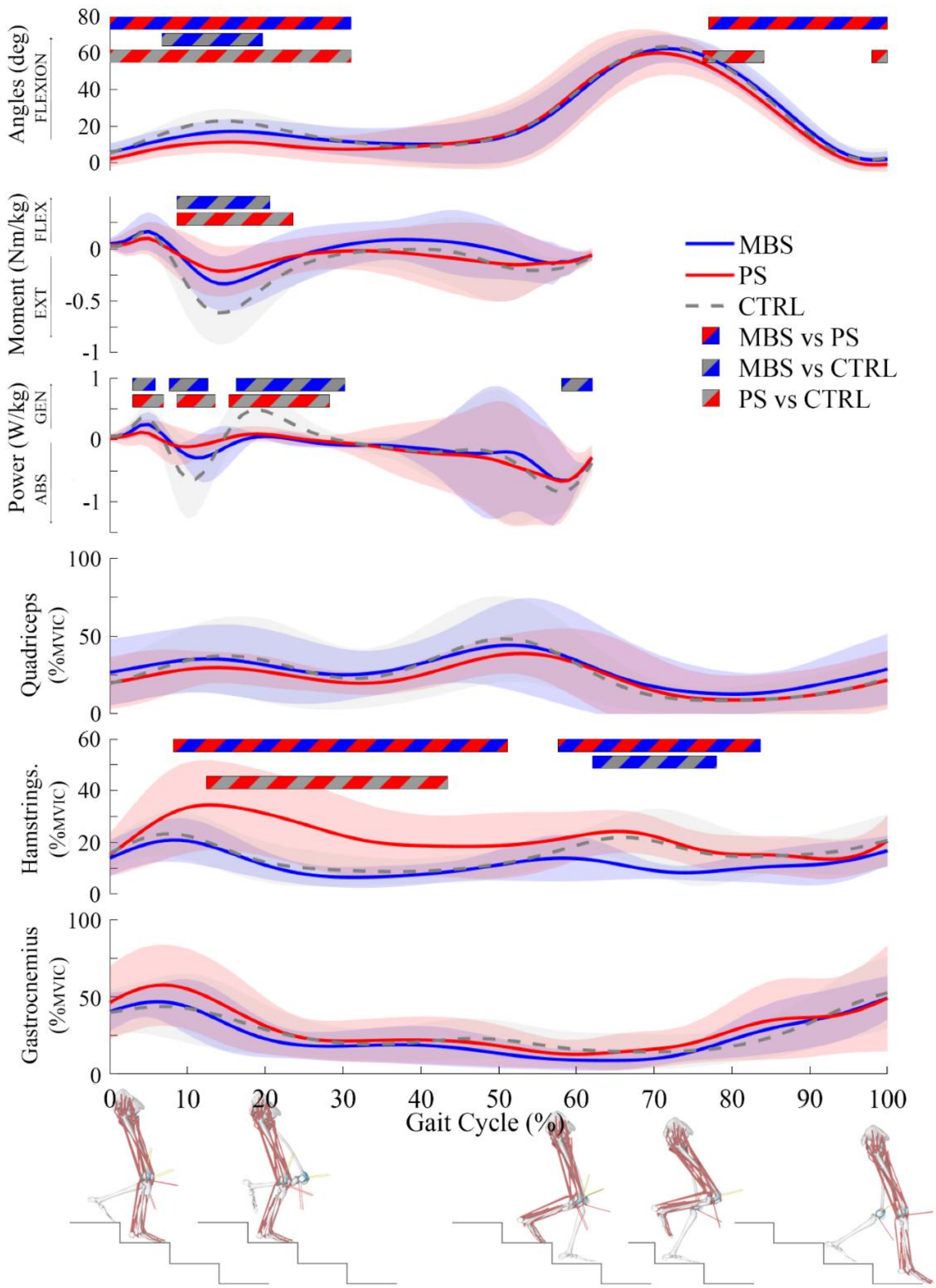

Figure 4.7.4: Group mean, and standard deviations (SD) for knee biomechanics and average linear envelopes of the quadriceps, hamstrings, and gastrocnemius muscles normalized over one stride during a stair descent task. The quadriceps signal was the average between rectus femoris, vastus medialis, and vastus lateralis; the hamstrings signal was the average between semitendinosus and biceps femoris; gastrocnemius was the average between the medial and lateral heads of the gastrocnemius. Shaded horizontal bars correspond to where in the stair cycle significant group differences occurred.

4.7.5 Discussion

One year after TKA, patients walked down the staircase with altered knee biomechanics compared to controls, regardless of whether they received a PS or MBS implant (Figure 4.7.4). Both groups had lower knee extension moments and powers when in single-limb support. However, the MBS group achieved closer knee flexion angles to the control than the PS, whereas the PS group had less knee flexion than both MBS and controls. These decreased knee flexion angles in the PS group were accompanied by increased hamstring muscle activity, which resulted in greater CCI. These findings suggest that the PS group descended the staircase with greater knee stiffness and co-contraction (McGinnis et al., 2013).

Although the MBS group had greater rectus femoris activity, the significance of this finding needs to be clarified. Superficial muscles, such as the rectus femoris, are contaminated by crosstalk from muscles adjacent to and beneath the target muscle (Talib, Sundaraj, Lam, Hussain, & Ali, 2019). Protocols that utilized indwelling and surface EMG electrodes to measure quadriceps muscle activity identified that the rectus femoris sEMG signal was contaminated due to crosstalk from the vastus intermedius muscle (Nene, Byrne, & Hermens, 2004). Vastus medialis peakEMG (Figure 4.7.2) and iEMG (Figure 4.7.3) were lower in the MBS group than in the CTRL. The reduced vastus medialis activity was possibly compensated by increased rectus femoris activity, as no group differences in total quadriceps activity were identified when all quadriceps muscles were averaged together (Figure 4.7.4).

During stair descent, the knee joint accepts the body weight is shifted from the contralateral limb and absorbs impact loads resulting from the lowering of the body. In healthy knees, the peak of vertical GRF forces reached 1.26 body weight (BW), and anterior peak forces reached 0.41BW during single-limb support (Lu & Lu, 2011). The cruciate ligaments and hamstring muscles work together to balance this anterior shear force, reaching peak force values around 0.5BW in the hamstrings and anterior cruciate ligament (ACL) and around 0.3BW in the PCL (Lu & Lu, 2011). However, both MBS and PS implants lack the passive knee stability provided by the cruciate ligaments, so knee joint stability is maintained by the surrounding muscles and implant design. The increased hamstring activity adopted by the PS group suggests they would have greater hamstring muscle and knee compressive forces than the MBS group. Therefore, a future study involving musculoskeletal modelling to estimate these forces is warranted.

Although coactivation between the quadriceps and hamstrings is necessary to reduce shear and strain across the tibiofemoral joint (Stevens-Lapsley, Balter, Kohrt, & Eckhoff, 2010), excessive coactivation may increase compressive forces and cause additional wear of the knee prostheses (Benedetti et al., 2003). Radiographic studies have shown that PS implants generate more wear particles than MBS implants (Minoda et al., 2003) and are often attributed to their differences in designs (Macheras, Galanakos, Lepetsos, Anastasopoulos, & Papadakis, 2017). MBS implants have a deeper, highly conforming medial compartment with a less congruent lateral compartment (Schmidt, Komistek, Blaha, Penenberg, & Maloney, 2003), compared to a femoral cam which articulates with a tibial post in PS implants (Ranawat, Komistek, Rodriguez, Dennis, & Anderle, 2004). The more conforming design of the MBS implant appears to provide better stability as patients achieved more knee flexion with less muscle activity, which may reduce implant wear.

No differences were found between the PS and MBS groups when compared using the KOOS (Table 4.7.2). Other studies found no clinically significant differences between implant designs when using patient-reported outcome measures (Kahlenberg, Lyman, Joseph, Chiu, &

Padgett, 2019), strengthening the argument to move beyond these qualitative tools to identify performance differences between implant groups.

In addition to the biomechanical and muscle activation differences, the MBS group achieved greater knee flexion strength than the PS group (Table 4.7.2). This may be related to the increased CCI experienced in the PS group (Table 4.7.3), as some evidence suggests that increased coactivation can significantly influence the resultant torque generated (Billot, Duclay, Simoneau-Buessinger, Ballay, & Martin, 2014).

This study only evaluated a single PS and MBS implant type, so the findings may not apply to other implants. The sample size was relatively small but aligned with or exceeded previous studies which evaluated muscle activity in TKA populations (Bolanos et al., 1998; Catani et al., 2003; Elkarif et al., 2021; Kelman et al., 1989; Wilson et al., 1996). All TKA participants were provided eight publicly-funded post-operative physiotherapy sessions. However, some patients may have continued beyond this, and this information was not recorded. Additional targeted physiotherapy may be beneficial, especially for patients with elevated coactivations, as some evidence suggests neuromuscular re-education is possible to reduce coactivations (Preece, Jones, Brown, Cacciatore, & Jones, 2016), but more research is necessary.

4.7.6 Conclusions

Twelve months following TKA, the PS and MBS groups had lower KOOS scores than the CTRL group on most KOOS subscales. Neither the MBS nor PS group achieved the same knee extension moments or knee powers as the control group, representing less knee loading in the TKA groups. However, the PS group had less knee flexion angle than the MBS group, showing that they descended the stairs with a stiffer knee gait pattern. Muscle activations indicated that the PS group activated their hamstrings more than MBS and control groups throughout single limb support. The MBS implant design may provide additional stability as patients required less muscle activity except for the rectus femoris than the PS group. Increased coactivations may increase compressive forces and cause additional wear of the knee prostheses.

4.7.7 References

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4.7.8 Appendix

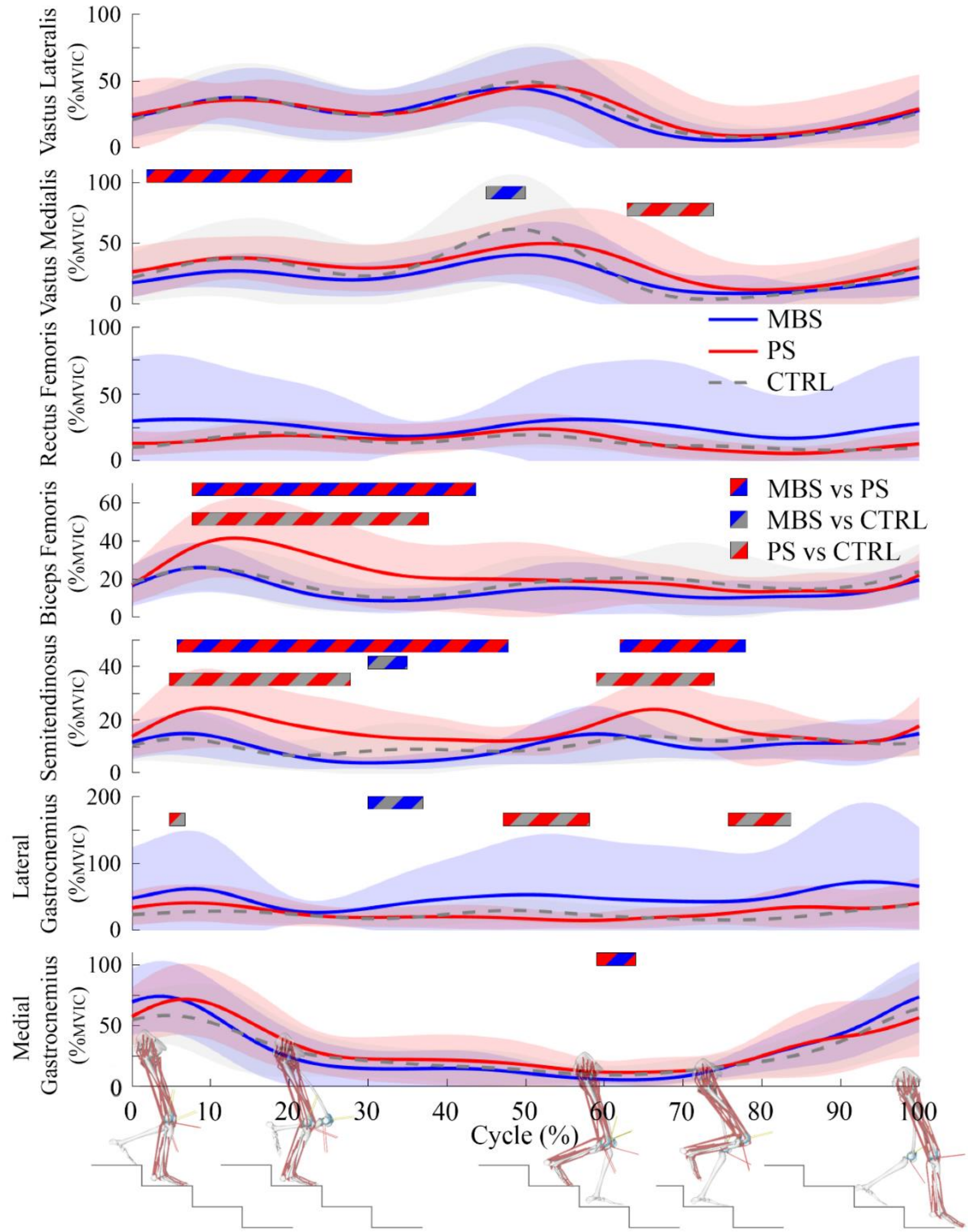

Figure 4.7.5: Group mean, and standard deviations (SD) for linear envelopes of all muscles normalized over one stride during a stair descent task. Shaded horizontal bars correspond to where in the stair cycle significant group differences occurred.

Table 4.7.4: Muscle activity parameters comparisons between PS, MBS, and CTRL measured during stair descent.

	PS	MBS	CTRL	ANOVA	PS vs MBS			PS vs CTRL			MBS vs CTRL		
				P value	Diff.	95% CI	*P* value	Diff.	95% CI	*P* value	Diff.	95% CI	*P* value
Muscle Onsets (%$_{SDC}$)													
VL	84.8 (20.6)	76.1 (19.9)	89.5 (3.8)	**<.001**	8.6	[1.6, 15.7]	**.011**	-4.7	[-12.3, 2.8]	.388	-13.4	[-20.8, -5.9]	**<.001**
VM	85.4 (18.9)	83.5 (22.6)	86.3 (12.7)	.685	1.9	[-6.3, 10.5]	1.000	-0.9	[-9.1, 7.3]	1.000	-2.8	[-10.6, 5.1]	1.000
RF	87.3 (23.7)	80.8 (23.1)	90.5 (9.2)	**.035**	6.5	[-2.6, 15.6]	.260	-3.2	[-12.2, 5.8]	1.000	-9.7	[-18.8, -0.6]	**.033**
BF	85.8 (28.0)	88.5 (18.0)	80.8 (15.0)	.132	-2.7	[-12.2, 6.9]	1.000	5.0	[-4.3, 14.4]	.576	7.7	[-1.7, 17.1]	.147
ST	74.8 (27.1)	72.5 (16.4)	73.1 (13.0)	.836	2.3	[-7.3, 12.0]	1.000	1.7	[-7.7, 11.1]	1.000	-0.6	[-9.5, 8.3]	1.000
GL	81.9 (21.2)	74.8 (19.5)	77.4 (12.9)	.106	7.1	[-1.1, 15.3]	.113	4.5	[-3.4, 12.4]	.511	-2.6	[-10.7, 5.5]	1.000
GM	87.1 (18.6)	74.3 (16.5)	76.2 (10.5)	**<.001**	12.8	[5.9, 19.7]	**<.001**	10.9	[4.4, 17.3]	**<.001**	-1.9	[-8.7, 4.8]	1.000
Total Duration (%$_{SDC}$)													
VL	68.5 (5.0)	70.6 (3.6)	68.6 (4.2)	**.008**	-2.1	[-3.9, -0.3]	**.016**	-0.1	[-1.9, 1.8]	1.000	2.0	[0.1, 3.9]	**.030**
VM	66.3 (5.4)	67.4 (12.7)	68.9 (4.7)	.253	-1.1	[-4.9, 2.7]	1.000	-2.6	[-6.4, 1.2]	.303	-1.5	[-5.1, 2.1]	.965
RF	68.3 (11.5)	73.9 (9.8)	73.1 (8.9)	**.008**	-5.6	[-10.3, -1.0]	**.012**	-4.8	[-9.4, -0.2]	**.038**	0.8	[-3.9, 5.5]	1.000
BF	55.8 (17.6)	61.6 (21.2)	61.9 (17.7)	.147	-5.8	[-14.3, 2.7]	.307	-6.1	[-14.4, 2.3]	.238	-0.3	[-8.7, 8.1]	1.000
ST	65.8 (17.8)	66.3 (18.9)	65.7 (17.9)	.982	-0.5	[-9.8, 8.7]	1.000	0.1	[-8.9, 9.2]	1.000	0.7	[-7.9, 9.2]	1.000
GL	62.8 (17.7)	67.0 (14.1)	76.0 (13.0)	**<.001**	-4.2	[-11.0, 2.6]	.411	-13.2	[-19.8, -6.6]	**<.001**	-9.0	[-15.7, -2.3]	**.004**
GM	49.5 (14.5)	52.0 (17.1)	58.6 (16.4)	**.004**	-2.6	[-9.7, 4.6]	1.000	-9.1	[-15.8, -2.4]	**.004**	-6.6	[-13.6, 0.5]	.075
PeakLE (%$_{MVIC}$)													
VL	57.9 (25.8)	55.4 (29.6)	57.2 (29.6)	.870	2.5	[-9.3, 14.3]	1.000	0.7	[-11.7, 13.1]	1.000	-1.8	[-14.0, 10.4]	1.000
VM	61.8 (32.3)	48.9 (27.5)	67.6 (45.1)	**.011**	12.9	[-2.9, 28.7]	.150	-5.8	[-22.0, 10.3]	1.000	-18.7	[-34.1, -3.4]	**.011**

RF	28.2 (12.7)	43.3 (51.1)	26.0 (9.9)	**.007**	-15.1	[-29.3, -1.0]	**.032**	2.2	[-11.7, 16.1]	1.000	17.3	[3.0, 31.6]	**.012**
BF	47.7 (21.7)	31.5 (14.4)	37.0 (14.8)	**<.001**	16.2	[8.2, 24.2]	**<.001**	10.8	[2.0, 19.5]	**.010**	-5.5	[-14.0, 3.0]	.365
ST	32.0 (10.7)	23.6 (12.8)	19.6 (9.0)	**<.001**	8.3	[2.7, 14.0]	**.002**	12.3	[6.4, 18.3]	**<.001**	4.0	[-1.5, 9.5]	.244
GL	48.0 (29.6)	51.8 (32.8)	39.8 (20.0)	.095	-3.8	[-18.6, 10.9]	1.000	8.2	[-4.4, 20.7]	.354	12.0	[-2.4, 26.3]	.134
GM	79.7 (24.6)	73.8 (29.7)	68.6 (25.1)	.104	5.9	[-7.7, 19.4]	.888	11.1	[-1.4, 23.6]	.101	5.2	[-8.0, 18.4]	1.000
iEMG													
VL	589.1 (210.1)	549.9 (352.6)	542.6 (300.0)	.633	39.2	[-83.5, 161.9]	1.000	46.6	[-82.1, 175.3]	1.000	7.3	[-119.2, 133.9]	1.000
VM	650.2 (349.8)	452.7 (266.0)	585.2 (400.4)	**.006**	197.5	[46.4, 348.6]	**.006**	65.0	[-89.5, 219.4]	.932	-132.5	[-279.4, 14.4]	.092
RF	313.3 (18.4)	434.4 (464.4)	258.5 (101.3)	**.007**	-121.1	[-256.5, 14.2]	.096	54.8	[-77.9, 187.5]	.957	175.9	[39.4. 312.5]	**.007**
BF	438.8 (321.4)	253.7 (125.8)	295.9 (133.6)	**<.001**	185.1	[86.3, 283.8]	**<.001**	142.9	[34.1, 251.6]	**.005**	-42.2	[-149.8, 65.4]	1.000
ST	320.6 (185.3)	174.8 (93.7)	154.8 (89.1)	**<.001**	145.8	[81.7, 209.9]	**<.001**	165.8	[98.4, 233.2]	**<.001**	20.0	[-42.4, 82.3]	1.000
GL	459.7 (346.2)	414.0 (309.3)	383.9 (187.5)	.378	45.8	[-108.6, 200.1]	1.000	75.9	[-55.7, 207.4]	.494	30.1	[-120.0, 180.2]	1.000
GM	555.3 (255.4)	423.0 (275.7)	439.1 (206.9)	**.017**	132.3	[6.7, 257.8]	**.035**	116.2	[0.5, 231.8]	**.049**	-16.1	[-139.1, 106.8]	1.000
CCI													
VMST	11.5 (5.2)	22.4 (8.2)	12.1 (5.5)	**<.001**	10.9	[7.2, 14.6]	**<.001**	0.6	[-2.7, 3.8]	1.000	10.3	[6.3, 14.3]	**<.001**
VLBF	17.8 (8.8)	24.6 (11.6)	20.2 (6.2)	**.001**	6.8	[2.4, 11.2]	**<.001**	2.4	[-2.7, 7.5]	0.783	4.4	[-0.8, 9.7]	.125
VMGM	17.8 (10.9)	26.1 (19.2)	18.2 (5.7)	**.007**	8.3	[1.1, 15.5]	**.019**	0.4	[-6.3, 7.0]	1.000	7.9	[1.2, 14.6]	**.015**
VLGL	21.8 (12.6)	25.4 (13.2)	23.7 (11.7)	.458	3.6	[-3.4, 10.6]	.641	1.9	[-4.9, 8.7]	1.000	1.7	[-4.4, 7.8]	1.000

4.8 Medial and lateral knee contact forces and muscle forces during sit-to-stand in patients one year after unilateral total knee arthroplasty

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Contents of this chapter prepared as an article for submission in *Clinical Biomechanics*

Kowalski, E., Pelegrinelli, A.R.M., Catelli, D.S., Dervin, G., & Lamontagne, M. Medial and lateral knee contact forces and muscle forces during sit-to-stand in patients one year after unilateral total knee arthroplasty. 2023. Not submitted.

4.8.1 Abstract

This study compared knee muscle forces and knee contact forces (KCF) during sit-to-stand in patients one year after unilateral total knee arthroplasty (TKA) with either a medial ball-and-socket (MBS) or posterior stabilized (PS) implant and compared them to a group of similarly healthy aged controls (CTRL). A musculoskeletal model and static optimization estimated lower limb kinematics, knee kinetics, muscle forces, and KCFs. The normalized sit-to-stand cycle was compared among the groups using statistical nonparametric mapping, and peak between-limb differences were compared using discrete statistics. The PS group required greater forward lean during the sit-to-stand task, causing greater spine flexion, posterior pelvic tilt, and decreased hip flexion on the operated limb. PS and MBS groups favoured their non-operated limb, resulting in less range of motion throughout the lower limb, lower knee kinetics, muscle forces, and KCFs on the operated limb. Compared to the controls, the MBS and PS groups had reduced medial compartment KCF. The control group did favour their dominant limb over their non-dominant limb. The consequence of MBS and PS patients favouring their non-operated limbs is not well understood. It may be related to the development of knee osteoarthritis on the non-operated limb. Post-operative rehabilitation should continue to promote greater use of the operated knee to have more symmetrical loading between operated and non-operated limbs and improve strength and mobility at the hip and ankle joints to reduce knee abduction.

4.8.2 Introduction

Total knee arthroplasty (TKA) reduces pain and improves the quality of life for individuals with severe knee osteoarthritis (OA) (Roos & Toksvig-Larsen, 2003). Despite the general success of TKA, 1 in 5 patients remain dissatisfied following surgery and continue with persistent knee pain or functional limitations in performing everyday activities of daily living

(ADLs) (Gunaratne et al., 2017; Noble, Conditt, Cook, & Mathis, 2006). Sixty percent of patients report no limitations while walking for two years post-surgery. However, more limitations exist during more demanding tasks, such as rising from a chair, as 49.4% and 7.8% of patients report mild or moderate limitations, respectively (Singh & Lewallen, 2014). These reported differences might be due to the different contact loads the knee experiences during various ADLS.

Studies which used instrumented knee implants to measure knee forces in vivo showed peak knee contact forces (KCF) between 2.2 and 3.5 body weights (BW) for most dynamic activities. While sit-to-stand and level walking have similar peak KCF (2.46-2.61 BW) (Moewis, Trepczynski, Bender, Duda, & Damm, 2022), the distribution of KCF between the medial and lateral components of the knee is not the same (Kutzner et al., 2017). During gait-type tasks (walking, climbing stairs, etc.), the medial KCFs are greater than lateral, whereas, during sit-to-stand tasks, the lateral KCFs are greater than medial KCFs (Kutzner et al., 2017; Moewis et al., 2022). Understanding how forces are transmitted through the knee after TKA is essential because, unlike biological tissues, the materials in a knee implant do not regenerate or remodel. The KCFs are directly implicated in wear and damage, especially to the polyethylene liner (D’Lima, Fregly, Patil, Steklov, & Colwell, 2012).

While KCF can be measured in vivo, they require costly, specialized, instrumented implants that limit their widespread use. Additionally, not all TKA implant designs have instrumented versions, which restrict our in vivo findings to cruciate retaining (D'Lima, Patil, Steklov, Slamin, & Colwell, 2006; D'Lima, Steklov, Patil, & Colwell, 2008) or ultra-congruent cruciate sacrificing designs (Kutzner et al., 2017; Kutzner et al., 2010; Moewis et al., 2022). Generic musculoskeletal (MSK) models have been developed to estimate KCFs and muscle forces during various ADLs. While early MSK models were developed and validated during gait (Delp et al., 2007), later models were developed and modified to accommodate a more extensive joint range of motions (Catelli, Wesseling, Jonkers, & Lamontagne, 2019), making them applicable to a more comprehensive range of tasks and populations (Bedo, Catelli, Lamontagne, & Santiago, 2020; Catelli et al., 2020).

One study evaluated patients with a posterior stabilized (PS) implant and identified that patients walked with a quadriceps avoidance pattern (Li et al., 2013). Compared to healthy controls, their vasti muscles contributed little to the knee extension moment during the stance phase of the gait cycle. Similar findings were identified during stair ascent as patients with a PS implant had similar KCF as controls but had different muscle force compensatory strategies during the loading and push-off phases. The reduced quadriceps force but greater knee flexor muscle forces ultimately resulted in the PS patients having similar KCFs as the controls (Rasnick, Standifird, Reinbolt, Cates, & Zhang, 2016). TKA patients had decreased quadriceps force during a stepping-down task compared to controls (Gaffney et al., 2016). During uphill walking, TKA patients primarily reduced total KCF on their operated limb by decreasing medial KCF (Thorsen, Wen, & Zhang, 2021). MSK models have also identified that patients load their non-operated knee more than their operated knee during uphill walking (Thorsen et al., 2021) and stationary cycling (Hummer et al., 2022). This compensation strategy causes higher loads to be sustained by the non-operated limb and may have long-term consequences (Hummer et al., 2022; Shakoor, Block, Shott, & Case, 2002; Thorsen et al., 2021).

To the best of our knowledge, no study compared KCF or muscle forces in patients who underwent TKA with a medial ball-and-socket (MBS) implant. Whereas PS implants cause paradoxical tibiofemoral motion (Schmidt, Komistek, Blaha, Penenberg, & Maloney, 2003), MBS implants were designed to reduce this paradoxical motion by having a single radius curvature femoral component with high conformity in the medial compartment about which it rotates (Atzori, Salama, Sabatini, Mousa, & Khalefa, 2015). The MBS design may benefit patients by changing how forces are transmitted through the knee. One study which compared peak knee contact pressures during a simulated squat with ultra-congruent and MBS implants found that MBS implants had lower peak pressure values (Putame et al., 2022). This study did show that the MBS achieved a lower peak pressure by distributing the force over a larger area compared to the ultra-congruent implant. However, it was done with a simulated squat that did not implement a patient’s motion capture data and provided no information regarding the muscle forces.

Additionally, to the best of our knowledge, no study estimated KCF and muscle forces during a sit-to-stand task in TKA patients to date. Sitting and standing are everyday tasks performed approximately 60 times per day by healthy adults (Dall & Kerr, 2010) and produce greater forces at the knee than walking (Kutzner et al., 2010; Mündermann, Dyrby, D'Lima, Colwell Jr., & Andriacchi, 2008; Zhao, Banks, D'Lima, Colwell Jr., & Fregly, 2007). It is also one of the ADLs with in vivo data, which creates greater peak KCF on the lateral component of the knee (Kutzner et al., 2017; Moewis et al., 2022), which may create difficulties in patients after TKA. Sit-to-stand also requires bilateral support in which both feet are in contact with the ground. If patients compensate with their non-operated limb, the task may be sensitive enough to evaluate movement asymmetries (Abujaber, Marmon, Pozzi, Rubano, & Zeni, 2015). This study aimed to compare knee kinematics, kinetics, muscle forces, and KCF on the operated and non-operated limbs during sit-to-stand in patients after TKA with either an MBS or PS implant and compare them to a group of similarly healthy aged controls (CTRL).

4.8.3 Methods

Participants

This study initially screened 86 individuals with end-stage knee OA (Kellgren and Lawrence, grade 4 (Kellgren & Lawrence, 1957)) who were scheduled to undergo TKA by a single orthopedic surgeon. Eligible participants were between the ages of 45 and 75 at enrollment and needed to be willing to complete the required study visits. To minimize soft tissues artefact, participants were excluded if they had a body mass index (BMI) and waist circumference measurement > 35 kg/m^2 and 102 cm respectively for men and > 35 kg/m^2 and 88 cm respectively for women; any past or present condition, which in the opinion of the investigators may impact gait; or previous joint replacement of the enrolled knee or other lower limb joint replacement. TKA participants were excluded if they had a degenerative condition (other than OA in the enrolled knee) impacting joints of the lower extremities. TKA participants were recruited from the surgeon’s practice, and the university and hospital ethics committees approved the study protocol. All participants included in this study provided written consent. The study was conducted by the principles of good clinical practice and the Declaration of Helsinki, and it is referenced in the clinical trials website: NCT02589197.

Thirty-two eligible participants underwent randomization to receive an MBS (MicroPort EVOLUTION® Medial Pivot System with cruciate sacrificing tibial inserts) or PS (Zimmer Biomet® NexGen® PS TKA system with PS inserts) TKA. Twenty-eight patients completed pre- and post-operative biomechanics lab visits, and this study only evaluated the postoperative time point. Several participants had to be excluded from the final analysis due to problems during the MSK framework's scaling process. These issues included problems with pelvis marker locations or pelvis with large anterior tilt. This caused muscles to cross the bones and resulted in poor muscle activations or joint contact loads. Eighteen patients were included in the final analysis (Appendix – Figure 4.8.3). Although the initial protocol had TKA patients complete the motion analysis protocol 12 ± 1 month after surgery, three PS patients were evaluated beyond the 13 months due to Covid-19 pandemic-related closures (two at 14 months, one at 20 months). These three participants were included in the final study cohort. The CTRL group had the same inclusion criteria and was excluded if they had any degenerative condition impacting joints of the lower extremities. All participants completed the Knee injury and Osteoarthritis Outcome Score (KOOS) questionnaire at the biomechanics lab visit.

All surgeries were performed by the same senior arthroplasty surgeon (GD). A midline incision and subvastus approach was performed for all patients (Hofmann, Plaster, & Murdock, 1991). Manual instruments were used with the goal of mechanical neutral alignment with the femur-first technique and the goal of the tibial component at a coronal neutral angle. The protocol required for resurfacing the patella and the posterior cruciate ligament was released in all patients. All components were cemented, and tourniquet use was restricted only to the time of cementation and deflated before closure. No patients required additional soft tissue release, and there were no complications or revisions within the surgical cohort. Following surgery, patients received eight publicly funded physiotherapy sessions. However, some participants may have continued with additional physiotherapy, and this information was not recorded.

Motion analysis

Participants warmed up on a cycle ergometer for five minutes and were invited to perform uninstructed stretching. They were outfitted with 45 retroreflective markers placed according to the University of Ottawa Motion Analysis Model marker set (Mantovani & Lamontagne, 2017). A height-adjustable bench was set for each participant to the height of their tibial tuberosity. Five sit-to-stand trials were performed at a self-selected pace, with feet positioned parallel at hip-width apart and the arms stretched out anteriorly since the beginning of execution. Marker trajectories were captured using a 10-camera infrared system sampled at 200 Hz (Vero, Vicon, Oxford Metrics), and ground reaction forces (GRFs) were captured using two embedded force platforms sampled at 1000 Hz (model FP4060, Bertec Corporation), with one foot on each force platform. Motion capture data were first labelled and then filtered using a Woltring filter with a mean, standard error of 15 mm. Force platform data was filtered using a 4th order (zero lag) Butterworth filter with a cut-off frequency of 10 Hz using Nexus 2.11.0 (Vicon). The start of the sit-to-stand cycle was defined by the first anterior displacement of the shoulder markers; the end was at maximum hip extension (Savelberg, Fastenau, Willems, & Meijer, 2007). All variables were time normalized to 101 points from the start to the end of each task. Kinematic variables of interest included sagittal spine, pelvis, hip, knee, and ankle angles. The included kinetic variables were sagittal and frontal knee angles and moments and knee joint powers.

Musculoskeletal modelling

A full-body generic musculoskeletal model (Bedo et al., 2020) was combined with updated passive muscle force curves and improved hip abductor paths (Uhlrich, Jackson, Seth, Kolesar, & Delp, 2022). The model included 80 lower-limb Hill-type muscle-tendon units (MTUs) with 37 degrees of freedom (Millard, Uchida, Seth, & Delp, 2013; Zajac, 1989) and 17 ideal torque actuators driving the upper body (Rajagopal et al., 2016). The modelled knee had flexion and extension rotations, and the anteroposterior and superior-inferior translations are described as a function of knee flexion, and frontal translation is locked (Lerner, DeMers, Delp, & Browning, 2015). The model allowed for estimating knee contact force for the medial and lateral compartments for tasks with high hip and knee flexion (Bedo et al., 2020; Lerner et al., 2015). The MSK model was used in an open-source musculoskeletal simulation software (OpenSim 3.3; Stanford University) (Delp et al., 2007).

Marker trajectories and GRF data sets were exported for the OpenSim format, and the models were scaled based on each patient's static anthropometric dimensions. The Batch OpenSim Processing Scripts (BOPS) Matlab toolbox was used to process inverse kinematics, inverse dynamics, muscle analysis, static optimization, and joint reaction analysis (Bedo et al., 2021). Static optimization was used to compute muscle forces, which minimized the sum of squared muscle activation (Buchanan & Shreeve, 1996; Todorov & Jordan, 2002). Joint reaction forces were computed using the resultant forces and moments activating on each articulating joint from all muscle forces and the internal and external loads applied to the model. Total vertical KCF was the sum of the medial and lateral compartment vertical contact forces. Muscle forces and KCF were normalized to body weight for each participant. The twelve MTUs which acted on the knee joint were selected for analysis. Each MTU was evaluated individually and included the four knee extensors (rectus femoris, vastus medialis, vastus intermedius, and vastus lateralis) and the eight knee flexors (biceps femoris long head (BFLH), biceps femoris short head, semitendinosus, semimembranosus, gastrocnemius medial head, gastrocnemius lateral head, gracilis, and sartorius.

Data Analysis

Five sit-to-stand trials were analyzed for the operated and non-operated limbs in the MBS and PS groups and dominant/non-dominant limbs in the CTRL group. The dominant limb was their preferred limb for kicking a ball (Chapman, Chapman, & Allen, 1987; Kowalski, Catelli, & Lamontagne, 2019). Statistical non-parametric mapping (SnPM) was performed to compare the kinematics, kinetics, muscle forces and KCF ($p \leq .05$) in the time-normalized (0% to 100%) sit-to-stand cycles in Matlab (Pataky, Vanrenterghem, & Robinson, 2015). The SnPM{t} representing the univariate pseudo-t-statistic was calculated at each waveform point; if it exceeded the critical threshold t, the difference between the groups was considered significant in that part.

A Wilcoxon signed-rank test was performed to compare between-limb differences. If significant between-limb differences existed within the CTRL group, an average of both limbs would be taken when comparing them to the MBS and PS groups (Kowalski et al., 2019). Mean values were compared using SnPM with Bonferroni corrections ($p \leq .05$). Demographics, isometric strength and KOOS were evaluated using a Kruskal-Wallis ANOVA followed by post hoc comparisons using Bonferroni corrections ($p \leq .05$) within SPSS (version 28, IBM).

4.8.4 Results

Mean demographics, isometric strength, and KOOS subscale values are presented in Table 4.8.1. No differences existed between the MBS or PS groups. The PS group (29.3±3.8 kg/m^2) had a larger BMI than the CTRL (25.1±3.0 kg/m^2) group (95% CI, 0.4 to 8.1 kg/m^2, p = .024). Knee extension, knee flexion, and ankle plantar flexion isometric strength was similar between all groups ($p > .05$). The operated limb (0.45±.012 BW) was weaker compared to the non-operated limb for the MBS group during knee extension (95% CI, .04 to .16 BW, $p = .013$). The MBS and PS group did not reach the level of CTRLs for any of the KOOS subscales ($p < .05$).

Table 4.8.1: Group means (SD) demographics, isometric strength, and knee injury and osteoarthritis outcome score (KOOS) subscale values.

	MBS	PS	CTRL
Number of participants (n)	9	9	9
Sex (female/male)	4/5	4/5	4/5
Age (years)	64.5 (6.5)	63.1 (8.6)	63.2 (4.4)
Height (m)	1.73 (0.10)	1.69 (0.12)	1.67 (0.09)
Mass (kg)	76.8 (12.6)	83.9 (12.6)	70.4 (14.8)
Body mass index (kg/m^2)	25.8 (2.7)	29.3 (3.8) †	25.1 (3.0)
Months post-surgery	12.6 (0.5)	13.7 (2.8)	---
Isometric Strength			
Operated Limb			
Knee extension strength (BW)	0.45 (0.12) ‡	0.43 (0.07)	0.56 (0.24)
Knee flexion strength (BW)	0.24 (0.08)	0.17 (0.09)	0.19 (0.07)
Ankle plantarflexion strength (BM)	0.80 (0.29)	0.55 (0.14)	0.84 (0.27)
Non-Operated Limb			
Knee extension strength (BW)	0.54 (0.13) ‡	0.46 (0.20)	0.58 (0.25)
Knee flexion strength (BW)	0.24 (0.08)	0.18 (0.10)	0.21 (0.08)
Ankle plantarflexion strength (BM)	0.80 (0.24)	0.49 (0.26)	0.79 (0.30)
Knee Osteoarthritis and Injury Score			
Symptoms	76.6 (14.1) †	66.7 (22.6) †	98.0 (3.5)
Pain	84.9 (11.2) †	81.2 (9.0) †	99.2 (2.1)
Activities of Daily Living	88.4 (13.5) †	90.9 (7.6) †	100.0 (0.0)
Sport & Recreation	75.0 (21.7)	55.6 (22.8) †	98.6 (2.4)
Quality of Life	68.1 (22.2) †	65.3 (15.6) †	98.2 (4.7)

BW – normalized to body mass; * significant MBS vs PS difference; † significant difference from CTRL; ‡ significant difference between operated and non-operated limbs.

Peak between-limb differences for kinematics, kinetics, muscle forces, and KCFs are presented in Table 4.8.2. MBS and PS groups favoured their non-operated limb as they had larger joint ranges of motion (ROM), peak joint moments, and power generation on their non-operated limb. Peak muscle forces for the knee extensor muscles were larger on the non-operated limb, along with medial, lateral, and total vertical KCF. For the CTRL group, sagittal joint angles were similar between dominant and non-dominant limbs. However, between-limb differences existed when comparing kinetics, knee extensor muscle forces, and lateral compartment KCF. Due to between-limb differences within the CTRL group, the average between both limbs was used for the remaining analyses compared to the MBS and PS groups.

Mean sagittal joint angles from the spine to the ankle are presented in Figure 4.8.1. The PS group performed the sit-to-stand movement with greater spine flexion. This resulted in greater spine flexion compared to the MBS from 0-55%, greater posterior pelvic tilt compared to the MBS group from 0-40%, less hip flexion than the MBS (0-53%) and CTRL (0-50%) groups, and ankle dorsiflexion than the MBS (0-55%) and CTRL (0-100%) groups. The MBS group had less spine flexion than the CTRL group throughout the movement (0-100%).

Mean knee joint angles, moments and powers are presented in Figure 4.8.2. Aside from a difference in knee flexion angle at the beginning of the movement (0-39%), no differences in knee joint angles, moments, or power existed between the MBS and PS groups. Compared to the CTRL group, the PS and MBS groups completed the sit-to-stand task with greater knee adduction angle (MBS vs CTRL: 0-59%; PS vs CTRL: 0-56%). The PS group had less knee extension moment than the CTRL group (46-58%). Knee power generation was lower in the PS group (45-67%) and MBS group (67-71%) compared to the CTRL group.

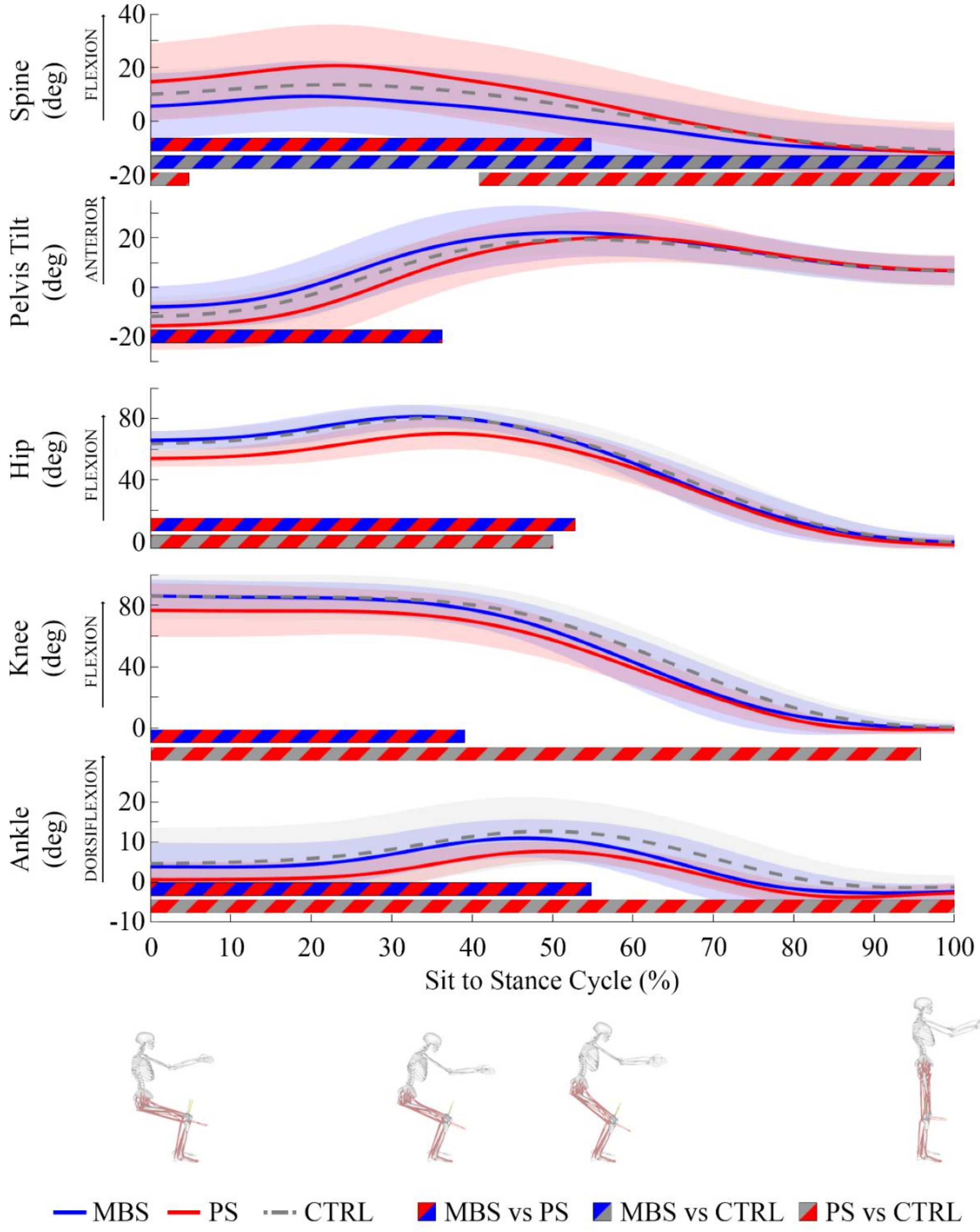

Figure 4.8.1: Sagittal plane joint angles for the spine, pelvis, hip, knee, and ankle during the sit-to-stand task for the operated limb in the MBS (blue) and PS (red) groups, and average between both limbs for the CTRL (grey) group. SnPM results are displayed below each figure and indicate significant ($p < .05$) differences between MBS and PS (red/blue), MBS and CTRL (blue/grey) and PS and CTRL (red/grey). MBS = medial ball-and-socket, PS = posterior stabilized, CTRL = control, SnPM = statistical non-parametric mapping, BW = body weight.

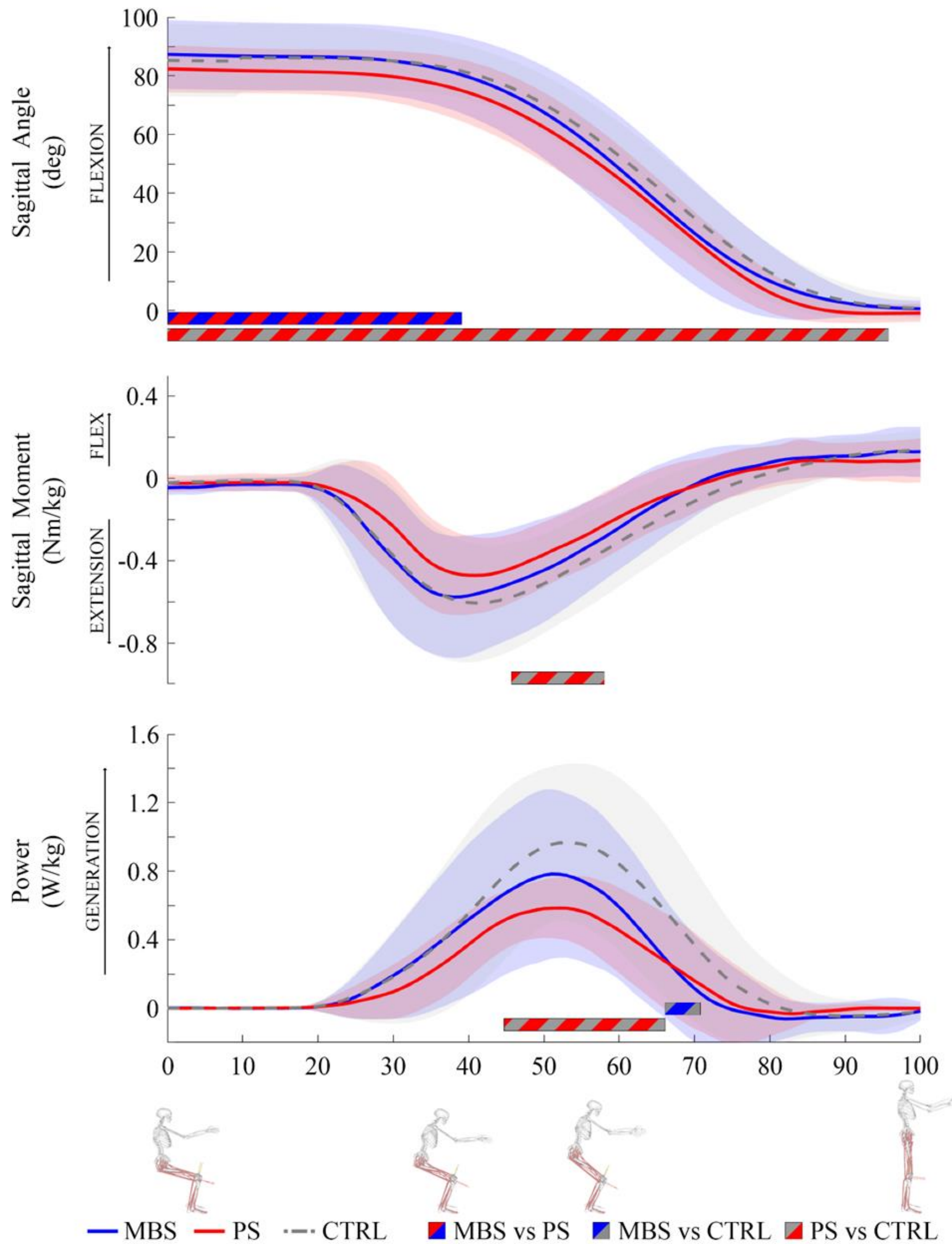

Figure 4.8.2: Knee joint angles, moments, and power during the sit-to-stand task for the operated limb in the MBS (blue) and PS (red) groups, and average between both limbs for the CTRL (grey) group. SnPM results are displayed below each figure and indicate significant ($p < .05$) differences between MBS and PS (red/blue), MBS and CTRL (blue/grey) and PS and CTRL (red/grey). MBS = medial ball-and-socket, PS = posterior stabilized, CTRL = control, SnPM = statistical non-parametric mapping, ABD = abduction, ADD = adduction, FLEX = flexion.

Knee muscle forces are presented in Figure 4.8.3. The vastus lateralis was the knee extensor muscle with the largest force, whereas the biceps femoris long head was the knee flexor with the largest force. No differences in the knee extensor muscles existed between any of the groups. The PS group had a significantly greater BFLH force than the CTRL group.

Vertical KCF for the medial and lateral compartments and total vertical KCF are presented in Figure 4.8.4. The PS group had less medial (17-27%) and total (20-22%) vertical KCF than the MBS group. The MBS and PS groups had similar total and lateral vertical KCF as the CTRL group. However, the MBS (43-62%) and PS (43-59%) had lower medial KCF than the CTRL group.

4.8.5 Discussion

To our knowledge, this is the first study to compare knee muscle and contact forces estimates using MSK modelling in patients who have undergone TKA with either an MBS or PS implant during a sit-to-stand task. Additionally, this study compared knee muscle and contact forces using a modified MSK model in patients who have undergone TKA with either an MBS or PS implant during a sit-to-stand task. Additionally, this study compared KCF for the medial and lateral compartments separately and evaluated both the operated and non-operated limbs.

MBS and PS participants favoured their non-operated limbs when performing the sit-to-stand task. Greater peak hip flexion, knee flexion, and ankle dorsiflexion angles were observed on the non-operated limb, as well as larger knee extension moments and power generation (Table 4.8.2). Greater knee extension moments and power generation was achieved by larger peak muscle forces produced by the vasti muscles on the non-operated limb ($p<.001$) and made greater total vertical, medial and lateral compartment KCFs (Table 4.8.2). The CTRL group had

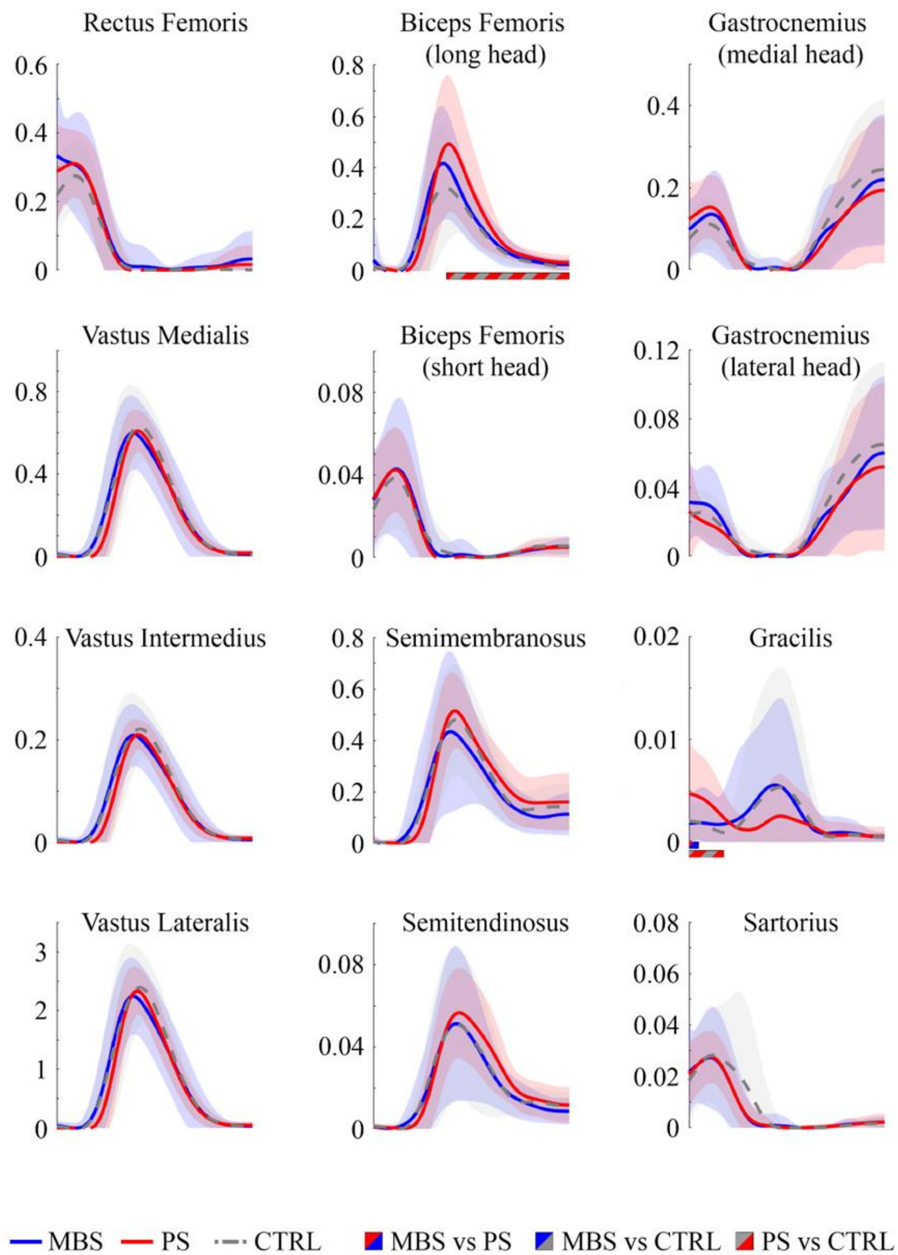

Figure 4.8.3: Muscle forces during the sit-to-stand task for the operated limb in the MBS (blue) and PS (red) groups, and average between both limbs for the CTRL (grey) group. SnPM results are displayed below each figure and indicate significant ($p < .05$) differences between MBS and PS (red/blue), MBS and CTRL (blue/grey) and PS and CTRL (red/grey). MBS = medial ball-and-socket, PS = posterior stabilized, CTRL = control, SnPM = statistical non-parametric mapping, BW = body weight.

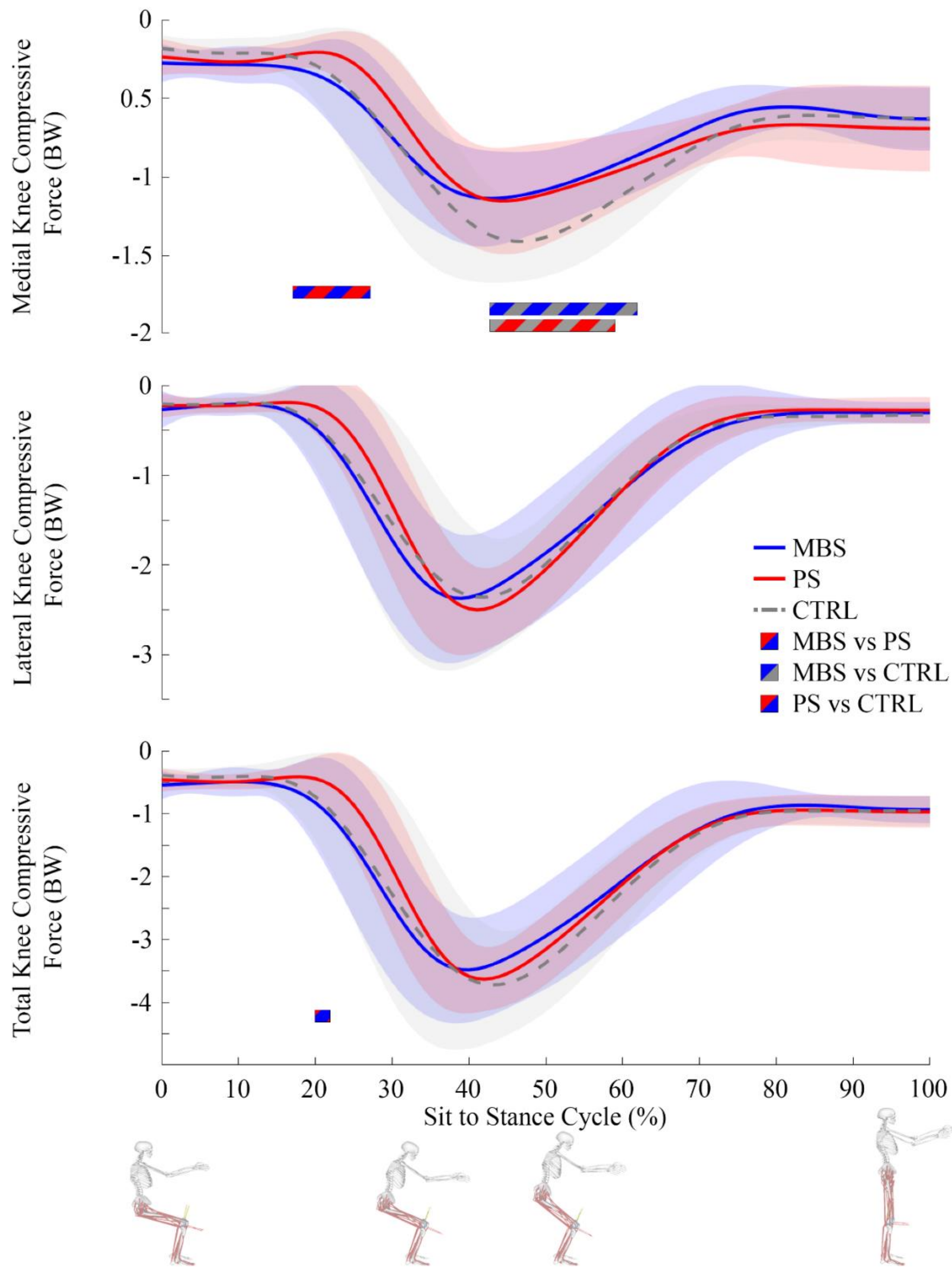

Figure 4.8.4: Knee contact forces during the sit-to-stand task for the operated limb in the MBS (blue) and PS (red) groups, and average between both limbs for the CTRL (grey) group. SnPM results are displayed below each figure and indicate significant ($p < .05$) differences between MBS and PS (red/blue), MBS and CTRL (blue/grey) and PS and CTRL (red/grey). MBS = medial ball-and-socket, PS = posterior stabilized, CTRL = control, SnPM = statistical non-parametric mapping, BW = body weight.

Table 4.8.2: Between-limb comparisons between operated and non-operated limbs in MBS and PS groups, and dominant and non-dominant limbs in CTRL groups. MBS – medial ball-and-socket, PS – posterior stabilized, CTRL – control, ROM – range of motion, ABD – abduction, ADD – adduction, GEN – generation, BW – body weight, KCF – knee contact force.

	MBS					PS					CTRL				
	Operated		Non-operated			Operated		Non-operated			Dominant		Non-dominant		
	Mean	SD	Mean	SD	*p*-value	Mean	SD	Mean	SD	*p*-value	Mean	SD	Mean	SD	*p*-value
Kinematics															
Max Hip Flexion (°)	83.3	6.2	84.5	6.0	.309	71.2	9.9	76.5	11.4	**.010**	81.9	8.1	82.6	9.6	.600
Max Hip Extension (°)	0.9	4.7	1.3	5.7	.589	2.6	4.2	0.8	4.1	.051	1.1	4.1	0.8	5.2	.406
Hip ROM (°)	84.2	7.9	85.8	8.4	.448	73.8	9.2	77.4	11.3	.061	83.1	7.7	83.4	7.6	.974
Max Knee Flexion (°)	86.2	10.8	92.0	7.2	**.005**	77.6	15.2	90.2	11.3	**<.001**	85.8	15.9	86.8	14.1	.628
Max Knee Extension (°)	0.8	3.2	0.0	3.1	.223	1.6	2.8	1.4	3.8	**<.001**	0.1	4.6	0.7	4.8	.208
Sagittal Knee ROM (°)	87.0	9.2	92.1	6.2	**.003**	79.3	16.7	88.8	11.6	**.006**	85.9	15.5	86.1	15.0	.968
Max Knee ABD (°)	25.0	14.9	15.5	9.1	**.002**	24.3	19.3	17.9	15.9	**.034**	17.1	14.5	15.7	13.3	.438
Max Knee ADD (°)	0.5	1.1	3.2	8.5	**.006**	2.8	6.5	3.6	5.2	.119	5.9	10.1	4.3	9.7	.090
Frontal Knee ROM (°)	24.5	14.5	18.7	6.8	.114	27.1	17.2	21.5	14.1	**.029**	23.0	11.7	20.0	12.8	.146
Max Ankle Dorsiflexion (°)	12.7	5.3	17.3	4.5	**<.001**	8.3	2.4	15.3	10.3	**<.001**	15.1	7.2	13.4	7.4	.242
Max Ankle Plantarflexion (°)	4.2	2.5	2.6	3.5	**.021**	4.8	2.0	2.1	3.3	**<.001**	2.9	4.1	3.9	5.6	.498
Ankle ROM (°)	16.8	3.9	20.0	4.0	**.001**	13.1	2.8	17.4	8.8	**.006**	18.0	4.9	17.3	4.2	.434
Kinetics															
Max Knee Flexion Moment (Nm/kg)	0.18	0.10	0.21	.011	**.042**	0.14	0.10	0.14	0.08	.959	0.15	0.09	0.21	0.10	**<.001**

Max Knee Extension Moment (Nm/kg)	0.70	0.23	0.87	0.24	**<.001**	0.52	0.20	0.71	0.21	**<.001**	0.75	0.27	0.64	0.28	**<.001**
Max Knee ABD Moment (Nm/kg)	0.05	0.07	0.02	0.02	**.030**	0.04	0.05	0.05	0.05	.364	0.04	0.04	0.06	0.05	.129
Max Knee ADD Moment (Nm/kg)	0.17	0.09	0.28	0.12	**<.001**	0.16	0.09	0.28	0.13	**<.001**	0.26	0.15	0.19	0.10	**.007**
Max Knee Power GEN (W/kg)	0.99	0.46	1.48	0.48	**<.001**	0.67	0.19	1.06	0.39	**<.001**	1.23	0.54	0.98	0.50	**<.001**
Muscle Forces															
Knee Extensors															
Rectus Femoris (BW)	0.41	0.20	0.41	0.12	.135	0.35	0.13	0.40	0.11	**.035**	0.31	.012	0.30	0.10	.346
Vastus Medialis (BW)	0.70	0.11	0.84	0.16	**<.001**	0.65	0.09	0.84	0.23	**<.001**	0.80	0.14	0.73	0.11	**.001**
Vastus Intermedius (BW)	0.24	0.03	0.29	0.06	**<.001**	0.22	0.03	0.29	0.08	**<.001**	0.28	0.05	0.25	0.04	**<.001**
Vastus Lateralis (BW)	2.62	0.36	2.99	0.42	**<.001**	2.49	0.36	2.83	0.45	**<.001**	3.02	0.49	2.78	0.35	**<.001**
Knee Flexors															
Biceps Femoris (Long Head) (BW)	0.48	0.22	0.45	0.20	.538	0.55	0.28	0.59	0.31	.200	0.42	0.23	0.44	0.23	.216
Biceps Femoris (Short Head) (BW)	0.05	0.03	0.06	0.03	.238	0.05	0.02	0.06	0.02	**.001**	0.04	0.03	0.05	0.03	.075
Semimembranosus (BW)	0.54	0.29	0.59	0.33	**.038**	0.57	0.15	0.56	0.24	.538	0.59	0.24	0.60	.017	.225
Semitendinosus (BW)	0.06	0.04	0.06	0.03	.075	0.06	0.02	0.06	0.03	.942	0.06	0.04	0.06	0.03	.286

Gastrocnemius (Medial Head) (BW)	0.27	0.14	0.38	0.18	**<.001**	0.26	0.17	0.35	0.17	**<.001**	0.26	0.15	0.31	0.16	**.001**
Gastrocnemius (Lateral Head) (BW)	0.07	0.04	0.10	0.05	**<.001**	0.06	0.05	0.08	0.06	**.001**	0.07	0.04	0.08	0.05	**<.001**
Gracilis (BW)	0.01	0.01	0.01	0.01	.756	0.01	0.01	0.01	0.01	.302	0.01	0.01	0.01	0.01	.068
Sartorius (BW)	0.04	0.02	0.04	0.01	.174	0.03	0.01	0.04	0.01	**<.001**	0.04	0.03	0.04	0.03	.413
Knee Contact Forces															
Medial KCF (BW)	1.31	0.23	1.65	0.40	**<.001**	1.25	0.32	1.56	0.24	**<.001**	1.66	0.32	1.50	0.21	.113
Lateral KCF (BW)	2.74	0.49	2.99	0.36	**.002**	2.66	0.45	2.96	0.62	**<.001**	3.00	0.54	2.77	0.42	**<.001**
Total KCF (BW)	4.05	0.48	4.64	0.69	**<.001**	3.91	0.49	4.52	0.69	**<.001**	4.66	0.74	4.27	0.44	**<.001**

differences between their dominant and non-dominant limbs for knee kinetics, vasti muscle forces, and total vertical and lateral compartment KCF. Therefore, the dominant and non-dominant limbs were averaged together compared to the TKA groups (Kowalski et al., 2019).

Previous MSK studies also reported that TKA patients favoured their non-operated knee during uphill walking (Thorsen et al., 2021) and stationary cycling (Hummer et al., 2022), as both studies found larger KCFs on the non-operated knee compared to the operated knee. During uphill walking, total vertical KCF was 16% lower on the operated knee compared to the non-operated knee (Thorsen et al., 2021). During stationary cycling, medial compartment KCF was 23% lower on the operated knee (Hummer et al., 2022). Similar differences were found in this study. Medial compartment KCF was 20.6% and 19.9% lower on the operated limb in the MBS and PS groups, respectively. In comparison, total vertical KCF was 12.7% and 13.5% lower on the operated limb in the MBS and PS groups, respectively.

The participant cohort was of similar age between our study and Hummer et al. (2022) and Thorsen et al. (2021) studies. However, the postoperative follow-up time was different. Thorsen et al. (2021) evaluated 25 patients with a mean follow-up of 22.1 ± 11.7 months (Thorsen et al., 2021); Hummer et al. (2022) compared fifteen patients 6 to 18 months post-TKA (Hummer et al., 2022), whereas this study evaluated eighteen patients between 12 and 20 months post-surgery. All three studies identified reduced KCF on the operated knee, suggesting that between-limb differences persist even after two years. KCFs have been recorded *in vivo* during a sit-to-stand task. At six weeks, peak vertical force was 1.5 BW (D'Lima et al., 2006) and reached almost 3 BW at a mean follow-up of 25 months (8 to 47) (Kutzner et al., 2017), suggesting patients gradually increase loading on their operated knee as they recover. However, it does not reach the same level as the non-operated knee during some activities (Hummer et al., 2022; Thorsen et al., 2021).

Having symmetrical loading between limbs during sit-to-stand is not a realistic expectation for patients after TKA. Even healthy young adults have asymmetrical lower limb joint moments and ground reaction forces when performing a sit-to-stand task (Lundin, Grabiner, & Jahnigen, 1995). The healthy older adults who formed the CTRL group in this study had similar sagittal joint angles between limbs. Still, between-limb differences existed in knee joint kinetics, muscle forces, and KCFs (Table 4.8.2). This suggests that everyone has a preferred limb when performing a sit-to-stand task. There may be some clinical concern with the MBS and PS groups loading their non-operated limb more.

The incidence of developing radiographic knee OA on the non-operated side is 62.5% within eight years (Aljehani et al., 2022) and 80% within 12 years (Metcalfe, Andersson, Goodfellow, & Thorstensson, 2012). Although the exact cause of contralateral knee OA development after TKA is not fully understood, increased loading and reliance on the non-operated knee during sit-to-stand and other daily activities may contribute to the risk of developing OA on the non-operated knee (Hummer et al., 2022; Shakoor et al., 2002; Thorsen et al., 2021). Abnormal gait biomechanics are known risk factors for primary knee OA (Aljehani et al., 2022). Reliance on the contralateral knee after primary TKA likely plays a role in contralateral knee OA development as individuals place greater force and have higher abduction moments on the contralateral knee (Alnahdi, Zeni, & Snyder-Mackler, 2016).

Recently, a study could not find biomechanical differences during gait between “progressors” and “non-progressors” for contralateral OA development 6-24 months after unilateral TKA. Peak knee joint angles and knee abduction moments during gait could not predict contralateral OA progression (Aljehani et al., 2022). Gait may not be a difficult or sensitive enough task to predict contralateral OA progression risk. Future studies could follow the same methodology and use more demanding activities of daily living such as climbing stairs, descending a ramp, or sit-to-stand. More research is necessary to understand the link between limb loading differences and knee OA development. Achieving better symmetry during sit-to-stand should continue to be a goal of post-operative rehabilitation as it may promote greater use of the operated limb during various activities of daily living (Pua et al., 2022).

Few differences in the patient-reported outcome measures (PROMS) between the PS and MBS groups were identified in this study. Both groups remained lower than the CTRL on all subscales of the KOOS (Table 4.8.1). The MBS group scored approximately 10 points higher on the ‘Symptoms’ and 20 points higher on the ‘Sport & Recreation’ subscale than the PS group. However, this difference did not reach significance, likely due to the small sample size and large standard deviation. A recent meta-analysis which included 1811 MBS and 1972 PS knees, found no differences in PROMs between the two implants (Shi et al., 2022), as PROMs cannot capture the objective functional deficits that remain following TKA (Stevens-Lapsley, Schenkman, & Dayton, 2011).

Although no differences in PROMs existed between the MBS and PS groups, some functional differences were identified during the sit-to-stand task. The PS group required greater forward lean to stand up, which resulted in greater spine flexion, posterior pelvic tilt, and reduced hip flexion compared to the MBS group (Figure 4.8.1). However, these differences did not result in differences in knee kinetics, muscle forces, or KCFs between the MBS and PS groups.

Few differences in knee muscle forces existed between the TKA groups and the controls. The exception was BFLH muscle force, where the PS group had significantly greater force than the CTRL group. PS implants use a post-cam system to guide knee movement and provide stability. However, the femur slides forward during mid-flexion, presenting a 'paradoxical anterior movement', causing mid-flexion knee instability (Shi et al., 2022). The PS group may have increased biceps femoris muscle force to help stabilize the knee. A previous MSK modelling study identified that PS groups required greater hamstring muscle force than controls during stair ascent (Rasnick et al., 2016). A future study including electromyography assessment during sit-to-stand could determine if PS patients require greater hamstrings-quadriceps co-contraction than MBS.

This study does have some limitations that should be addressed. First, we acknowledge that the small sample size and increasing the number of participants would have resulted in higher predictive power. Second, participants only received a single MBS and PS implant, so the findings may be generalizable to only some TKA implants. Third, the geometry of each implant was not parameterized in the MSK model since it relied on a scaled generic model based on healthy knee definitions (Walker et al., 1988), which may influence KCF outputs. Fourth, this study scaled a generic model and used static optimization. Utilizing a subject-specific model may provide more accurate estimations of KCFs (Knarr & Higginson, 2015). However, the approach used in this study provided estimates in line with in vivo measurements with a shorter processing time than a subject-specific framework would have required. Finally, a previous study showed that the generic MSK model used in this study overestimates total vertical KCF by ~ 20% compared to in vivo data. However, the KCF estimates followed the same pattern and were similar in shape compared to the *in vivo* data (Pelegrinelli et al., 2023 – Under Review), so the estimated KCF can provide insights into how the knee is loaded during the sit-to-stand task.

This study provided insights into the movement patterns, muscle, and knee forces one year after unilateral TKA with either an MBS or PS implant during a sit-to-stand task. The PS group required greater forward lean to complete the task, and both groups favoured their non-operated limb. Lower limb joint ROM, knee kinetics, muscle forces, and KCFs were all lower on the operated limb compared to the non-operated limb for both MBS and PS groups. Compared to the controls, the MBS and PS groups had reduced medial compartment KCF. Post-operative rehabilitation should continue to promote greater use of the operated knee to have more symmetry between operated and non-operated limbs but also target improving strength and mobility at the hip and ankle joint.

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4.8.7 Appendix

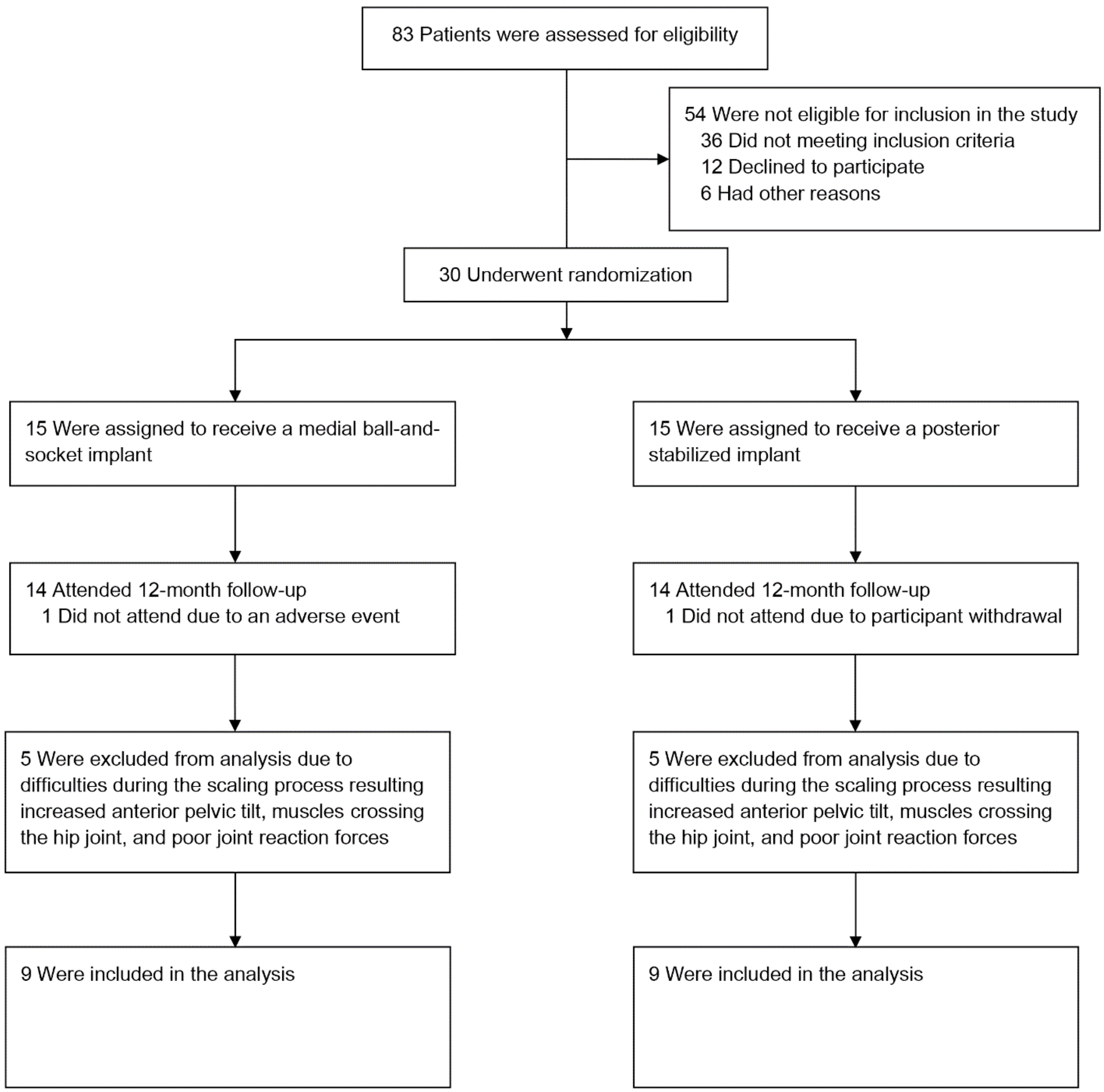

Figure 4.8.5: Consolidated Standards of Reporting Trials (CONSORT) flow diagram for enrolled patients.

5 Discussion

Summary | Clinical Implications | Limitations | Future Directions | Conclusions

5.1 Summary

The ultimate goal of this thesis was to better understand the biomechanical knee function after a total knee replacement with either a medial ball-and-socket (MBS) or posterior stabilized (PS) implant at around a one-year follow-up during activities of daily living (ADLs). This research program was structured to examine knee kinematics, kinetics, muscle activity, and muscle and knee contact forces during several daily tasks. Several gaps in the literature were revealed in Chapter 2, and several main areas for research development were highlighted:

1. Examine the changes of gait variability in knee OA patients after receiving a TKA with either a PS or MBS implant.
2. Enhance the understanding of the post-operative effects of TKA with either MBS and PS implants on knee biomechanics and muscle activities during level walking, as well as more demanding tasks such as descending a ramp or staircase.
3. Simulate the dynamic knee joint loads in post-operative TKA patients with either PS or MBS implants during closed-chain, bilateral tasks such as sit-to-stand.

Previous biomechanical studies were performed mainly in postoperative patients. Only a few studies compared MBS and PS implants but were limited primarily to level walking conditions and overlooked tasks that produced more demand on the knee joint. MBS and PS implants have different designs that may affect the knee joint's stability. Therefore, the basis of this doctoral study was to assess the function and stability of MBS and PS implants by evaluating the gait variability, knee kinematics, kinetics, muscle activation parameters, and muscle and knee contact forces in patients before and after surgery during various ADLs.

This thesis compared patients with knee OA before and after TKA with healthy controls of similar body mass index, sex, and age. Several methods exist when comparing gait data between groups, including analyzing a single leg or taking an average of both limbs. The first study (Chapter 4.1) evaluated the limb-matching method's impact on the statistical outcomes when comparing biomechanical variables in healthy individuals. During gait, the limb-matching method did not have a direct effect on biomechanical outcomes as (i) average of both limbs; (ii) dominant; (iii) non-dominant; and (iv) random limb had similar outcomes. Gait is cyclical, so limb symmetry is often assumed. However, during more complex tasks, the limb-matching method becomes more crucial. During the sit-to-stand task (Chapter 4.8), controls favoured their dominant limb as they had more significant knee contact force (KCF) than their non-dominant limb.

The KOOS was used to evaluate patient-reported outcome measures (PROMs) one-year after surgery with the MBS and PS implants. Both groups significantly improved on all KOOS subscales one-year after surgery (Chapter 4.2). However, no differences between the MBS or PS groups could be detected with the KOOS. Neither the PS nor MBS group reached the level of control for the Symptoms, Function in Sport and Recreation, and Quality of Life sub-scales. This was in line with a recent meta-analysis which compared 1811 MBS and 1972 PS knees and found no differences in PROMs between the two implants (Shi et al., 2022). Although PROMs continue to increase in popularity in clinical practice and clinical trials (Kluzek et al., 2022), they suffer from ceiling effects and cannot detect differences in from well-performing groups (Eckhard et al., 2021). Functional differences were identified between the groups during various ADLs.

Gait is a basic requirement of daily life, and a major determinant of independence and quality of life (Schmid et al., 2007). For this reason, level walking was involved in five of the eight manuscripts included in this thesis. Level walking was included in the limb-matching method (Chapter 4.1), to compare MBS and PS groups before and after TKA (Chapter 4.2), and in the development and analysis with the waveform-level variance equality test (Chapter 4.3 to 4.5)). The differences identified in the first five studies were essential to better understand differences that occurred at more challenging ADLs evaluated in the later chapters.

During level walking (Chapter 4.2), the MBS group achieved a gait pattern closer to the control group with no differences in knee joint angles or moments. In contrast, the PS group walked with wider steps, less knee flexion angle and less extension moment than the control group. However, MBS and PS groups had less knee power absorption before toe-off. The MBS implant provided some benefits over the PS implant during gait. However, the advantages became less clear during more demanding activities.

Knee joint instability is a cause for concern after TKA as it can cause a loss of balance and lead to an eventual fall (Nevitt et al., 2016). Dynamic stability of the knee has been evaluated by the variability of temporospatial and biomechanical measures during gait (Fallah Yakhdani et al., 2010; Lewek et al., 2006). Gait variability refers to the natural variations that occur in an individual's walking pattern over time that is measured by the variability in various biomechanical parameters. Many methods have been used to provide a measure of global variability, including standard deviation (Owings & Grabiner, 2004), coefficient of variation

(Winter, 1984), or Lyapunov exponent (Yakhdani et al., 2010). A review on gait variability in patients with knee OA highlighted that variability remains lower than healthy controls before and after a TKA (Smith, 2014). The limitation with previous methods to measure gait variability are that they integrate the variability across the time domain to yield a single scalar value and provide no indication of where in the movement the variability differences occurred. Therefore, a series of studies were conducted to develop a statistical method to identify where in the movement cycle variability differences occurred.

The first study of this series (Chapter 4.3) developed a waveform-level variance inequality test which was not previously used in the literature. This test utilized the free and open-sourced 'SPM1D' software package to use an F-test to compare group variances across an entire waveform (Pataky, 2016). Common calculations of the critical F value, which depend on the degrees of freedom and alpha, could not be used because multiple F values are calculated across time. Therefore, one-dimensional random field theory was used to calculate F-critical. This function required several inputs, including alpha ($\alpha = 0.05$), degrees of freedom, the number of discrete field nodes and the field smoothness. Field smoothness was estimated using the full-width-half-maximum function within SPM.

This test was then used to evaluate where differences in gait variability occurred throughout the gait cycle in healthy controls (Chapter 4.4) and then before and after TKA (Chapter 4.5). Knee OA patients had lower knee moment variability throughout the stance phase than controls (Chapter 4.5). Following TKA, knee moment and power variability decreased at single-limb support, and neither the MBS nor PS implant provided movement variability like controls, representing reduced knee joint stability. Knee stability is achieved by the implant design and muscles, so these findings warranted further exploration of knee biomechanics, muscle activations, and muscle and knee contact forces through various ADLs to understand if one implant can provide better function and outcomes for patients.

When descending a ramp (Chapter 4.6), various gait patterns, characterized by different knee moment patterns, were observed in controls and knee OA patients. Individuals either had an initial knee flexion moment followed by a knee extension moment or maintained a knee extension moment throughout the ramp descent cycle. All TKA patients adopted the same movement pattern postoperatively, regardless of how they walked pre-operatively or which implant they received. MBS and PS groups adopted a ‘cautious gait pattern’ characterized by a continuous knee extension moment and increased knee power absorption throughout the stance phase. When descending a staircase (Chapter 4.7), the PS group had reduced knee flexion angle and greater hamstring activation throughout single limb support than the MBS and control groups. Knee joint moments and powers were similar between the MBS and PS groups, but neither reached the level of the control group.

Isometric strength normalized to body weight during knee extension, knee flexion, and ankle plantar flexion was assessed between the groups. PS and MBS groups achieved similar isometric strength as the control for all movements, and the MBS had greater knee flexion isometric strength than the PS group (Chapter 4.7). However, compared to their non-operated limb, the MBS group was weaker in knee extension, whereas the PS group had similar isometric strength on operated and non-operated limbs (Chapter 4.8).

The final study (Chapter 4.8) evaluated knee joint loading in both the operated and non-operated limbs during a sit-to-stand task. MBS and PS groups had similar performance, and both had reduced KCF on the medial side compared to the control group. All groups favoured one limb; the MBS and PS groups favoured their non-operated limb, and the control group favoured their dominant limb. Lower limb joint ROM, knee kinetics, muscle forces, and KCFs were all lower on the operated limb compared to the non-operated limb for both MBS and PS groups. Compared to the controls, the MBS and PS groups had reduced medial compartment KCF. Post-operative rehabilitation should continue to promote greater use of the operated knee to have more symmetry between operated and non-operated limbs but also target improving strength and mobility at the hip and ankle joints. The favouring of the non-operated limb may lead to clinical implications on the non-operated limb over time.

5.2 Clinical Implications

5.2.1 Patient-Reported Outcome Measures

TKA with either a PS or MBS implant improved the quality of life of the patients who previously suffered from knee OA. Patients reported improved pain, function, and overall quality of life. Postoperative KOOS values significantly increased in all subscales measured for both groups compared to their preoperative scores (Table 4.2.1). This improvement exceeded the minimum clinical change and substantial clinical benefit change (Lyman, Lee, McLawhorn, Islam, & MacLean, 2018). However, the Symptoms, Function in Sport and Recreation, and Quality of Life subscales did not reach the level of the control group for either the MBS or PS group. Although PROMs are reliable, valid, and easy to administer, they are subjective measures suffering from ceiling effects which cannot detect differences between well-performing groups (Eckhard et al., 2021). The KOOS was not sensitive enough to detect differences between the MBS and PS groups, which agreed with a recent meta-analysis that evaluated differences between these implants using multiple PROM tools (Shi et al., 2022).

Pearson's correlations were performed to determine if a relationship existed between KOOS scores, strength measurements, and discrete biomechanical variables. Various variables were significantly ($p < .05$) correlated with one another. However, the Pearson's correlation coefficient (r) was between -0.6 and 0.6, representing a poor to fair correlation (Akoglu, 2018). The only variables with stronger correlations were with the five KOOS subscales, which were all correlated with one another, achieving moderate ($r \geq 0.7$) or very strong (≥ 0.8) correlations. This correlation implied that if a patient scored poorly on one subscale of the KOOS, such as symptoms, there was a strong likelihood they also scored poorly on the other subscales. The reliance on PROMs within the TKA literature has increased over time (Lan et al., 2020). This thesis has highlighted the necessity to move beyond qualitative tools to identify performance differences between implant groups, as they are subjective measurements which suffer from ceiling effects and cannot detect differences between well-performing groups (Eckhard et al., 2021).

5.2.2 Total Knee Prosthetic Design

Several implant designs exist, however, their effects on biomechanical knee joint function are not well understood. This thesis investigated two types of TKA implant designs: MBS and PS. MBS and PS implants share common elements, including fixed bearing, cruciate sacrificed, and cemented metal femoral components on polyethylene tibial trays (Marques et al., 2021). However, they differ in design. The PS design incorporates a post-cam system to substitute the cruciate ligaments and provide anteroposterior stability (Castellarin et al., 2023). The MBS implant has a concave medial compartment which functions as a ball and socket joint, and the lateral component rotates around the medial compartment (Chang et al., 2021). Previously reported advantages of the MBS over the PS design included better restoration of native knee kinematics, improved contact stresses, reduced polyethylene wear, and increased mid-flexion stability (Chang et al., 2021; Mannan & Scott, 2009; Samy et al., 2018).

Both implants lack cruciate ligaments, so the remaining collateral ligaments, muscles, and implant design achieve knee joint stability (Abulhasan & Grey, 2017). The differences in design between the PS and MBS implants resulted in observed differences in knee joint stability throughout various ADLs. The groups differed in gait variability, knee kinematics and kinetics, muscle activity, and muscle and knee contact forces, which highlight the effects of the implant design on knee joint stability.

5.2.3 Knee Joint Dynamic Stability

Dynamic stability of the knee has been assessed by evaluating the variability of temporospatial and knee biomechanical measures during gait (Yakhdani et al., 2010; Lewek et al., 2006). TKA patients remain with lower variability than controls after surgery (Yakhdani et al., 2010; Kiss et al., 2012; J. W. Smith, 2014), suggesting a less stable knee that is less capable of adapting to perturbations, which may potentially increase fall risk (Chan et al., 2018; Liu et al., 2020). However, previous measures of gait variability were completed with methods that only provided a measure of global variability, so it was unknown where variability differences occurred in the gait cycle. This thesis significantly contributed to evaluating gait variability by developing the Equal Variance Test (Chapter 4.3) to determine where in the gait cycle groups differ in variability (Chapter 4.4). The Equal Variance Test has been implemented into the SPM1D package (*eqvartest*, as of Version 0.4.8, 2021-09-09) and is accessible for free to anyone to use with the Matlab or Python SPM1D package.

During gait, both MBS and PS had a reduction in knee moment and knee power variability compared to preoperative (Figure 4.5.3) and remained lower than the control group (Figure 4.5.4). These differences were most significant during the transition from double- to single-limb (~15% gait cycle) and single- to double-limb (~50% gait cycle) support. Greater kinematic and kinetic variability is favourable as it reflects adaptability (Hausdorff, 2007), suggesting that TKA patients would be less stable and less capable of adapting to unanticipated situations while walking. Between 7-38% of TKA recipients experience a fall within the first 12 months after surgery (Chan et al., 2018; Liu et al., 2020). The reduction in gait variability may partially explain this fall-risk.

Patients with knee OA adapt their movement patterns to reduce pain while awaiting TKA (Farquhar, Reisman, & Snyder-Mackler, 2008). Several strategies to reduce variability have been identified, including stiffening the knee joint through co-contraction, walking slower, or paying more attention to how they walk (Yakhdani et al., 2010). These strategies do not imply conscious cognitive involvement, as it is suspected that patients do not know how they are adopting or why they are doing so (Dijksterhuis & Aarts, 2009; Yakhdani et al., 2010). The patients in this study may have adopted some strategies to reduce variability while awaiting TKA, which were further reduced after surgery. PS and MBS groups had altered knee biomechanics and muscle activations while completing various ADLs, which suggest different adaptations to maintain knee joint stability.

Compared to healthy individuals, variability is lower in patients with knee OA, and continues to decrease after TKA (Smith, 2014). This may be due to movement restrictions imposed by the implant after TKA. A healthy knee has six degrees of freedom (DoF) – three translations and three rotations (Grood & Suntay, 1983). While a replaced knee provides similar

DoF, dual fluoroscopic imaging studies have shown that MBS and PS knees have different *in vivo* knee kinematics than healthy knees (Tan et al., 2021). Both MBS and PS implants lacked femoral rollback in early-stance, and the tibia internally rotated in both groups, compared to external rotation in healthy knees (Tan et al., 2021). In healthy knees, knee flexion and femoral rollback in early-stance increases the quadriceps' lever arm and stabilizes the body during weight acceptance (Draganich, Piotrowski, Martell, & Pottenger, 2002). The absence of this femoral rollback could reduce the lever arm of the quadriceps, compromising the knee extensor's effectiveness, which may cause feebleness and instability during walking (D'Lima et al., 2001). The lack of tibial external rotation may be due to the resection of the ACL, which leads to altered knee kinematics and a loss of the screw-home mechanism (Ng et al., 2013). These differences between healthy, MBS, and PS knees may ultimately cause differences in variability and knee biomechanics during various ADLs.

5.2.4 Knee Kinematics and Kinetics

Differences in knee kinematics and kinetics were identified between the MBS and PS groups during level walking (Chapter 4.2) and stair descent (Chapter 4.7). In contrast, during ramp descent, both groups adopted a similar movement pattern (Chapter 4.6). A limitation of previous TKA gait studies was that they only evaluated patients after surgery (Esposito et al., 2020; Kramers-de Quervain et al., 1997), so it is unclear if gait adaptations were developed before or as a result of the surgery or implant. The pre-operative evaluation in this thesis ensured patients were generally similar, except for the PS patients walking slower pre-operatively; no other pre-operative differences in age, BMI, KOOS, temporospatial measures, or knee biomechanics existed between the MBS and PS groups during level walking (Chapter 4.2). Postoperatively, the PS group walked with wider steps than MBS and the control group. This adaptation increased their stability by widening their base of support. It can also decrease knee abduction moment, reducing the loading of the medial compartment of the knee (Sample, Thorsen, Weinhandl, Strohacker, & Zhang, 2020).

The MBS achieved a gait pattern that resembled the control group, with no significant differences in knee joint angles or moments. The PS group also walked with less knee flexion angle and extension moment than the controls (Figure 4.2.4). This may represent a pain avoidance strategy to minimize knee loading (Briem & Snyder-Mackler, 2009). This knee extension moment decreased from single-limb support on the affected limb to double-limb support. During this period, the MBS and PS groups had a reduced knee power absorption compared to the control, suggesting both groups utilize a stiff knee gait pattern to avoid loading their operated knee (Dorr, Ochsner, Gronley, & Perry, 1988). The findings suggest that the MBS implant has some advantages over the PS during level walking, but the results become less clear during more demanding ADLs.

When descending a ramp (Chapter 4.6), all TKA patients adopted a 'cautious gait pattern' to increase stability, regardless of how they walked preoperatively (Figure 4.6.2) or which implant they received. The MBS reduced their walking speed, and both groups widened their steps after TKA (Figure 4.6.5). This 'cautious gait pattern' was also characterized by a knee extension moment and increased power absorption. The increased knee extension moments and power absorption are caused as patients generate larger shear (i.e., braking) forces to improve stability (McVay & Redfern, 1994; Redfern & DiPasquale, 1997). In contrast, half the controls and some knee OA patients descended the ramp with a knee flexion moment during single limb support (Figure 10.2). This gait pattern has smaller braking forces and larger anteroposterior GRF (Redfern & DiPasquale, 1997), resulting in a faster walking speed (Figure 4.6.8).

Much of the metabolic cost of walking occurs at the transition from single- to double-limb support. The individuals who walk with the knee flexion moment pattern generate power at the knee at the start of single limb support (~10% gait cycle), efficiently transferring power from one limb to the other as the centre of mass is moved forward. In contrast, the MBS and PS groups walked with a knee extension moment at this point in the gait cycle and constantly absorbed power at the knee joint which indicate a ‘cautious gait pattern’. This effectively caused a ‘controlled fall’ from one foot to the other as they descended the ramp, which increased stability but decreased efficiency (Hunter, Hendrix, & Dean, 2010; Monsch, Franz, & Dean, 2012). When walking downhill, gravitational energy can assist with propulsion, but at the expense of stability. Patients who adopt the ‘cautious gait pattern’ do not take advantage of the propulsion provided by gravity to decrease the energetic cost but instead opt for a more stable but metabolically more costly gait pattern (Hunter et al., 2010; Monsch et al., 2012).

Although MBS and PS groups adopted a ‘cautious gait pattern’, the MBS group achieved greater knee flexion angles, larger knee extension moments, and power absorptions at the start of single limb support (Figure 4.7.6). These findings suggest that the PS group used greater knee joint stiffness than the MBS group to reduce loading at the knee (Briem & Snyder-Mackler, 2009; McGinnis, Snyder-Mackler, Flowers, & Zeni, 2013). This difference in knee joint stiffness is likely due to increased co-contraction between the hamstrings and quadriceps muscles (McGinnis et al., 2013). Differences in muscle activity were identified during ramp descent when comparing different implant types (Simon et al., 2018). The findings of this study warranted the evaluation of muscle activity during a stair descent task. Like ramp descent, descending a staircase is an eccentric task where the knee must maintain stability as the body’s centre of mass is lowered. This may uncover neuromuscular differences between the MBS and PS groups.

5.2.5 Muscle Activation Parameters

During the stair descent task (Chapter 4.7), the MBS and PS groups had similar knee joint moments and powers (Figure 4.7.4), but neither reached the control group level, representing less knee loading. The PS group descended the stairs with reduced knee flexion angle and lower quadriceps/hamstrings co-activation ratio (Table 4.7.3), due to greater normalized hamstring muscle activation throughout single limb support compared to the MBS and control group (Figure 4.7.4), suggesting greater knee stiffness (McGinnis et al., 2013). Except for the rectus femoris muscle, the MBS group had lower total muscle activity for most muscles evaluated (Figure 4.7.3), even if they were activated earlier or for a longer duration of the stair descent cycle (Table 4.7.3).

Neither the PS nor MBS implant has cruciate ligaments, so the remaining ligaments, muscles, and implant design maintain knee joint stability. The more conforming design of the MBS implant appears to provide better stability as patients achieved more knee flexion with less muscle activity. The PS group increased knee joint stiffness and may have consequently increased knee contact forces (KCF) which can cause additional wear of the knee prosthesis (Benedetti et al., 2003; McGinnis et al., 2013; Stevens-Lapsley et al., 2010). Radiographic studies have identified that PS implants generate more wear particles than MBS implants (Minoda et al., 2003). Evaluating how the different implant designs affect muscle and knee contact forces is necessary. Additionally, this thesis was primarily focused on adaptations on the operated limb. Understanding how the contralateral knee is loaded after TKA is essential, as the incidence of radiographic knee OA progression in the contralateral knee was 62.5% based on the Kellgren Lawrence score after eight years (Moiyad Saleh Aljehani et al., 2022).

5.2.6 Muscle and Knee Contact Forces

The estimated total vertical KCF reached a similar peak force of 3.5 BW during sit-to-stand for MBS and PS groups, and there was greater KCF in the lateral compartment of the knee (Figure 4.8.3). The main muscle contributors to knee contact force were the vastus lateralis for knee extensors and the long head of the biceps femoris for the knee flexors. PS and MBS groups compensated more with their non-operated limb, as total vertical, medial and lateral compartment KCF was greater on the non-operated limb (Table 4.8.2). Knee extensor muscle forces were also greater on the non-operated limb (Table 4.8.2).

TKA patients gradually increase the load on their operated knee as they recover after surgery. A peak KCF of 1.5 BW was recorded *in vivo* at six weeks post-TKA (D'Lima et al., 2006) and increased up to 3 BW at a mean of 25 months (8 to 47) post-surgery during a sit-to-stand task (Kutzner et al., 2017). Indications of TKA patients favouring their non-operated limbs have been reported during uphill walking (Thorsen et al., 2021) and stationary cycling (Hummer et al., 2022). However, no differences were identified during level walking (Thorsen et al., 2021). This reduced loading on the operated limb was achieved primarily by reducing medial KCF compared to their non-operated limb (Thorsen et al., 2021). Similar differences were found in this thesis, as medial KCF was 21% and 20% lower in the operated limb in the MBS and PS groups, respectively. In comparison, total vertical KCF was 13% lower on the operated limb in both groups.

Having symmetrical loading between limbs during sit-to-stand is not a realistic expectation for patients after TKA. A previous study with healthy young adults showed asymmetrical lower limb joint moments and GRFs when performing a sit-to-stand task (Lundin, Grabiner, & Jahnigen, 1995). The older controls in this study had similar sagittal joint angles between limbs. However, between-limb differences existed in knee joint kinetics, muscle forces, and KCF (Table 4.8.2). These findings suggest that everyone has a preferred limb when performing a sit-to-stand task. However, the loading of non-operated limbs by the MBS and PS groups may raise some clinical concerns.

The incidence of developing radiographic knee OA on the non-operated side is 62.5% within eight years (Aljehani et al., 2022) and 80% within 12 years (Metcalfe, Andersson, Goodfellow, & Thorstensson, 2012). Increased loading and reliance on the non-operated knee during sit-to-stand and other ADLs may contribute to this risk of non-operated knee OA development (Hummer et al., 2022; Shakoor et al., 2002; Thorsen, C. Wen, & S. Zhang, 2021). Although abnormal biomechanics are known risk factors for primary knee OA, a study by Hummer et al. (2022) did not find any biomechanical differences between patients who were “progressors” and “non-progressors” for contralateral knee OA development 6-24 months after unilateral primary TKA. The variables they selected in their study; peak knee joint angles and peak knee abduction moment during gait; could not predict contralateral OA progression (Aljehani et al., 2022). Gait may not be a difficult or sensitive enough task to predict contralateral OA progression risk. Patients could have begun loading their non-operated limb more as a defensive mechanism through the progression of knee OA. Their operated knee was guarded at the expense of the non-operated knee, which did not resolve after TKA. Future studies could follow a similar methodology and use more demanding ADLs such as climbing stairs, descending a ramp, or sit-to-stand.

5.3 Limitations

5.3.1 Participants

Most studies in this thesis consisted of a relatively small sample size of 14 patients in each group. They may have been underpowered to identify differences between our groups, potentially leading to a type II error. However, the sample size was in line with or exceeded previous studies which evaluated biomechanics (Simon et al., 2018; Wen et al., 2022; Wiik et al., 2015) and muscle activity (Bolanos et al., 1998; Catani et al., 2003; Elkarif et al., 2021; Kelman, Biden, Wyatt, Ritter, & Colwell, 1989; Wilson et al., 1996) in TKA populations.

Except for three PS participants, patients were compared at 12 ± 1 months after TKA surgery. Some evidence suggests that patients continue to recover beyond this period, as studies which compared patients at 24 months found fewer differences between groups (Kulshrestha et al., 2020; Stolarczyk et al., 2022).

The decision was taken to exclude obese (BMI>35kg/m^2) individuals. While many TKA recipients are overweight and are represented within the general patient population, excessive adiposity can affect the accuracy of gait analysis (Silva-Hamu et al., 2013). To limit the effect of other potential gait-influencing factors, the study only included patients who were to undergo primary TKA (no revisions) and who did not have any other lower limb joint replacements.

Clinical and radiographic parameter were not evaluated as part of this thesis. Surgeons clinical evaluate various functional parameters such as passive and active knee ROM, varus/valgus stability, or translational/rotations stability of their patients. They also use radiographs to assess tibial slope, alignment of the limb, and restoration of the joint line. These variables may have provided greater clinical relevance to the overall thesis findings.

5.3.2 Surgical Approach

All TKA patients in this thesis underwent surgery by the same surgeon using the same surgical approach: mini-invasive incision, subvastus approach, with a mechanical alignment. This approach provides some theoretical benefits of less blood loss and greater respect for the capsule, tendon, and muscle structures, but long-term benefits need to be clarified (Sanna et al., 2013; Sidhu et al., 2021). A kinematic alignment compared to mechanical alignment may improve patient-reported outcomes (Liu et al., 2022), but few differences in temporospatial, knee angles or knee moments during gait exist between kinematic and mechanical alignments (McNair et al., 2018; Yeo et al., 2019). The selected approach was based on the surgeon's skill and experience (Sanna et al., 2013). As such, the findings of this study may not be generalizable to those performed by another surgeon or when using a different approach.

Most orthopedic companies offer an MBS and PS implant, which may vary slightly in design. This thesis only included a single type of MBS (MicroPort EVOLUTION® Medial Pivot System with cruciate sacrificing tibial inserts) and PS (Zimmer Biomet® NexGen® PS TKA system with PS inserts) implant. The findings uncovered between the MBS and PS implants in this thesis may not be generalizable to all implants.

5.3.3 Post-Operative Rehabilitation

All patients were provided with eight publicly funded physiotherapy sessions. Some patients may have continued beyond these sessions or not completed all their visits, and that information was not recorded. Treatment may have differed between clinics. Additionally, targeted physiotherapy may benefit patients, especially patients with elevated coactivations, as some evidence suggests neuromuscular re-education can reduce coactivations (Preece, Jones, Brown, Cacciatore, & Jones, 2016). These potential differences in post-operative rehabilitation may have created some variability within our results.

5.3.4 Muscle Activations

To compare muscle activation between participants, the EMG signals were normalized by the MVIC signal obtained from the strength assessments on the instrumented knee flexion/extension machine and Biodex. Although verbal encouragement was continuously provided throughout the MVICs, it is uncertain if actual maximal signals were obtained. Pain, lack of motivation, discomfort, or difficulty performing the task may have prevented participants from generating a true MVIC. Since PeakLE was reported as a percentage of MVIC, this could have resulted in an overestimated PeakLE and iEMG.

To limit variability in electrode placement, the same researcher placed the EMG electrodes on all participants according to SENIAM guidelines. Before securing the electrodes with tape, placements were verified by evaluating the signal on the computer as participants contracted their muscles.

5.3.5 Modelling Parameters

MSK models provide a simplified representation of the body. They consist of rigid body structures, simplified joint kinematics, and can remove some degrees of freedom at the joints. Muscle lines of action are not always physiologically representative as it is limited to one or more segments of a motor tendon unit (MTU). Multiple MTUs cannot attach to the same tendon (e.g., triceps surae), and MSK models consider separate tendons for each MTU. The Hill-type muscle models used in OpenSim also ignore several phenomena (Hicks, Uchida, Seth, Rajagopal, & Delp, 2015).

To calculate muscle forces, static optimization was used, and it considers each time-frame separately. It uses musculoskeletal geometry and joint moments calculated from inverse dynamics to estimate individual muscle forces at each instant. An optimal neuromuscular strategy is assumed as muscle forces are resolved by minimizing the sum of squared muscle activations (Heintz & Gutierrez-Farewik, 2007). It also assumes that the activation and force production occur simultaneously, which is not representative of the electromechanical delay. Other methods include EMG-driven models that combine joint kinematics and EMGs as inputs to estimate individual muscle forces and joint moments. EMG-driven and static optimization estimated muscle force profiles show broad agreement (Trinler et al., 2018). Muscle force signals are often qualitatively compared against EMG measurements, which provides a general indication of whether muscle force patterns are reasonable (Hicks et al., 2015). However, it offers limited use in the magnitude of force estimates as there is no simple relationship between the EMG signal and the magnitude of force generation within the muscle (Trinler et al., 2018).

This thesis used a modified MSK model as opposed to a subject-specific model. Subject-specific models require medical images to create individualized geometries and properties for each participant, creating a more complex, costly, and time-consuming workflow (Kainz et al., 2021; Modenese et al., 2018). Although some improvements can be expected with subject-specific models (Kainz et al., 2021; Knarr & Higginson, 2015), the additional medical images were unavailable to utilize such a framework. Additionally, the generic MSK model did not incorporate the geometries of the MBS or PS implant since it relied on a scaled generic model based on healthy knee definitions (Walker et al., 1988), which may influence KCF outputs.

The modified MSK model used for the sit-to-stand task did overestimate total KCF compared to *in vivo* data. Whereas the estimated total vertical KCF was between 3 to 3.5 BW, in vivo total KCF was around 3 BW (Kutzner et al., 2017; Kutzner et al., 2010). The model was previously compared with *in vivo* data and compared against other generic models (Pelegrinelli et al., 2023 – Under Review). In their findings, the model did overestimate total and lateral KCF by ~20% and underestimated medial KCF by a similar magnitude. However, estimated and *in vivo* KCF followed the same pattern and were similar in shape, so the estimated KCF can still provide insights into how the knee was loaded during a sit-to-stand task.

5.4 Future Directions

Based on the findings of this thesis, long-term postoperative analysis, the effect of other types of implants, the impact of kinematic and mechanical alignment, and developing a better understanding of the role of the non-operated limb during a variety of ADLs should be investigated. Studies which evaluated patients at 24 months found fewer differences between groups (Kulshrestha et al., 2020; Stolarczyk et al., 2022). It should be explored if certain implant types, whether MBS, PS or some other, provide greater function and limit the development of knee OA on the non-operated limb.

Smart implants are already here (Persona IQ, Zimmer Biomet), which can remotely monitor functional progress throughout recovery. As this technology becomes cheaper and more widely used, large data sets which include patients' cadence, stride length, knee ROM, walking distance, step count, and walking speed, will become available. Since the transmitter is located within the tibial stem, future iterations may include other implant designs. This would provide researchers and clinicians with real-world data to evaluate how different implants affect function over time and determine if a particular implant design provides a better long-term function for patients.

6 Conclusion

This doctoral thesis comprised eight studies investigating the biomechanical changes that occurred in patients with knee OA after receiving a TKA with either an MBS or PS implant. This was completed using motion capture, EMG, and MSK modelling approaches to understand better how biomechanical knee function changes after a total knee replacement with either an MBS or PS implant after a one-year follow-up during various ADLs.

One year after surgery, MBS and PS groups improved on all metrics evaluated by the KOOS compared to their preoperative values. However, both groups did not reach the level of control on Symptoms, Function in Sport and Recreation, and Quality of Life sub-scales, suggesting that some impairments persist after one year.

In this thesis, a novel method was used to identify where differences in gait variability occur during a movement cycle. MBS and PS groups exhibited reduced knee moment and power variability compared to control group during single limb support. This suggests that patients may experience reduced knee joint stability after TKA.

The MBS group had a gait pattern closer to the control group during level walking, whereas the PS group walked with a stiffer knee. However, during more demanding ADLs, the differences were less apparent. During ramp descent, knee joint stability issues became apparent as MBS and PS groups adopted a ‘cautious gait pattern’, widening their base of support and stiffening their knee to reduce loading. During stair descent, the MBS implant provided increased stability as it required less muscle activity than the PS, requiring greater hamstring muscle activation.

During sit-to-stand, MBS and PS groups favoured their non-operated knee as they had reduced total vertical, medial, and lateral KCF on their operated knee compared to their non-operated side. This may be due to compensatory strategies developed through the progression of knee OA and may increase the risk of developing knee OA on the non-operated limb.

A doctorate thesis is expected to make a significant contribution to the academic community by addressing a research gap or providing novel insights into an existing problem. To our knowledge, this thesis contributed to the TKA literature in several novel ways. First, the eqvartest function was developed, enabling researchers to assess the variation between two groups throughout an entire waveform. It was then applied to the TKA population to identify that TKA patients have less variability after surgery, regardless of MBS or PS implant, and that variability differences occurred primarily near the transitions from double- to single-limb support. Additionally, this function has been implemented into the SPM1D statistical package and it is readily available to all researchers in the public domain. Second, this thesis improved our understanding of MBS and PS biomechanics during various ADLs. Comparisons of these implants were limited to a few tasks and discrete statistical comparisons. This thesis utilized SPM to compare MBS and PS groups to identify where the movement cycle differences would occur. Third, the EMG study added to our understanding of muscle activations in patients after TKA with either an MBS or PS implant by including surface EMG from seven muscles around the knee joint. This study evaluated the onsets, total activation times, peakEMG, and iEMG and identified that PS patients require greater hamstring muscle activations to stabilize the knee joint. Finally, the MSK study was the first to evaluate patients with an MBS implant during the sit-to-stand task to assess patients after TKA. It highlighted that patients still favour their non-operated limb one year after surgery by having larger muscle forces and KCFs on their non-operated limb.

The findings of this thesis could assist clinicians in selecting the most suitable implant for their patients and provide guidance in designing rehabilitation programs that can enhance patient function following TKA.

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8 Appendix

8.1 Ethics

University of Ottawa Health Sciences and Science Research Ethics Board

File Number: H11-15-27 **Date (mm/dd/yyyy):** 01/13/2022

Université d'Ottawa
Bureau d'éthique et d'intégrité de la recherche

University of Ottawa
Office of Research Ethics and Integrity

Ethics Approval Notice

Health Sciences and Science REB

Principal Investigator / Supervisor / Co-investigator(s) / Student(s)

First Name	Last Name	Affiliation	Role
Geoffrey	Dervin	Medicine / Medicine	Principal Investigator
Mario	Lamontagne	Health Sciences / Human Kinetics	Co-Principal Investigator
Johanna	Dobransky	Health Sciences / Human Kinetics	Co-investigator
Erik	Kowalski	Health Sciences / Human Kinetics	Co-investigator
Danilo	Catelli	Health Sciences / Human Kinetics	Research Assistant
Kristina	Vogel	Health Sciences / Others	Research Assistant

File Number: H11-15-27

Type of Project: Professor

Title: Muscle Activation and Biomechanical Function of the Lower-Limb Joints Following Total Knee Arthroplasty with Medial-Pivot or Posterior-Stabilized Implants: An Electromyography Study

Renewal Date (mm/dd/yyyy)	Expiry Date (mm/dd/yyyy)	Approval Type
10/15/2021	10/15/2022	Renewal

Special Conditions / Comments:

Revised Informed Consent Form (correction to version date) approved Jan 13, 2022.

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Ottawa Health Science Network Research Ethics Board

Ottawa Hospital Research Institute / Institut de recherche de l'Hôpital d'Ottawa

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University of Ottawa Heart Institute / Institut de cardiologie de l'Université d'Ottawa

Ottawa Health Science Network Research Ethics Board/ Conseil d'éthique de la recherche du Réseau de science de la santé d'Ottawa

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July 9, 2019

Re: Protocol # 20150690-01H Muscle Activation and Biomechanical Function of the Lower-limb Joints Following Total Knee Arthroplasty with Medial-Pivot or Posterior-Stabilized Implants: An Electromyography Study

Thank you for the letter of May 27, 2019 from Kristina Vogel. I am pleased to inform you that the following documentation is approved, effective May 30, 2019:

- Amendment Form, dated May 14, 2019
- Revised Protocol, version 4.0, dated May 14, 2019

Ethical approval remains in effect until October 22, 2019.

The OHSN-REB operates in compliance with, and is constituted in accordance with, the requirements of the Tri-Council Policy Statement: Ethical Conduct for Research Involving Humans (TCPS 2); International Council for Harmonisation of Technical Requirements for Pharmaceuticals for Human Use; Integrated Addendum to ICH E6 (R1): Guideline for Good Clinical Practice E6 (R2) Part C, Division 5 of the Food and Drug Regulations Part 4 of the Natural Health Products Regulations Part 3 of the Medical Devices Regulations and the provisions of the Ontario Personal Health Information Protection Act (PHIPA 2004) and its applicable regulations. OHSN-REB is qualified through the CTO REB Qualification Program and is registered with the U.S. Department of Health and Human Services (DHHS) Office for Human Research Protection (OHRP).

Yours sincerely,

Chairman
Ottawa Health Science Network Research Ethics Board

/HMc

8.2 Knee Injury and Osteoarthritis Outcome Score - KOOS

Knee Injury and Osteoarthritis Outcome Score (KOOS)

Source: Roos EM, Roos HP, Lohmander LS, Ekdahl C, Beynnon BD. Knee Injury and Osteoarthritis Outcome Score (KOOS)--development of a self-administered outcome measure. *J Orthop Sports Phys Ther*. 1998 Aug;28(2):88-96.

The Knee Injury and Osteoarthritis Outcome Score (KOOS) is a questionnaire designed to assess short and long-term patient-relevant outcomes following knee injury. The KOOS is self-administered and assesses five outcomes: pain, symptoms, activities of daily living, sport and recreation function, and knee-related quality of life. The KOOS meets basic criteria of outcome measures and can be used to evaluate the course of knee injury and treatment outcome. KOOS is patient-administered, the format is user-friendly and it takes about 10 minutes to fill out.

Scoring instructions

The KOOS's five patient-relevant dimensions are scored separately: Pain (nine items); Symptoms (seven items); ADL Function (17 items); Sport and Recreation Function (five items); Quality of Life (four items). A Likert scale is used and all items have five possible answer options scored from 0 (No problems) to 4 (Extreme problems) and each of the five scores is calculated as the sum of the items included.

Interpretation of scores

Scores are transformed to a 0–100 scale, with zero representing extreme knee problems and 100 representing no knee problems as common in orthopaedic scales and generic measures. Scores between 0 and 100 represent the percentage of total possible score achieved.

Knee Injury and Osteoarthritis Outcome Score (KOOS)

Pain

P1 How often is your knee painful?	☐ Never	☐ Monthly	☐ Weekly	☐ Daily	☐ Always

What degree of pain have you experienced the last week when…?

P2 Twisting/pivoting on your knee	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
P3 Straightening knee fully	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
P4 Bending knee fully	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
P5 Walking on flat surface	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
P6 Going up or down stairs	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
P7 At night while in bed	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
P8 Sitting or lying	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
P9 Standing upright	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme

Symptoms

Sy1 How severe is your knee stiffness after first wakening in the morning?	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
Sy2 How severe is your knee stiffness after sitting, lying, or resting later in the day?	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
Sy3 Do you have swelling in your knee?	☐ Never	☐ Rarely	☐ Sometimes	☐ Often	☐ Always
Sy4 Do you feel grinding, hear clicking or any other type of noise when your knee moves?	☐ Never	☐ Rarely	☐ Sometimes	☐ Often	☐ Always
Sy5 Does your knee catch or hang up when moving?	☐ Never	☐ Rarely	☐ Sometimes	☐ Often	☐ Always
Sy6 Can you straighten your knee fully?	☐ Always	☐ Often	☐ Sometimes	☐ Rarely	☐ Never
Sy7 Can you bend your knee fully?	☐ Always	☐ Often	☐ Sometimes	☐ Rarely	☐ Never

Activities of daily living

What difficulty have you experienced the last week…?

A1 Descending	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A2 Ascending stairs	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A3 Rising from sitting	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A4 Standing	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A5 Bending to floor/picking up an object	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A6 Walking on flat surface	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A7 Getting in/out of car	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A8 Going shopping	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A9 Putting on socks/stockings	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A10 Rising from bed	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A11 Taking off socks/stockings	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A12 Lying in bed (turning over, maintaining knee position)	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A13 Getting in/out of bath	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A14 Sitting	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A15 Getting on/off toilet	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A16 Heavy domestic duties (shovelling, scrubbing floors, etc)	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
A17 Light domestic duties (cooking, dusting, etc)	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme

Sport and recreation function

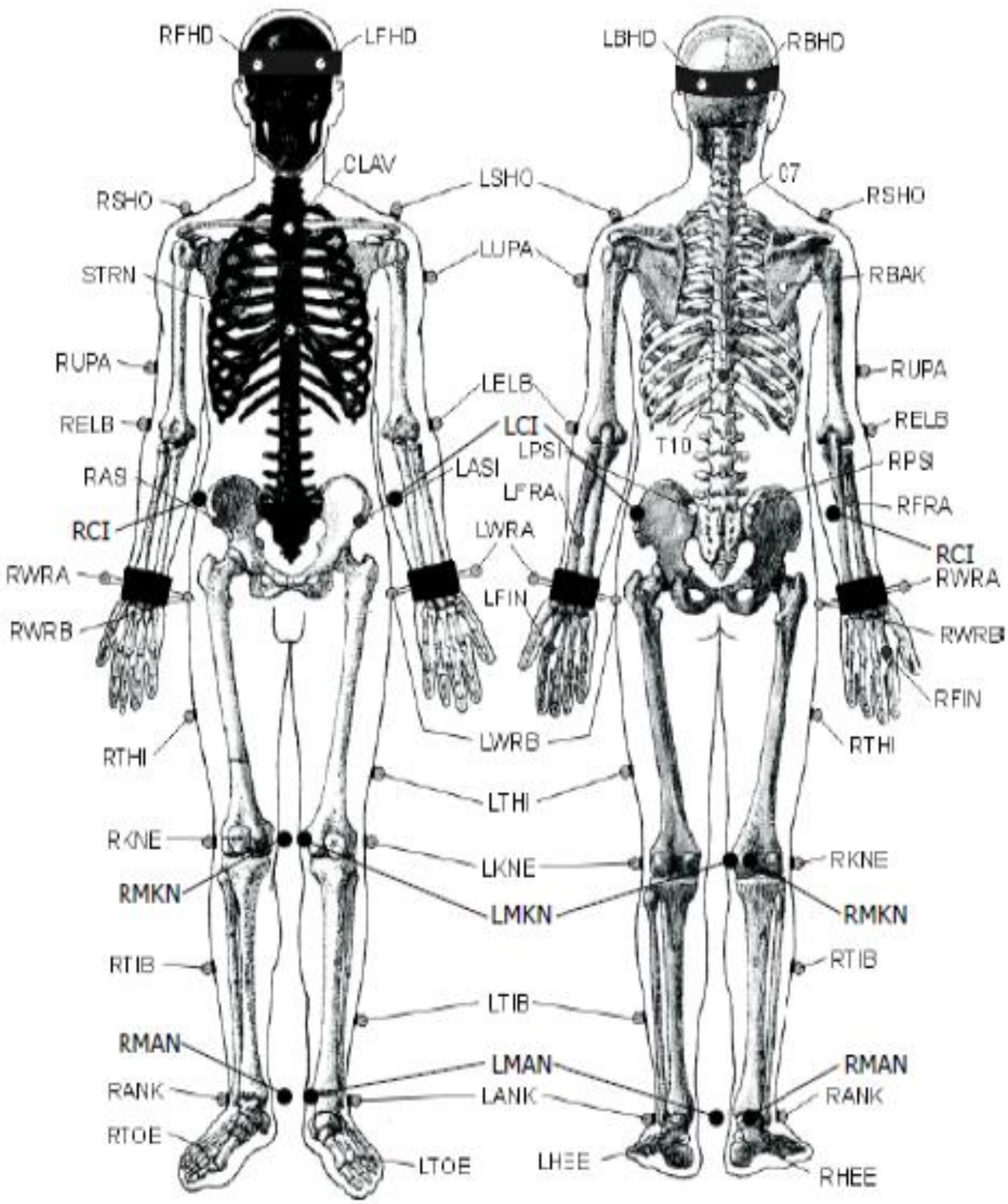
What difficulty have you experienced the last week…?

Sp1 Squatting	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
Sp2 Running	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
Sp3 Jumping	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
Sp4 Turning/twisting on your injured knee	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme
Sp5 Kneeling	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme

Knee-related quality of life

Q1 How often are you aware of your knee problems?	☐ Never	☐ Monthly	☐ Weekly	☐ Daily	☐ Always
Q2 Have you modified your lifestyle to avoid potentially damaging activities to your knee?	☐ Not at all	☐ Mildly	☐ Moderately	☐ Severely	☐ Totally
Q3 How troubled are you with lack of confidence in your knee?	☐ Not at all	☐ Mildly	☐ Moderately	☐ Severely	☐ Totally
Q4 In general, how much difficulty do you have with your knee?	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ Extreme

8.3 University of Ottawa Motion Analysis Model - UOMAM



Head	
LFHD & RFHD	Left & Right Temple
LBHD & RBHB	Left & Right Back of Head
Torso	
C7	7th Cervical Vertebrae
T10	10th Thoracic Vertebrae
CLAV	Jugular Notch
STRN	Xiphoid Process
RBAK	Middle of Right Scapula
Arms	
LSHO & RSHO	Left & Right Acromio-Clavicular Joint
LUPA & RUPA	Left & Right Upper Arm
LELB & RELB	Left & Right Lateral Epicondyle
LFRA & RFRA	Left & Right Forearm
LWRA & RWRA	Left & Right Wrist Bar Thumb Side
LWRB & RWRB	Left & Right Wrist Bar Pinkie Side
LFIN & RFIN	Left & Right Dorsum of the Hand Head of 2nd Metacarpal
Pelvis	
LASI & RASI	Left & Right Anterior Superior Iliac Crest
LPSI & RPSI	Left & Right Posterior Superior Iliac Crest
LCI & RCI	Left & Right Iliac Crest
Legs	
LTHI & RTHI	Left & Right Lateral Thigh
LMKN & RMKN	Left & Right Medial Epicondyle of the Knee
LKNE & RKNE	Left & Right Lateral Epicondyle of the Knee
LTIB & RTIB	Left & Right Lateral Shank
Feet	
LANK & RANK	Left & Right Lateral Malleolus
LMAN & RMAN	Left & Right Medial Malleolus
LTOE & RTOE	Left & Right 2nd Metatarsal Head of Foot
LHEE & RHEE	Left & Right Posterior Calcaneus

9 List of Contributions

9.1 Journals

9.1.1 Published

1. Kowalski, E., Catelli, D. S., & Lamontagne, M. (2019). Side does not matter in healthy young and older individuals - Examining the importance of how we match limbs during gait studies. Gait & posture, 67, 133–136. https://doi.org/10.1016/j.gaitpost.2018.10.008
2. Kowalski, E., Catelli, D. S., & Lamontagne, M. (2021). A waveform test for variance inequality, with a comparison of ground reaction force during walking in younger vs. older adults. Journal of biomechanics, 127, 110657. https://doi.org/10.1016/j.jbiomech.2021.110657
3. Kowalski, E., Catelli, D. S., & Lamontagne, M. (2022). Gait variability between younger and older adults: An equality of variance analysis. Gait & posture, 95, 176–182. https://doi.org/10.1016/j.gaitpost.2022.04.022

9.1.2 Submitted

4. Kowalski, E., Catelli, D. S., Dervin, G., & Lamontagne, M. Knee biomechanics before and after a total knee arthroplasty with either a medial ball-and-socket or posterior stabilized implant. 2022. Submitted for publication to *Gait & Posture*, *October 2022*.
5. Kowalski, E., Catelli, D. S., Dervin, G., & Lamontagne, M. Knee biomechanics variability before and after total knee arthroplasty: a prospective study comparing patients with different implant designs. 2023. Submitted for publication to *The Knee, April 2023*.
6. Kowalski, E., Catelli, D. S., Dervin, G., & Lamontagne, M. Total knee arthroplasty recipients descend a ramp with a cautious gait pattern twelve months after surgery independent of implant design. 2023. Submitted for publication to *Journal of Orthopedic Research, April 2023*.
7. Kowalski, E., Pelegrinelli, A.R.M., Ryan, N., Dervin, G., & Lamontagne, M. Muscle activity and biomechanics while descending a staircase after total knee arthroplasty: A study comparing different posterior stabilized and medial congruent designs. 2023. Submitted for publication to *Journal of Arthroplasty, April 2023*.

9.1.3 Additional

8. Kowalski, E., Pelegrinelli, A.R.M., Catelli, D.S., Dervin, G., & Lamontagne, M. Medial and lateral knee contact forces and muscle forces during sit-to-stand in patients one year after unilateral total knee arthroplasty. 2023. Prepared for submission to *Clinical Biomechanics*.

9.1.4 Collaborations

9. Lukas, K. J., Verhaegen, J. C. F., Livock, H., Kowalski, E., Phan, P., & Grammatopoulos, G. (2023). The effect of ethnicity on the age-related changes of spinopelvic characteristics: a systematic review. Bone & joint research, 12(4), 231–244. https://doi.org/10.1302/2046-3758.124.BJR-2022-0335.R1
10. Pelegrinelli, A. R. M., Kowalski, E., Ryan, N. S., Moura, F. A., & Lamontagne, M. (2022). Lower limb inter-joint coordination in individuals with osteoarthritis before and after a total knee arthroplasty. Clinical biomechanics (Bristol, Avon), 100, 105806. https://doi.org/10.1016/j.clinbiomech.2022.105806
11. Kowalski, E., Catelli, D. S., & Lamontagne, M. (2021). Comparing the Accuracy of Visual and Computerized Onset Detection Methods on Simulated Electromyography Signals with Varying Signal-to-Noise Ratios. Journal of functional morphology and kinesiology, 6(3), 70. https://doi.org/10.3390/jfmk6030070
12. Catelli, D. S., Kowalski, E., Beaulé, P. E., & Lamontagne, M. (2021). Muscle and Hip Contact Forces in Asymptomatic Men With Cam Morphology During Deep Squat. Frontiers in sports and active living, 3, 716626. https://doi.org/10.3389/fspor.2021.716626
13. Catelli, D. S., Ng, K. C. G., Wesseling, M., Kowalski, E., Jonkers, I., Beaulé, P. E., & Lamontagne, M. (2020). Hip Muscle Forces and Contact Loading During Squatting After Cam-Type FAI Surgery. The Journal of bone and joint surgery. American volume, 102(Suppl 2), 34–42. https://doi.org/10.2106/JBJS.20.00078
14. Catelli, D. S., Ng, K. C. G., Kowalski, E., Beaulé, P. E., & Lamontagne, M. (2019). Modified gait patterns due to cam FAI syndrome remain unchanged after surgery. Gait & posture, 72, 135–141. https://doi.org/10.1016/j.gaitpost.2019.06.003
15. Catelli, D. S., Kowalski, E., Beaulé, P. E., & Lamontagne, M. (2019). Increased pelvic mobility and altered hip muscles contraction patterns: two-year follow-up cam-FAIS corrective surgery. Journal of hip preservation surgery, 6(2), 140–148. https://doi.org/10.1093/jhps/hnz019
16. Pelegrinelli, A. R. M., Kowalski, E., Ryan, N. S., Catelli, D.S., Lamontagne, M. & Moura, F. A. Tibiofemoral contact forces and muscle demands in knee osteoarthritis patients during the stand-to-sit and sit-to-stand tasks. Prepared for submission to *Journal of Biomechanics*.

9.2 Conferences

9.2.1 Published

1. Pelegrinelli, A. R. M., Kowalski, E., Catelli, D. S., Lamontagne, M., & Moura, F. A. (2023) Knee contact forces differences in knee osteoarthritis patients during the sit-to-stand task. Brazilian Journal of Motor Behavior. Volume 17, Supplement
2. Kowalski, E., Dervin, G., Lamontagne, M. (2022) Patient-reported outcome measures and biomechanical indicators of success in total knee arthroplasty. Orthopaedic Proceedings, 104 (Supp_12), 17-17
3. Kowalski, E., Catelli, D., Lamontagne, M., Dervin, G., (2021) Gait variability before and after total knee arthroplasty: a comparison of medial pivot and posterior stabilized implants. Orthopaedic Proceedings, 103 (SUPP_13) 90-90
4. Kowalski, E., Lamontagne, M., Catelli, D. Beaule, P.E., (2020) Surgical correction for femoroacetabular impingement: motion analysis during stairs task at two-year follow-up. Orthopaedic Proceedings, 102 (SUPP_6), 109-109
5. Lamontagne, M., Kowalski, E., Galmiche, R., Dervin, G. (2019) Muscle activity in TKA patients while ascending and descending a ramp. Bone and Joint Journal, Orthopaedic Proceedings Supplement 101-B(5)
6. Catelli, D.S., Ng, G.K.C., Kowalski, E., Beaulé, P., Lamontagne, M. (2018) Hip muscle and contact forces in post-surgical cam femoroacetabular impingement during gait. Orthopaedic Proceedings, 100 (SUPP_15), 117-117

9.2.2 Accepted

1. Ryan, N., Kowalski, E., Grammatopoulos, G., Lamontagne, M. (2023) The effect of hip capsule preservation on gait biomechanics following total hip arthroplasty. Orthopedic Research Society, Dallas, TX.
2. Kowalski, E., Pelegrinelli, A. R. M., Ryan, N., Dervin, G., Lamontagne, M. (2023) Comparing muscle forces and knee contact forces between implant designs after total knee arthroplasty – preliminary outcomes. Orthopedic Research Society, Dallas, TX.
3. Kowalski, E., Catelli, D., Dervin, G. Lamontagne, M. (2022) Biomechanical comparison of medial pivot and posterior stabilized implants during ramp and stair descent following total knee arthroplasty. International Society for Technology in Arthroplasty 2022: The 33rd International Congress, Maui, HI
4. Pelegrinelli, A.R.M., Kowalski, E., Dervin, G., Moura, F.A., Lamontagne, M. (2022) The inter-segmental coordination and knee motion during gait in total knee arthroplasty. North American Congress on Biomechanics 2022, Ottawa, Canada
5. Kowalski, E., Pelegrinelli, A.R.M., Ryan, N., Dervin, G., Lamontagne, M. (2022) Electromyographic activity of knee muscles during ramp and stair descent in patients

after total knee arthroplasty: A study comparing different implant designs. Ottawa Musculoskeletal Symposium, Ottawa, Canada

6. Kowalski, E., Pelegrinelli, A.R.M., Ryan, N., Lamontagne, M. (2022) Subject-specific modelling of muscle force during gait in total knee arthroplasty. North American Congress on Biomechanics 2022, Ottawa, Canada
7. Pelegrinelli, A.R.M., Kowalski, E., Ryan, N., Dervin, G., Moura, F.A., Lamontagne, M. (2022) Differences in lower limb intersegmental coordination in total knee arthroplasty during the gait. ICORS 2022, World congress of orthopaedic research, Edinburgh, Scotland.
8. Kowalski, E., Pelegrinelli, A.R.M., Ryan, N., Dervin, G., Lamontagne, M. (2022) Biomechanical indicators of success after total knee arthroplasty predicted with pre-operative clinical measurements. ICORS 2022, World congress of orthopaedic research, Edinburgh, Scotland.
9. Catelli, D., Kowalski, E., Beaulé, P.E., Lamontagne, M. (2022) Squat kinematic variability in asymptomatic cam-type femoroacetabular impingement. 24th EFORT Annual Congress, Vienna, AT
10. Lamontagne, M., Kowalski, E., Arora, I., Pelegrinelli, A.R.M., Catelli, D., Dervin, G. (2022) Muscle activity during ramp walking after a total knee arthroplasty with either a medial pivot or posterior stabilized implant: preliminary results. 24th EFORT Annual Congress, Vienna, AT
11. Kowalski, E., Catelli, D., Dervin, G., Lamontagne, M. (2022). Knee biomechanics variance inequality following TKA with either a medial pivot or posterior stabilized implant: a randomized control study. 2022 Orthopedic Research Society Annual Meeting, Tampa Bay, FL
12. Kowalski, E., Catelli, D., Lamontagne, M. (2021) Visual detection on simulated electromyography signals with varying signal-to-noise ratios: A training tool to enhance onset identification. International Society of Biomechanics Conference
13. Kowalski, E., Catelli, D., Lamontagne, M., Dervin, G. (2021) Gait Variability Analysis Before And After A Total Knee Arthroplasty: Medial Stabilized Versus Posterior Stabilized Design. 2nd Virtual EFORT Congress
14. Lamontagne, M., Kowalski, E., Catelli, D., Dervin, G. (2021) Muscle activation during walking on inclined surface in total knee arthroplasty patients: medial pivot compared to posterior stabilized implants. Orthopedic Research Society 2021 Annual Meeting, Virtual.
15. Lamontagne, M., Kowalski, E., Dervin, G. (2020) Knee biomechanics during ramp walking after a total knee arthroplasty with either a medial pivot or posterior stabilized implant. 1st Virtual EFORT Congress
16. Lamontagne, M., Kowalski, E., Catelli, D., Grammatopoulus, G., Beaulé, P. (2020) The effect of lower limb length and hip offset differences, and surgical approaches on gait mechanics in total hip arthroplasty. 2020 Orthopedic Research Society Annual Meeting, Pheonix, AZ
17. Lamontagne, M., Kowalski, E., Catelli, D., Grammatopoulus, G., Beaulé, P. (2020) Is it surgical approach or restoration of surgical parameters that lead to improved gait

following THA? A motion analysis study. 2020 American Academy of Orthopaedic Surgeons Annual Meeting, Orlando, FL

18. Galmiche, R., Kowalski, E., Lamontagne, M., Beaulé, P. (2019) The effect of reconstruction parameters and surgical approaches after total hip arthroplasty on hip biomechanical function during gait. Orthopedic Research Society 2019 Annual Meeting, Austin, TX
19. Catelli, D., Kowalski, E., Cruaud, W., Lamontagne, M., Beaulé, P. (2019) An in-silico muscle forces comparison between hip dislocation and arthroscopic approaches for FAI corrective surgery during a deep squatting task. Orthopedic Research Society 2019 Annual Meeting, Austin, TX
20. Kowalski, E., Dervin, G., Lamontagne, M. (2019) Knee biomechanics during stair climbing after a total knee arthroplasty with either a medial pivot or posterior stabilized implant. 20th European Federation of National Associations of Orthopedics and Traumatology Congress 2019, Lisbon, Portugal
21. Kowalski, E., Catelli, D., Lamontagne, M. (2019) Gait biomechanics after total hip arthroplasty: using statistical parametric mapping to identify differences between various surgical approaches. 20th European Federation of National Associations of Orthopedics and Traumatology Congress 2019, Lisbon, Portugal
22. Catelli, D., Wesseling, M. Ng, G., Kowalski, E., Jonkers, I., Beaulé, P., Lamontagne, M. (2019) Deep Squat 2-Years after Hip Preservation of Cam Femoroacetabular Impingement: Muscle and Hip Contact Forces Accessed by Computational Modelling. 20th European Federation of National Associations of Orthopedics and Traumatology Congress 2019, Lisbon, Portugal
23. Lamotagne, M., Catelli, D., Kowalski, E., Beaulé, P. (2019) Surgical Correction for FAI: Motion Analysis during Stairs Task at 2-Year Follow-Up. 2nd International Combined Meeting of Orthopaedic Research Societies, Montreal, Canada