DOES GAIT RETRAINING HAVE THE POTENTIAL TO SLOW OA DEVELOPMENT AND PROLONG THE BENEFITS OF KNEE REALIGNMENT SURGERY?

Thesis submitted in fulfilment of the requirement for the degree of Doctor of Philosophy in Biomedical Engineering

By

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DECLARATION AND STATEMENTS

DECLARATION

This work has not been submitted in substance for any other degree or award at this or any other university or place of learning, nor is being submitted concurrently in candidature for any degree or other award.

Signed ……………………….…... (**Jake Bowd**) Date: 18th January 2022

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“A person who never made a mistake never tried anything new.”

Albert Einstein

ABSTRACT

Medial knee osteoarthritis (mKOA) and varus knee alignment is associated with altered knee loading. Patients may be offered high tibial osteotomy surgery (HTO) to reduce medial compartment knee joint loading and pain. Gait retraining to alter dynamic knee alignment is proposed as a non-invasive approach to offload the medial compartment. This approach has the potential to provide symptomatic relief prior to surgery and optimise gait along with knee loading post-HTO. This thesis explored the biomechanical gait changes pre-to-post HTO with (i) standard gait to define a baseline change in knee loading following surgery and (ii) with an altered gait style (wide stance, medial thrust, toe out) to provide recommendations that will feed into future gait retraining development.

A systematic review revealed a paucity of information on the consequences of altering gait for patients with mKOA and associated effects on adjacent joints. Gait biomechanics and knee loading features were determined using Inverse Dynamic Modelling where primary metrics of medial compartment knee loading were the two external knee adduction moment peaks (EKAM1 and EKAM2) and predictive musculoskeletal modelling via the Concurrent Optimisation of Muscle Activations and Kinematics (COMAK) to quantify the magnitude and location of internal knee joint contact forces. In addition to the above, Principal Component Analysis and the Cardiff Classifier was used to define the baseline change in gait and knee loading because of HTO surgery.

HTO surgery resulted in biomechanical changes in all three planes at the hip, knee, and ankle joints. Post-HTO, medial knee loading was reduced by ~10% and ~16% when assessed using COMAK at both peaks in stance. HTO surgery reduced the classification belief in mKOA for 20 out of 22 patients, indicating biomechanical improvement occurs due to realignment surgery.

Toe out gait pre- and post-HTO reduced EKAM2 (~12% pre and -11% post) and second half of stance internal medial loading peak (~12% pre and ~7% post). Pre-HTO, adopting a toe out gait increased medial internal joint loading in early stance by ~6%.

Wide stance gait also reduced medial compartment loading in late stance when adopted pre or post HTO (10-13% reduction in EKAM2 and ~7% reduction in medial internal tibiofemoral joint loading).

Medial thrust gait reduced EKAM pre- and post-HTO. The reductions in EKAM were met with significant alterations at the hip and ankle joint moments and kinematics. Contrary to EKAM, medial thrust resulted in a reduced gait speed and conflicting findings with predictive internal joint loading.

This study is the first to investigate the influence of gait alteration on medial compartment loading pre-to-post HTO surgery. It reveals a set of novel clinically important findings and provides preliminary data supporting future development of patient specific gait retraining aimed at clinical translation.

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LIST OF ABBREVIATIONS

Name	Acronym
Activities Of Daily Living	ADL
Biomechanics And Bioengineering Research Centre Versus Arthritis	BBRCVA
Body Mass Index	BMI
Centre Of Mass	COM
Centre Of Pressure	COP
Concurrent Optimisation of Muscle and Secondary Kinematics	COMAK
Confidence Intervals	CI
Degree Of Freedom	DOF
External Knee Adduction Moment	EKAM
Finite Element Analysis	FEA
Foot Progression Angle	FPA
Ground Reaction Forces	GRF
High Tibial Osteotomy	HTO
Hip-Knee-Ankle	HKA
Human Motion Analysis	HMA
Kellgren And Lawrence System for Classification	KL
Knee Angular Adduction Impulse	KAAI
Knee Contact Forces	KCF
Lateral Wedge Insoles	LWI
Magnetic Resonance Imaging	MRI
Medial Knee Contact Force	mKCF
Medial Knee Osteoarthritis	mKOA
Musculoskeletal	MSK
Musculoskeletal Biomechanics Research Facility	MSKBRF
National Health Service	NHS
National Institute for Health and Clinical Excellence	NICE
Non-Steroidal Anti-Inflammatory Drugs	NSAIDs
Osteoarthritis	OA
Osteoarthritis Research Society International	OARSI
Percentage Body Weight Multiplied by Height	%BW×HT
Principal Component Analysis	PCA
Quality Of Life	QOL
Range Of Motion	ROM
Standard Deviation	std
Standard Mean Difference	SMD
Total Knee Arthroplasty	TKA
Total Knee Replacement	TKR
Uni-Compartment Knee Replacements	UKR
United Kingdom Knee Osteotomy Registry	UKKOR
World Health Organisation	WHO

CHAPTER 1: INTRODUCTION

1.1 Scope

This PhD focuses on the potential benefit of a non-surgical intervention (alteration of gait) compared to a surgical intervention (High Tibial Osteotomy (HTO)) to reduce medial knee joint loading on a cohort of individuals with medial knee osteoarthritis (mKOA). There are various non-surgical interventions proposed to reduce medial knee joint loading, including: orthosis, knee braces, walking canes and gait retraining. Gait alterations/retraining aimed at altering joint loading is the cheapest approach which has shown promise (Simic *et al.*, 2011; Richards *et al.*, 2017).

This PhD has three overall focuses. First, to evaluate the potential biomechanical merits of undergoing an HTO. This will establish whether surgery offloads the diseased part of the knee, which could potentially prevent further deterioration of the knee joint. Second, to evaluate the potential biomechanical merits of an altered gait intervention prior to undergoing an HTO. This will establish whether gait alterations prior to surgery offloads the diseased part of the knee, which could potentially prevent further deterioration of the knee joint. If this is the case, there is potential for gait alterations to offer relief to patients whilst they await surgery or offer an alternative non-surgical option in some cases. Third, to evaluate the biomechanical merits of gait alterations following HTO in view of prolonging the benefits of the HTO surgery; assessed at 12 months post-HTO.

The use of Human Motion Analysis (HMA) to calculate lower limb biomechanics during gait has been adopted by numerous studies to characterise biomechanical changes during mKOA disease progression and changes following HTO (Lind et al., 2013; McClelland et al., 2007; Tanamas et al., 2009; Whatling et al., 2019). This has been applied further through implementing gait retraining strategies aimed to reduce joint loading (Simic *et al.*, 2011). However, the following three questions remain unanswered; (a) whether individuals with mKOA and varus deformity could benefit from altered gait strategies whilst awaiting surgery? (b) whether the altered gait strategies can be used instead of HTO? (c) whether an altered gait could be used to compliment the surgery further?

Novelty in this thesis is multifaceted. First, informing a non-invasive intervention that reduces medial knee compartment loading indicating that it may be able to provide (a) relief from deterioration of knee joint loading whilst waiting for a HTO, or (b) prolong the benefits of surgery which may delay the need for further surgical interventions. Second, to quantify the biomechanical changes at the hip and ankle joints due to the interventions. Third, applying state-of-the-art modelling techniques to estimate internal tibiofemoral joint contact forces to this cohort of individuals for the very first time.

This PhD uses the cumulation of data that has been collected between 2009-2019 as part of the Biomechanics and Bioengineering Research Centre Versus Arthritis (BBRCVA) (*Versus Arthritis was formerly named Arthritis Research UK*) based at Cardiff University, UK. Since receiving their Research Passport and being trained on the protocol in March 2018, the author of this thesis collected 20 out of the 24 motion capture visits that are included in this thesis. The sessions that were not collect by the author of this thesis was due to their 6 weeks visit to KU Leuven. The BBRCVA is a multidisciplinary Centre of Excellence investigating how everyday movements can influence the health of joints and consequently disease onset. As part of the Centre, this PhD aims to answer a sub-section of the aims and objectives of Work-Package 2, which is titled '*High Tibial Osteotomy: An in-vivo human knee osteoarthritis model and a route to optimising joint realignment*'.

Due to the interdisciplinary nature of the BBRCVA and the objectives and milestones within, the work afforded from this thesis also directly contributes to Work-Package 2 study 2.1 which has a research hypothesis '*Structural joint changes correlate with biomechanical changes associated with joint realignment following HTO'*. This PhD was funded to directly address the questions related to the potential benefits of altered gait pre- and post-HTO as an alternative or to compliment interventions, respectively. This PhD is specifically focused on study 2.2 of Work-Package 2, which has a research hypothesis of '*gait retraining has the potential to slow OA development and prolong the benefits of HTO*'.

The research group (*Musculoskeletal Biomechanics Research Facility (MSKBRF), School of Engineering, Cardiff University*) has collected HMA on individuals pre- and post-HTO since 2009 and has led to international conference proceedings and peer-reviewed publications (Bowd et al., 2019; Holt et al., 2017, 2016; Whatling et al., 2019; Whelton et al., 2017). From this work it is evident that HTO has been shown to restore frontal and transverse knee loading to unaltered non-pathological levels (Whatling *et al.*, 2019) as well as potentially influencing biomarkers of pain, inflammation, and pathology (Holt *et al.*, 2017). These findings suggest that the current standard of treatment for varus deformed mKOA (i.e., HTO) lowers medial compartment knee loading as measured by surrogate measurements of joint loading. However, HTO is an invasive operation that requires the patient to spend a considerable amount of time rehabilitating and having time off work to recover. In a young and active population with mKOA or varus deformity, an HTO allows patients to return to work. However, patients with high-intensity occupations may be absent from work longer than those with lesser physically demanding occupations (Agarwalla *et al.*, 2019). However, one major issue after HTO is the recurrence of varus deformity despite a successful operation (Lee, Lee and Lee, 2018). HTO is not a definitive treatment option as nearly 40% of patients underwent knee arthroplasty by ~6 years post-operatively (Agarwalla *et al.*, 2019). Therefore, a less invasive and cheaper alternative would be of great merit. The cheapest potential alternative would be simply altering an individual's gait to reduce medial compartment knee joint loading. However, there is less understanding on the role of altered gait in reducing knee joint loading before and after HTO. Additionally, there is a lack of understanding of how these changes biomechanically effect adjacent joints to the knee.

Younger patients with mild to moderate mKOA can find themselves in a treatment gap since they are not candidates for a replacement joint. In this case, orthopaedic surgeons may decide to wait for the patients' symptoms to progress, or they may be offered HTO surgery. In the newly established United Kingdom Knee Osteotomy Registry (UKKOR), out of 1,776 cases of osteotomy surgery registered between 1 December 2014 and 1 December 2017, 35% underwent surgery and 65% were either waiting for surgery or had no operative data entered on the registry (Palmer *et al.*, 2018). Importantly, surgeries have been delayed due to COVID-19 and patients are therefore not receiving symptomatic relief. As explained in **Chapter 2**, changing the way someone walks alters their knee loading which may reduce pain. In addition, offloading the diseased side of the knee may arrest the further development of mKOA (Andriacchi and Mündermann, 2006), associated tissue damage, inflammation, and pain. Therefore, gait retraining would in theory, provide symptomatic relief leading up to surgery by distributing knee loading towards the healthy side of the joint.

The three different gait styles assessed in this thesis were determined at the start of the BBRCVA back in 2009 and were selected on the evidence and recommendations at the time of inception of the BBRCVA. The gait styles that were chosen for investigation were: (1) wide stance gait style, (2) medial thrust gait style, and (3) toe out gait style, which were assessed on patients just prior undergoing surgery (as close to the operation date as feasibly possible) and at ~12 months post-HTO. These altered gait styles were chosen as they are proposed to elicit different mechanisms to reduce medial joint loading which are explained in **Chapter 2**. This PhD utilises this complete dataset for the first time to generate recommendations that will inform future development of gait retraining regimes.

Another novel aspect of this thesis is the implementation of a state-of-the-art computational modelling technique used in this specific cohort for this first time to establish whether an altered gait has the potential to slow mKOA development and prolong the benefits of HTO. This model (Concurrent Optimisation of Muscle and Secondary Kinematics (COMAK) (Lenhart *et al.*, 2015; Smith *et al.*, 2018)) estimates tibiofemoral joint compartment (medial, lateral, total) contact forces and pressure distributions and has already shown the potential effects of varying degrees of valgus/varus deformity at the tibiofemoral joint (van Rossom *et al.*, 2019). However, van Rossom et al. assessed healthy individuals and computationally altered the coronal and transverse planes alignment. Therefore, the biomechanics of the lower limb joints were that of healthy individuals and not those with mKOA.

1.2 Aims and objectives

The primary aim of this research was to explore the biomechanical consequences of trying to lower medial knee joint loading for patients before and 12 months post-HTO through performing three separate altered gait styles, along with quantifying their effects on the adjacent joints to the knee. Two different biomechanical approaches to modelling human movement were used to fulfil the aims. **Chapter 4** outlines the methodology undertaken within this thesis. This PhD had the following four objectives:

Objective 1 (**Chapter 3**): A systematic review to address the following question: Does gait retraining have the potential to reduce medial compartmental loading in individuals with mKOA while not adversely affecting the other lower limb joints?

Objective 1 Addressed the following:

Whether gait styles and gait retraining can reduce medial knee loading as assessed by first and second peak EKAM.

Consequences of gait retraining on the biomechanics of the ankle and hip as well as trunk and pelvis biomechanics.

Objective 2 (**Chapter 5**): Quantified the biomechanical differences of knee joint loading between the following three groups: non-pathological healthy cohort (control), pre-HTO unaltered level gait cohort, and a 12-month post-HTO unaltered level gait cohort.

Objective 2 provided data that can be used as biomechanical targets for gait retraining. The following three analyses were performed:

First, lower limb biomechanical differences between the control group, pre-HTO unaltered level gait, and 12 months post-HTO unaltered level gait using a Visual 3D (C-Motion) command pipeline. The pipeline was designed in-house to extract discrete metrics to understand knee joint loading in the form of external moments as well as joint rotations. Additionally, hip and ankle external moments and rotations are assessed to determine the consequences of HTO surgery on the adjacent joints to the knee.

Second, an enhanced musculoskeletal model was used to predict internal joint loading differences between the control group, pre-HTO unaltered level gait, and 12 months post-HTO.

Third, waveform analysis using Principal Component Analysis (PCA) and the Cardiff Classifier were used to better understand biomechanical factors affecting varus deformity of the knee and fundamental differences between the control group, pre-HTO unaltered level gait, and 12 months post-HTO groups.

Objective 3 (**Chapter 6**): Quantified the biomechanical differences of knee joint loading between the control group, pre-HTO unaltered level gait, and pre-HTO altered gait styles.

Objective 3 was addressed using the following two approaches:

First, lower limb biomechanical differences were determined between the control group, pre-HTO unaltered level gait, and pre-HTO altered gait styles in the form of a toe out gait, wide stance gait and a medial thrust gait. Visual 3D (C-Motion) was used to extract discrete metrics of external joint moments to understand knee joint loading. The consequences of altering gait on the rotations and moments of adjacent joints to the knee were also determined.

Second, an enhanced musculoskeletal model was used to predict internal joint loading differences between a control group, pre-HTO unaltered level gait, and pre-HTO altered gait styles in the form of a toe out gait, wide stance gait and a medial thrust gait.

Objective 4 (**Chapter 7**): Quantified the biomechanical differences of knee joint loading between post-HTO unaltered level gait and post-HTO altered gait styles.

Objective 4 was addressed using the following two approaches:

First, lower limb biomechanical differences were identified between the control group, post-HTO unaltered level gait, and post-HTO altered gait styles in the form of a toe out gait, wide stance gait and a medial thrust gait. Visual 3D (C-Motion) was used to extract discrete metrics to understand knee joint loading in the form of external moments. Additionally, hip and ankle external moments and rotations were assessed to determine the consequences of altering gait on the adjacent joints to the knee.

Second, an enhanced musculoskeletal model was used to predict internal joint loading differences between the control group, post-HTO unaltered level gait, and post-HTO altered gait styles in the form of a toe out gait, wide stance gait and a medial thrust gait.

In **Chapter 8** the main findings are discussed alongside recommendations for translation to clinical practice as well as identifying the limitations within this thesis before concluding with ideas for future work.

1.3 Motivation

The motivation for this thesis was to make recommendations that have the potential to optimise patient care in terms of therapy and surgery. This PhD will produce recommendations that can be translated to MSK rehabilitation clinic environment and advise future therapy.

This PhD studentship was supported by Versus Arthritis and the work was carried out within the BBRCVA. The interdisciplinary research centre involves close collaborations with surgeons, engineers, biomedical scientists, and physiotherapists to investigate non-pathological joint biomechanics and to determine how this is influenced by arthritis to inform clinical interventions.

CHAPTER 2: LITERATURE REVIEW

In this Chapter, background knowledge linking mKOA, varus malalignment and the mechanical environment is outlined. From which, an overview of potential interventions to reduce medial knee compartment loading is considered for individuals with mKOA, including the current literature with regards to gait retraining and HTO surgery.

This thesis aims to establish whether altering gait has the potential to offload the diseased compartment of the tibiofemoral joint without causing considerable alterations elsewhere at the knee, hip, or ankle joints. Additionally, this thesis aims to establish whether altered gait can support patients following HTO by optimising their biomechanics to complement surgery further.

By the end of this chapter, it will be clear that research assessing altered gait to reduce medial knee joint loading is lacking for individuals with varus deformity and mKOA. By the end of this chapter, it will also be clear that there is a severe lack of understanding of the consequences of altering gait has at the hip and ankle joints. By targeting this research gap in the literature for the use of altered gait styles pre-HTO and post-HTO, the aim and hypothesis of this research thesis will conclude Chapter 2.

2.1 Osteoarthritis as a serious disease

Osteoarthritis (OA) is a painful, disabling disease which can result in significant reductions in an individuals' quality of life (QOL) (Salaffi *et al.*, 2005) which cannot be prevented effectively and so management essentially focuses on alleviating the joint pain (Simic *et al.*, 2011).

In 2016, the Osteoarthritis Research Society International (OARSI) put forward a white paper outlining that OA should be considered as a serious disease to the U.S. Food and Drug Administration (*Osteoarthritis: A serious disease*, 2016). Historically, OA was deemed a simple mechanical disease, with degeneration caused by "wear and tear" of the joint (Mora, Przkora and Cruz-Almeida, 2018). However, this is not supported by the epidemiology of OA (Ayhan, Kesmezacar and Akgun, 2014). While OA has been described as a failure of the joint caused by mechanical factors, it is now understood that other non-mechanical factors also play an important role (Ayhan, Kesmezacar and Akgun, 2014).

Initiation of damage to the articular cartilage can be because of trauma, habitual loading, metabolic factors, or anatomic deformities. In recent years, it has been shown that the aetiology of OA is multifactorial, with contributions from structural and biological pathways.

mKOA represents a complex MSK disorder with multiple genetic, constitutional, and biomechanical risk factors (Chen et al., 2012). In the past, mKOA was thought to be mainly driven by degeneration of the articular cartilage within the synovial joint. However, over time, it has been proven that not only cartilage, but also the subchondral bone, menisci, ligaments, the synovial fluid, muscles, and neural tissues are involved in the complex initiation and progression of the mKOA. OA is, therefore, a whole joint disease rather than simply degeneration of the cartilage, as indicated in Figure 1 below. Consequently, patients complain of joint pain, reduced range of joint movement, stiffness, instability, swelling, muscle weakness, and alterations in proprioception (Hunter, Mcdougall and Keefe, 2008). These symptoms significantly restrict the individual's physical capacity in activities of daily living (ADL), such as walking. This results in loss of independence, reduced QOL and ultimately high health-related costs. There is currently no cure or specific treatment for mKOA (Tanamas *et al.*, 2009). As such identification of risk factors to inform prevention strategies is paramount. There is a strong body of evidence emerging which is highlighting the potential effectiveness of non-invasive interventions in the form of altering an individual's gait with the aim of slowing down the onset and/or the progression of mKOA (Simic *et al.*, 2011; Richards *et al.*, 2017; Shaw *et al.*, 2017).

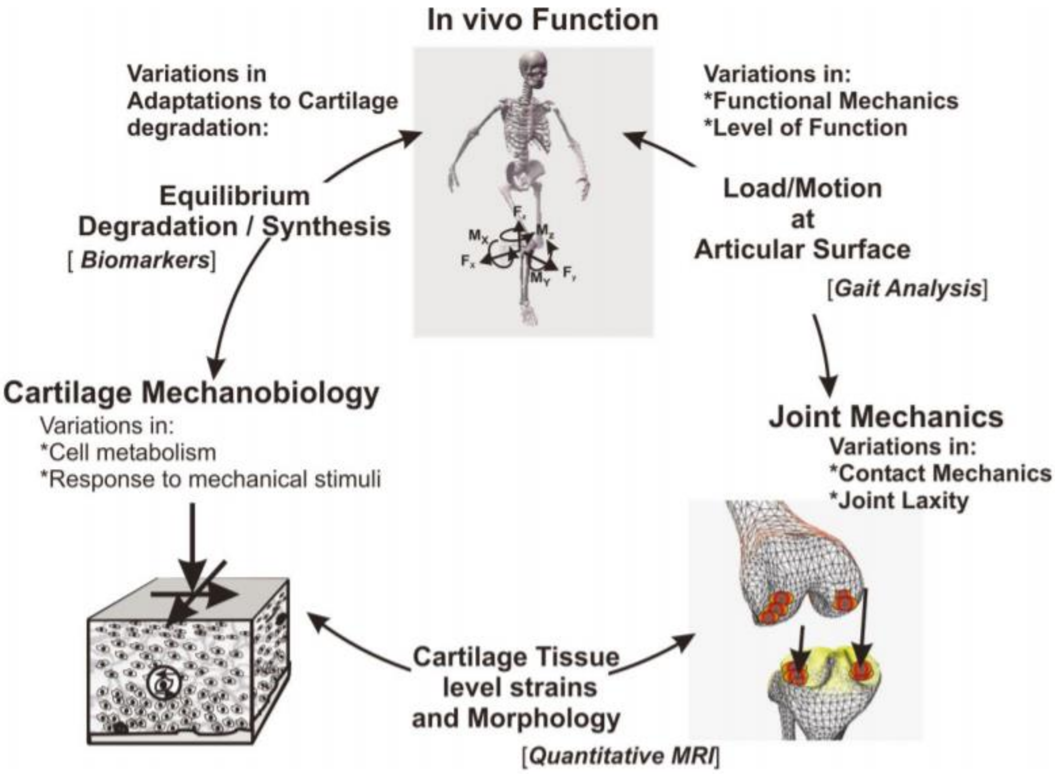


Figure 1 The interrelationship of the different pathways involved in knee osteoarthritis

In-vivo response of articular cartilage to its physical environment requires an integrated view of the problem that considers functional, anatomical, and biological interactions **(Andriacchi *et al.*, 2004)**. Figure extracted from Andriacchi et al. (2004).

mKOA is the most common form of joint disease and disability in older people which has a significantly higher incidence rate in the over 45 years old, as shown in Figure 2, and ranks amongst the top 5 causes of disability (Murray and Lopez, 1997). In England alone, KOA is a leading cause of long-term physical disability affecting over 4 million adults (> 45 years) (Arthritis Research UK, 2018). Newly diagnosed cases of OA in England occur in 9 in 1000 at risk adults each year, which is consistent with other international estimates (Yu *et al.*, 2015) (Figure 3). Surprisingly, there are no published studies regarding direct or indirect costs of KOA in the UK (Chen et al., 2012). However, what is paramount is that such costs are very substantial and are continuing to rise (Chen et al., 2012) as shown in Figure 4 indicating an increasing economic burden to the UK. It is therefore important to establish effective intervention plans to reduce this burden and to improve the QOL for those individuals.

Risk factors of mKOA in young patients are related to an unfavourable biomechanical condition, such as overloading caused by obesity, overuse, and malalignment (Heijink *et al.*, 2012) (Figure 5). Of them, the malalignment of lower limb (as depicted in Figure 5) is postulated to substantially influence load distribution across the articular surface of the knee joint, which is therefore considered as a critical risk factor for mKOA progression (Heijink *et al.*, 2012). In this chapter, varus malalignment is introduced, and the relationship to mKOA and biomechanical loading is outlined (shown in Figure 5). As individuals with varus knee deformity are more susceptible to developing mKOA and once developed, the mKOA environment is exacerbated, it is critical to establish interventions that are effective in realigning the lower limbs; either statically (i.e., surgical correction) or dynamically (i.e., altering gait).

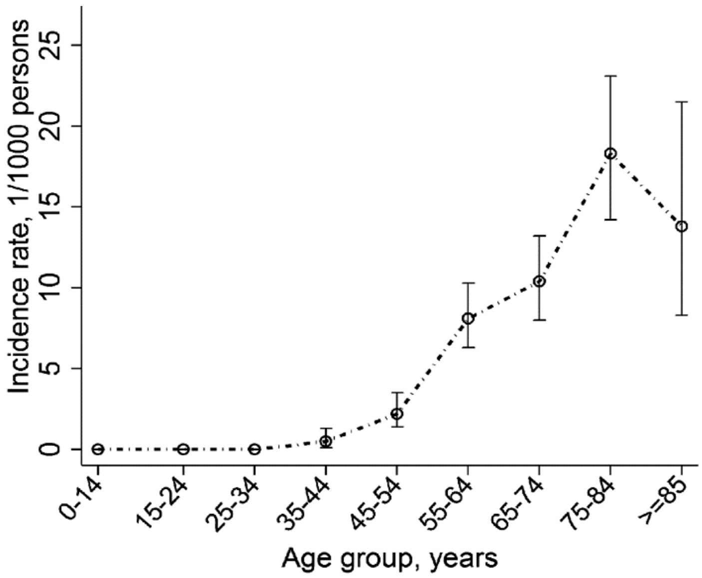


Figure 2 Annual age-specific consultation incidence of knee osteoarthritis in England

Extracted from **(Yu *et al.*, 2015)**

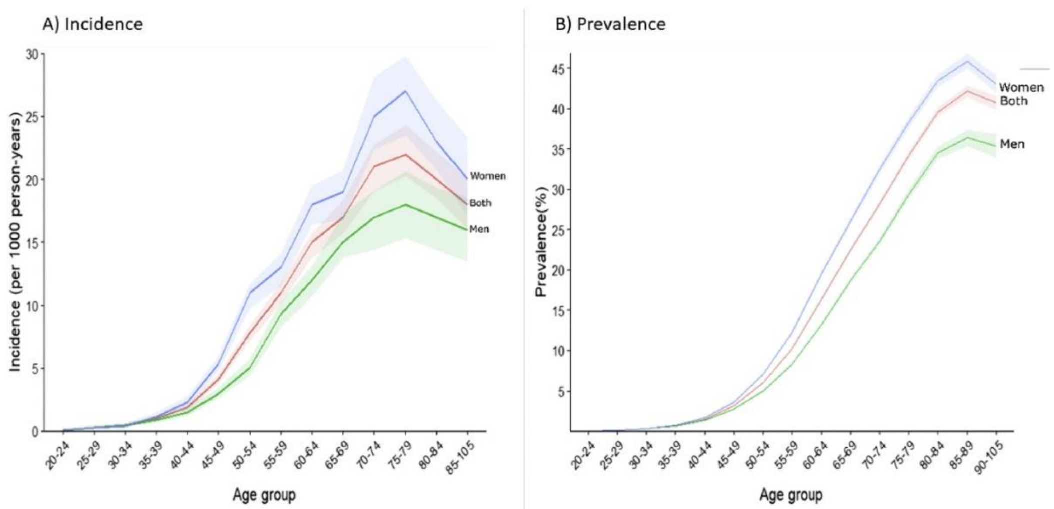


Figure 3 Age specific incidence (A) and prevalence (B) of osteoarthritis for men, woman and combined in the UK in 2017

Extracted from **(Swain *et al.*, 2020)**

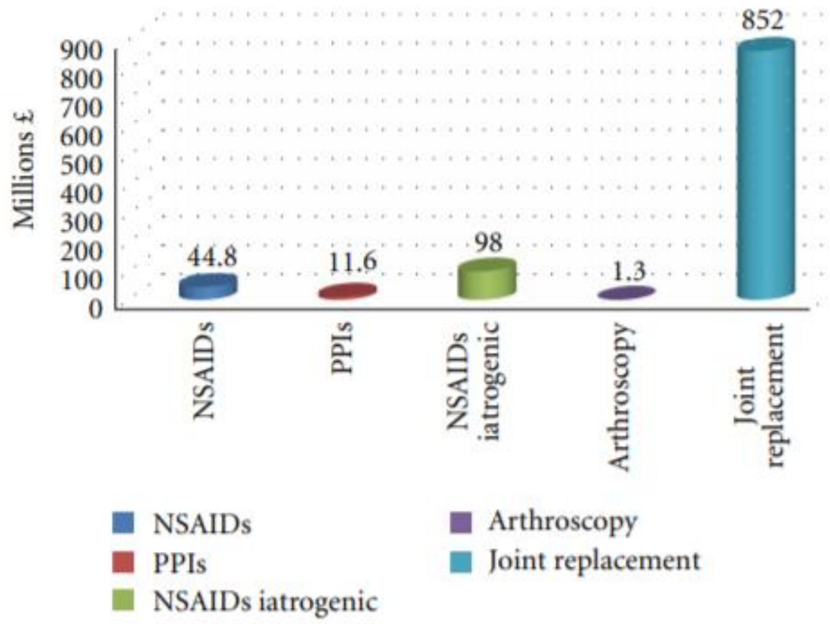


Figure 4 Direct cost of osteoarthritis in the United Kingdom

Extracted from (Chen et al., 2012)

2.1.1 Diagnosis of medial knee osteoarthritis

Historically, the diagnosis of mKOA has most often been based on radiographic appearance, rather than clinical features. Radiographic criteria were proposed by Kellgren and Lawrence in 1957 (Kellgren and Lawrence, 1957), and was accepted by the World Health Organisation (WHO).

In 1981, the American Rheumatism Association asked the Diagnostic and Therapeutic Criteria Committee to establish a sub-committee on mKOA. Developing on from Kellgren & Lawrence, Altman et al. defined mKOA as a heterogenous group of conditions that lead to joint symptoms and signs which are associated with defective integrity of articular cartilage, in addition to related changes in the underlying bone and at the joint margins (Altman *et al.*, 1986). Although articular cartilage is poorly innervated and defects in cartilage are not, in themselves, symptomatic, a clinical syndrome of symptoms, which often includes pain, may evolve from such defects.

As opposed to viewing OA as a single disease, it is now considered to represent the net effect of a collection of diseases with different causes and potential treatments ((Lane *et al.*, 2011). Kraus et al.'s (2015) review on the terminology surrounding the classification and diagnosis of OA identified a need for the standardisation of the definition of OA. Of the four primary draft definitions collated, the main elements are as follows:

1. OA is a complex disease involving movable joints, which is difficult to diagnose and define.

2. OA subjects are a very heterogeneous group, with large variations in clinical symptoms and outcomes.

3. The specific causes of OA are unknown – it is believed to occur because of both mechanical and molecular events.

4. OA is characterised pathologically by cell stress, extracellular matrix degradation and tissue remodelling due to maladaptive repair response, including pro-inflammatory pathways and disruption of the homeostasis of catabolic and anabolic processes.

5. The initial stages of the disease are characterised by abnormal joint tissue metabolism, which eventually leads to macroscopic changes such as joint inflammation, cartilage degeneration, and osteophyte formation, particularly around the joint margins.

6. The clinical condition is characterised by joint pain, tenderness, crepitus, movement limitations, inflammation, and occasional effusion.

A systematic review (Chapple et al., 2011) assessing patient characteristics that predict the progression of mKOA found age, varus knee alignment, presence of OA in multiple joints, and radiographic features had strong evidence as predictors of mKOA progression (Chapple *et al.*, 2011). Accordingly, attention to the mechanical loading environment and in particular, magnitudes and distributions within the joints has been suggested by many to be a critical piece towards the understanding of mKOA pathophysiology and improvements in treatment (Andriacchi and Mündermann, 2006). Furthermore, attention should be given to individuals with varus knee alignment, who have been shown to have higher risks of mKOA than the general public and represents a promising candidate as an intervention target (Moisio *et al.*, 2011). This thesis will focus on this specific cohort of individuals, to establish the extent interventions can offload the medial compartment of the tibiofemoral joint; without having adverse biomechanical effects at the ankle and hip joints.

Lower-limb alignment is typically defined as the hip-knee-ankle (HKA) angle which represents a key determinant of load distribution in the knee joint (Coventry, 1984). A neutrally aligned knee is indicated with a HKA equal to 180°, whereas a varus knee position is typically indicated by a HKA <180°, i.e., 175° of HKA angle equal to 5° varus (Figure 5).

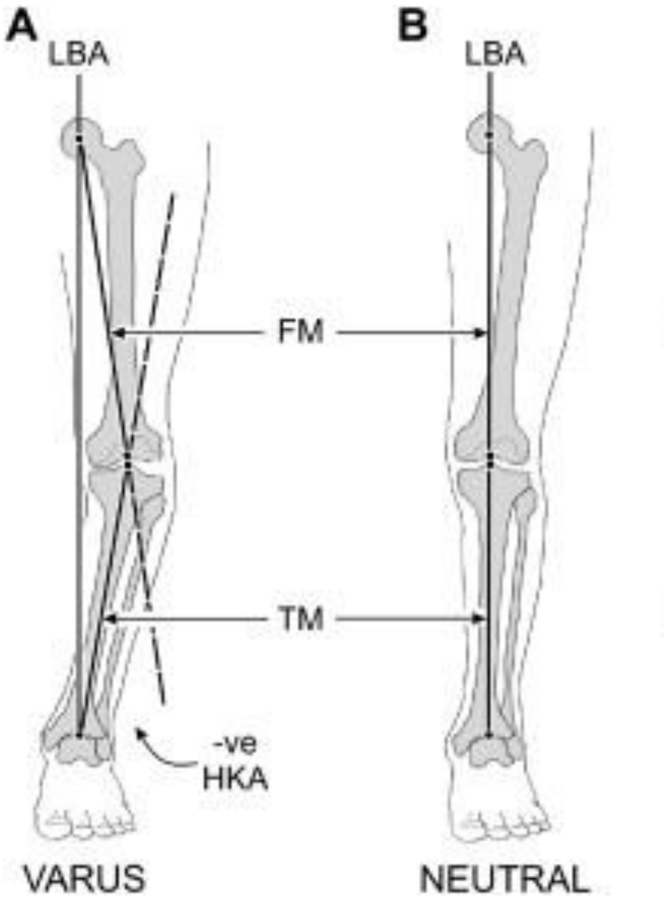


Figure 5 Diagram of the lower limb alignment as indicated by the hip-knee-ankle angles

Showing varus (-ve HKA) and neutral alignment. In neutral the femoral (FM) and Tibial (TM) axes are co-linear with the load bearing axis (LBA). Extracted from (Cooke, 2007)

Knee alignment is one risk factor for mKOA that has been commonly investigated (Tanamas *et al.*, 2009). Malalignment of the lower leg, in either the valgus or varus direction, has been found to influence the distribution of load across the articular surfaces of the knee joint (Tetsworth and Paley, 1994). Although the medial compartment of the knee joint has been reported to bear approximately 60 –80% of the compressive loads in the neutrally aligned knee (Schipplein and Andriacchi, 1991), minor alterations in knee alignment have been shown to result in abnormal load distribution across the joint. This has recently been evidenced in a simulated alteration of frontal plane alignment in an otherwise healthy cohort (van Rossom *et al.*, 2019) (Figure 6).

Coronal plane knee malalignment significantly affected the medial-lateral force distribution beyond three degrees additional varus compared to the reference position (van Rossom *et al.*, 2019). Additional varus resulted in increased medial condyle loading, with a simultaneous load reduction on the opposite condyle. As a measure of medial condyle loading, EKAM showed a clear dependence of coronal plane knee alignment on the load distribution in terms of knee contact forces as well as EKAM. This indicates that coronal plane malalignment could impose excessive stress on the articular cartilage and subchondral bone and could therefore potentially contribute to mKOA disease initiation and progression.

Furthermore, mKOA patients with increased internal tibia rotation, an indication of transverse plane malalignment, were found to walk with increased EKAM, suggesting increased loading and compression of the medial condyle as a potential contributor to mKOA (Krackow *et al.*, 2011).

It is these increases in medial compartment loading that are thought to increase stress on articular cartilage and other joint structures, subsequently leading to degenerative changes. Therefore, any method to decrease this load or laterally shift the load may result in medial compartment joint preserving alterations.

In the abnormally aligned ambulating knee, load is disproportionately transmitted to the medial compartment (Schipplein and Andriacchi, 1991) due to (1) alignment, where less than 2° is considered as normal, and (2) the different anatomical structure between the medial and lateral compartment. Therefore, it is reasonable to associate the increased incidence of mKOA with mechanical loading. The excessive compressive loading on the medial compartment may impair joint repair and maintenance. Varus deformity at the knee significantly increases the likelihood of an individual developing mKOA (van Rossom *et al.*, 2019) (Figure 6). The sub-group of individuals who have varus deformity are an in vivo example of altered joint loading (Bhatnagar and Jenkyn, 2010). This thesis focuses on understanding the knee joint loads for individuals with varus deformity before and 12 months post knee realignment surgery and altered loading introduced through changes to altered gait styles.

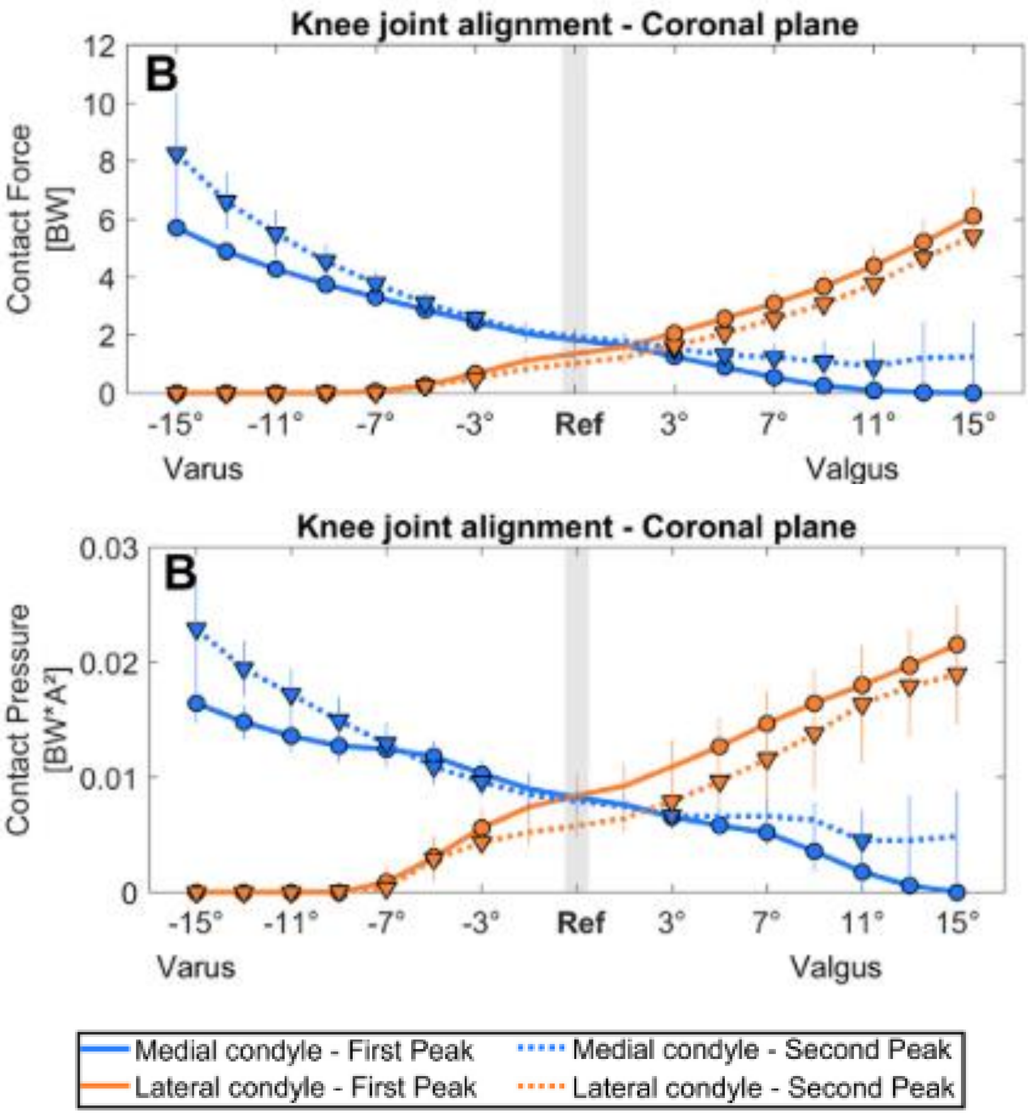


Figure 6 Effect on the contact force and contact pressure distribution the effect of an altered joint geometry in the coronal plane

A significant difference compared to the contact force during the reference simulation (grey bar) is indicated by a solid dot (first peak) and a solid triangle (second peak). Figures extracted from (van Rossom *et al.*, 2019)

2.2 The normal gait cycle

The term 'gait cycle' describes the period from the heel strike of one foot, ending at the subsequent heel strike of the same foot (Perry, 1992). The gait cycle comprises of two general phases; (1) 'stance', when the foot is in contact with the floor, and (2), 'swing' which describes the forward swinging motion of the limb. This can be further divided into four parts (1) early-stance (0-20% of the gait cycle), (2) mid-stance (21-40% of the total gait cycle), (3) late-stance (41-60% of the total gait cycle) and (4) swing phase (61-100% of the total gait cycle). It is during the stance phase when the lower limb joints are loaded.

During a typical gait cycle, external ground reaction forces (GRF) act on the lower limbs due to the foot striking or pushing off from the ground and acceleration or deceleration of the body creating moments. A moment can be defined as a turning force created by a force applied at a distance from a turning point (Richards *et al.*, 2018). The use of 3D gait analysis allows better understanding of biomechanics and any alterations that may take place in the presence of mKOA. Gait speed is an important consideration when measuring gait parameters based on the magnitude of the GRF and segmental accelerations. More rapid accelerations of the centre of mass (COM) of the body may result in a higher GRF and higher joint moments, and both healthy (Fukuchi et al., 2019) and mKOA individuals (Mündermann *et al.*, 2004) experience increases in joint moments when walking at faster speeds.

Larger joint moments are indicative of increased joint loads, and increased joint loads are often implicated in the disease progression of mKOA (Miyazaki *et al.*, 2002). Numerous biomechanical alterations are present in mKOA patients, meaning several significant changes to the normal gait cycle can be observed. mKOA is associated with adjustments to normal biomechanics in gait, balance, muscle strength, and muscle co-contraction to accommodate the condition and decrease pain.

2.3 Loads at the knee

The medial tibiofemoral compartment of the knee is frequently affected by mKOA (Simic *et al.*, 2011) and this predilection probably reflects the loading experienced during daily locomotor activities (Reeves and Bowling, 2011).

During normal walking, the activity that imposes the largest number of loading cycles in daily life, the tibiofemoral joint is subjected to two peak loads during the stance phase of every step (Amis, 2013). The first peak is caused by the large quadriceps force that is necessary to arrest the descent of the body mass, when the weight transfers from the leg that is pushing off from the ground to the leg that is accepting the load shortly after the heel strikes the ground. The second peak is when the knee and hip are extended, the heel is raised from the floor and the forefoot is pushing off, propelling the body forwards.

Assessment of effectiveness of an intervention to reduce medial compartment tibiofemoral loading is difficult since medial compartment contact force cannot be measured in vivo under normal circumstances (D'lima *et al.*, 2012). Given the complexity of calculating the joint forces in the case of co-contraction and the sensitivity of the calculation to the length of the muscle moment arms; net joint moments around the knee are more commonly used in the field of mKOA biomechanics, to indirectly express joint load, although we know that it is only the very first approximation (Kutzner *et al.*, 2013).

2.3.1 External knee adduction moment

Researchers have identified the EKAM as a surrogate measure for medial compartment contact force (Simic *et al.*, 2011). The EKAM results in the tibia rotating medially with respect to the femur in the frontal plane and therefore, a large proportion of the force is transferred by the medial compartment of the knee (Figure 7) (Shelburne *et al.*, 2008; Reeves and Bowling, 2011). The peak value of this moment during stance phase has been correlated with poorer outcomes following HTO surgery as well as with pain, disease severity, and the rate of disease progression in non-operated patients (Prodromos, Andriacchi and Galante, 1985).

Treatment methods should aim to decrease the load on the joint to potentially slow disease progression and lessen the symptoms experienced. The knee joint is subjected to both internal and external moments, knee implants are an accurate and reliable method of measuring knee forces, however they are invasive. Therefore, the EKAM has been identified as an indirect measurement of medial load distributions (Schipplein and Andriacchi, 1991; Kutzner *et al.*, 2013) and the presence, severity, and rate of progression (Miyazaki *et al.*, 2002) of mKOA.

Additionally, the EKAM has been correlated with higher levels of pain in individuals with mKOA, and reduction of medial loading may result in pain relief (Miyazaki et al., 2002; Sharma et al., 1998). The EKAM can be defined in simple terms as the product of the GRF and the perpendicular moment arm from the knee joint centre to the line of action of the force. This method of calculating the EKAM, known as the lever arm method, gives an indication of the magnitude of the EKAM. However, it is now generally accepted that the

EKAM should be calculated using inverse dynamics which is based on the Newton-Euler equations of motion and uses both linear (∑ m = Iα) and angular acceleration (∑ m = Iα). Where 'F' is force, 'm' is mass, 'α' is acceleration, 'm' is moment, 'I' is the mass moment of inertia and 'α' is angular acceleration.

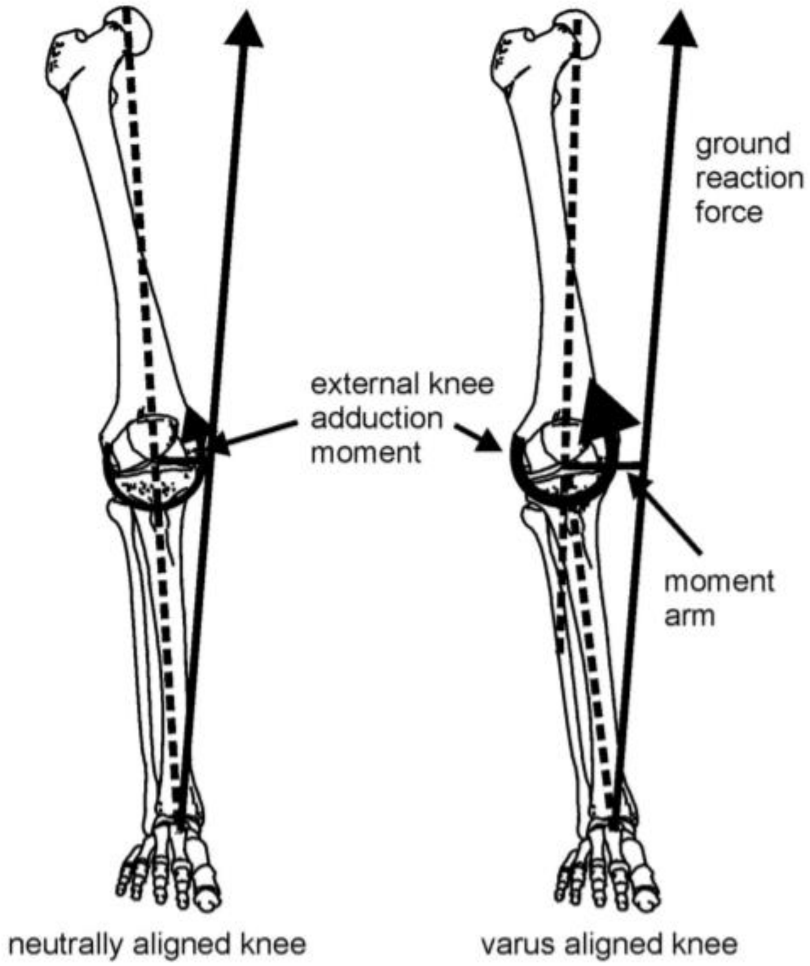


Figure 7 External knee adduction moment for the neutral and varus aligned knee (Mündermann et al. (2008))

Coronal plane external moments i.e., EKAM, along with additional external forces acting on the knee can be calculated using a motion analysis system and a force platform. The inertial and mass properties of the segment are calculated by the external forces acting on each body segment, the joint motion of the segment using kinetic data, as well as anthropometric data. To achieve stability and equilibrium during movement, the EKAM must be balanced by an equal internal moment. Consequently, the net internal moment in the knee joint, produced predominantly by muscle, soft tissue and contact forces is equal and opposite to the EKAM. In the absence of reduced antagonist muscle activity, a larger EKAM can be attributed to a larger contact force.

Varus alignment was identified as increasing both the risk of mKOA, and the progression of the disease (Sharma *et al.*, 2001). Suggesting that the magnitude of the EKAM can be associated with radiographically identified joint space narrowing of the medial compartment of the knee (Sharma *et al.*, 2001). Not surprisingly therefore, extensive previous investigations have reported the EKAM to be larger in mKOA patients, when compared with healthy subjects during early stance. An adaptive mechanism that is found in individuals who develop a greater EKAM is to increase lateral trunk sway or pelvis lean alterations towards the stance leg to reduce the moment lever arm and therefore the EKAM, providing some pain relief (Huang *et al.*, 2008). Conversely, several contradictory findings concerning EKAM, and late stance exist (Chapman *et al.*, 2015). Some report EKAM in mKOA patients to be like healthy subject groups (Mündermann et al., 2005; Huang et al., 2008).

Since the excessive loading caused by varus moment may increase the magnitude of compressive stress and strain in the cartilage, varus deformity can hasten progression of mKOA. Methods to offload the medial compartment include gait adaptation, mechanical walking aids and surgical re-alignment. However, reducing the EKAM has become the main objective of early conservative treatment of mKOA with many researchers using this surrogate measure as a target for biomechanical treatments.

Although the EKAM is a time-series over the stance phase of the gait cycle, studies reporting on the EKAM tend to focus on discrete points. Due to the characteristic double peaked shape, the two peaks of the EKAM are often chosen for analysis. The EKAM is often used as a representation of the loading in the knee joint since the loading itself cannot be measured in vivo. Assessing the magnitude of EKAM peaks can therefore be used as an indirect measure of medial compartment joint loading and is measured during activities of daily living.

Shelburne et al. stated that the high incidence of patients with mKOA was due to an existing EKAM and a concomitant increase in load in the medial compartment of the knee (Shelburne *et al.*, 2008). Shelburne et al. (2008) used a MSK simulation technique (Shelburne *et al.*, 2004). For the unaltered gait cohort, the total force acting between the femur and tibia peaked at 2.7 times body weight at contralateral toe-off. Most of this force was transmitted by the medial compartment; peak medial contact force was 2.4 times body weight. The force acting in the lateral compartment was much lower; peak force on the lateral side was only 0.8 times bodyweight. The pattern of force in the medial compartment resembled the shape of the EKAM. Peak EKAM in the model was 3.5 %BW.h. Peak EKAM occurred just before contralateral toe-off when maximum force was transmitted by the medial compartment. Medial compartment load increased linearly with the adduction moment applied at the knee during stance.

Contrariwise, several studies have reported early-stance peak EKAM to be comparable in patients with varying severities of mKOA to healthy individuals of matching age and gender, perhaps due to compensatory mechanisms such as trunk lean or pelvic list towards the stance leg to lower the EKAM by decreasing the moment lever arm (Huang *et al.*, 2008). Huang et al. (2008) tested the hypothesis that patients with mild and severe mKOA adopt different compensatory gait patterns to unload the diseased knee, in both the frontal and sagittal plane. The mild mKOA group successfully reduced the extensor moment and maintained normal abductor moment at the diseased knee mainly through listing and anterior tilting the pelvis. With extra compensatory changes at other joints and increased hip abductor moment, the severe group successfully reduced the knee extensor moment but failed to reduce the abductor moment.

Furthermore, in mild mKOA (KL grade 1-2), conflicting results have emerged concerning late-stance peak EKAM, with Mündermann et al. (Mündermann, Dyrby and Andriacchi, 2005) reporting it to be significantly smaller when compared with both age and gender matched patients with severe mKOA, and healthy individuals. Additionally, Huang et al. (2008) observed the EKAM to be similar between mKOA patients and healthy participants. Most investigations have identified an increase in the EKAM in patients with mKOA compared to healthy participants, and therefore the EKAM can be considered a reliable indication of mKOA.

A study from our research group (Whatling *et al.*, 2019) performed gait analysis on 18 participants (19 knees) with mKOA and varus alignment pre- and post-HTO, along with 18 controls. The primary aim of this study was to determine the key changes in knee kinetics, in three clinical planes, during the stance phase of gait in patients before and after HTO. It was hypothesised that pre-operatively, elevated frontal plane loading would be accompanied by altered loading in the sagittal and transverse planes relative to controls. Furthermore, it was hypothesised that HTO surgery would restore frontal plane loading to that of healthy controls whilst also affecting loading in the sagittal and transverse planes. This study highlighted how 1st and 2nd peak EKAM were reduced post-HTO (3.0 %BW.h vs 2.1 %BW.h and 2.5 %BW.h vs 1.5 %BW.h, for 1st and 2nd peaks respectively). Whatling et al. (2019) concluded that HTO surgery restored frontal and transverse plane knee loading to normal levels and improved PROMs. This study concluded by stating that the gait adaptations known to reduce knee loading employed pre-HTO were not present post-HTO.

The present thesis aims to develop on from these findings to establish whether (a) the same conclusions can be taken from a larger cohort of individuals undergoing HTO, (b) establish the alterations at the hip and ankle joints because of these knee loading parameters (c) whether these findings are in line with the findings from using an enhanced MSK simulation pipeline that predicts medial and lateral tibiofemoral joint contact forces and pressure distributions.

Asides from invasive methods that measure in vivo contact forces which leaves the knee in an unnatural state to compare to individuals who have their knee intact, and asides from using EKAM as a surrogate measure of medial compartment knee joint loading, researchers are afforded MSK models to predict muscle behaviours during gait and then can predict knee joint contact forces.

2.3.2 Simulating tibiofemoral joint contact forces

Knee contact forces (KCF) can be directly measured in vivo in patients who have received instrumented total knee arthroplasty (TKA) (Mündermann *et al.*, 2008). Mündermann et al. (2008) assessed in vivo knee loading characteristics during activities of daily living as measured by an instrumented total knee replacement. However, it is challenging to infer articular loading for individuals with and without mKOA from these measurements, owing to the procedure involving the articular surface replacement, changing the bone structure, and the realignment of the mechanical knee axis.

Normative joint contact forces for a range of ADL, including unaltered level gait, are estimated to be ~2 times body weight, and for most activities ~2.5 times body weight (Mündermann et al., 2008). Most activities placed a greater load on the medial compartment than the lateral compartment. The results demonstrate that the forces and motion sustained by the knee are highly activity-dependent and that the unique loading characteristics for specific activities should be considered for the design of functional and robust total knee replacements, as well as for rehabilitation programmes for patients with mKOA or following total knee arthroplasty.

To corroborate these findings further, other studies that have assessed individuals with instrumented total knee prosthesis have concluded that peak KCF are between 1.9 and 3.5 times body weight for walking at self-selected speed (Kim et al., 2009; Mündermann et al., 2008; Zhao et al., 2007). Higher KCF of approximately 4.5 times body weight have been reported in healthy individuals when assessed by computational approaches (Richards and Higginson, 2010). This may indicate that individuals undergoing a TKA may walk slower and have protective mechanisms to prevent such high tibiofemoral compressive forces.

Although it is challenging to infer articular loading for with and without mKOA from an instrumented TKA, in vivo measurements of the tibial compressive loads are essential to validate computational models. Alternatively, to direct measurement of knee contact force, MSK modelling in combination with simulations of motions might be used to calculate knee contact force. Different from in vivo measurements, computational approaches are non-invasive and can be applied to a larger number of participants. Therefore, computation of knee contact force has received much attention (Lenhart et al., 2015; Richards & Higginson, 2010; Smith et al., 2016; van Rossom et al., 2019). Knee contact force not only accounts for the external forces but also account for muscle and ligament forces.

Resultant knee forces (knee contact forces, muscle forces, ligament forces and external forces) are calculated based on the dynamic equilibrium, in which the sum of all the forces acting on a body is equal to the product of the body mass and the linear acceleration (Newton's second law). The resultant knee moment is the sum of all the moments acting (internal and external) about the joint. To be able to estimate knee contact forces, muscle forces must be calculated first. The major problem for the estimation of muscle forces acting around MSK joints is the problem of redundant muscles (Meireles, 2017).

2.3.2.1 Muscle redundancy problem

Muscle redundancy results from the higher number of muscles compared to the degree of freedoms (DOF) of the joint. As a result, there is no unique solution for the muscle force distribution and hence for predicting knee contact forces. Optimisation methods in a static or dynamic configuration have commonly been used to resolve this redundancy by assuming the human movement is produced by optimising some performance criterion (Anderson and Pandy, 2001). Although static optimisation neglects muscle activation and contraction dynamics, which are accounted for by dynamic approaches, static optimisation results in similar muscle force solutions as dynamic optimisation for gait (Anderson and Pandy, 2001).

Briefly, static optimisation determines the set of muscle forces producing net joint moments while minimising a cost function based on a certain performance criterion at a discrete time within certain muscle force limits. Previous research (Challis, 1997) has shown that minimising effort, by minimising the sum of squared muscle activations, yields muscle activation patterns like those observed experimentally and this performance criterion is therefore largely used (Anderson and Pandy, 2001).

2.3.3 State of the art: Predicting knee joint contact forces

Recently, some patients have received instrumented knee prostheses, and the outputs from these have been implemented to validate the computer models used to calculate forces in normal knees (Kinney et al., 2013). There has been agreement that the peak forces are approximately three times body weight when walking.

Few studies have used computational modelling to calculate knee contact forces during walking in patients with mKOA. Using a statically determinant model, Henriksen et al. (2006) compared knee contact force estimations between mKOA and healthy subjects and found significant differences. The average peak knee contact force calculated during early single limb support was 1.8 body weight for mKOA subjects and 2.4 body weight for healthy subjects, and 1.6 and 1.9 body weight during late single limb support for mKOA and healthy subjects, respectively. However, they grouped all patients with radiographic evidence of mKOA into one group and compared them to a healthy control group (Henriksen *et al.*, 2006). Previous examples where knee contact forces have been calculated through simulation work is in the work of Richards & Higginson (2010); however, this was not using the COMAK approach. Richards & Higginson did not find significant differences in the first peak knee contact forces between healthy subjects and those with varying degrees of mKOA (all groups presented peak knee contact force between 4-4.5 body weight). However, the severe mKOA group showed a very different knee contact force pattern compared to healthy subjects, and both mKOA groups presented reduced second peak knee contact force. Kumar et al. (Kumar, Manal and Rudolph, 2013), on the other hand, found increased first peak medial knee contact force in established mKOA subjects (2.6 body weight) with radiographic signs of joint structural changes in the Kellgren and Lawrence (KL) system for classification (KL ≥ 2) compared to healthy subjects (2.4 body weight) but not in terms of total knee contact force (3.7 body weight and 3.5 body weight, respectively, healthy and mKOA participants). While compartmental knee contact force has been reported by Kumar et al. (2013) for patients with severe mKOA, there is still a lack of information regarding patients in the early stages of mKOA both in terms of total knee contact and, more importantly, contact forces on the medial compartment of the knee joint.

MSK models can estimate a participants muscles forces, joint moments, and joint reaction forces as well as joint kinematics by solving the muscle redundancy problem. MSK modelling method would afford a wealth of understanding on the influence of gait patterns and HTO surgery on joint force magnitudes. This is currently severely lacking in the present literature and is needed to better inform intervention outcomes to reduce medial compartment joint loading and to preserve the knee joint. Computational modelling of joint degradation can help to estimate the patient-specific progression of mKOA, with the goal to aid clinicians' ability to estimate a suitable time for surgical intervention in OA patients; as well as determine the success of surgical and non-surgical interventions.

It has been shown by Richards et al. (2018) that for patients with mKOA, the EKAM is a strong predictor of the medial knee contact force (mKCF) at the first peak during normal walking, which was previously indicated by Kutzner et al. (2013). It was also reported in Richards et al. that walking with toe in gait style or wide step gait style, modified the first peak EKAM, but no reduction in mKCF was observed. However, the ratio of mKCF to total KCF, which represents the distribution of the loading, was reduced. Richards et al. (2018) demonstrated the potential of MSK models to identify markers that increase tibiofemoral joint loading, thereby triggering more degeneration, and to define modified gait strategies to reduce the joint loading in an already degenerated joint. In Meireles et al. (2017) it was shown that the medial-lateral force and contact pressure distributions were already altered in early stages of mKOA during unaltered gait. Indicating that MSK models can be used to differentiate between populations with mKOA. Interestingly, in Meireles et al. (2017) the control group (~64 years old), early OA (~63 years old) and established OA (~67 years old) groups walked with non-significant differences in gait speed. The more established OA group had a varus alignment of ~4° compared to the near-to-neutral alignment in the control group. Only patients with established KOA showed significantly higher peaks and higher minimum total KCF during the single support phase when compared to controls. No significant difference in total KCF was found between early KOA and control group. Both patient groups presented higher peak medial KCF compared to controls. Maximum contact pressure was significantly higher for subjects with established KOA (25.78 MPa) compared to the control (15.02 MPa) and early KOA (19.72 MPa) groups. In subjects with early KOA, the medial compartment centre of pressure (COP) at the time instant of the first peak medial KCF was shifted from central to a significantly more posterior region, while a significantly more posterior-lateral location of the COP was found in subjects with established OA. No significant differences were found in peak KFM or EKAM between the three groups. No differences in peak EKAM adds to the debate as to whether EKAM really does reflect the internal medial compartment joint loading.

An approach developed by Lenhart et al. (2015), uses an enhanced static optimisation technique, the COMAK algorithm (Lenhart et al., 2015; Smith, Lenhart, et al., 2016) to simultaneously solve for ligament forces, muscle forces, and contact forces in the medial and lateral compartment of the knee joint. COMAK estimates secondary knee kinematics, muscle forces, ligament forces, and contact pressures based on minimising a certain cost function while satisfying dynamic equations of motion. This cost function is defined as the weighted sum of squared muscle activations and the net cartilage contact elastic energy. The contact pressures are derived from an elastic foundation model based on the theory developed by Bei and Fregly (Bei and Fregly, 2004) implemented in the articular cartilage of the knee. The specifics of this approach will be detailed in the methods section of this thesis.

The COMAK technique is the first-time MSK simulations can be used to predict 12 degrees of freedom kinematics and contact forces for the tibiofemoral and patellar-femoral joints. It is a novel aspect of this thesis to undergo analysis using the COMAK simulation approach on individuals who have varus deformity and have radiographic medial compartment knee osteoarthritis who undergo HTO. It is also a novel aspect of this thesis to undergo analysis using the COMAK simulation to assess the effectiveness of three altered gait styles pre- and 12 months post-HTO on offloading the medial compartment of the tibiofemoral joint.

Relatively small alterations in coronal plane knee alignment significantly affect the knee loading distribution (van Rossom *et al.*, 2019). In neutral alignment, more loading is taken up by the medial condyle. Coronal plane knee malalignment significantly affected the medial-lateral force distribution beyond 3° additional varus or valgus compared to the reference position. A higher varus angle resulted in increased medial condyle loading. Indicating a clear dependence of coronal plane knee alignment on the load distribution in terms of KCF as well as EKAM. This suggests that coronal plane malalignment could impose excessive stress on the articular cartilage and subchondral bone and could therefore potentially contribute to mKOA disease initiation and progression.

It follows that people with a varus alignment are at increased risk of developing mKOA. Varus tibiofemoral alignment has been reported as one of the best predictors of a high knee EKAM (Liu *et al.*, 2019). Studies in patients with mKOA have calculated the coefficient of correlation as 0.5 (Wada *et al.*, 2001), 0.6 (Hunt *et al.*, 2008) or 0.8 (Hurwitz *et al.*, 2002) between varus knee alignment and the peak EKAM during walking. In line with this association, patients with moderate to severe mKOA show an increased varus tibiofemoral alignment of between 2° and 6° compared to patients with mild to moderate symptoms (Hurwitz et al., 2002; Mündermann et al., 2005, 2004; Thorp et al., 2006; Wada et al., 2001). This can therefore indicate that HTO may have limited success for a percentage of the cohort who are considered for the operation and so an alternative intervention in the form of altering gait may produce more favourable results.

Although EKAM has been reported as a surrogate for knee contact force, it is well suited to predict the medial force ratio throughout the whole stance phase or medial force during the early stance phase (Kutzner *et al.*, 2013). However, EKAM was not sufficient to predict joint loading at the end of the stance, where EKAM contributed substantially to the loading, especially in early mKOA (Meireles et al., 2016; Richards, Andersen, et al., 2018). Some findings suggested that the knee contact force predicted by a novel MSK simulation routine provides a more helpful metric than the EKAM (Meyer *et al.*, 2013). Lerner et al. (2015) found that each 1° of tibiofemoral alignment deviation altered the first peak medial KCF by 51 N, whilst each 1 mm of medial-lateral translation of the compartment contact point position altered the first peak medial knee contact force by 41 N. Knee contact force may be used to identify early mKOA development prior to the onset of radiographic evidence.

Recently, the COMAK algorithm (Lenhart et al., 2015; Smith, Lenhart, et al., 2016) was introduced to simultaneously solve for muscle and soft tissue loading during functional movement. In COMAK, inverse kinematic measurement techniques (Lu and O'Connor 1999) are first used to compute the coordinates, speeds, and accelerations of the *primary model DOF*. Thereafter, numerical optimisation is performed to simultaneously solve for the secondary kinematics, muscle, ligament, and articular contact forces that generate the primary joint accelerations while minimising a cost function that resolves inherent muscle redundancy.

To study soft tissue loading and internal joint mechanics of the knee, the one DOF knee joint in the Arnold model (Arnold *et al.*, 2010) was replaced with a knee that has six DOF at both the tibiofemoral and patellofemoral joints. Removal of the artificial kinematic constraints required the force contributions of the passive structures and articular contact to be explicitly modelled and calculated.

To obtain the geometries of these structures, segmentation of the bone, ligament, and cartilage of a healthy young adult subject from high resolution magnetic resonance imaging (MRI) was used (Lenhart et al. 2015). Each ligament was represented in the model as a bundle of nonlinear springs. The ligament force-strain relationship was assumed quadratic at low strains and linear at high strains to capture the nonlinear effects of collagen crimp straightening and fiber elongation (Huiskes and Blankevoort, 1991).

The articular contact forces are computed using an elastic foundation model (Bei and Fregly, 2004). The articular surfaces are represented by high resolution triangular meshes that do not deform but are allowed to interpenetrate. Contact pressure on each triangle face is computed based on the local penetration depth, cartilage thickness, and material properties.

2.4 Appreciation of the use of electromyography to understand mKOA muscle activity

Measuring the forces applied to a joint and estimating how these forces are partitioned with respect to surrounding muscles, ligaments, and articular surfaces is fundamental to understanding joint function, injury, and disease. Muscle forces have been proposed as the primary determinants of joint contact forces (Herzog et al., 2003), with correctly predicted muscle forces assuming to result in sensible estimates of joint contact loads. However, to-date, accurate measurement and prediction of individual muscle forces are still a major challenge. Muscle electromyography (EMG) has been used for decades to evaluate neuromuscular responses due to KOA pathology (Benoit et al., 2003). Lloyd and Buchanan (2001) investigated the activation strategies used by individuals to support adduction/abduction moments and the muscle loading patterns that result from these activation schemes during highly controlled isometric tasks. Wilson et al. (2012) associated EMG patterns of the knee periarticular musculature with post-operative tibial implant migration. Higher muscle co-contractions have been linked to KOA severity (Hubley-Kozey et al., 2009; Metcalfe et al., 2013), presumed to be linked with higher muscle forces (Hubley-Kozey et al., 2008), to compensate for joint instability.

There is now a large body of evidence demonstrating that patients with KOA exhibit excessive muscular co-contraction (simultaneous activation of the quadriceps and hamstrings) during walking and other functional tasks. This co-contraction increases compressive loads at the knee joint surface, accelerates structural progression of the disease. Elevated loading may also increase the stress on articular structures, such as the joint, bone, synovium/joint capsule, and periarticular structures, resulting in increased pain.

Gait retraining interventions assumes that knee joint overloading, especially at the medial compartment during gait is an important cause of progression of KOA. It targets the distribution of force between the medial and lateral compartments of the tibiofemoral joint. Although many studies have shown that gait retraining is effective in reducing the EKAM, the influence of changes in muscle activation, an important determinant for knee joint loading, is often neglected in these studies. This is because concurrent activation of agonist and antagonist muscles will cancel out each other's contribution to the joint moment but add in their contribution to the knee reaction force. While co-contraction will enhance stabilisation of the knee joint, it increases knee loading, which is not reflected by the EKAM. Consequently, muscle co-contraction is an important outcome parameter that should be considered in interventions that target KOA progression such as gait retraining.

Six studies have investigated muscle activation changes during gait retraining (Rutherford et al., 2010; Ogaya et al., 2015; Lynn and Costigan, 2008; Shull et al., 2015; Charlton et al., 2018; Uhlrich et al., 2017). These studies focused on changing the foot progression angle, of which four studies taught participants Toe in gait and toe out gait. Toe in gait showed an increased medial hamstring activation during stance, higher co-contraction between the lateral (Uhlrich et al., 2017) and medial (Charlton et al., 2018) quadriceps and hamstrings and a trend of higher medial to lateral hamstring activity.

The study by Booij et al. (2020) showed that wide stance gait was the most successful gait modification. Wider Steps did not lead to increases in muscle activation or co-contraction, implying no confounding increase of knee joint loading was shown. Furthermore, in terms of the number of steps reaching the target EKAM reduction of 10%, Booij et al. (2020) saw that walking with a wide stance was equally as feasible as walking with toe in gait. Patients walked for three minutes per modification, which may not have been sufficient for the gait patterns to be executed naturally, without novelty co-contraction. After the training session, participants indicated that the learned gait patterns still felt unnatural, so this might have affected their motor control, i.e., the muscle activation level.

2.5 Medial knee osteoarthritis gait

As mentioned previously, certain alterations exist between the normal gait of healthy individuals and the gait of individuals with mKOA. Individuals with mKOA across varying radiographic severities of the disease have been reported in the literature to adopt slower walking speeds (Zeni and Higginson, 2009). This is associated with; shortened step lengths, larger double support times, decreased hip and knee range of motion (ROM) angles, reduced cadence and stride length and increased stance times when compared to age matched healthy populations (Zeni and Higginson, 2009).

The effect of walking speed is a fundamental concern in gait studies when measurements are based on the level of GRF and acceleration because of walking speed on the EKAM and the subsequent impact on knee joint loading (Zeni and Higginson, 2009; Wilson, 2012).

The load on the knee joint will increase due to an increase of the dynamic GRF that is proportional to the walking speed (Foroughi, Smith and Vanwanseele, 2009; Zeni and Higginson, 2009). Zeni and Higginson, (2009) identified variances in gait parameters to be due to slower walking speeds, when walking speeds were freely chosen in a study, rather than a result of mKOA disease progression. Mundermann et al. (2004) implies that the reduction in walking speed observed in mKOA populations to be an adaptation to reduce the load on the knee joint.

Mundermann et al. (2005) observed secondary gait alterations among mKOA patients indicating an adaptive strategy to shift the body's mass more hastily from the contralateral limb to the support limb. This alteration appears successful in reducing the load at the knee in patients with mild to moderate mKOA. The overloading of lower extremity joints could possibly lead to rapid progression of mKOA symptoms and the onset of OA in joints contiguous to the knee joint (Mundermann et al., 2005). This finding indicates the importance of thorough research into possible mKOA interventions. Interventions should be assessed not only on their ability in the treatment of mKOA, but for their effects on surrounding lower limb joint mechanics (Mundermann et al., 2005).

An increase in walking speed results in surplus forces acting on the knee joint and therefore requiring a higher amount of shock absorption in the knee, shifting the knee into greater flexion. Therefore, lower knee flexion is an adaptive strategy adopted by mKOA patients to reduce pain and maintain functional activity (Kaufman *et al.*, 2001). The drawn-out mid stance knee extension moment may increase stability during gait due to amplified biceps femoris activity (Al-Zahrani and Bakheit, 2002). Conflicting results have emerged regarding mid stance knee extension moments, with Huang et al (Huang *et al.*, 2008) reporting a decrease, Al Zahrani and Bakheit (2002) implying an increase, and Messier et al. (Messier *et al.*, 2005) declaring the mid stance knee extension moment remains constant in comparison to healthy subjects (Mündermann, Dyrby and Andriacchi, 2005).

Muscle co-contraction, which greatens the compressive forces acting on the knee joint, is increased in mKOA (Lewek, Rudolph and Snyder-Mackler, 2004) and has been found to increase further when subjects increase their walking speed. mKOA patients are therefore likely to reduce walking speed, as a slower walking speed requires lower levels of knee flexion and therefore lower levels of shock absorption are required to help reduce the load on the knee joint (Mündermann *et al.*, 2004) which can be described as an adaptive mechanism, providing some symptom relief (Lewek, Rudolph and Snyder-Mackler, 2004).

2.6 Interventions to offload the medial compartment of the knee

Current treatments for medial compartment mKOA aim to relieve symptoms enabling QOL to be maintained or improved, focusing on reducing pain, maintaining/improving joint ROM and mobility, and decreasing functional impairment. Treatments can be generally divided into three categories (Jordan *et al.*, 2003):

2.6.1 Pharmacological interventions

The most frequently prescribed treatment are analgesics, non-steroidal anti-inflammatory drugs (NSAIDs) and COX-2 inhibitors which are effective in reducing mild to moderate pain (Bradley *et al.*, 1991; NICE, 2013). Long term use of these drugs can cause severe side effects in some patients, such as gastrointestinal ulceration and bleeding, electrolyte imbalances, abnormal results in liver function tests, and hypertension. Results of the Machado et al. (Machado *et al.*, 2015) study have prompted the National Institute for Health and Clinical Excellence (NICE) to review their advice regarding the use of paracetamol as an analgesic for the treatment of mKOA in the form of a planned full review of evidence on the pharmacological management of mKOA.

The above medications may provide pain reduction, an increase in QOL and allow an increase in activity level. However, they fail to address the biomechanical causes and only manage symptoms of the disease possibly leading to increased joint loading and accelerated disease progression due to an increase in pain free activity.

2.6.2 Surgical interventions: High Tibial Osteotomy vs uni-compartment

For individuals with mKOA, two common surgical procedures can be offered: either (1) uni-compartment knee replacements (UKR), or (2) HTO. Debate remains whether HTO or UKR is more beneficial for the treatment of uni-compartmental mKOA (Santoso and Wu, 2017).

Below, both procedures will be briefly described in relation to the research surrounding the operations and their biomechanical success. It will then be shown that for individuals of a certain age, healthy and varus malalignment, HTO is the operation of choice to restore knee joint functionality. Finally, novel approaches to understanding the success of HTO will be addressed. After which, the potential benefits of non-surgical interventions and altering gait, will be addressed.

UKR was first introduced in the 1970s (Marmor, 1979) as an alternative to total knee replacement (TKR) or HTO for single compartment KOA. UKR is a joint resurfacing procedure in which the affected degenerative compartment is treated with an implant prosthesis, while the nonaffected compartment is preserved. UKR allows knee bone stock preservation and offers patients a less invasive procedure with a faster recovery time to TKR (Bruni *et al.*, 2013). Studies that compare the outcomes of HTO and UKR and their effects are lacking; thus, the relative merits of the two procedures are still under debate (Santoso and Wu, 2017).

The most important finding from Santoso & Wu (2017) was that both HTO and UKR are satisfactory operative treatment options for symptomatic mKOA. Patient selection is generally stricter for individuals undergoing HTO than for those receiving UKR. However, mKOA patients selected for HTO experience many benefits. Ideal indications for HTO include (1) young and active patients (age < 65 years), (2) normal-range body mass index (BMI), (3) mild articular destruction (no more than grade 2 Ahlbäck classification), (4) no patellofemoral arthrosis, and good ROM and a stable joint. Age, BMI, and pre-operative state KOA are key factors that optimise clinical outcomes and survival in patients undergoing HTO.

Previous studies have reported that a pre-operative BMI higher than 27.5 is a significant risk factor for early failure (Akizuki *et al.*, 2008). HTO and UKR share similar indications that include the following: age 55–65 years, moderately active, non-obese, presenting with mild varus malalignment and moderate uni-compartmental arthrosis, no joint instability, and good ROM (Dettoni *et al.*, 2010).

HTO is an extra-articular procedure, and hence entirely joint preserving. As with UKR, it is indicated for isolated medial tibiofemoral arthrosis, but aims to alleviate mKOA symptoms, and delay or prevent further progression, by offloading pressure in the diseased tibiofemoral joint. This is achieved by hinging open, or closing, an incomplete saw cut in the proximal tibial metaphysis, to alter leg alignment, and hence the direction of load across the knee joint. The concept of HTO is supported by the strong relationship between leg malalignment and the development of knee arthrosis.

Gait studies suggest that HTO can also normalise stride length, walking speed, and knee flexion (Lind *et al.*, 2013). By avoiding the cost of an expensive knee prosthesis, HTO is cheaper than knee arthroplasty (Konopka *et al.*, 2014). Given the benefits of HTO for younger patients, and UKR in patients over 60 years old, versus the high rate of dissatisfaction with TKR, these joint sparing procedures currently appear to be underutilised. Below, the reasons as to why HTO and UKR are underutilised will be outlined.

A meta-analysis comparing HTO and UKR found that survival between 9-12 years follow-up was 84% for HTO and 87% for UKR (Spahn *et al.*, 2013). Although, after 12 years follow-up, HTO tended to be revised more frequently than UKR (survival 70% v 78% respectively) (Spahn *et al.*, 2013).

The concern is that revision rate alone is a blunt tool for measuring outcome; patients with a poor outcome who do not undergo further surgery, are classified as a success. And this is particularly relevant when comparing UKR or HTO with TKR because the threshold to revision is different for each procedure. This is supported by New Zealand joint registry data demonstrating that a UKR is more likely to be revised than a TKR with the same patient reported outcome score (*The New Zealand Joint Registry*, 2018). Age is another confounder when comparing revision rates because the rates are higher in patients under 55 years of age, and the mean age for UKR and HTO is lower than TKR.

2.6.2.1 High Tibial Osteotomy

HTO aims to re-align the lower limb and consequently shift the knee joint contact forces laterally in the operative leg (Lind *et al.*, 2013), with some degree of longevity (Weidenhielm, Svensson and Broström, 1992). The basic mechanical principles of osteotomy have been known for many years, relating to the realignment of deformities following fractures, and to the unloading of localised arthritic lesions (Amis, 2013).

Currently, the most common osteotomy is the medial opening wedge osteotomy of the proximal tibia. There are various anatomical and surgical factors that support this choice, irrespective of the mechanics (Amis, 2013). In the absence of any better evidence, the realignment usually aims towards having a straight ‘mechanical axis’ that passes (in the coronal plane with the knee in full extension) from the centre of the hip, through the knee, to the centre of the tibiotalar joint. For some patients, HTO results in decreased pain, improved function, and a decreased rate of disease progression, supporting the hypothesis that reducing medial compartment contact force has disease modifying potential.

There are limited studies regarding the effects of HTO on other areas of lower extremity. Kazemi et al. investigated the changes of tibiotalar joint following HTO (Kazemi *et al.*, 2017). It was found that HTO can significantly decrease the shearing forces exerted on the ankle joint. Little is known at whether HTO has detrimental effects at the ankle and hip joints and whether non-surgical interventions do not have such consequences. If HTO is found to have potential adverse effects on the hip and ankle joints, the justification for the surgery may be limited.

HTO is considered as an option to treat isolated mKOA in varus knees, which was reported by Jackson in 1958. This surgery was not popular until Coventry reported good results in 1973 (Coventry, 1984). HTO has become more popular in young active patients after improvement in surgical technique, fixation devices, and patient selection with fewer complications (Lobenhoffer and Agneskirchner, 2003). A publication from our research group documented how KOA alters peri-articular knee muscle strategies during gait (Ghazwan *et al.*, 2022). Ghazwan et al. (2022) investigated the variation in neuromuscular control mechanisms and joint biomechanics for three subject groups including those listed for HTO surgery (pre-HTO, n = 10). Compared to the control, the peak gastrocnemius muscle force reduced by 30% pre-HTO, and the peak force estimated for hamstring muscle increased by 25% for pre-HTO. Higher quadriceps and hamstring forces suggest that co-contraction with the gastrocnemius could lead to higher joint contact forces (Ghazwan *et al.*, 2022). Combined with the excessive loading due to a high EKAM this may exacerbate joint destruction.

Balancing loads between medial and lateral compartments is an important factor in improving the long- or short-term success rates of the knee joint post-HTO (Amendola and Bonasia, 2010). Ideally, an appropriate correction achieves a minimum overcorrection from baseline alignment necessary for adequate medial unloading, whilst avoiding overloading on the lateral compartment cartilage. The patient's gait pattern after HTO is modified based on the limb alignment, which would further influence the EKAM and medial-lateral contact forces and consequently the contact stresses of the cartilage on the medial-lateral compartments of the tibiofemoral joint.

Biomechanical environment of HTO is crucial for understanding the complications of HTO and improving surgical accuracy. Currently, there is a lack of biomechanical studies on HTO in assessing the effectiveness of HTO on gait analysis, joint kinematics, and joint contact mechanics at the tibiofemoral, ankle and hip joints. The biomechanical relationships between the alignment and plate breakage, cartilage degeneration, non-union, and others are still unclear. The "safety corrective range" is still unknown. Integration of gait analysis, musculoskeletal dynamics modelling, and finite element analysis (FEA) will help comprehensively understand in vivo patient-specific biomechanics information of HTO.

Surgical intervention of KOA is costly, and of great expense to the National Health Service (NHS). Surgery also impacts on the individual in terms of recovery time and functional independence (Griffin *et al.*, 2007). An alternative to invasive surgical interventions is more conservation non-surgical interventions.

While the HTO treatment has successfully proven to have the great short-term effect on pain reduction, the clinical outcome (e.g., survivorship) deteriorate with time. Despite several studies available, comparison and pooling of the clinical outcome are somewhat challenging because of the different evaluation systems and techniques used. The evaluation of clinical outcome of HTO has centred on survival analysis in which failure was defined as the need for conversion of HTO to TKR or if the osteotomy fails to reduce pain, follow-up evaluation system (various scoring system of knee joint) and radiography for tibiofemoral angle measurement. Although some extraordinary high survival rate of HTO have been reported (Akizuki *et al.*, 2008), most studies showed good results within the first 5 years and poor results after 15 years.

Surgical procedures are invasive and have multiple disadvantages. Some mKOA patients are often not suitable for surgery (too young, for medical reasons, or no access to NHS funding), or do not want surgery. Complications of surgery can arise such as deep venous thrombosis and wound and infection complications following HTO and knee replacement surgeries (Griffin *et al.*, 2007). Surgery also requires constant revision depending on the age and activity level of the patient. KOA surgery is expensive and as mentioned previously, is only used as the last line of treatment (Griffin *et al.*, 2007).

Therefore, more conservative methods of treatment are needed. For these reasons, considerable research has been invested into more conservative treatments of mKOA as non-invasive methods are considered valuable approaches. It is important to understand which conservative techniques bring the most benefits and improvements in mKOA symptoms to the most patient types both in terms of pain reduction and improvements in functional independence. If patients report improvements in pain and functional improvement with the use of conservative techniques, they may delay or negate the need for surgery altogether.

2.6.3 Non-invasive interventions to reduce medial knee loading

Methods for gait modification can be learned, meaning the patient are taught/instructed to walk differently, or assisted, meaning an assistive object is used. The use of lateral wedge insoles has been extensively researched to reduce medial compartment tibiofemoral joint loading (Shaw *et al.*, 2017). Learned modifications that reduce the peak EKAM include, but not limited to walking with decreased speed, increased stance width, toes pointed outward, and knees medialised (Simic *et al.*, 2011). The goal of reducing the peak EKAM is to reduce peak medial contact force during stance.

Different conservative approaches exist for treating mKOA including exercise, alterations to gait (Simic *et al.*, 2011), knee bracing (Moyer *et al.*, 2011, 2015; Toriyama *et al.*, 2011; Dessery *et al.*, 2014) and footwear modification (Shaw *et al.*, 2017) to realign the weight-bearing load, providing symptom relief (Reeves and Bowling, 2011). Advantages of conservative treatment include cost and recovery duration benefits, meaning costly surgery is delayed due to slowing of disease progression and recovery times are rapid due to the non-invasive approach.

As there is no cure for mKOA (Tanamas *et al.*, 2009), management essentially involves alleviating the symptoms. Clinical guidelines stress the importance of conservative nonpharmacologic management (Kolasinski *et al.*, 2020), as drug therapies are often associated with adverse side effects. It is therefore not surprising that there has been great interest in understanding the benefits associated with surgical and non-surgical interventions.

Non-surgical interventions are proposed to reduce EKAM and reduce pain (Pereira *et al.*, 2021). Pereira et al. (2021) investigated the relationship between changes in EKAM induced by non-surgical biomechanical interventions and consecutive changes in pain and/or physical function in patients with mKOA and compared this relationship for different interventions. Fourteen papers reporting 11 studies were identified. Braces were tested in 6 studies, insoles in 5 studies, shoes in 3 studies and gait retraining in 2 studies. Methodological differences were large among studies. Large effect sizes (≥0.8) changes in pain/function were observed with interventions having at least a small EKAM effect size (≥0.2), suggesting an association between EKAM and pain/function changes (Pereira *et al.*, 2021). A linear trend was observed between inter-intervention EKAM and VAS pain effect sizes, based on 4 studies. No firm conclusions could be drawn for the different intervention types.

2.6.3.1 Lateral wedge insoles

Lateral wedge insoles (LWI) are inexpensive, discreet, self-administered mechanical interventions used as a conservative form of treatment of mKOA comprising of a shoe insert with a thicker border on the lateral side compared to the medial side with good adherence to treatment (Shaw *et al.*, 2017). LWI are a management technique advocated by NICE for the conservative treatment of mKOA (NICE, 2013). The simplicity of LWI means they can be easily and safely used by mKOA patients, and are accessible to the majority of people, due to their low cost.

Shoe-worn foot orthotic devices are an inexpensive intervention for potentially altering knee joint biomechanics. The predominant mechanism responsible for the decrease in the EKAM observed with LWI is a lateral shift in the COP which has the effect of reducing the moment arm of the GRF around the knee in the frontal plane. This lateral shift in the COP also means that the ankle eversion moment increases (Shaw *et al.*, 2017). This increase could have implications for patients with mKOA who have acute ankle sprains or chronic ankle instability.

Importantly, LWIs reduce certain biomechanical risk factors of mKOA progression such as the EKAM. However, despite shoe-worn insoles such as LWIs reducing EKAM values, a randomised trial reported that LWI provide no additional clinical improvements in pain when compared to braces (van Raaij *et al.*, 2010). One limitation of most biomechanics' studies examining shoe-worn insoles has been the focus on changes in knee biomechanics, predominantly the EKAM, data are generally lacking on the effect of the insoles at other joints or on other biomechanical outcomes (Shaw *et al.*, 2017). Given that shoe-worn insoles evoke changes directly at the foot/shoe interface with anticipated changes experienced more proximally at the knee joint a thorough understanding of their effects on joints other than the knee is needed to best guide their use in the clinical management of mKOA (Shaw *et al.*, 2017).

Shaw et al. (2017) examined the larger kinetic chain, not just the knee. Importantly, this approach has shown that use of shoe-worn insoles, LWI, has implications on the biomechanics of all joints of the lower limb. Given the potential for adverse effects at the hip and ankle joints, regardless of any beneficial loading outcome at the knee joint, clinicians must complete a thorough lower limb assessment when prescribing shoe-worn insoles to people with mKOA to minimise the potential for patient harm (Shaw *et al.*, 2017).

2.6.3.2 Walking aids

For patients with mKOA, the use of a cane or walking stick in the hand contralateral to the symptomatic knee reduced the peak EKAM by 10% (Kemp et al., 2008). Patients must, however, be careful not to use their cane in the hand on the same side as the symptomatic leg, as this technique can increase the EKAM (Chan et al., 2005). Finally, this type of gait modification requires that the patient uses an external device, which may not be practical or feasible in day-to-day activities.

2.6.3.3 Valgus knee braces

Valgus knee braces secured around the thigh and lower leg and worn throughout the day have been suggested as a conservative treatment strategy for patients with mKOA (Reeves and Bowling, 2011). The underlying rationale for use of a valgus knee brace is the application of a valgus moment (knee abduction moment) to the knee joint, which could reduce the EKAM during walking and unload the medial compartment of the knee.

Although reports of this approach have been mostly positive, studies of valgus knee braces in patients with mKOA have not conclusively demonstrated an improvement in the EKAM (Reeves and Bowling, 2011). The EKAM either decreased (Self, Greenwald and Pflaste, 2000; Petersen *et al.*, 2016), showed a tendency to decrease (Gaasbeek *et al.*, 2007), or did not change (Hewett *et al.*, 1998; Pollo *et al.*, 2002) when a valgus brace was used compared with an unbraced condition. However, these results refer to the EKAM only. The valgus brace itself exerts a moment that opposes the EKAM. Valgus knee braces are patient administered, load modifying devices used for the conservative treatment of mKOA. Valgus knee braces aim to realign the knee joint, and therefore reduce a proportion of the load acting on the medial compartment, providing pain and OA symptom relief.

The literature provides evidence of poor patient acceptance of valgus knee braces compared to other devices, such as LWI used for the conservative treatment of medial compartment KOA (Jones *et al.*, 2013). Jones et al., (2013) compared the biomechanical effects of both LWI and valgus knee braces, establishing that valgus knee braces were worn for less than 4 hours per day by 71% of users within the trial. Conversely, LWI were worn for longer than 4 hours per day by 71% of users. The LWI were deemed more comfortable, and more easily accepted by individuals within the trial, with the valgus knee braces presenting adherence issues by users.

Although valgus bracing may achieve effective unloading of the medial compartment of the knee and offers potential for improving the clinical outcome in patients with mKOA, the success of this intervention relies upon the patient being prepared to wear the knee brace continually. Valgus knee braces are bulky, potentially uncomfortable and might not be a practical daily solution for many patients.

2.6.3.4 Gait retraining

Gait retraining approaches have been advocated to delay the progression of mKOA and are a frequently used conservative strategy in the clinic that offers promise in managing mKOA (Bowd et al., 2019; Richards et al., 2017; Simic et al., 2011). Teaching a patient with mKOA to modify their gait may be beneficial in reducing dynamic medial knee load, although it is presently unclear which gait retraining are most likely to be successful. It is also unclear whether gait retraining reduces in vivo medial compartment tibiofemoral contact forces; and if they do, what the consequences are to the joint biomechanics of the ankle and hip joints.

If gait retraining can reduce the peak EKAM, it may provide one of the few conservative treatment options with disease modifying potential (Simic *et al.*, 2011). Furthermore, it could fill an important treatment ''hole'' for patients in their 40s and 50s who no longer achieve sufficient pain relief through pharmacological means and yet want to delay a TKR. Gait retraining represents a simple and inexpensive treatment strategy that may be employed by a range of health professionals to reduce medial knee load. Some of the gait retraining reported to benefit mKOA include increases in toe out angle (Whelton *et al.*, 2017), toe in angle (P Shull *et al.*, 2013; van den Noort *et al.*, 2014), lateral trunk lean (Hunt *et al.*, 2011; Gerbrands *et al.*, 2017; Anderson *et al.*, 2018), reductions in walking speed (Simic *et al.*, 2011), medialisation of the knees (Fregly, D'lima and Colwell, 2009; Gerbrands *et al.*, 2017), and a wider stance gait (Richards *et al.*, 2018).

The modifications of the abovementioned strategies have the potential to alter the EKAM magnitude. Although a range of modifications offer prospection for reducing medial knee load, the efficacy of gait retraining for mKOA remains unknown. Rynne et al. (2022) undertook a systematic review and meta-analysis to understand the Effectiveness of gait retraining interventions in individuals with hip or knee osteoarthritis (Rynne *et al.*, 2022). The meta-analysis pooled effect demonstrated significant improvements for EKAM [SMD, −1.10; 95% CI. −1.85, −0.35] in favour of gait retraining than a control intervention [SMD, −0.86; 95% CI. −1.33, −0.39]. The systematic concluded that gait retraining may be beneficial for improving biomechanics and symptoms in KOA, however due to the high heterogeneity and limited studies in the analysis, further research is required. This thesis will be utilising 3 different approaches of altered gait: (1) toe out gait (2) wide stance gait (3) medial thrust gait. However, before specifically outlining the literature on these three strategies, a general overview on the literature regarding altering gait for individuals with mKOA will be discussed.

Booij et al. (2020) studied the effect of walking with a modified gait on activation patterns of the knee spanning muscles in people with mKOA. The objective of the study was to evaluate muscle activation patterns and co-contraction around the knee in response to walking with modified gait patterns in patients with mKOA (Booij *et al.*, 2020). 40 mKOA patients walked on an instrumented treadmill. Surface EMG activity from seven knee-spanning muscles (gastrocnemius, hamstrings, quadriceps), kinematics, and ground reaction forces were recorded. When walking with ≥10% EKAM reduction, medial thrust gait (EKAM−31%) showed increased flexor activation (24%), co-contraction (17%) and knee flexion moment (35%). Isolated wide step gait also reduced the EKAM (−26%), but to a smaller extent, but without increasing muscle activation amplitudes and co-contraction. Gait modifications that are most effective in reducing the EKAM also yield an increase in co-contraction, thereby compromising at least part of the effects on net knee load (Booij *et al.*, 2020).

Perhaps the simplest gait modification is to reduce the speed of walking since the magnitude of the ground reaction force and hence the peak EKAM is associated with walking speed, with higher walking speed leading to a higher magnitude peak (Wilson, 2012). However, reducing the walking speed increases the duration of the loading and hence causes an increase in the EKAM impulse; a parameter which is linked to increased progression of mKOA. Furthermore, it can be argued that encouraging people with mKOA to walk at reduced speeds is not functional and therefore not clinically recommended (Simic *et al.*, 2011).

In 2011, Simic, et al. conducted a systematic review assessing the use of gait retraining strategies to modify the EKAM. At that time, they found 24 studies met the inclusion criteria, with 14 different strategies investigated. While the results were not completely homogenous across different studies, reducing toe out (increasing toe in angle), reducing walking speed, walking with medial knee thrust or increased trunk lean were generally effective strategies for reducing the first peak EKAM, whilst walking with increased toe out, increased speed, increased step width and medial knee thrust were generally effective for reducing the second peak EKAM.

Since the publication of this review, many new studies have been published in this field including the first longitudinal studies investigating the effects on the EKAM of gait training with toe in or toe out modification over a longer period, both of which showed positive effects on the EKAM after a period of 6 weeks and 10 weeks (Hunt & Takacs, 2014; Shull et al., 2013).

2.6.3.4.1 Lateral trunk lean gait style

In individuals who lean towards the side of the weightbearing limb as they walk, the body's COM shifts laterally and moves closer to the COP under the weightbearing foot. As the GRF tends to act through the COM, this approach changes the angle of the GRF, shifting it towards the knee joint centre. The outcome of this shift is a reduction in the moment arm of the GRF that, in turn, reduces the EKAM.

Lateral trunk lean has, therefore, been suggested as a compensatory strategy to reduce the EKAM in patients with mKOA (Hunt et al., 2008; Mündermann et al., 2008). The extent of lateral trunk lean is inversely correlated with the magnitude of the EKAM in individuals with mKOA (Hunt *et al.*, 2008). As a compensatory strategy to unload the affected knee, greater degrees of lateral trunk lean have been reported in patients severely affected by the disease compared with those experiencing mild symptoms (Hunt *et al.*, 2010). In healthy participants, walking with an exaggerated lateral trunk sway reduced the EKAM by 65% compared with unaltered walking (Mündermann et al., 2008). Although this compensatory strategy could be effective for reducing the EKAM, it should be treated with caution when considered as an 'imposed' intervention in patients owing to the risk of falling associated with excessive upper body sway. Lateral trunk sway is an effective gait alteration that decreases the early stance peak of EKAM, it seems to be uncomfortable and difficult to maintain, increases the energy cost, and may affect balance (Shull et al., 2013; Takacs et al., 2014; van den Noort et al., 2013).

2.6.3.4.2 Toe in gait style

Toe in gait has been identified as a promising non-surgical treatment option for patients with mKOA (Shull et al., 2013). Toe in gait which can be defined as a decreased foot progression angle from baseline through internal foot rotation has been found to significantly reduce the first peak EKAM during walking. Shull et al., (2013) required patients to undertake a six-week gait retraining programme and reported a decrease in the EKAM and an improvement in symptoms and pain. Six weeks of gait retraining resulted in an average 20% reduction in EKAM1 (3.1 (1.4) %BW.h vs 2.6 (1.5) %BW.h, post-training compared to baseline and no change in EKAM2. Furthermore, Simic et al., (2013) observed a reduction in EKAM1 when patients walked with a modified, toe in gait (3.7 Nm/(BW × HT)% (3.3, 4.2 95% CI) vs 3.5 Nm/(BW × HT)% (3.0, 4.0 95% CI). Greater results were detected in patients with more varus knees.

A recent meta-analysis (Wang *et al.*, 2020) was the first meta-analysis to analyse the effects of foot progression angle (FPA) modification during walking on the EKAM peaks and knee angular adduction impulse (KAAI) between healthy individuals and patients with mKOA. The study found that toe in gait reduced EKAM1 but increased EKAM2. The subgroup effects of FPA modification were inconsistent. For healthy individuals, toe in gait lowered EKAM1 and KAAI, and for patients with mKOA, and toe in gait did not affect EKAM or KAAI. It was concluded that age, BMI, and knee alignment might also affect the outcome of FPA modification. Toe in gait reduced EKAM1 peak (standard mean difference (SMD): -0.8; 95%CI: -1.1~-0.5) and KAAI (SMD: -0.5; 95%CI: -0.9~-0.1) comparing mKOA patients and healthy individuals.

2.6.3.4.3 Increased step width gait style

Limited research has been undertaken assessing the feasibility of an increased step width as an altered gait intervention to reduce medial compartment knee joint loading. Two single-subject studies evaluated the effect of increased step width, achieved by increasing the frontal plane distance between feet during consecutive steps (Fregly et al., 2008; Reinbolt et al., 2008).

Favre et al. (Favre *et al.*, 2016) highlighted the interactions of a general combination of gait modifications (increasing step width, toeing-in, and increasing trunk sway) associated with reductions in EKAM first peak. These interactions are particularly important because, as shown in this study, some gait variables are difficult to modify without inducing involuntary secondary changes in other gait variables. Understanding that gait retraining is isolated to a particular gait measure will aid in the design of gait retraining programs and in the guidelines provided to the participants of these programs, as it demonstrates the importance of considering an overall scheme of altered walking mechanics.

When asked to walk normally, participants walked with a mean progression angle of ~11˚, step width of 0.036 m, speed of 1.4 m/s, and trunk sway of 1.7˚. Participants successfully followed the instructions to modify gait when they were instructed to do so. The four instructions to modify gait also induced involuntary secondary changes: instructions to modify progression angle influenced step width; instructions to modify step width had effects on progression angle, walking speed, and trunk sway; instructions to modify walking speed had an effect on progression angle; and instructions to modify trunk sway had an effect on step width.

Thus, retraining programs should not instruct participants to modify a particular gait variable without considering secondary changes to other gait variables. Interestingly, the larger secondary changes were the increase in step width induced by the instructions of decreased progression angle or increased trunk sway.

2.6.3.4.4 Medial thrust gait style

Fregly (2008) evaluated the ability of a medial thrust gait to reduce mKCF in the knee. The effectiveness of gait pattern was evaluated using internal contact force data collected from a single patient with a force-measuring knee replacement. It was hypothesised that both gait patterns would produce the largest reductions in mKCF near 25% and 75% of stance phase, which are approximately the locations of peak contact force.

Medial thrust during gait has been identified as reducing the EKAM (Fregly *et al.*, 2007; Fregly, D'lima and Colwell, 2009; Gerbrands *et al.*, 2017), and was found to be the most effective EKAM reducing gait modification in 43% of participants in a study by Gerbrands et al. (Gerbrands, Pisters and Vanwanseele, 2014) which compared the reduction in EKAM using various gait alteration strategies (trunk lean, medial thrust, lateral trunk sway, and toe out) in 37 healthy participants. The aim of this study was to determine the most effective gait retraining strategy to reduce EKAM peak and impulse, to determine if the same strategy is the most effective for each participant and if the efficiency of the strategy is related to how well the subjects can follow the instructions. The overall EKAM peak was reduced significantly by trunk lean, medial thrust, and reduced vertical acceleration, and EKAM impulse by medial thrust, trunk lean and toe out. Trunk lean and medial thrust affected both overall peak and impulse and showed the greatest EKAM reduction. This suggests that dynamically reducing the knee joint frontal plane lever arm has the highest potential to reduce both peak and cumulative knee load during gait. For these two conditions Gerbrands et al. (2014) results regarding EKAM peaks fall within the range of findings in literature.

2.6.3.4.5 Toe out gait style

Biomechanical factors affecting the EKAM include walking with a greater toe out angle (Chang *et al.*, 2007; Whelton *et al.*, 2017), which shifts the GRF vector closer to the knee joint centre decreasing the moment arm and thereby reducing the EKAM (Gerbrands, Pisters and Vanwanseele, 2014). Chang et al., (2007) identified greater toe out angle was inversely related to the EKAM during the late stance of gait in subjects with both healthy and osteoarthritis-stricken knees.

The unaltered FPA is approximately 5°, therefore indicating that the toes point slightly outward during unaltered gait (Shull et al., 2013). The toe out angle of the foot was found to increase during walking in patients with mKOA (Baliunas *et al.*, 2002; Chang *et al.*, 2007; Whelton *et al.*, 2017) and has been found to reduce the EKAM during walking. The toe out angle is proposed as a compensatory mechanism to unload the knee, achieved by transforming a proportion of the EKAM into a flexion moment in early stance phase and therefore partially shifting the load at the knee joint away from the medial compartment to other structures (Jenkyn *et al.*, 2008). The toe out occurs with lateral placement of the COP which shifts GRF nearer to the knee joint centre. This leads to a decreased GRF moment arm length at the knee, which in turn reduces the EKAM (Hurwitz *et al.*, 2002).

The reduction in the second peak of the EKAM observed with a toe out gait occurs mainly because of a decreased moment arm, caused by a lateral shift in the path of the COP. This shift occurs in late stance as the COP moves towards the toes and causes the GRF to pass closer to the knee joint centre, presumably without changing the angle of the GRF vector with the ground. This mechanism of the toe out gait seems to be specific to the second peak of the EKAM, as the COP travels further towards the toes to generate the second peak when the foot is externally rotated. By contrast, the COP is located closer to the heel (that is, positioned more medially) during the first peak of the EKAM and would, therefore, be far less affected than the second peak by a toe-out gait.

Studies have reported a reduction in the second (but not the first) peak of the EKAM in patients with mKOA when the foot is externally rotated by between 10° and 21° beyond the natural foot position during walking (Chang *et al.*, 2007; Jenkyn *et al.*, 2008; Whelton *et al.*, 2017).

The lack of consensus regarding the reductions in EKAM1 achieved with a toe out gait could relate to the fact that this foot position can result from external rotation at either the ankle or hip. The mechanism to reduce EKAM1 requires external rotation of the knee joint axis, which can only be achieved via external rotation at the hip.

With respect to the long-term influence of toe out gait, an increased baseline toe out angle was associated with a reduced likelihood of disease progression in patients with mKOA over an 18-month follow up period (Chang *et al.*, 2007)**.** Although data relating to the long-term effects of toe out gait are scarce, its immediate effect is to consistently reduce EKAM2, with less consistent effects upon the first peak of this parameter. Despite this strategy being relatively simple and not requiring any equipment, it does require permanent adoption of an altered gait by the patient. nevertheless, if patients can adhere to this strategy, it offers potential for reducing the progression of mKOA.

Individuals who naturally self-select a toe out gait have been shown to have reduced risk of disease progression in a longitudinal cohort study (Chang *et al.*, 2007), indicating the importance of toe out angle in the study of mKOA. Cochrane et al. (Cochrane, Takacs and Hunt, 2014) explored the biomechanics of toe out gait, as well as the differences between older individuals with mKOA and young, healthy individuals without mKOA. This study was the first to examine changes in external rotation throughout the lower limb during a toe out gait modification and to compare self-selected and modified external rotation angles during gait in young, healthy individuals with older individuals with mKOA. It was found that toe out gait alteration was achieved primarily from rotation of the shank and foot, and to a lesser extent the thigh. Surprisingly, when comparing groups, it was observed that young, healthy individuals performed self-selected and toe out gait in a similar manner as their mKOA counterparts. This latter finding suggests that performance of toe out gait retraining may not be influenced by the presence of mKOA. The study by Uhlrich et al. (2020) evaluated the importance of personalisation when selecting FPA modifications that aim to reduce the peak EKAM in individuals with mKOA. Sixty-six percent of individuals reduced their larger EKAM peak by at least 5% with a personalised FPA modification, which is more than ($p<0.001$) the 23% of individuals who reduced it with a uniformly assigned toe out modification (Uhlrich *et al.*, 2020).

During toe out gait alteration, the shank and foot exhibited greater changes in external rotation than the thigh. This finding provides further evidence to support the link between shank and foot biomechanics when performing a toe out gait retraining. A recent meta-analysis (Wang *et al.*, 2020) concluded that toe out gait reduced EKAM2 and KAAI. The subgroup effects of FPA alterations were inconsistent. Toe out gait reduced EKAM2, but not the first peak or the KAAI. For patients with mKOA, toe out gait was found to reduce EKAM2 and KAAI. As previously stated, age, BMI and knee alignment might also affect the outcome of FPA gait retraining.

2.7 Acknowledging the exclusion of Electromyography

This PhD does not incorporate EMG data that was collected within the longitudinal data collected as part of the work package in the Biomechanics and Bioengineering Research Centre Versus Arthritis. There are several reasons for this. First, a previous PhD candidate within the MSKBRF has incorporated EMG into their work evaluating muscle contraction changes due to HTO (Ghazwan, 2017). As discussed previously in this thesis, colleagues within the research group of the author of this thesis has published work on muscle activation pre-HTO (Ghazwan et al., 2022). This work as well as Ghazwan's PhD thesis has highlighted the different muscle strategies during the different stages of mKOA. Second, the opportunity of learning the COMAK pipeline at KU Leuven was delivered in such a way that EMG data was not included within the pipeline. the key idea behind learning the pipeline was to get a thorough appreciation of technique before adapting accordingly to include EMG. Third, time restrictions were a key reason as to why EMG data was not included within this work.

Notwithstanding the above, the author of this thesis has full appreciation of EMG and the importance of the work in better understanding MSK modelling. The author of this work along with MSKBRF foresee using EMG in the future building on the work from this thesis. During this PhD, research had been undertaken by Booij et al. (2020) which outline muscle activation patterns when adopting different gait styles. This was the first study that studied gait retraining and the effect on muscle activations. Future work that is planned from this PhD thesis is to undertake studies to compare the EMG data that has been collected to the findings presented from the Concurrent Optimisation of Muscle Activations and Kinematics (COMAK) framework.

2.8 Combining non-surgical and surgical interventions

This thesis focuses on assessing two proposed methods of reducing medial knee joint loading in individuals with mKOA: and thus, reducing the progression of mKOA. These two proposed methods can be broadly categorised as (a) surgical intervention, in the form of HTO, and (b) non-invasive approach of using an altered gait style to reduce knee joint loading; namely 3 altered gait styles (b1) toe out gait (b2) wide stance gait (b3) medial thrust gait.

This PhD has three overall aims. First, to evaluate the potential biomechanical merits of an altered gait intervention prior to undergoing an HTO. This will establish whether gait alterations prior to surgery offloads the diseased part of the knee, which could potentially prevent further deterioration of the knee joint. If this is the case, there is potential for gait alterations to offer relief to patients whilst they await surgery or potentially offer an alternative non-surgical option in some cases. Second, to evaluate the potential biomechanical merits of gait alterations following HTO in view of prolonging the benefits of an HTO; assessed at 12 months post-HTO. Third, to produce a set of recommendations for future work based on the findings from this thesis.

Preliminary findings by our group aimed to establish differences in EKAM1&2 between toe out and HTO (Whelton *et al.*, 2017). These treatments aim to offload the medial compartment with dynamic or anatomical compensation of varus knee deformity. The study aim was to compare these adaptations, and to identify if any benefit in EKAM can be attained with toe out after HTO, to compare the benefits of this gait adaptation after the deformity is corrected. Additionally, the aim was to assess toe out as an alternative to surgery or an addition to surgery by its effect on EKAM, which had not previously been reported. The Preliminary findings by Whelton et al. (2017) demonstrated that a reduction in second peak EKAM, but a small increase in the first peak with toe out gait for the varus aligned individuals. However, this study was the first report of understanding the effects of an altered gait pre and post HTO for individuals with varus deformity. This study focused on the biomechanical changes at the knee only, disregarding any potential effects at the ankle and hip joints.

This literature review chapter has addressed what current literature is available in understanding knee joint loading in individuals with mKOA. It is clear from this literature review that load and motion alterations lead to mKOA and once developed progresses the condition further. From the literature there are several key mechanical and gait variables of interest have been proposed to impact and/or been shown to be surrogate measures of cartilage damage. These include, but not limited to, spatial-temporal parameters such as gait speed, variables that reflect medial compartment knee joint loading in the form of EKAM and medial contact forces of the tibiofemoral joint, medial-lateral tibiofemoral joint contact forces, net moments at the knee joint, and knee joint kinematics. Chapter 3 will focus on better outlining what variables have been researched in this field for ankle and hip biomechanics and how it is important to have an appreciation of these joints when adopting altered gait styles to offload the knee joint.

2.9 Principal Component Analysis and the Cardiff Classier

2.9.1 Principal Component Analysis

The collection of HMA data results in an extensive amount of temporal information. Generally, gait variables are normalised using 101 data points to a percentage of stance phase or the entire gait cycle.

To allow a meaningful statistical analysis to be performed, these temporal waveforms must be summarised using a smaller number of discrete variables. This has resulted in an extensive application of data reduction techniques to HMA data (Chau, 2001). A common method of reducing data is to define discrete parameters of the waveform. For example, during the swing phase of gait, the knee must flex to achieve toe clearance as the limb progresses forward. A reduction in this angle might be related to an indication of an increased risk of trips or falls. Choosing which discrete parameter to calculate, however, is subjective and may be discarding valuable information. While consistent peaks and troughs may be identifiable in healthy subjects, often the waveforms of pathological subjects will have completely different characteristics. Furthermore, by completely discarding the rest of the waveform, important information regarding inter-subject variability can be lost. Deluzio et al. (1999) demonstrated that PCA was a useful technique in the reduction of temporal biomechanical data. The study found that principal component scores were sensitive to gait changes associated with KOA, as well as changes following a partial knee replacement. PCA has since been successfully applied at Cardiff University to help distinguish between OA and non-pathological control groups and hence objectively measure changes in gait parameters following TKR surgery (Jones & Holt, 2008; Metcalfe et al., 2013; Whatling et al., 2008). Principal Component Analysis is a multivariate data analysis technique which applies an orthogonal transformation of an ‘n’ dimension dataset of potentially correlated variables, to arrive at a new n dimension dataset of linearly uncorrelated variables. The first dimension of the new dataset will represent the greatest amount of variance in the dataset, and so forth until the nth dimension, which will often end up representing an extremely small amount of the total variance.

2.9.2 The Cardiff Classifier

The Cardiff Classifier is a novel approach for generating an overall index of gait function. Previous applications of the Cardiff Classifier include the differentiation of pathologic gait function seen in individuals with KOA and healthy controls, and to monitor postoperative recovery following TKR (Biggs et al., 2019; Biggs et al., 2019; Jones et al., 2006; Metcalfe et al., 2013) and patients who undergo total hip arthroplasty (Biggs et al., 2021; Whatling et al., 2008). To date, application of the Cardiff Classifier in patients undergoing HTO, and whether it has any predictive value for post HTO outcomes, remains unknown.

Simplifying 3D gait analysis data into a single metric describing the overall gait pattern would be of great value in clinical practice to discern whether the overall gait function is affected and to what extent, and to inform healthcare providers and patients what can be expected in terms of change in gait patterns. Further, knowledge on whether it is patients with the greatest perceived recovery who also have the best biomechanical outcomes, and vice versa, is limited. A comprehensive metric, accounting for interdependencies of biomechanical variables, would facilitate interpretation of results of 3D gait analyses among clinicians, and facilitate monitoring over time and following interventions.

Finally, to the author of this thesis' knowledge, biomechanical HTO research mainly focuses on discrete metrics from a waveform. It would therefore be of interest to establish whether waveform analysis and the introduction of the Cardiff Classifier yield similar results to the conventional method of analysis.

2.10 Novelty from this thesis

Recent literature has suggested that gait retraining has the potential of reducing medial knee joint loading. However, this work has not focused on individuals with varus deformity as well as not appreciating what the biomechanical consequences are at the ankle and hip joints. This literature review has highlighted several gaps in research that exist in the current literature regarding interventions specific to individuals with varus tibiofemoral alignment:

First, there is a lack of understanding on whether altered gait can be used instead of HTO to dynamically align the knee and to reduce medial compartment joint loading.

Second, whether altering an individual's gait could have the potential to compliment surgery to prolong the benefits of HTO and to slow down the progression of mKOA.

Third, most of the research into gait retraining assesses EKAM as the sole indication of medial compartment joint loading. There is limited work that aims to simulate and predict internal joint loading on mKOA patients, and no research on individuals that have undergone HTO.

Fourth, almost all the research undertaken within this literature review has focused on discrete metrics. As well as the conventional method of comparing peaks and troughs, this thesis introduces PCA and the Cardiff Classifier to the HTO cohort for the first time. No research has used full waveform analysis to identify the merits of HTO as options to reduce medial compartment knee joint loading. This thesis introduces the use of PCA and the Cardiff Classifier as a new approach to understanding the waveforms as a whole and the level of belief in OA.

Finally, there is a lack of understanding on what the biomechanical effects are of altered gait to the ankle and hips joints. This is further highlighted in Chapter 3 of this thesis.

CHAPTER 3: SYSTEMATIC REVIEW

Title: Does Gait Retraining Have the Potential to Reduce Medial Compartmental Loading in Individuals with Knee Osteoarthritis Whilst Not Adversely Affecting the Other Lower Limb Joints? A Systematic Review

3.1 Background

As highlighted in the Literature Review chapter of this thesis, there is considerable research surrounding gait retraining and altering gait to reduce medial compartment knee joint loading. The purpose of this chapter was to undertake a systematic review to better understand what research was available in understanding what the consequences of knee joint loading is on hip and ankle joint loading. By the end of this chapter the reader should be aware that there is a severe lack of research in identifying adjacent joint biomechanics to the knee when altering gait to reduce medial knee joint loading. This chapter is based on the published systematic review by the author of this PhD thesis (Bowd et al., 2019).

3.2 Rationale

Medial knee compartment overloading is associated with mKOA progression (Miyazaki *et al.*, 2002) and radiographic disease severity (Sharma et al., 1998). EKAM has been the most used variable to assess/indicate medial knee joint loading (Simic *et al.*, 2011). EKAM acts to force the tibia into varus and has been validated as a reliable indicator of medial knee load (Birmingham *et al.*, 2007). Therefore, this moment is said to reflect medial-to-lateral knee joint load distribution during gait (Chang *et al.*, 2015). In the presence of an increased EKAM, the medial compartment of the tibial-femoral joint is hypothesised to experience increased load (Simic *et al.*, 2011).

The following altered gait styles are proposed to reduce medial knee joint loading as indicated by a decrease in EKAM values: wide stance gait (Reinbolt et al., 2008), toe out gait (Whelton *et al.*, 2017; Hunt *et al.*, 2018), toe in gait (Simic *et al.*, 2011), medial thrust gait (Fregly, D’lima and Colwell, 2009; Barrios, Crossley and Davis, 2010), trunk lean gait (Simic *et al.*, 2012), and medial foot weight transfer of the foot (Erhart-Hledik *et al.*, 2017).

Simic et al.'s systematic review (Simic *et al.*, 2011) analysed gait retraining strategies for altering medial knee joint load and concluded that different gait alterations exert different effects on dynamic knee load at varying points throughout the gait cycle. Of the 14 gait retraining modifications identified, sufficient data was not available to address whether there are any changes at other lower extremity joints with the implementation of gait alterations to reduce EKAM (Simic *et al.*, 2011). It has been suggested that an increased loading rate in the lower extremity joints may lead to a faster progression of existing mKOA and to the onset of mKOA at joints adjacent to the knee (Simic *et al.*, 2011). Therefore, any interventions for mKOA should be assessed for their effects on the mechanics of all joints of the lower extremity. Therefore, the current systematic review aimed to establish the body of evidence on how changes to EKAM effects adjacent joints to the knee because of modifying an individual's gait. The interaction between hip, knee and ankle biomechanics is not well understood when modifying gait in mKOA patients and needs to be reviewed to make clinical decisions on the role of gait retraining in reducing knee joint pain and discomfort (Richards *et al.*, 2018); justifying the necessity of a systematic review of the current literature.

Previous research has indicated that patients with mKOA experience abnormal loads of their major weight bearing joints bilaterally, and abnormalities persist despite treatment of the affected limb (Metcalfe et al., 2013). Further treatment may be required if we are to protect the other major joints following joint preserving interventions, such as HTO for individuals with varus deformity and mKOA. No systematic review has established what effects changing knee joint loading via gait retraining has on the other ipsilateral and contralateral joints in the lower limbs as well as trunk biomechanics. To lower knee joint loading, altered gait styles will undoubtedly change the kinematics and/or kinetics at the neighbouring joints, e.g., for toe in gait the foot is at a more inverted position throughout the gait cycle. The clinical benefit of reducing the EKAM variables is questionable if there are detrimental consequences to other joints of the lower body. If the goal of gait retraining is to alleviate pain and to slow down the deterioration of medial joint loading at the knee itself whilst not adversely affecting hip and ankle joint function, then an appreciation of what biomechanical changes is occurring at the hip and ankle joints is fundamental.

Allowing patients to select their own gait modifications, may have some benefits over using prescribed modifications. However, this may also result in modifications which reduce gait energy efficiency or increase loads on other joints of the lower limb or trunk which may have adverse biomechanical consequences (at other joints). It is important therefore to evaluate the loading at the hip and ankle joints. The effect of self-defined gait modifications on the hip and ankle joint moments remains an important unanswered question. Previous studies on gait retraining in mKOA patients have largely neglected the effects of modifications at the hip and ankle joints. Reductions in EKAM may cause an increased hip adduction moment may increase loading at the hip, a risk factor for development of hip OA. The clinical relevance of changes at the ankle and hip joint are unclear. This thesis has taken the viewpoint that when adopting an altered gait style or undergoing an HTO that if the hip and ankle joint kinematic and moment changes go further away from the control cohort may lead to adverse consequences over a prolong period of time.

3.2.1 Objectives

The following objectives are the same as published in Bowd et al. (2019). The following 3 objectives of this systematic review were to:

(1) identify the consequences of gait retraining on the biomechanics of the ankle and hip as well as trunk and pelvis biomechanics, and

(2) establish whether gait styles and gait retraining can reduce medial knee loading as assessed by first and second peak EKAMs. Additionally,

(3) a third objective was to outline patient/participant reported outcomes on how easy the gait retraining style was to implement. This would aid the clinical translation of aforenamed gait retraining techniques.

3.3 Methods

3.3.1 Protocol and registration

This work followed preferred reporting items for systematic reviews and meta-analysis (PRISMA) guidelines (Moher *et al.*, 2009) and was registered with the International Prospective Register of Systematic Reviews (PROSPERO) on the 23rd January 2018 (registration ID: CRD42018085738) (available at https://www.crd.york.ac.uk/prospero/display_record.php?RecordID=85738).

3.3.2 Eligibility criteria

This systematic review was restricted to peer-reviewed published articles that were relevant to the research question.

To be eligible for this systematic review, articles had to:

1. evaluate the effect of a gait retraining technique on EKAM, and
2. evaluate at least one biomechanical variable at the ankle and/or hip.

There was no restriction on whether the participants of a study had to be clinically diagnosed as having mKOA. The reason for including studies involving gait retraining on healthy participants was due to the anticipated lack of studies using participants with symptomatic KOA, as evidenced in previous systematic reviews on similar topics (Simic *et al.*, 2011; Richards *et al.*, 2017). In the interpretation of results, healthy and mKOA cohorts are presented separately to establish any biomechanical differences between them when adopting a gait style. A novelty to this thesis is the unique cohort having correctable varus deformity and mKOA and so although it is important to understand the changes in healthy individuals, it is of more importance to understand the interventions to a cohort of individuals with mKOA.

3.3.3 Intervention

For this systematic review, gait retraining was defined as any researcher-initiated alteration of natural gait without the use of any devices or walking aids. Studies were included if they used 3D motion analysis and force-plate derived data during both natural and modified gait conditions as well as providing EKAM data. The altered gait style (intervention variable) was compared to the individual's natural level gait (control variable).

Studies evaluating post knee operations such as total knee replacements as well as studies that included participants with specific diseases and conditions which can affect the participant's gait were excluded.

3.3.4 Information sources

The following 12 databases were searched by the author of this thesis with the assistance of two experienced librarians up to January 2019 on the following databases: Cumulative index to Nursing and Allied Health (CINAHL, 1982-2019), EBSCO MEDLINE (MEDL) (1966-2019), Ovid Allied and Complementary Medicine Database (AMED) (1995-2019), Ovid EMCare (1995-2019), Ovid Joanna Briggs Institute (JBI) (1991-2019), Web of Science (1900-2019), BIOSIS Citation Index (Web of Science) (1926-2019), Scopus (1960-2019), Cochrane Library (Cochrane Library, DARE and Central), ProQuest British Nursing Index (BNI) (1994-2019), Turning Research Into Practice Pro (TRIP PRO) (1997-2019), British

Library e-theses online service (EThOS) (all years until 2019) and ProQuest Dissertations & Theses (1986-2019). Additionally, PROSPERO was searched for ongoing or recently completed systematic reviews.

3.3.5 Search

The search strategy was designed by following the PICO model (patient, intervention, comparison, and outcome) (Huang et al., 2006) and was purposely broad in approach to maximise saturation.

The electronic databases were searched through using the combination of key search terms organised into sets and combined with the operators ‘AND’ and ‘OR’ (Appendix A).

3.3.6 Study selection

Titles were assessed by the author of this thesis. The principal investigators for each ClinicalTrials.gov identifier number (NCT number) were contacted to ascertain what peer-reviewed papers had been published from these clinical trials. The author of this thesis and Dr Paul Biggs, a Postdoctoral Research Associate at the MSKBRF School of Engineering Cardiff University, assessed the abstracts of the remaining articles independently. To ensure consistency and for expert advice, articles that were included in the systematic review were collectively reviewed by three of the authors of the publication (Bowd et al., 2019). During a meeting, the key data that was to be extracted from each study was determined.

3.3.7 Data collection process

The author of this thesis extracted the data for the following items: study design, sample size, participant characteristics, gait retraining technique used, EKAM parameters evaluated, study duration, ankle and/or hip biomechanical analysis that was undertaken, and the main study findings.

3.3.8 Risk of bias in individual studies

Risk of bias was assessed using the Downs and Black quality index (Downs and Black, 1998) which is a validated index for non-randomised trials (Richards *et al.*, 2017) consisting of 27 items used to assess reporting quality (items 1-10), external validity (items 11-13), internal validity (14-26) and study power (item 27).

The tool has been used in various modified forms for gait focusing on interventions aimed at individuals with mKOA (McClelland, Webster and Feller, 2007; Simic *et al.*, 2011; Moyer *et al.*, 2015; Shaw *et al.*, 2017; Richards *et al.*, 2018). Piloting of the tool and agreeing on interpretation of the questions was undertaken by the author of this thesis and Dr Paul Biggs.

Risk of bias scores for individual studies were rated in line with previous systematic reviews on similar topics (Simic *et al.*, 2011; Richards *et al.*, 2017). Neither review ((Simic *et al.*, 2011; Richards *et al.*, 2017)) explicitly defined their boundaries in their papers and so it was inferred that 10-14 and 15-20 correspond with fair and moderate scores respectively.

3.3.9 Summary measures

The principal summary measure from each article was the within-group mean differences in hip and/or ankle data between natural level gait and the gait retraining intervention presented as a percentage difference from natural level gait. Standardised mean difference (SMD) effect sizes were also calculated for these metrics. The SMD using the hedges' g effect size is calculated as follows:

$$Hedges'g = \frac{M_1 - M_2}{SD *_{pooled}}$$

Where:

- $M1 – M2$ = difference in means.
- SD^{*}_{pooled} = pooled and weighted standard deviation.

EKAM has been used widely in the gait retraining literature as a surrogate measurement of medial knee joint loading (Simic *et al.*, 2011). For this review, 'natural level gait' is defined as an individual assessment of walking without any instruction to alter their ordinary walking pattern when being assessed in a motion capture laboratory. Finally, any data presented regarding participant perceptions on task difficulty was extracted to consider the practicality of translation to a clinical setting.

3.3.10 Changes from the original protocol

After analysing the data from the 11 studies that met the inclusion criteria, there was enough evidence to also include trunk and pelvic biomechanical data within the analysis and the review. Additionally, the decision was made after the databases were searched to include any information on how easy the gait retraining was to implement.

3.3.11 Synthesis of results

The narrative synthesis explores the relationship of the findings between the included studies by way of gait style comparisons and methodological quality. The standardised mean difference (SMD) using the hedges' g effect size was calculated for the change in EKAM and hip/ankle kinetic metrics (equation shown above). The SMDs were standardised according to small (0.2–0.5), medium (.51–0.8), and large (>0.8) in accordance with Cohen (1992) (Cohen, 1992).

3.3.12 Statistical analysis

Downs and Black scoring agreement between two reviewers were assessed using a Cohen's kappa coefficient (k) statistic, with reference to Landis and Koch's criteria where κ values >0.81 represent 'almost perfect' agreement (Landis and Koch, 1977). The Cohen's kappa is a statistical coefficient that represents the degree of accuracy and reliability in a statistical classification. It measures the agreement between two raters who each classify items into mutually exclusive categories.

$$K = \frac{p_o - p_e}{1 - p_e}$$

where p_o is the relative observed agreement among raters, and p_e is the hypothetical probability of chance agreement.

To interpret Cohen's kappa results you can refer to the following guidelines (Landis & Koch, 1977).

- 0.41 – 0.60 moderate agreement
- 0.61 – 0.80 substantial agreement
- 0.81 – 1.00 almost perfect or perfect agreement

kappa is always less than or equal to 1. A value of 1 implies perfect agreement and values less than 1 imply less than perfect agreement.

To estimate the SMD, the mean and standard deviation values were used. If mean and standard error mean (SEM) data were provided in the studies, standard deviation was calculated as SEM multiplied by the square root of the sample size. Standardised mean differences were calculated using the Hedges' g effect size. All results are presented as Forest Plots. The 95% confidence interval (CI) was calculated and presented for each effect size. A 95% confidence interval is a range of values that you can be 95% certain contains the true mean of the population.

3.4 Results

3.4.1 Study selection

Figure 8 outlines 184 studies identified; from which 11 were included within this review. The reviewers showed almost perfect (as defined above) agreement in assessing the quality of each included study, $k = 0.89$. The 11 included articles focused on assessing the effects of gait retraining on reducing EKAM as well as documenting biomechanical variables for the pelvis, hip, and ankle joints. All data presented in this systematic review is from the mKOA ipsilateral limb for the patients.

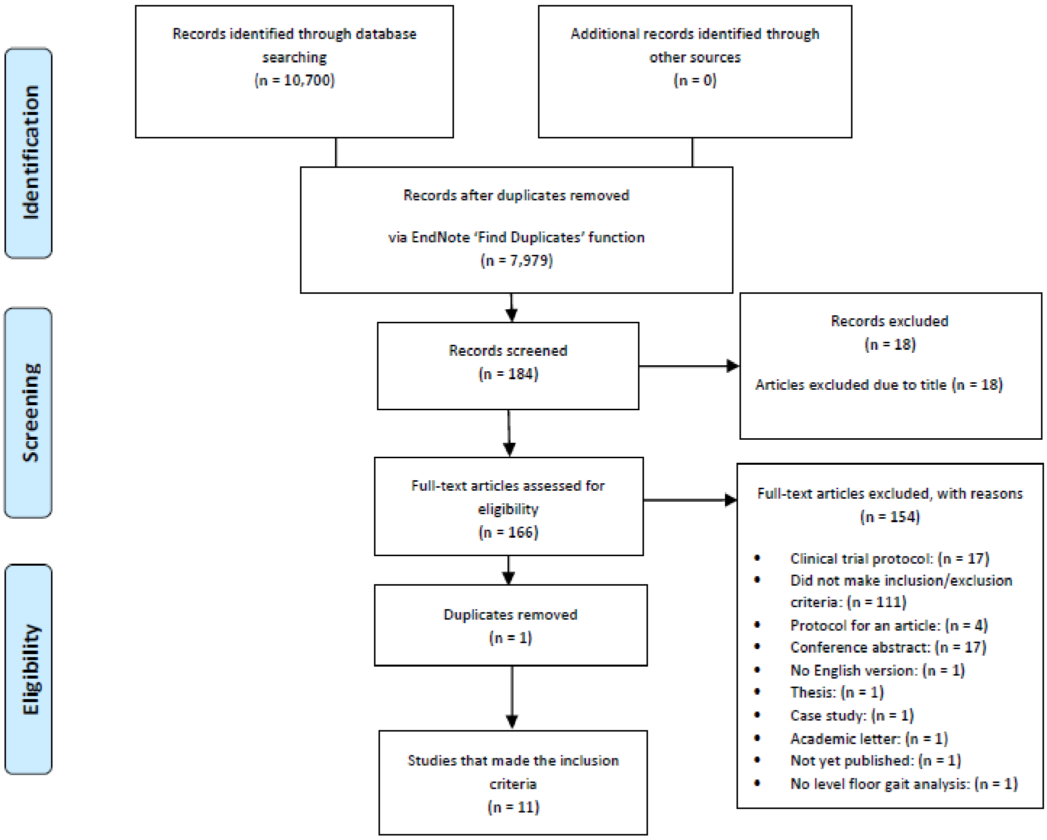


Figure 8 Flow chart

Figure 8 is extracted from (Bowd et al., 2019)

3.4.2 Study characteristics

Group demographics are shown in Table 3-1. Excluding Barrios et al. (2010), all other papers included in this systematic review utilised a within-subject design. The majority of included papers evaluated the immediate within-session effect as well as the potential benefits of gait retraining. The sample size of the papers included within this systematic review ranged from 8-40 participants. In terms of the population type, 6 of the 11 studies assessed healthy non-pathological participants, whilst 5 papers included KOA participants.

Table 3-1 Systematic Review: Group Demographics

Authors and year	Population	Gait retraining modification	Gait speeds (m/s) mean (std)	Over ground/treadmill walking	n (M: F)	Age (years) (mean ± (std))	Height (m) (mean ± (std))	Mass (kg) (mean ± (std))	BMI (mean ± (std))
Shull et al. (2013)	Symptomatic KOA (K/L grade ≥1)	• T-I	1.23 (0.21)	Instrumented treadmill	12 (7: 5)	59.80 (12.00)	1.71 (0.80)	77.70 (18.00)	26.50 (4.20)
Richards et al. (2018)	Symptomatic KOA	• Self-selection combination of T-I, WS and MT	N-R	Instrumented treadmill	40 (15: 25)	61.70 (6.00)	1.73 (0.10)	77.20 (11.00)	25.60 (2.50)
Erhart-Hledik et al. (2017)	Symptomatic KOA and physician-diagnosed radiographic medial compartment KOA (K/L grade ≥ 1)	• Medial weight transfer at the foot	Control [unaltered speed (1.28 (0.14)); fast speed (1.53 (0.18)))]; active feedback [unaltered speed (1.31 (0.12)); fast group (1.50 (0.15)).	Overground	10 (9:1)	65.30 (9.80)	NR	NR	27.80 (3.00)
Gerbrands et al. (2017)	Symptomatic KOA; physician-diagnosed with radiographic and fulfilment of the criteria by the American College of Rheumatology	• LT; • MT	Comfortable walking (1.21 (0.10)); MT walking (1.02 (0.19)); TL walking (1.08 (0.15)).	Overground	30 (10: 20)	61.00 (6.20)	1.71 (0.10)	75.70 (13.10)	NR
Charlton et al. (2018)	Radiographic medial compartment KOA (K/L grade ≥2)	• T-I • T-O	1.22 (0.15)	Overground and a treadmill	15 (6:9)	67.90 (9.40)	1.67 (0.11)	75.60 (15.00)	NR
Barrios et al. (2010)	Healthy	• HIR strategy	1.46 (2.50)	Overground	8 (7:1)	21.40 (1.60)	1.75 (0.07)	71.70 (8.80)	NR
Hunt et al. (2011)	Healthy	• LT	TL (1.42 (0.18)); small TL (1.36 (0.19)); medium TL	Overground	9 (3:6)	18.60 (0.7)	1.71 (0.11)	65.20 (13.80)	NR

			(1.36 (0.19)); large TL (1.40 0.19)).						
Mündermann et al. (2008)	Healthy	• Increased medio-lateral trunk sway	Unaltered gait (1.48 (0.17)); medio-lateral trunk sway (1.44 (0.15)).	Overground	19 (12: 7)	22.80 (3.10)	1.75 (0.97)	70.50 (16.30)	NR
Van den Noort et al. (2015)	Healthy	• HIR feedback	1.00 (0.09)	Instrumented treadmill	17 (8: 7)	28.20 (7.60)	1.78 (0.07)	71.60 (12.50)	NR
Dunphy et al. (2016)	Healthy	• Contralateral pelvic drop	1.31 (0.12)	Instrumented treadmill	15 (7: 8)	25.00 (2.65)	1.73 (0.08)	76.70 (16.50)	25.70 (5.06)
Khan et al. (2017)	Healthy	• T-O; • T-I	Slow (0.85); unaltered (1.18); fast (1.43)	Overground	20 (8: 12)	29.00 (4.10)	1.65 (0.11)	59.30 (10.40)	NR

HIR = hip internal rotation; LT = lateral trunk lean; T-I = toe in gait; EKAM = knee adduction moment; WS = wide stance gait; MT = medial thrust gait; T-O = toe out gait; BMI = body mass index; K/L grade = Kellgren and Lawrence system; m: metre; NR = not reported; M: male; F: female; std: standard deviation.

Table 3-1 is extracted from (Bowd et al., 2019)

Table 3-2 Systematic Review: Disease Severity

Authors and year	**Population**	**K/L grade**	**PROMS**
Shull et al. (2013)	Symptomatic KOA	II: 4, III: 7, IV: 1	WOMAC pain (mean ± SD): 74.20 (19.00) [max. 100], WOMAC Function (mean ± SD): 81.70 (21.60) [max. 100]
Richards et al. (2018)	Symptomatic KOA	I: 19, II: 8, III: 9, IV: 4	WOMAC pain (mean ± SD): 5.35 (3.13) [max. 20], WOMAC Function (mean ± SD): 19.10 (12.08) [max. 68], WOMAC stiffness: 3.25 (1.96) [max. 8], Baseline pain: 3.05 (2.16) [max. 10]
Gerbrands et al. (2017)	Symptomatic KOA	NR	KOOS Pain (%): 57.50 (13.40), KOOS Function (%): 62.30 (14.10)
Erhart-Hledik et al. (2017)	Symptomatic KOA	All above I.	Daily pain score: 3.20 (3.60)
Charlton et al. (2018)	Radiographic KOA	II: 7; III: 8	WOMAC pain (mean ± SD): 4 (2.20) [max. 20], WOMAC stiffness (mean ± SD): 3.00 (1.30) [max. 8], WOMAC Function (mean ± SD): 15.40 (8.00) [max. 68]
Hunt et al. (2011)	Healthy	NR	NR
Barrios et al. (2010)	Healthy	NR	KOOS-SR score (mean ± SD): 0.70 (0.90) [max. 20]
Mundermann et al. (2008)	Healthy	NR	NR
Van den Noort et al. (2015)	Healthy	NR	NR
Dunphy et al. (2016)	Healthy	NR	NR
Khan et al. (2017)	Healthy	NR	NR

PROMS = Patient-reported outcome measures; K/L grade = Kellgren and Lawrence system; WOMAC = The Western Ontario and McMaster Universities Osteoarthritis Index; KOOS = Knee injury and Osteoarthritis Outcome Score; NR = not reported; OA = osteoarthritis. Barrios et al. (2010) used the KOOS-SR score (Function in Sport and Recreation) which ranged from 0-20, a score of 0 indicating no difficulty. Shull et al. (2013) measured WOMAC levels on the day of assessment, with the scale ranging from 0-100 with 100 indicating no pain and perfect function (Bellamy et al., 1988). Richards et al. (2018) measured WOMAC levels on the day of assessment, evaluating the pain and function of the participant in the past week, with the lower the scoring of pain out of 20 equating to the lower the pain, and the lower the score out of a maximum of 68 being the better the function of the participant. Gerbrands et al. (2017) assessed pain and function using the Knee injury and Osteoarthritis Outcome Score (KOOS), scores are presented as a percentage, where 0% represents extreme problems and 100% represents no problems. Table 3-2 is extracted from (Bowd et al., 2019)

3.4.2.1 Hip kinetic biomechanics

The two studies that reported peak external abduction moment data did so when assessing trunk lean as a gait retraining intervention. These two studies showed a null to small effect (Hunt *et al.*, 2011). The results presented for the peak external abduction moment when adopting a trunk lean has somewhat of a dose-response effect; with the largest trunk leans resulting in the small effect (trunk lean of ~12° resulted in a standardised mean difference (SMD) of 0.23 with a confidence interval (CI) of -0.69 to 1.16). These findings are in comparison to a large increase due to adopting a similar trunk lean (10°) in Mündermann et al. (2008) (SMD 0.89 CI 0.23 to 1.56). These two studies assessed the trunk intervention in a healthy population as opposed to a mKOA cohort. Therefore, these findings lack external validity and impacts on any clinical findings that can be taken from the research.

Only one study assessed peak external hip adduction moment (EHAM) (Richards *et al.*, 2018). This paper evaluated the effect of real-time feedback in a population of individuals with mKOA. This work involved several different feedback mechanisms to reduce EKAM1 and indicated a null effect (SMD <0.2).

Three trunk lean studies reported findings on first peak EHAM with conflicting data (Gerbrands et al., 2017; Hunt et al., 2011; Mündermann et al., 2008). The differences in findings may be due to two distinct groups being assessed; one paper studied an OA cohort group (Gerbrands *et al.*, 2017) (indicating a small effect increase (SMD 0.36 CI -0.15 to 0.87). The other two papers assessed healthy cohorts (Hunt et al., 2011; Mündermann et al., 2008). These last two papers indicate a small and a large effect size decrease in late stance EHAM.

Only two studies assessed the effects of EHAM changes in the second half of stance when adopting a trunk lean intervention (Gerbrands et al., 2017; Hunt et al., 2011). The findings were that the greater the trunk lean implemented, the lower the reduction in late stance peak EHAM with increasingly higher effect size associated with the change accordingly to the increase in trunk lean angle. However, caution must be had because only one study assessed a patient population and the other assessed a healthy cohort who did not have KOA (Hunt *et al.*, 2011). These findings in late stance peak EHAM for a trunk lean intervention is not the same as to when adopting a medial thrust gait style. Adopting a medial thrust gait suggests that second peak EHAM has a small effect size increase (SMD 0.25 CI -0.26 to 0.75).

Peak external hip flexion moment was reported in one study and indicated that a null effect was a result of all four different feedback mechanisms (SMD <0.2) (Richards *et al.*, 2018). Barrios et al. (2010) was the only study that assessed maximum hip axial loading rates which also indicated a null effect (SMD -0.08 CI -0.72 to 0.55).

It can therefore be stated that there is a severe lack of hip kinetic data being reported in KOA gait retraining studies for which future work should address. As a result of the severe lack of hip kinetics being reported and the lack of patient-specific studies being performed, caution must be taken when interpreting these results. On a final note, the 95% CI was large for all variables assessed, with most metrics 95% CI measured crossing the line of null effect. This means that no firm conclusions can be made from the data available.

3.4.2.2 Ankle kinetic biomechanics

Only Gerbrands et al. (2017) assessed first and second peak external inversion moment. Gerbrands et al. assessed the effectiveness of trunk lean at reducing EKAM. The findings of which are that first half external inversion moment had a null effect for trunk lean (SMD 0 CI -0.51, 0.51) but potentially increasing when adopting a medial thrust gait (SMD 0.49 CI -0.02, 1.01). In late stance, Gerbrands reported a null effect for trunk lean (SMD 0.15 CI -0.66, 0.36) and a small effect in a reduction in peak external inversion moment when adopting a medial thrust gait (SMD 0.33 CI -0.84, 0.18). Although this is only reported in one study, the study was rated as moderate (15/25) and assessed an OA population.

Only Richards et al. (2018) reported on peak frontal and sagittal plane external moments. There was a high standard deviation in the frontal plane results and so the interpretation and clinical significance is difficult to establish. In terms of the sagittal plane moment, there was a null effect for the interventions. This study was rated as moderate (15/25) and assessed an OA population.

Charlton et al. (2018) was the only study that reported findings on peak external ankle eversion/inversion and plantarflexion/dorsiflexion moments; all of which had a 95% CI crossing the line of null effect. This makes the interpretation of results difficult to establish. Again, peak external ankle plantarflexion/dorsiflexion moment impulses crossed the line of null effect resulting in difficulty in interpreting the results (Charlton *et al.*, 2018).

When adopting a toe out gait, peak external ankle eversion moment impulse reduced. This is in comparison to adopting a toe in gait which resulted in a null effect. There is a large effect size for peak external ankle inversion moment impulse. This indicated that when adopting a toe in gait there is an increased load experienced (SMD of 1.43 [0.6, 2.26]). This study was rated as moderate (15/25) and assessed an OA population.

Shull et al. (2013) which assessed toe in gait as an intervention was the only study which reported findings on the COP at EKAM1 and EKAM2 (Shull et al., 2013); both of which indicating no effect size (SMD < 0.2) when adopting a toe in gait style. Erhart-Hledik et al. (2017) found a large effect size increase in the first half of stance COP due to their intervention and small size increase in the second half of stance COP (SMD of 0.85 and 0.28 respectively). Caution should be had when interpreting these findings as the 95% CI for these two variables cross the line of null effect. Only one study (Barrios et al., 2010) assessed maximum ankle axial loading rates, and again, indicated a null effect (SMD -0.15 CI -0.79, 0.49).

All ankle kinetic data presented above utilised an mKOA population within their studies, with varying methodological scores (14-17 out of 25); having scored low on external validity. The findings above should be approached with caution as the metrics crossed the line of null effect.

3.4.2.2.1 Trunk & pelvis biomechanics

Out of the 11 papers included in the systematic review, six reported pelvic/trunk biomechanics data (Dunphy et al., 2016; Gerbrands et al., 2017; Hunt et al., 2011; Mündermann et al., 2008; Shull et al., 2013; van den Noort et al., 2014).

Shull et al. (Shull et al., 2013) assessed the effectiveness of a toe in gait in reducing EKAM as well as reporting findings on lateral trunk sway at first or second peak EKAM. The results being that there were no significant differences between the altered gait style and the unaltered natural gait. This contrasts with the Gerbrands et al. paper (Gerbrands *et al.*, 2017) which reported a significant increase in peak trunk angle between natural gait to both trunk lean and medial thrust gait modifications. Two papers presented trunk biomechanics (Hunt et al., 2011; Mündermann et al., 2008). Van den Noort et al. (van den Noort *et al.*, 2014) outlined trunk and hip changes with and without hip internal rotation feedback on hip internal rotation. The Dunphy et al. paper (Dunphy *et al.*, 2016) assessed the influence of contralateral pelvic drop. Dunphy et al. documented differences in pelvic drop angle between natural gait and contralateral pelvic drop gait style.

Table 3-3 Systematic Review: Joint Kinematics When Adopting a Gait Style

	Trunk and pelvis	Hip	Ankle, foot and COP
Shull et al. (2013)	• N-S LT sway between T-I gait (0.20 (2.00)) and unaltered gait (0.50 (2.30)) at first peak KAM, p = 0.44; • N-S LT sway between T-I gait (0.40 (1.30)) and unaltered gait (0.60 (1.20)) at second peak KAM, p = 0.48; • N-S peak lateral trunk sway angle between unaltered gait (1.50° (1.60)) and T-I gait (1.30° (0.50)), p = 0.49.	• N-S findings for peak HIR angle between unaltered gait (3.20° (3.80)) and T-I gait (4.10° (4.10)), *p* = 0.18;	• Significant difference between unaltered gait FPA at first (3.30° (4.50)) and second (3.90° (4.60)) peak EKAM compared to FPA for T-I gait at first (-2.60° (6.30)) and second (-1.40° (6.40)) peak KAM; • Early stance, the CoP shifted laterally from unaltered gait (27 (77) mm) compared to 33 (79) mm), *p* = 0.04; • Late stance CoP did not significantly change between unaltered gait (30 (83) mm) and TI gait (30 (83)), *p* = 0.96.
Richards et al. (2018)	• N-R	• N-S changes in the peak HAM, *p* = 0.083; • N-S changes in peak HFM between unaltered gait and gait modifications, *p* = 0.182.	• Peak AAM was significantly increased compared to baseline during the second peak EKAM visual feedback trial and the final retention trial, *p* < 0.001; • N-S in peak AFM for any condition, *p* > 0.058; • FPA significantly more internally rotated during second EKAM visual feedback and retention trials, *p* < 0.001; • Patients significantly increased their step widths during all trials.
Gerbrands et al. (2017)	• During the MT the peak trunk angle significantly increased to 5.5° (3.7) and during the TL the peak trunk angle significantly increased to 16.1° (5.5) compared to unaltered walking trunk angle of 3.4° (1.8), p < 0.05.	• Early stance peak hip flexion angle significantly increased from unaltered walking (15.3° (37.7)) to 18.2 (37.2) during TL, *p* < 0.05. N-S in early stance peak hip flexion angle between unaltered walking (15.3 (37.7)) and MT (10.2 (21.1)), *p* > 0.05; • N-S findings in HAM between baseline walking trials and neither the TL, or MT gait retraining trials at both the first and second peak KAM, *p* > 0.05.	• Significant reductions were found for late stance peak ankle inversion moment of 3% during MT gait compared to unaltered walking (p < 0.05). Peaks did not increase significantly for plantar and dorsal ankle moments between the two different walking styles.
Erhart-Hledik et al. (2017)	• N-R	• N-R	• N-S changes in peak ankle eversion angle in stance between control (13.9° (5.4)) and active

			feedback (14.7° (5.3)), p = 0.193 for unaltered walking speed. • Average foot CoP in the first half of stance phase in the medial/lateral direction was significantly different between control (43.1 mm (5.6)) and active feedback (49.0 mm (7.6)), p = 0.011 for unaltered walking speed. Average foot CoP in the second half of stance phase was significantly different between control (28.3 mm (9.5)) and active feedback (31.8 mm (13.7)), p = 0.079; • Average foot CoP in the first half of stance phase was significantly different between control (43.9 mm (6.0)) and active feedback (47.5 mm (6.7)), p = 0.006, for fast walking speed. NS CoP findings in the second half of stance phase for fast walking speed.
Charlton et al. (2018)	• N-R	• N-R	• T-I 10° significantly increased rearfoot inversion angles by 68%, 139%, and 289% for ZR, T-O 10° and T-O 20°, respectively. T-O 20° resulted in significantly decreased rearfoot inversion angles by -57% compared to natural gait. • Significant peak frontal plane rearfoot angles during stance. T-I 10° significantly decreased rearfoot eversion by -48%, -57%, and -61% compared to all the other conditions. Significant differences in frontal plane ankle rearfoot excursion was observed. T-I 10° significantly increased frontal plane rearfoot excursion by 20%, 32%, and 50% compared to all the other conditions. Also, ZR resulted in significantly increased frontal plane rearfoot angle excursion

			by 25% compared to T-O 20°. • Significant differences for sagittal plane ankle angles at IC was observed. Angles at IC during T-I 10° were significantly more dorsiflexed by 129% compared to T-O 10°. Additionally, T-O 20° was significantly more dorsiflexed by 138% and 136% compared to ZR and T-O 10°. No main effects could be detected for peak sagittal plane ankle angles during stance or for sagittal plane ankle angle excursion. • The foot rotation conditions resulted in different EKAM magnitudes, evidenced by the significant main effect for early and late stance peak EKAM. • N-S findings for ankle eversion moment impulse after post-hoc correction. No main effect for ankle inversion moment impulse could be detected. • A main effect for step width was found across conditions (p = .001). Pairwise comparisons revealed that T-I 10° increased step width compared to all the other conditions.
Barrios et al. (2010)	• N-R	• Significant increase between baseline natural gait peak HIR: 5.3° (7.4); post-training modified peak HIR: 13.5° (8.5); 1-month post modified peak HIR: 12.8° (9.2); • N-S change in peak hip adduction angle (*p* = 0.073); baseline natural gait hip adduction angle: 9.2° (2.4).	• N-R
Hunt et al. (2011)	• Unaltered gait TL 2.61° (1.64); • Small TL 5° (0.87); • Medium TL 8.34° (1.61);	• Significant early stance peak HAM differences were observed between all TL conditions (5.22 (0.99), 4.61 (0.65), 4.09 (0.61) for small, medium and large TL	• N-R

	• Large TL 12.88° (1.91).	respectively) compared to unaltered walking (5.72 (0.90), with greater early stance peak HAM reductions associated with increasing amounts of TL, *p* < 0.001; • N-S differences in late stance peak HAM for any TL gait modification compared to unaltered gait (4.16 (1.13), *p* > 0.05; • N-S differences observed in peak hip abduction moment for any TL gait modifications compared to unaltered gait (1.38 (1.10)).	
Mundermaan et al. (2008)	• Increased medio-lateral trunk sway (10° (5)).	• N-S differences were observed for the maximum axial loading rates at the hip joint for unaltered gait (1286 (488) %Bw/s) and trunk sway (1250 (371) %Bw/s), *p* = 0.763; • Significant increase in maximum hip abduction moment of 55.3% between unaltered gait (2.0 (1.1)) and increased trunk sway (3.1 (1.3)), *p* < 0.001; • First peak HAM was significantly reduced by 57.1% for the increased medio-lateral trunk sway trial (1.8 (1.5)) compared to unaltered gait (4.2 (1.4)), *p* < 0.001.	• N-S differences we observed for the maximum axial loading rates at the ankle joint for unaltered gait (1280 (490) %Bw/s) and trunk sway (1214 (356) %Bw/s), *p* = 0.568.
van den Noort et al. (2014)	• Pelvis lift decreased by more than 5° in six participants (N-S at group level), pelvis protraction increased (4-6º, only significant for graph p = 0.03), and ipsilateral trunk sway decreased (2-3°, p < 0.01 except for colour); • With HIR feedback, maximal hip extension decreased (5-6°, p < 0.05 for bar and polar), and pelvis protraction increased by more than 5° in six participants (but N-S at group level).	• Hip angle feedback, HIR in the early stance phase increased significantly compared with baseline levels (bar 8°, *p* < 0.01; polar 10°, *p* < 0.01; colour 8°, *p* < 0.01, graph 7°, *p* < 0.01). The bar, polar and colour showed the largest change in late stance [9° (*p* = 0.01), 11° (*p* < 0.01) and 8° (*p* = 0.03), respectively]; • The kinematic changes that occurred while visual feedback on EKAM was provided included a decreased hip adduction (5°, polar p = 0.01, graph *p* = 0.02) and a maximal hip extension decrease (4-5°, *p* < 0.03 except for colour).	• Kinematic changes that occurred while visual feedback on EKAM was provided included an increased T-I angle of more than 5° in eight participants (on average: 2-7° at group level but N-S), an increased step width (6-7 cm, *p* < 0.03 for all feedback conditions); • While HIR feedback was provided, apart from significant changes in the HIR, participants also showed a significant increase in WS (7-10 cm). Furthermore, six participants showed an increased T-I angle of more than 5°, and five participants showed an increased T-O angle (on

			average 3-7° increase in T-I angle in group level, but N-S).
Dunphy et al. (2016)	• Significant differences were observed in maximum pelvic drop angle between unaltered gait (3° (1)) and contralateral pelvic gait (7° (1)), $p < 0.001$; • The correlation between change in pelvic drop and change in EKAM peak was $r = 0.88$ ($p < 0.001$).	• Significant differences were observed in maximum hip adduction angle between unaltered gait (0° (2)) and contralateral pelvic gait (4° (2)), $p < 0.001$; • The correlation between change in peak hip adduction angle and change in EKAM peak was $r = 0.83$ ($p < 0.001$); • N-S differences in hip flexion/extension between unaltered gait and contralateral pelvic drop gait trials.	• N-R
Khan et al. (2017)	• N-R	• Through the entire range from T-I to T-O, the hip joint's contribution to the total limb work decreased significantly at slow speed from 35.00% to 22.00%; • The hip joint increased its contribution at unaltered gait speed (26%–37%) through T-I to T-O. • At T-O, significant increase of hip joint's contribution from 22% to 37% in slow to unaltered walking speeds; • At T-I, the contribution of hip joint decreased from 35% to 26% in slow to unaltered walking speeds.	• The mean (SD) of self-selected FPAs for ST, TO and TI were 12.91 cm (4.78), 31.56 cm (7.51) and 13.43 cm (3.39) respectively; • N-S findings in ankle joint contribution by the speed transitions, except at T-I in slow to fast gait speeds. The ankle joint's contribution remained consistent except at slow speeds (decreased from 43.00% to 37.00%) from T-I to T-O gait.

KAM: knee adduction moment; T-I: toe-in gait; HIR: hip internal rotation; HAM: hip adduction moment; AAM: ankle adduction moment; AFM: ankle flexion moment; COP: centre of pressure; MT: medial thrust; T-O: toe-out gait; T-L: trunk lean; ZR: N-R: not reported; N-S: non-significant. Table 3-3 is extracted from (Bowd et al., 2019),

3.4.2.2.2 External knee adduction moment

The biggest EKAM reductions were reported in work that assessed trunk lean (~ 10°) as a gait retraining intervention (Mündermann, Asay and Mündermann, 2008) (SMD -1.99 CI -2.72, -1.18). This was also reported in other studies that assessed the effectiveness of trunk lean in reducing EKAM1 (Gerbrands *et al.*, 2017) (SMD -1.18 CI -2.24, -0.11), (Hunt *et al.*, 2011) (SMD -0.45 CI -1.12, 0.24). Interestingly, there was a dose response influence of adopting a trunk lean gait. The larger the trunk lean adopted, the larger the reduction in EKAM1.There was also a medium to large effect size on the reduction of EKAM1 for the following gait retraining styles: hip internal rotation (Barrios, Crossley and Davis, 2010) (SMD -1.24 CI -2.31, -0.17), medial thrust (Gerbrands *et al.*, 2017) (SMD -0.66 CI -1.17, -0.13), toe in gait (SMD -0.57 CI -1.29, 0.17) (Charlton *et al.*, 2018), and a self-selection of a combination of toe in, wider stance and medialisation of the knee position whilst receiving visual direct feedback on EKAM (SMD -0.54 CI -0.98, -0.09).

The finds for EKAM2 were not as effective and documented. Two studies reported a medium effect size reduction. This was shown in using polar visual feedback on hip internal rotation (SMD -0.60 CI -1.28, 0.09) and toe out gait (~ 20°) (Charlton *et al.*, 2018) (SMD -0.50 CI -1.23, 0.22). These studies assessing a gait style on the reduction of EKAM2 had a CI that crossed the line of null effect resulting in difficulty in their interpretation.

3.4.2.2.3 Ease of adapting gait style

It was decided after making the review protocol available online to extract additional information to establish the ease of adopting a given gait style. Out of the 11 studies included within this systematic review, 5 studies included subjective commentary on how easy the gait retraining was to implement (Barrios, Crossley and Davis, 2010; Hunt *et al.*, 2011; P Shull *et al.*, 2013; van den Noort *et al.*, 2014; Charlton *et al.*, 2018). Barrios et al. (2010), Charlton et al. (2018), van den Noort et al. (2014) asked the participants in their studies for their feedback/views on the ease of adopting the specific gait style. Barrios et al. (Barrios, Crossley and Davis, 2010) reported that improvement in how natural the gait style felt improved as the training sessions went on. In van den Noort et al. (2015) (van den Noort *et al.*, 2014), the intuitiveness of the type of feedback was verbally tested after each trial by a subjective score on the question: “how well were you able to modify your gait pattern?”. There were no significant differences between subjective scoring of the intuitiveness for all visual feedback trials. Therefore, the type of visual feedback is not of primary concern when aiming to modify gait (van den Noort *et al.*, 2014). Adopting a toe in gait was adopted with ease and without discomfort (Charlton *et al.*, 2018). Trunk lean was shown to introduce at least some degree of difficulty (Hunt *et al.*, 2011). The Shull et al. (2013) paper commented on the ease of learning toe in gait only within the paper's discussion section and indicated that participants in the study appeared to walk naturally with toe in gait (P Shull *et al.*, 2013).

3.4.2.2.4 Study quality assessment

The methodologic quality of the 11 studies were considered fair to moderate. Two studies were rated fair, whilst the remaining 9 studies were rated moderately (Table 3-4). It was shown that the studies lacked external validity and internal validity (confounding). All studies did not attempt to control for gait speed and step length, inadequate standardisation of gait retraining, and small sample sizes.

Table 3-4 Systematic Review: Risk of Bias Within Studies

Authors and year	Population	Reporting (n = 1-10)	External validity (n = 11-13)	Internal validity: bias (n = 14-20)	Internal validity: confounding (n = 21-26)	Power (n = 27)	Methodological score (/25 or /28)
Shull et al. (2013)	Symptomatic KOA	9	0	4	0	1	14/25
Richards et al. (2018)	Symptomatic KOA	8	0	4	2	1	15/25
Gerbrands et al. (2017)	Symptomatic KOA	9	0	4	1	1	15/25
Erhart-Hledik et al. (2017)	Symptomatic KOA	9	1	4	2	1	17/25
Charlton et al. (2018)	Radiographic KOA	9	0	4	1	1	15/25
Barrios et al. (2010)	Healthy	10	0	4	3	1	18/28
Hunt et al. (2011)	Healthy	9	0	4	2	0	15/25
Mundermann et al. (2008)	Healthy	8	0	4	2	1	15/25
Van den Noort et al. (2015)	Healthy	7	0	4	3	0	14/25
Dunphy et al. (2016)	Healthy	9	0	4	2	0	15/25
Khan et al. (2017)	Healthy	6	0	4	1	1	12/25

Table 3-5 Systematic Review: Percentage (%) Change in EKAM Parameter Measured Between Unaltered Gait and Gait Retraining Intervention

	1st peak EKAM values (presented as %BW*H unless otherwise stated)	**2nd peak EKAM values (%BW*H)**	**% Change in 1st peak KAM**	**% Change in 2st peak EKAM**
Shull et al. (2013)	Baseline: 3.28 (1.37); T-I: 2.90 (1.38) **	Baseline: 1.98 (1.14); T-I: 1.94 (1.09)	T-I: -13%	N-S
Richards et al. (2018)	Combination of WS, T-I and MT gait modifications with real-time feedback. Baseline:3.29 (1.00); visual feedback with self-selected combination of WS, T-I and MT gait: 2.82 (0.71) **; retention: 3.00 (0.77) **	N-R	Visual feedback: -14.28% Retention: -8.81%	N-R
Gerbrands et al. (2017)	Baseline: 0.24 (0.12); TL:0.15 (0.10) **; MT: 0.17 (0.09) **	Baseline: 0.19 (0.12); TL:0.15 (0.10) **; MT: 0.17 (0.10)	TL: -38% MT: -29%	TL: -21% MT: N-S
Erhart-Hledik et al. (2017)	Baseline: 2.41 (1.10); medial weight transfer at the foot: 2.26 (1.04) ** Baseline, fast walking: 2.90 (1.28); medial weight transfer at the foot, fast walking: 2.63 (1.35) **	Baseline: 1.71 (1.01); medial weight transfer at the foot, unaltered gait: 1.47 (0.96) ** Medial weight transfer at the foot, fast gait: 1.50 (1.13)	Medial weight transfer at the foot: -6.22% Medial weight transfer at the foot, fast gait: -9.31%	Medial weight transfer at the foot, unaltered gait: -14.04% Medial weight transfer at the foot, fast gait: N-S
Charlton et al. (2018)				
Barrios et al. (2010)	Baseline visit: 0.426 (0.07) (N m/kg); post-training: 0.34 (0.66) * (N m/kg); 1-month post: 0.34 (0.073) * (N m/kg)	N-R	Post-training: -20% 1-month post: -20%	N-R
Hunt et al. (2011)	Baseline: 4.07 (1.64); small lean: 3.82 (1.77); medium lean: 3.37 (1.72) *; large lean: 3.26 (1.64) *	Baseline: 1.89 (0.77); small lean: 1.64 (0.96); medium lean: 1.64 (1.02); large lean: 1.60 (0.90)	Small lean: N-S Medium lean: -21% Large lean: -25%	N-S
Mundermann et al. (2008)	Baseline: 2.00 (0.70); increased trunk sway: 0.70 (0.60) **	N-R	Increased trunk sway: -65%	N-R
van den Noort et al. (2015)	Baseline: 2.14 (0.20); HIR colour feedback: 1.92 (0.25); HIR polar feedback: 1.73 (0.24)	Baseline: 1.91 (0.29); HIR colour: 1.60 (0.34); HIR polar: 1.14 (0.32) **	HIR colour: N-S HIR polar: N-S	HIR colour: N-S HIR polar: -40.32 %

Dunphy et al. (2016)	Baseline: 0.41 (0.03); contralateral pelvic drop: 0.56 (0.04) *	N-R	Contralateral pelvic drop: +36.6%	N-R
Khan et al. (2017)	Slow, ST: 1.81 (N-R); slow, T-I: 1.82 (N-R); slow, T-O: 2.28 (N-R) *; Unaltered, ST: 1.96 (N-R); unaltered, T-I: 1.80 (N-R) ***;** unaltered, T-O: 2.81 (N-R) * fast, ST: 2.70 (N-R); fast, T-I: 2.23 (N-R) *****; fast, T-O: 3.08 (N-R) *	Slow, ST: 1.28 (N-R); slow, T-I: 1.64 (N-R) *; slow, T-O: 1.13 (N-R) *; Unaltered, ST: 1.42 (N-R); unaltered, T-I: 1.70 (N-R) *; unaltered, T-O: 1.06 (N-R) *; Fast, ST: 1.56 (N-R); fast, T-I: 1.60 (N-R); fast, T-O: 1.22 (N-R) *	Slow, T-I: N-S; Unaltered, T-I: -8.88%; Fast, T-I: -21% Slow, T-O: +25.97%; Unaltered, T-O: +43.37%; Fast, T-O: +14.07%	Slow, T-I: +21.90%; Unaltered, T-I: + 19.72%; Fast, T-I: N-S Slow, T-O: -11.72%; Unaltered, T-O: -25.35%; Fast, T-O: -21.79%

KAM: knee adduction moment; baseline: unaltered gait; Hunt et al. (2001): small lean (4 °), medium lean (8 °), large lean (12 °); S-T: straight-toe gait; T-I: toe-in gait; HIR: hip internal rotation; WS: wide stance gait; MT: medial thrust; T-O: toe-out gait; T-L: trunk lean; N-R: not reported; N-S: non-significant, $p > 0.05$; %BW*H: % body weight multiplied by height*: $p < 0.05$; ** $p < 0.01$. Table 3-5 is extracted from (Bowd et al., 2019).

3.4.3 Discussion

3.4.3.1 Summary of evidence

The purpose of this systematic review was to evaluate the consequences of different gait retraining techniques on hip and ankle joints. This systematic review has evidenced, for the first time, that there is a severe lack of reporting of hip and/or ankle joint biomechanics when adopting a gait retraining technique to reduce knee joint loading. With the current evidence available, it is not possible to report whether there is an adverse effect on adjacent joints to the knee when adopting a gait style. This is because there is a lack of research, as well as conflicting evidence presented. Therefore, there is a need to establish whether biomechanics of the hip and ankle joints are altered as a direct consequence of adopting a gait retraining intervention aimed at reducing medial compartment knee joint loading.

One key finding from this systematic review is that different gait retraining strategies may have different knee joint loading alterations. Trunk lean, hip internal rotation and medial thrust gait styles reduced first peak EKAM the most (Table 3-5). It is important to note that any conclusions made from this systematic review are based on a limited number of studies. This therefore emphasise that more exploratory studies need to be undertaken. As well as having a limited number of included papers, the quality of the trunk lean gait style and medial thrust gait style studies was only moderate. The present systematic review aligns with Simic et al. (2011) in terms of their medial thrust and trunk lean showing the highest reductions in early stance EKAM (Table 3-5). A profound finding is that all studies lacked external validity. Therefore, any conclusions of the included studies cannot be generalised to other populations. Further work is required to assess the effect of gait retraining styles on an mKOA population group; more importing those with mKOA and varus alignment.

The clinical translation of any gait retraining technique for this population will depend greatly on changes in the loading of joints, ligaments and muscles throughout the kinematic chain, a potential increase of energy expenditure and the aesthetics of the resulting gait (Gerbrands *et al.*, 2017). Caldwell et al. (2013) and Takacs et al. (2014) both indicate that there is potential limitation when adopting to a trunk lean gait. This is due to the increased energy expenditure associated with this gait adaptation, leading to fatigue and discomfort (Caldwell, Laubach and Barrios, 2013; Takacs *et al.*, 2014). Therefore, it is important to establish even where a gait style can reduce EKAM, there may be changes in terms of energy expenditure that may be counterproductive.

This systematic review has exemplified the severe lack of evidence in the biomechanical effect of gait retraining on the hip and/or ankle joints. Future research should therefore assess the adverse effects of the proposed gait retraining strategies. This is essential before gait retraining can be recommended as a clinical intervention especially in KOA individuals.

As a result of the lack of hip and ankle data and lack of external validity, any clinical recommendations made from this systematic review must be made with caution. The study by Hunt et al. (2011) emphasises the pathway towards clinical translation of their findings, is with examining the biomechanical effects at other joints and overcoming potential barriers to using this intervention in individuals with mKOA. Future research is recommended to focus on a gait retraining technique that aims to modify gait patterns to the extent that a clinically significant reduction in the EKAM, *and not necessarily a maximum reduction*, is achieved. In addition to this, the gait pattern should be sustainable and implemented by mKOA patients daily (van den Noort *et al.*, 2014). This is further exemplified in Erhart-Hledik et al. (2017) where it was stated that a gait retraining programme should be one that is sustainable and tolerable for long-term clinical implementation and requires future consideration.

3.4.4 Short term vs long term gait retraining

This systematic review included 11 studies. Out of these studies, only one assessed gait retraining over multiple visits and weeks (Shull et al., 2013). Shull and colleagues found that a 6-week gait retraining intervention reduced EKAM by 20% from baseline, and WOMAC pain and function scores improved 29% and 30% respectively. Barrios et al. (2010) showed that in healthy controls gait retraining reduced EKAM by 19%, and the pattern persisted up to 1 month post retraining. Shull and colleagues (2013) showed that pain is reduced, and function is increased in patients with KOA following a gait retraining study that targeted reduction in the EKAM. Another study that assessed gait retraining over multiple visits/times was Cheug et al. (2018). The purpose of this randomised controlled trial was to evaluate the effectiveness of an EKAM gait retraining programme in a group of patients with early KOA up to 6 months post-training. They hypothesised that patients receiving gait retraining would present lower EKAM during walking, which would in turn improve KOA related symptoms. At present, there is no optimum recommended timeframe to undergo a gait retraining programme. However, what is known is that the changes induced by gait retraining retained over a longer time of assessment (Shull et al., 2013). The purpose of this PhD is to determine the influence of altered gait styles on knee loading. Should this be adopted within a retraining programme then optimum training programme and follow up would need to be designed using this research as guidance. Gait retraining is a safe and effective intervention to reduce EKAM during walking and to improve the symptoms of patients with early KOA in the short term. A structured gait retraining programme is therefore recommended as an intervention for this patient cohort. Further study with a longer follow-up and evaluation of gait biomechanics outside the laboratory is warranted.

3.4.5 Limitations

A limited number of papers were included within this systematic review (n = 11). Each paper had varying biomechanics reported for the hip and ankle joints and so conclusive interpretation is limited. Future work should incorporate the consequences an altered gait has on the hip and ankle joints when considering a gait alteration for a clinical purpose.

The majority of the 11 included studies within this systematic review had a low number of participants and involved only one single visit. As well as this, papers did not always include a patient population and assessed a healthy cohort instead. Therefore, the translation of the findings to mKOA patients is limited. Future studies must evaluate gait retraining interventions on individuals with mKOA and establish the longer-term effects of the adopted gait style. It would also be of great use to better understand the participant’s perspective on how difficult the gait retraining style is to perform.

3.4.6 Conclusion

For the first time, this systematic review has focused on assessing gait retraining and the effects of such on the hip and/or ankle biomechanics. Several key findings arise from this work. There is a lack of studies that have included hip and/or ankle biomechanical consequences when altering an individual’s gait with the objective of lowering knee joint loading. Studies have also lacked external validity and were scored at best moderate in their study quality.

There were 3 objectives to this systematic review. First, the consequences of gait retraining on the biomechanics of the ankle and hip as well as trunk and pelvis biomechanics. Second, establish whether gait styles and gait retraining can reduce medial knee loading as assessed by first and second peak EKAM. The third objective was to outline patient/participant reported outcomes on how easy the gait retraining style was to implement. This would aid the clinical translation of aforenamed gait retraining techniques.

In terms of objective one, it was clear that only 11 studies included some variable that was related to hip, ankle, or pelvis biomechanics. There was no real pattern of what variables are considered important when monitoring the effect of gait retraining to reduce medial knee joint loading. Objective 2 resulted in showing that different gait retraining techniques resulted in different EKAM alterations. Due to a lack of studies included in the work, it was not possible to have a collective and sound summary of findings. The findings that were present from objective 3 shows that very limited research is available which assesses how easy gait retraining styles are to implement. It is clear from this systematic review that caution should be had at present on the recommendations on gait retraining strategies until more is known about what alterations are occurring at the hip and ankle joints when adopting these strategies.

Future research should consider this work when designing a gait retraining intervention to reduce medial tibiofemoral joint contact forces in mKOA patients, over multiple visits as well as analysing the potential changes of the gait retraining strategy to the other lower limb joints. This will aid the clinical relevance of the findings that arise from the research and develop the clinical translation of the interventions.

CHAPTER 4: METHODS

4.1 Background

This chapter aims to outline the experimental design and methodology for the objective assessment of lower limb function during gait using 3D motion analysis for a control cohort during unaltered gait and for one time point for patients at pre-surgery and 12 months post-HTO who walk with four gait styles (1) unaltered gait (2) toe out gait (3) wide stance gait (4) medial thrust gait.

For each of these aspects, methods for data collection and generation of variables for analysis are outlined. Study design and methodology described in this chapter are common to multiple chapters within this thesis, however specific chapter data analysis methods will be outlined within the chapter methods sections. Specifically, this applies to Chapter 5 when PCA and the Cardiff Classifier are introduced and used on a sub-cohort to better understand what biomechanical parameters recover post-HTO.

All motion capture data was collected at the MSKBRF motion analysis laboratory at Cardiff University as part of ongoing research between the MSKBRF, School of Engineering, Cardiff University, and the Biomechanics and Bioengineering Research Centre Versus Arthritis (BBRC VA). The recruitment of patient participants and healthy control volunteers was approved by the Wales Research Ethics Committee 3 (10/MRE09/28) and Cardiff and Vale University Health Board. This permitted the collection and assessment of clinical data and access to historical patient records regarding the involved condition and assessment of patient gait using approved 3D motion analysis protocols. The following courses were attended/documents were obtained for the approval of all methodology carried out with NHS patients during this study:

- Good clinical practice training and certificate.
- Informed consent training and certificate.
- Disclosure and Barring Service (DBS) check.
- Honorary research contract issued by Cardiff and Vale University Health Board.
- Phlebotomy Training Level 2 certificate.

Written informed consent was obtained from each participant prior to data collection. At the time of writing this thesis, the collection of gait biomechanics pre- and at 3 months, 6 months, and 12 months post-HTO surgery had been ongoing for 12 years. Throughout this period, several different trained operators have carried out the data collection following the MSKBRF Standard Operating Procedure (SOP) of which is outlined below.

Due to the natural progression of research methods and technology, the SOPs in terms of tracking markers have evolved over time, however the instructions given to the participants have stayed consistent. There have been hardware changes over this period. The most notable being the upgrades of the motion capture cameras from eight ProReflex cameras (Qualisys, Sweden), to nine Oqus 3 cameras (Qualisys, Sweden), to 14 Oqus 700+ cameras, and the change in the number of force plates used to collect force data changing from 2 to 4, then to 6 (Bertec Corporation, Ohio, USA). As a result of moving to a new motion analysis laboratory, any data collected before June 2017 was collected at 120 Hz camera frequency and 1080 Hz force plate frequency. From June 2017 onwards, data was collected at 200 Hz camera frequency and 2000 Hz force plate frequency. The resultant effect of these hardware changes on the accuracy and precision of the motion capture data is not quantified within this thesis.

The data collections also incorporate several different elements, besides level gait, e.g., such as additional ADLs and electromyography (EMG). The latter two are not used further in this thesis. Therefore, the outline of data collection methods will only be mentioned briefly. The methodologies and validity of data processing for the ADL and EMG shall not be further described.

4.2 Data collection

4.2.1 Recruitment: Non-pathological controls

Volunteers were recruited via email and poster advertisements throughout Cardiff University and the wider South Wales community. The criteria for inclusion in the study as a control cohort volunteer was as follows:

- No self-reported OA, or pain in the foot, ankle, knee, hip or back.
- No known difficulty performing ADLs.

- No history of musculoskeletal conditions which required medical treatment e.g., ligament or meniscal tear.
- No other musculoskeletal, neurological, or visual condition which might affect the way they move.
- An ability to give informed consent.
- Within the age range of 18-80.

Suitable volunteers were given a healthy volunteer information sheet for motion capture analysis to read, outlining the study protocol and additional ethical considerations, before agreeing to take part in the study. Willing candidates were then consented using the healthy volunteer informed consent form.

4.2.2 Recruitment: High Tibial Osteotomy patients

The recruitment of NHS patients with mKOA was approved by the Research Ethics Committee for Wales and Cardiff and Vale University Health Board. These patients all had mKOA with correctable varus deformity and had been listed for HTO surgery. The mKOA cohort presented KL grade II – IV KOA in the medial knee compartment and were diagnosed with radiographic static knee malalignment. KL grading is defined as (Brandt et al., 1991):

- KL grade II – 'definite joint space narrowing (JSN) and possible osteophytic lipping'.
- KL grade III – 'multiple osteophytes, definite JSN, sclerosis, possible bony deformity'.
- KL grade IV – 'large osteophytes, marked JSN, severe sclerosis and definite bone deformity.

Patients were invited to attend the motion analysis laboratory just prior to surgery, at 3 months, 6 months, and 12 months post-surgery. This thesis focused on analysing data pre-HTO and at 12 months post-surgery. The reason for these two time points being of particular interest to this thesis is explained in the Literature Review. The criteria for inclusion in the study as a patient volunteer was as follows:

- An ability to walk 10m without a walking aid.
- An ability to give informed consent.

- No unrelated musculoskeletal, neurological, or visual condition which might severely affect the way they move.
- Within the age range of 18-75.

Before taking part in any aspect of the study, a member of NHS staff would explain the study and fill out a ‘Permission to Contact’ form and pass this information onto the research team for recruitment. After which an ‘Information Sheet’ was sent to the volunteer. If the patient volunteer was still interested in taking part, they were asked to sign a patient consent form at the time of their assessment session.

4.3 Gait assessment

The infrared cameras, force-plates and EMG electrodes were all connected to a 56-channel analogue board which allowed simultaneous recording of kinematic, kinetic and EMG data with synchronisation using a Delsys trigger module (Delsys Inc, USA) that sent out a TTL pulse sent via a single trigger, ultimately used to start and to stop recording.

4.3.1 Calibration

Before the volunteer arrived, the IR cameras were calibrated such that the QTM software could calculate their position and orientation relative to the origin and orientation of an orthogonal global coordinate system (GCS) defined in the laboratory. The laboratory (Figure 13) was calibrated to define a GCS and the force platforms were calibrated and defined with respect to the GCS.

The GCS was defined by placing an L-Frame (as shown in Figure 9), where the long hand of the frame represents the x-axis (direction of gait travel), the shorthand the y-axis (left to the direction of travel), and the vertical and mutually orthogonal axis is called the z-axis. The definition of the GCS defined the coordinate system within which all the data was then described by the camera system. Figure 13 is a photograph of the MSKBRF Clinical Motion Capture Laboratory.

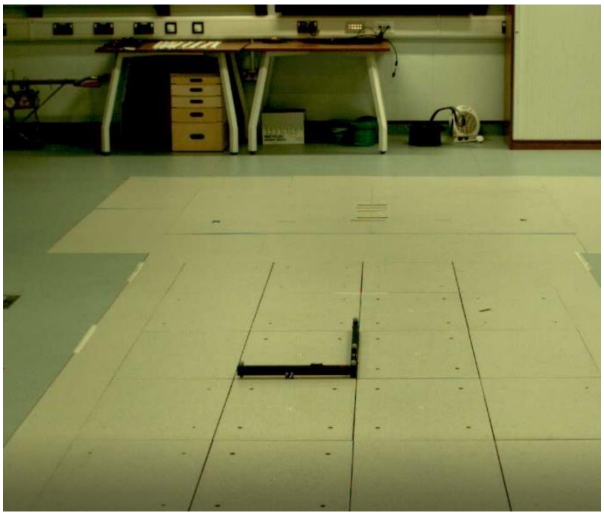


Figure 9 L-frame positioning

Each of the four markers are located directly over the corners of the force plate (Figure 10). The X and Y coordinates of these markers are used to define the location and orientation of the force plates within the GCS and the Z-offset is defined in Qualisys Track Manager. This step is crucial in identifying accurate COP coordinates, and hence calculating knee kinetics.

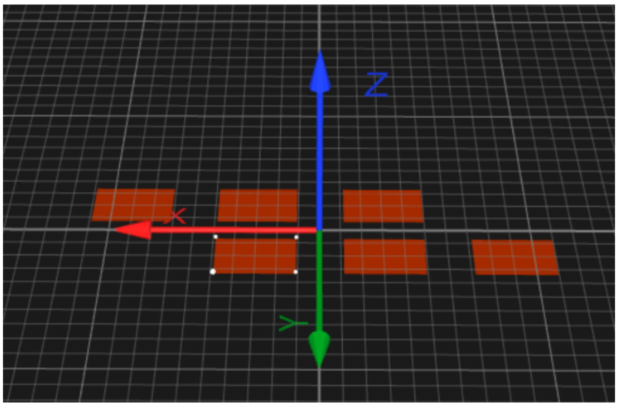


Figure 10 Locating force plates corners

With the L-Frame in place, the camera system was calibrated by waving a calibration wand (Figure 11), which contained two markers (circled red in Figure 11) that are a known distance apart, through the intended volume of capture for 60 seconds. During this process, each camera recorded the position of the two markers relative to the L-Frame. Once the calibration recording was completed, 3D positions of the markers were reconstructed in Qualisys, and errors were returned for every camera representative of the residuals between them for estimated dynamic wand marker location which is typically between 0.4-0.8mm. If residual errors exceed 1.0mm, the calibration procedure was repeated to avoid poor marker tracking during the subject recordings. Finally, the calibrated volume is visually assessed to determine any major gaps in which the calibration wand was not moved through that could result in poor marker tracking for that given area in the field of analysis.

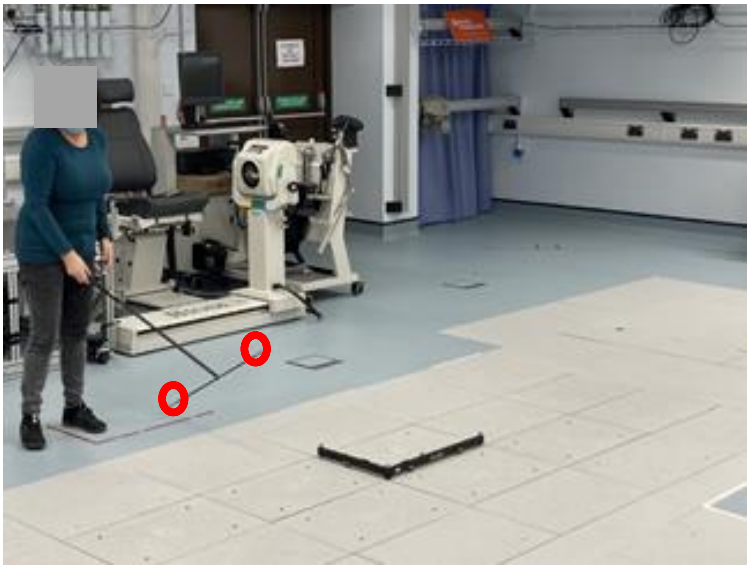


Figure 11 Calibration wand with markers at known distance apart (red circles), and frame located at the origin of the GCS

The 3D coordinate positions of the force platforms corners were defined relative to the GCS using markers attached to brackets positioned on each force platform. After June 2017, the same approach was undertaken but with markers placed in the corners of the force plates in a known position without the use of a bracket (Figure 12). This step is crucial in identifying accurate COP coordinates, and hence calculating knee kinetics.

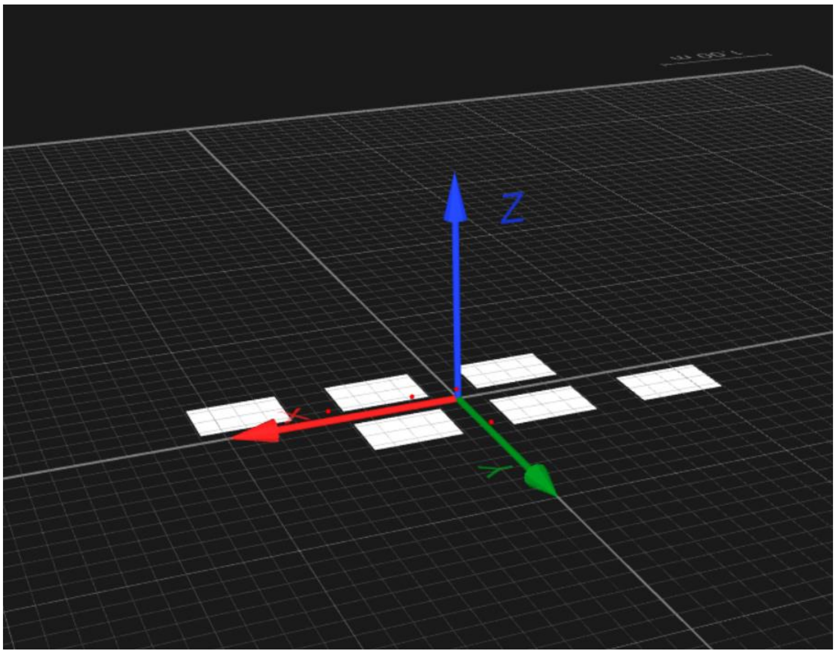


Figure 12 Global coordinate system and force plate locations

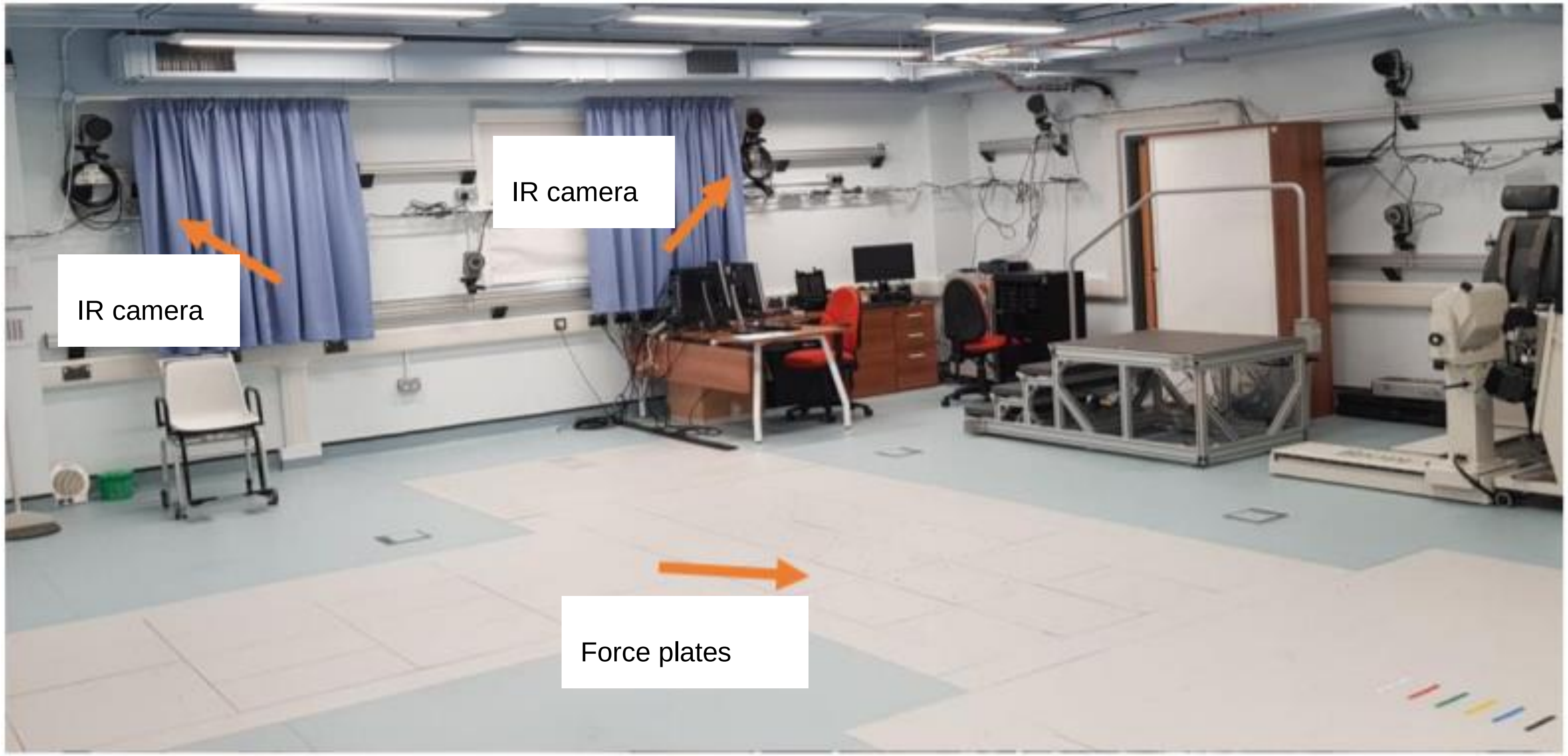


Figure 13 Musculoskeletal Biomechanics Research Facility Motion Analysis Laboratory

4.3.2 Informed consent

The participants were given an information sheet at least 48 hours before their first assessment, which explained the purpose of the study, what would be expected from them, and how the data would be anonymised and stored. During the assessment, the researcher explained the study and once it was established that the participant had read and understood the information sheet, they were asked to sign an informed consent form, which was also signed by one of the lead researchers [patient information sheet and inform consent example in Appendix B].

4.3.3 Questionnaires

The participant was asked to fill in relevant questionnaires regarding their knee pain and function, as detailed in the below list of 5 questionnaires. These 5 questionnaires were collectively chosen at the onset of the longitudinal study by the research team including input by clinicians. These questionnaires are briefly outlined below but are not included in the analysis with this thesis, apart from the Oxford Knee Score being used in the Cardiff Classifier analysis in Chapter 5:

1. **Oxford Knee Score** (OKS) (Dawson *et al.*, 1998): Designed and developed to assess function and pain. It is short, reproducible, valid, and sensitive to clinically important changes.
2. **Knee Outcome Survey** (KOS) (Irrgang *et al.*, 1998): Provides a percentage of disability during everyday activities (activities of daily living subscale) or sports (sports activity subscale).
3. **Knee Injury and Osteoarthritis Outcome Score** (KOOS) (Roos *et al.*, 1998): Knee-specific instrument, developed to assess the patients' opinion about their knee and associated problems.
4. **Pain audit collection system** (PACS)
5. **Western Ontario and McMaster Universities Osteoarthritis Index** (WOMAC) (Bellamy *et al.*, 1988): Self-administered health status measure used in assessing pain, stiffness, and function in patients with OA of the hip or knee.

4.3.4 Clothing

All patients and control subjects were treated the same in relation to motion capture analysis, and thus the following protocol encompasses all data collection methods. Subjects were asked to wear loose-fitting, non-reflecting clothing to allow full mobility and reduce false marker detection artefacts induced by undesired reflection of IR-light, respectively. Retro-reflective markers must be visible to cameras and were attached to skin, therefore clothing typically consisted of shorts for males and females, as well as a loose-fitting top or sports bra for females. The assessment was carried out without footwear to standardise the effect of different footwear.

4.3.5 Anthropometrical measures

Initially, the participants stature and mass were recorded. The participant was then sat with their knees at 90° to identify the medial and lateral knee joint space and the femoral epicondyles (both of which were marked with a pen). The participants thigh girth, medial-lateral and anterior-posterior knee widths measurements were taken and recorded.

4.3.6 Assessment preparation

55 retro-reflective markers in total were used for motion capture data collection. The markers were specifically designed for motion analysis and consisted of a plastic hollow spherical base with a reflective coating. No preparation was required other than applying double-sided tape to the markers prior to attaching them to the skin. Placement of markers was consistent with the Helen Hayes marker-set protocol (Kadaba, Ramakrishnan and Wootten, 1990; Collins *et al.*, 2009) which allows accurate tracking of each segment (i.e., hip, femur, tibia, and foot) in six DOF, but has been modified to include extra markers. These extra markers include medial knee, medial malleolus and iliac crest which permitted improved calculation of joint centre of rotations using regression methods for more accurate estimation of joint centres (Cereatti, Croce and Cappozzo, 2006).

4.3.7 Electromyography

Seven lower limb muscle targets were selected for analysis from previous recommendations of muscles with minimal crosstalk for the accurate detection of lower limb surface signals during gait (Hermens *et al.*, 2000). This thesis does not use this data for any analysis. Although the aims of the work in this thesis do not include assessment of muscle activation and EMG, future work may adopt the outputs from the assessment, so the method of EMG data collection is outlined in this Chapter.

Placement of EMG electrodes was based on a modified version of the Surface Electromyography for the Non-invasive Assessment of Muscles (SENIAM) guidelines (Hermens, Hägg and Freriks, 1997) and shown in Figure 14. This firstly involved locating target muscle bellies, achieved by a combination of asking the patients to tense individual sets of muscles whilst visually inspecting, as well as palpation of the muscle region. Muscle bellies that were difficult to palpate or not visible were placed using a repeatable method using limb measurement estimations based on SENIAM guidelines (Hermens *et al.*, 2000). Skin preparation consisted of first dry shaving skin around placement area to remove hair, skin exfoliation to remove dead skin and application of electro-gel to improve the conductivity of muscle signals. Alignment of individual muscles was pre-determined based on anatomic diagrams and EMG electrodes were placed aligned with the muscle belly. To reduce inter-operator reproducibility error, all muscle belly locating, skin preparation and placement of electrodes was performed by a single researcher. Finally, electrodes were secured using elastin tubing (Tubigrip).

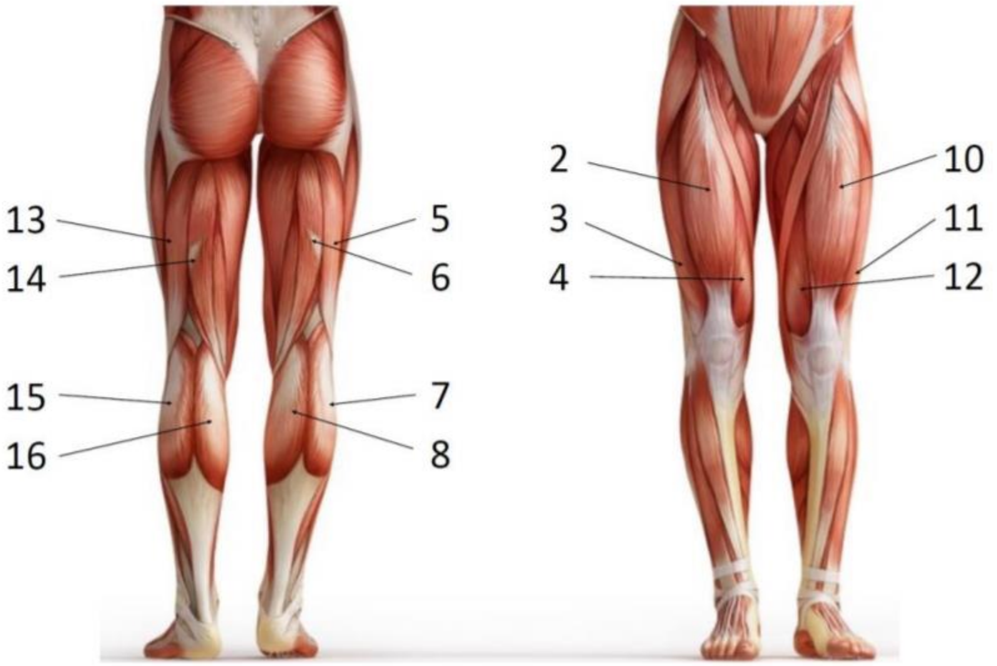


Figure 14 Electromyography electrode placement protocol

Numbers represent electrode labels. Rectus femoris (2, 10); Vastus lateralis (3, 11); Vastus medialis (4, 12); Biceps femoris (5, 13); Semitendinosus (6, 14); Gastrocnemius lateral (7, 15); Gastrocnemius medial (8, 16). Imagine extracted from (Kathib, 2018)

4.3.7.1 Acknowledging the exclusion of EMG

This PhD does not include EMG in any analysis. The reader is directed to the Literature Review of this thesis for an explanation. In short, the author acknowledges the usefulness of EMG in better understanding MSK movement and future work should aim to incorporate EMG analysis.

4.3.8 Retro-reflective marker placement and marker-set

Marker placement using a modified Cleveland Clinic Marker set as shown in Figure 15. Anatomical markers (blue in Figure 15) defined in the protocol were used to define segment dimensions, anatomical axes and centre of rotations using the static trial, whereas tracking markers (red in Figure 15) were used for calculations of how segments translate and rotate in relation to each other. Four tracking markers were used per segment, since at least 3 are required for calculation of joint rotations and translations in the three planes (i.e., sagittal, frontal and transverse) which permits for single marker drop-outs due to obstructions from clothing or equipment.

For anatomical markers, all specified bony landmarks corresponding to segment definitions were palpated to find body protrusions which were defined as the optimal location for placement for repeatability. Tracking markers were placed around the central region of tracked segments for the thigh and shank in a consistent manner using visual cues, and foot tracking markers were located using a combination of palpation of bony landmarks, as well as visual cues.

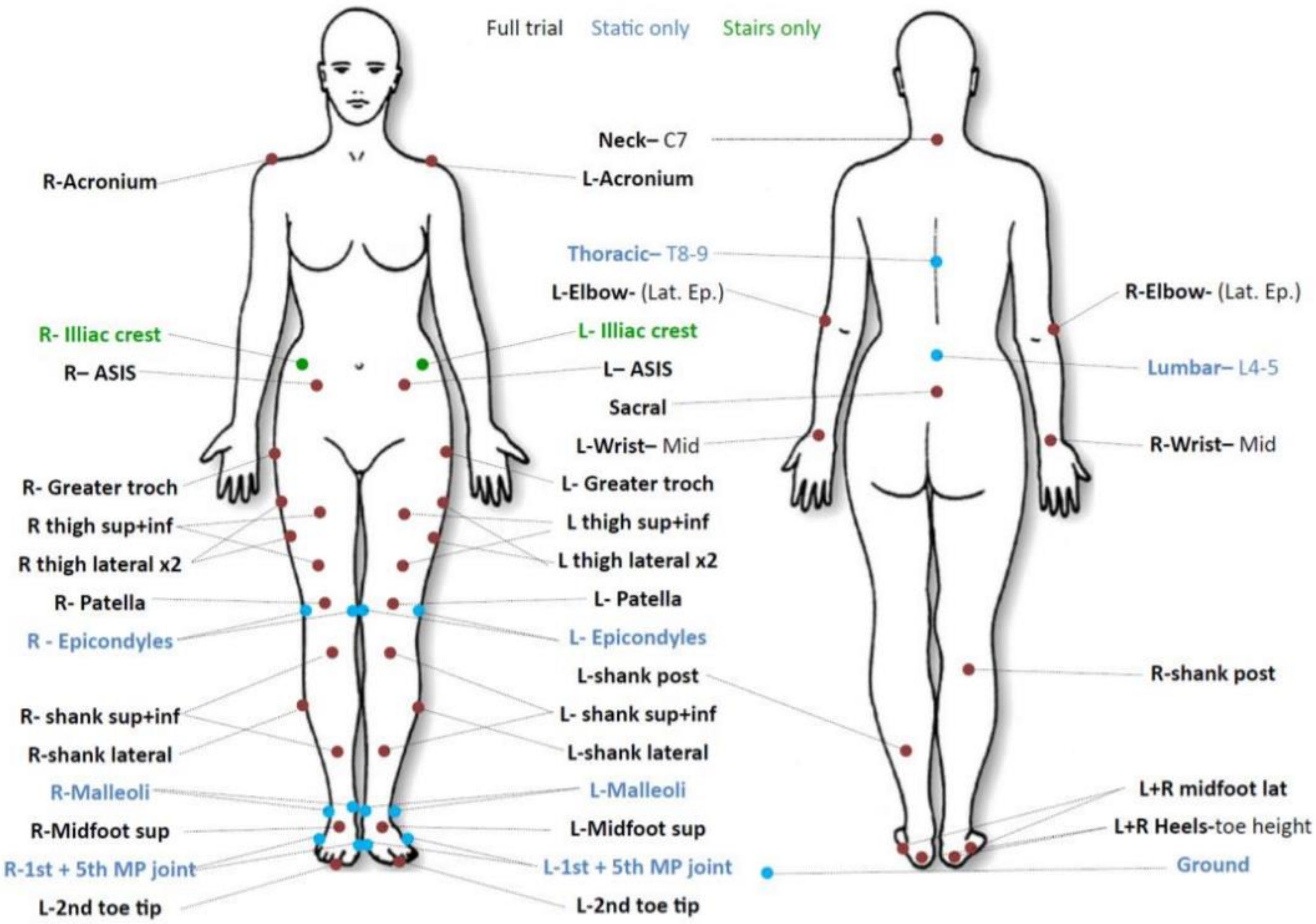


Figure 15 Motion capture marker protocol

4.3.9 Assessments

Participants were asked to walk barefoot at a self-selected pace over a 10m walkway. The force platforms were located within the middle third of the walkway and patients were unaware of their location. If clean force platform readings were not recorded, the participants' starting point was adjusted accordingly to ensure heel-force plate contact and to avoid force plate targeting. This whole process was repeated until there were at least six clean force platform readings for each leg for four different gait styles. The gait cycle was defined between two consecutive heel strikes (HS) of the same leg. Initial heel strike was defined from the force platform when initial contact force exceeded 20N. Visual inspection was undertaken for each gait cycle to ensure an individual made full contact within a force plate during the whole of the stance phase. If for example, the participant made contact partially with a force plate, this trial was classed as an unsuccessful trial and the process was repeated until at least 6 clean gait cycles were saved. The participant was not made aware of the existence or location of the force platforms; however, they may have been conscious that something was different about this area of flooring.

4.3.10 Calibrations

4.3.10.1 Static calibration

The volunteer was asked to remain still with arms by their sides whilst recording for at least 3 seconds. The static calibration is required for data processing in both Visual 3D (C-Motion, Maryland) and with using the COMAK MSK pipeline (Lenhart *et al.*, 2015; Smith *et al.*, 2018). Anatomical landmarks were defined using markers placed on anatomical landmarks and captured during the static calibration trials.

4.3.10.2 Dynamic calibration

Hip, knee, and ankle dynamic calibrations were recorded. However, these recordings were not utilised for this thesis and the definitions and explanations of such techniques will not be expanded up further.

4.3.11 Walking trials

Four gait styles were recorded in a session and the reasons for each are outlined in Chapter 2. The four gait styles were (1) unaltered level gait, (2) toe out gait, (3) wide stance gait, and (4) medial thrust gait. Whilst recording data, two researchers were present to observe successful force-plate hits which involved the placement of all contact areas of the foot within the centre, at least ~2cm away from the edge of the plate. One researcher observed alignment of the foot along the length of the walkway, whilst the other observed foot placement perpendicular to the walkway.

4.3.11.1 Determining gait style was achieved

For each gait style, the participant was given visual and verbal instructions on how to successfully complete the specific gait style. Visual and verbal instructions given are outlined below. The participant was then given as much time as they needed to practice the specific gait style before the operator then recorded the trials. During the practice trials, participants were observed by the lead researcher of the session to ascertain whether the desired gait style was undertaken. At this point, no metric determination was undertaken to determine whether the gait style was reached. This was done determined during the post-processing of the data collected.

4.3.11.2 Unaltered level gait

After a period of practice to allow familiarity, the participants performed 6 walking trials with satisfactory force plate strikes with their natural gait and self-selected speed.

4.3.11.3 Wide stance gait

Increasing the step width lateralises the COP, allowing the resultant GRF to pass closer to the knee joint centre. A verbal and visual demonstration of wide stance gait was provided, and participants were then asked to walk with their feet wider apart to a comfortable degree. The width achieved was therefore self-selected but within the boundaries of what would be achievable and tolerated by patients in clinical practice. Figure 16 below is a visual example of a wide stance gait being performed.

Figure 16 Wide stance gait

4.3.11.4 Toe out gait

A verbal and visual demonstration of toe out gait was provided, and participants were then asked to walk with their feet turned outwards, to a comfortable degree. The angle achieved was therefore self-selected but within the boundaries of what would be achievable and tolerated by patients in clinical practice. Figure 17 below outlines a toe out gait style being performed.

Figure 17 Toe out gait

4.3.11.5 Medial thrust gait

Increasing medial knee thrust reduces the knee varus angle during stance and hence decreases the frontal plane moment arm. Patients were instructed to bring their knees closer together (medialising) during the stance phase, while trying to avoid an excessive increase in knee flexion. Patients were encouraged to medially rotate their hip to medialise their knee at point of heel strike. Figure 18 below shows a medial thrust gait style being performed.

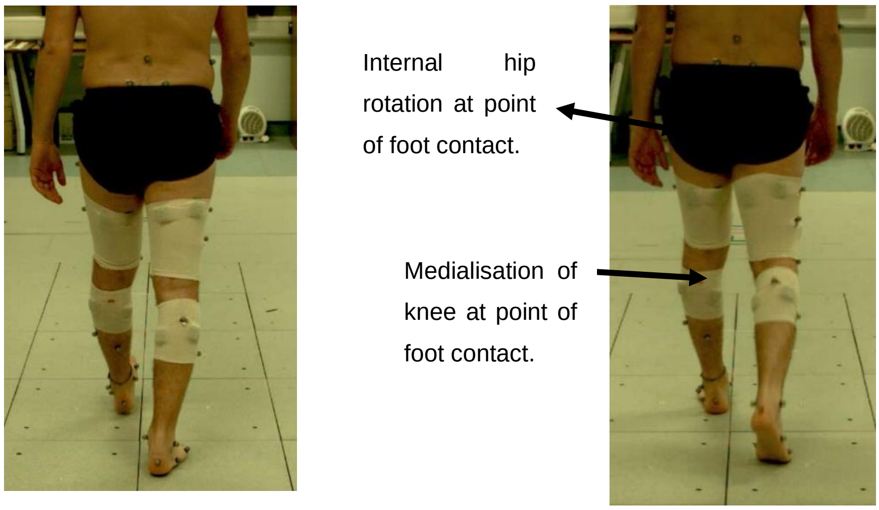


Figure 18 Medial thrust gait

4.3.12 Activities of daily living

Following the assessment of gait, the participants were also asked to perform other ADLs. The following two activities were undertaken: (1) stair ascent and descent, and (2) sit-to-stand. This data is not presented in this thesis and the detail of methodology is not expanded upon further.

4.3.13 Marker tracking

Marker coordinate data was captured during motion trials in QTM (Qualisys, Sweden). The markers were initially manually tracked and labelled on the first dynamic trial and then used to define an Automatic Identification of Markers (AIM) model which can be applied to other walking trials and static calibrations Once the QTM files were labelled they were exported as .c3d for analysis in Visual3D (C-motion, USA).

4.3.14 Sources of error in data collection

Errors in defining the location of the force plates relative to the GCS due to incorrect placement force plate calibration frame (pre-2017) or markers put in specified know positions on top of the force plate (2017 onwards) can result in errors in reconstructions of the force vectors in relation to the applied COP. If the generated force vector is inconsistent with the relative marker location data, inaccuracies will be present in the calculations of kinetic parameters around the examined joint. Accurate and reliable kinematics and kinetics data are essential to the appropriate application of movement analysis data for clinical and research purposes. Proper laboratory calibration includes the accurate determination of the positions of the force platform(s) and cameras in the laboratory coordinate system, as well as correct setting of force platform parameters. Any errors in the parameter settings or calibration measurements will lead to incorrect values of kinetic calculations that rely on the force data. Routine checks were undertaken to estimate the force platform locations by implementing a Cal Tester and following company guidelines on standards to follow.

Misalignment of EMG electrodes with muscle bellies can result in poor signal from the muscle causing low signal-to-noise ratios which may increase errors in final calculation of normalised muscle activation waveforms, as well as increase crosstalk from surrounding muscles, thus reducing specificity of the signal. Whereas misplacement of retro-reflective markers on skin, particularly for anatomical landmarks, leads to inaccurate calculation of joint centres. This can lead to large errors in the calculation of joint rotations and moments from musculoskeletal models.

The goal of motion capture is to accurately record the movement of bone segments relative to each other, to allow estimations of segment rotations and joint moments. However, as a subject moves, soft tissues including muscle, fat and skin move independently of the bone due to muscle contractions and inertial effects on tissues. As markers are attached to the skin, this can lead to inaccuracies in the calculation of segment kinematics. This is a well-recognised limitation of motion analysis which has been validated by the disparity of kinematic calculations using skin markers compared to bone pins, however currently there is no solution to eliminate this in skin-marker based motion capture (Benoit *et al.*, 2006). However, it is possible to reduce the overall influence on calculation by placement of tracking markers on differing anatomic locations of the segments including the lateral, anterior and posterior shank and thigh as demonstrated in the marker protocol.

4.4 Calculation pipelines for kinematics and inverse dynamics in Visual 3D

4.4.1 Background

Visual 3D (C-motion, USA) is an academic and commercial biomechanics research software that provides in-software tools used for the generation of biomechanical models, calculation of biomechanics data (including spatial-temporal parameters or kinematic and kinetic waveform data), processing of waveform data (such as normalisation, filtering, and rectification) and parameterisation of waveform data (such as calculation of maximum or minimum values). In-software pipeline tools also allows for the generation of scripts to run a sequence of functions to eliminate the requirement of manual processing. Marker data and force data are recorded and saved within QTM as a '.qtm' file and is exported as '.c3d' for compatibility with Visual 3D for processing.

Joint kinematics, kinetics and temporal parameters were calculated within Visual 3D (C-Motion, USA) using a custom model of the lower limbs and thorax. A Butterworth fourth order Filter was used on raw marker coordinate data with a cut-of frequency of 7 Hz.

4.4.2 Biomechanical model generation

An in-house Visual 3D pipeline that has been implemented in previous research (Whelton *et al.*, 2017; Whatling *et al.*, 2019) was used in the present thesis. Alterations that were made are documented within this thesis.

A six DOF model (three rotations, three translations for every joint) were generated for each participant, which included generation of eight segments from anatomical markers. Anatomical joint axes and their positional relationship to tracking markers used to define segment motion, were calculated from a measurement of static standing. Knee and ankle centres were defined as the midpoint of the epicondyles and malleoli, respectively. Hip joint centres were defined relative to the markers on the pelvis using the Bell regression model (Bell, Brand and Pedersen, 1989). The thorax axis origin was defined between virtual iliac crest markers created from the position of the anterior superior iliac spine (ASIS) and trochanter markers. Local coordinate systems were defined to coincide with anatomic axes and segments defined as rigid bodies with inertial properties estimated. As a result of additional markers being placed on segments, additional tracking markers were added to the tracking marker-set accordingly.

4.4.3 Calculating gait events

The gait cycle was defined between two consecutive heel strikes (HS) of the same leg. Initial heel strike was defined from the force platform when initial contact force exceeded 20N. Automatic gait events were defined in Visual3D (C-Motion, USA) using a proprietary function, a gait recognition algorithm that calculates the second heel strike based on the axial and anterior posterior position of the proximal end of the foot. This method is called Target Pattern Recognition and requires a clean force strike with the foot (Stanhope *et al.*, 1990).

4.4.4 Kinematics

Joint angles were calculated using the Cardan/Euler x, y, z sequence, equivalent to the Grood and Suntay definitions (Grood and Suntay, 1983). Kinematics of the hip and ankle are calculated using the Cardan-Euler sequence (X-Y-Z) based on ISB recommendations (Wu *et al.*, 2002) as well as axis definitions. Angles are defined as the orientation of the distal segment with respect to the reference proximal segment, except for the thorax, pelvis, and feet where they are calculated with respect to a GCS aligned with the direction of gait.

The rotations were described using the default Visual 3D Cardan sequence system, which uses the ordered sequence of rotation (X, Y and Z) that assumes the X axis is the medial/lateral direction (flexion-extension), Y axis is in the anterior/posterior direction (abduction/adduction) and Z is the axial direction (longitudinal rotation), which is based on ISB recommendations (Kadaba, Ramakrishnan and Wootten, 1990). The output of these rotational calculations for each joint in the motion capture recordings is in the form of x, y and z rotations in relation to the reference segment for each frame of the captured recording, which presented as a time-series makes up kinematic biomechanical waveforms.

The direction (i.e., positive, negative) of the calculated angles were described such that flexion, adduction and internal rotation are positive:

- Flexion (+) / Extension (-) angles
- Adduction (+) / Abduction (-) angles
- Internal rotation (+) / external rotation (-) angles

The number of captured data points in the time-series was normalised to % gait cycle so that one point on the waveform was equal to a single percent. For the Cardiff Classifier work, joint angle data used for waveform analysis were broken down into individual angle waveforms representative of rotations for a single gait cycle.

4.4.5 Kinetics

Ground reaction force (GRF) vectors in the x, y and z plane are measured within calibrated strain-gauge components of the force-plates which allowed the calculation of vertical, mediolateral, and anteroposterior force vectors. All GRF were normalised to bodyweight to make meaningful comparisons within inter-subject and inter-group comparisons.

The external joint moment is the rotational force acting at the joint created by the ground reaction force in each plane produced during locomotion, which is counteracted by the internal moments that are produced by muscles and ligaments to keep the joint stable. External moments are often used to describe knee function, since they can act as surrogate measures to the forces acting locally at each joint and can be used to understand the function of active and passive stabilisers. Joint moments in each plane are calculated as a product of the effect of inertial forces, the planar GRF and the shortest distance (moment- or lever-arm) between the centre of joint rotation and the GRF vector, which depends on the COP, centre of mass and mechanical axis alignment of the joint.

Joint kinetics were calculated using the inverse dynamics approach using Visual3D to calculate moments around the joints. The moments were resolved in the LCS of the distal joint. GRF are normalised to weight and shown as percentage of body weight (%BW). Moments were normalised for weight and height and expressed as %BW.Height.

There is no standard reference frame for expression of joint moments. Moments can be expressed using the distal segment coordinate system (Kaufman *et al.*, 2001; Gök, Ergin and Yavuzer, 2002), the proximal segment coordinate system (Schache and Baker, 2007), or the joint coordinate system, which is a combination of both systems.

Like angles, joint moments were calculated at each frame and converted into time-series data for waveform analysis. Moments were also normalised to %BW*Height since calculations consider GRFs and the moment lever arm which is affected by limb length. This is to allow meaningful relevant inter-subject comparisons. The direction of the moments acting at the joint were described similarly to angles, such that:

- Flexion (+) / Extension (-) moments.
- Adduction (+) / Abduction (-) moments.

- Internal rotation (+) / external rotation (-) moments.

4.4.6 Calculation of spatial-temporal parameters

Visual 3D pipeline tools were also used to calculate several spatial-temporal parameters:

- **Gait speed**: Distance of heel-strike to heel-strike divided by time of a single gait cycle.
- **Stance time**: Time of heel-strike to toe-off.
- **Swing time**: Time of toe-off to heel strike.
- **Double limb support time**: Time whereby both feet at in contact with the ground.
- **Cycle time**: Computed speed uses the average of all the parts of the gait cycle which are seen and sums up the parts. This is more accurate since more parts are used in the computation and can provide a measure even if a full stride is not present in the data (right/left heel strike to right/left heel strike).
- **Step length**: Distance between proximal end position of the contralateral foot at the previous contralateral heel strike to the proximal end position of the ipsilateral foot at the ipsilateral heel strike. This is calculated as the distance in the walking path direction.
- **Step time**: Heel strike of one foot subtract the heel strike of the ipsilateral foot.
- **Stride length**: Distance between proximal end position of the foot at ipsilateral heel strike to the proximal end position of the foot at the next ipsilateral heel strike. Stride length and width requires a walking direction to be computed, so at least 1 completed stride needs to be present.
- **Stride width**: Medio-lateral Distance between proximal end position of the foot at ipsilateral heel strike to the proximal end position of the foot at the next contralateral heel strike. Calculated by taking a stride vector, and the step in between, and computing the cross product (distance between the stride vector and the opposing step (heel) position.

4.4.7 Calculation of discrete metrics in Visual 3D

4.4.7.1 Knee metrics

4.4.7.1.1 External knee adduction moment peaks 1 and 2

The two peaks of the EKAM were calculated for the first and second half of stance phase (EKAM1, EKAM2), along with the EKAM trough defined at 50% stance. Since not all EKAM moments were bi-phasic, EKAM1 and EKAM2 were defined as peaks within (17% and 34%) and (62% and 86%) of stance phase, respectively. Specific to this thesis, these ranges were defined as the mean percentage in stance ± 2 standard deviation from 1178 trials from pre-HTO and 12-month post HTO visits for normal level gait, wide stance, toe out, and medial thrust gait styles which showed clear double peaks.

Discrete metrics for EKAM were calculated for:

- Peak EKAM value during stance
- Peak EKAM value during the first and second half of stance (EKAM1 and EKAM2, the window for which these peak discrete points are extracted from are explained above)
- EKAM value at midstance

4.4.7.1.2 External knee adduction angular impulse metrics

Knee adduction angular impulses (KAAI) were calculated as the integral of the positive and negative regions of the moment profiles separately, in three clinical planes, to provide information on average loading over the stance phase. In the frontal plane, KAAI was also calculated during the first and second half of stance and for four additional portions of the stance phase to identify when the largest effect on loading occurs following intervention; altered gait styles and/or HTO surgery.

Expanding on from Whatling et al. (2019), which incorporated discrete metrics for KAAI that were presented in previous research (L. E. Thorp et al., 2006), the following was calculated:

- Peak KAAI value during stance

- Peak KAAI value during the first half of stance
- Peak KAAI value during the second half of stance
- Peak KAAI value between heel strike to 16% of stance
- Peak KAAI value between 17% of stance and midstance
- Peak KAAI value between midstance to 83% of stance

In addition to the numerous discrete metrics during KAAI, peak knee abduction angular impulse value for both the first and second half of stance were extracted.

4.4.7.1.3 Miscellaneous knee metrics

To determine if medial thrust gait style was achieved during post-processing, varus angle ROM and frontal plane knee joint velocity during the loading phase from heel strike to 16% stance (Thorp et al., 2006) were computed to give an indication of frontal plane knee thrust. In addition, peak adduction angle during the first half of stance was calculated. This is a metric of medial thrust that has been previously reported (Gerbrands, Pisters and Vanwanseele, 2014).

Knee sagittal, frontal, and transverse planes range of motion (ROM) as well as maximum flexion, adduction and internal angles were calculated for the respective limb of interest during stance.

In addition to discrete metrics for EKAM, peak flexion and extension moments and peak internal and external moments during stance were extracted.

4.4.7.2 Hip metrics

The following hip external moment discrete metrics were calculated for three different regions of stance phase (0-100% of stance; 0-50% of stance, and 50-100% of stance):

- Peak external hip flexion and extension moments
- Peak external hip adduction and abduction moments
- Peak external hip internal and external moments

In terms of hip kinematic discrete metrics, the following were calculated:

- Peak hip flexion and extension angles during stance
- Peak hip adduction and abduction angles during stance
- Peak hip internal and external angles during stance
- Hip sagittal, frontal, and transverse planes ROM

4.4.7.3 Ankle metrics

The following ankle external moment discrete metrics for were calculated from three different regions of stance phase (0-100% of stance; 0-50% of stance, and 50-100% of stance) to identify:

- Peak dorsi-flexion moment
- Peak plantar-flexion moment
- Peak inversion moment
- Peak eversion moment
- Peak internal rotation moment
- Peak external rotation moment

In terms of ankle kinematic discrete metrics, the following were calculated:

- Peak ankle dorsiflexion and plantarflexion angles during stance
- Peak ankle inversion and eversion angles during stance
- Peak ankle external and internal angles during stance
- Ankle sagittal, frontal, and transverse planes ROM

4.4.7.4 Remaining Visual 3D metrics

To establish if toe out gait was achieved during post-processing, FPA was calculated as the angle between the long axis of the foot and the line of forward progression. The FPA at heel strike was used to determine the FPA due to the different gait style interventions and changes due to HTO.

4.5 Concurrent Optimisation of Muscle Activations and Kinematics pipeline

4.5.1 OATech+ network work placement

External joint moments may not represent the true medial-lateral joint loading distribution of the tibiofemoral joint (Kutzner *et al.*, 2013). As a result, it is important to use enhanced MSK models to predict tibiofemoral joint contact forces (van Rossom *et al.*, 2019).

The author of this thesis was awarded a £3,000 travel grant by the OATech Network+ to travel to KU Leuven's Human Movement Biomechanics Research Group which is led by Professor Ilse Jonkers. The purpose of the visit was to undergo training on the use and adaptation of an enhanced MSK modelling pipeline that predicts medial-lateral compartment tibiofemoral joint contact force and pressure distributions.

The project aimed to quantify knee joint medial-lateral contact forces in patients undergoing HTO and thus data collected at the MSKBRF at Cardiff University's School of Engineering was used. For the calculation of the medial-lateral contact force distribution expertise from the Human Movement Biomechanics Research Group at KU Leuven was needed, in terms of their musculoskeletal modelling (MSM) workflow. This workflow (COMAK) originates from Professor Darryl Thelen's group at the University of Wisconsin-Madison.

Inverse dynamics modelling is limiting as it has been suggested that external knee moments may not tell the exact picture when it comes to internal joint loading. Therefore, the purpose of the OATech Network+ travel bursary was to adopt the COMAK pipeline to predict actual joint loading. This approach has contributed to the understanding knee joint loading distribution (Brandon et al., 2018; Lenhart et al., 2015; Smith, 2017; Smith et al., 2018; van Rossom et al., 2019), which cannot be obtained from just assessing external joint moments and inverse dynamics modelling. Clinically, this is critically important to understand as patient specific outcomes are often quantified by external joint loading to determine success of surgical intervention, which may not accurately describe the internal knee joint loading.

4.5.2 Modelling the knee as a 12 degree of freedom joint

Generating subject-specific MSK geometries from imaging data is often an intensely laborious process therefore a more common approach is to scale a generic model based on subject-specific anthropometric measures. In the methods described below, the Arnold et al. (2010)

model was used, along with an integrated 12 DOF knee joint (Lenhart *et al.*, 2015), as a generic musculoskeletal geometry of the lower limb (as shown in Figure 19).

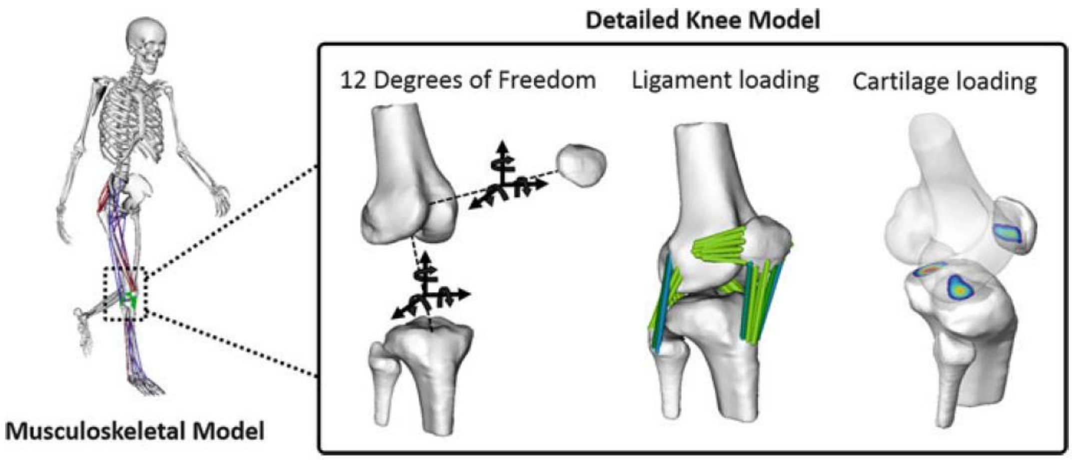


Figure 19 Twelve degree of freedom knee is incorporated into a lower extremity musculoskeletal model. Figure extracted from (Brandon et al., 2018)

To analyse the loading of soft tissues, the definition of the joint of interest must be refined to remove the artificial constraints and include explicit representation of the soft tissues (Brandon et al., 2018). To study soft tissue loading and internal joint mechanics of the knee, COMAK replaces a one DOF knee joint model (Arnold *et al.*, 2010) with a 12 DOF knee model (Lenhart et al., 2015). Removal of the artificial kinematic constraints of a one DOF knee joint model requires that the force contributions of the passive structures and articular contact are explicitly modelled and calculated. To obtain the geometries of these structures, the authors of the COMAK algorithm segmented bone, ligament, and cartilage of a healthy young adult subject from high resolution MRI (Lenhart et al., 2015).

4.5.3 Background to Concurrent Optimisation of Muscle Activations and Kinematics framework

The COMAK algorithm (Figure 20) was introduced to simultaneously solve for muscle and soft tissue loading during functional movement (Brandon et al., 2018; Lenhart et al., 2015; Smith, 2017)**.** In COMAK, inverse kinematic measurement techniques (Lu and O'Connor, 1999) are first used to compute the coordinates, speeds, and accelerations (q, $\dot{q}$, $\ddot{q}$) of the *primary model DOF*. Then, numerical optimisation is performed to simultaneously solve for the secondary kinematics (secondary kinematics consist of tibiofemoral translations and non-sagittal rotations), as well as muscle, ligament, and articular contact forces that generate the primary joint accelerations while minimising a cost function that resolves inherent muscle redundancy. The cost function that is integrated within the COMAK framework incorporates a weighted term given to each muscle, muscle volume and the muscle activation squared. Smith et al. (2016) found that penalising biarticular muscles (i.e., gastrocnemii and rectus femoris) is necessary to predict tibiofemoral contact forces consistent with measurements from instrumented implants during walking.

COMAK integrates an extended knee model, that allows 6 DOF patellofemoral and 6 DOF tibiofemoral movement, in a generic full-body model (Lenhart et al., 2015). Each leg included 44 musculotendon actuators spanning the hip, knee, and ankle and 14 bundles of non-linear springs that represent the major knee ligaments and posterior capsule. A non-linear elastic foundation formulation was used to calculate the cartilage contact pressures, based on the penetration depth of the overlapping surface meshes of the contact model (Lenhart et al., 2015; Smith, 2017; Smith et al., 2018). The ligament force-strain relationship was assumed quadratic at low strains and linear at high strains to capture the nonlinear effects of collagen crimp straightening and fiber elongation (Huiskes & Blankevoort, 1991).

The cartilage was modelled with a uniformly distributed thickness of 4mm tibiofemoral and 7mm patellofemoral (Eckstein *et al.*, 2001; Hudelmaier *et al.*, 2003) and the elastic modulus and Poisson's ratio were assumed as 10MPa and 0.45, respectively (Adouni & Shirazi-Adl, 2014; Li et al., 2001). The articular surfaces are represented by high resolution triangular meshes that do not deform but are allowed to interpenetrate. Contact pressure on each triangle face is computed based on the local penetration depth (d), cartilage thickness (h), and material properties (the elastic modulus and Poisson's ratio were assumed as 10MPa and 0.45, respectively).

This model was implemented in Software for Interactive Musculoskeletal Modelling (SIMM) with the Dynamics Pipeline (Musculographics Inc., Santa Rosa, CA) and SD/Fast (Parametric Technology Corp., Needham, MA) to generate the multibody equations of motion. SIMM is a powerful tool kit that facilitates the modelling, animation, and measurement of 3D musculoskeletal systems. In SIMM, a MSM consists of representations of bones, muscles, ligaments, and other structures.

4.5.3.1 MATLAB-OpenSim scripting environment

MATLAB (The Mathworks, Inc., USA) is a common analysis tool used for data manipulation, signal processing and function integration. These features can be used in conjunction with simulation tools provided by the OpenSim interface. The processing of each stage of the simulations within this thesis was undertaken using MATLAB-OpenSim scripting environment.

4.5.4 Validation of Concurrent Optimisation of Muscle Activations and Kinematics framework

The knee model performance has previously been validated (Lenhart et al., 2015). As kinematic validation, the predicted joint kinematics in the secondary DOF of the knee were validated against joint kinematics measured using dynamic MRI and are reported in the study of Lenhart et al. (Lenhart et al., 2015). As dynamic validation, the calculated knee contact force was compared with instrumented implant data provided through the Grand Challenge Competition to Predict in vivo knee loads, a subject-specific data set that allows researchers to validate muscle and contact forces estimated in the knee. When comparing between the measured and calculated knee contact forces, the joint contact load prediction errors for root mean square (rms) was 0.33 BW (Smith, Vignos, *et al.*, 2016; Smith *et al.*, 2018). These findings were comparable to those observed from a unique optimisation approach, termed force-dependent kinematics, introduced by the 2014 “Grand Challenge” winner (Marra *et al.*, 2015) (rms error = 0.26 BW). Finally, they were slightly better than those that have been obtained using traditional optimisation or forward dynamic simulations (Kinney et al., 2013; Shakoor & Block, 2006).

4.5.5 Observed kinematics: Primary vs secondary kinematics

Optical motion capture enables the measurement of segment kinematics during functional Movement. However, because skin and soft tissue motion prevent direct observation of the underlying bones, motion capture is limited in its ability to quantify DOF which undergo small excursions during full body movement (Li *et al.*, 2012).

The differentiation between measurable DOFs of high (primary) and low (secondary) confidence is a key concept in COMAK. The algorithm solves for the muscle and soft tissue loads necessary to generate the measured motion of the primary DOFs while simultaneously predicting a dynamically consistent set of secondary kinematics.

Three-dimensional hip rotations, tibiofemoral flexion and ankle flexion are the primary DOFs. The tibiofemoral translations, non-sagittal rotations, and all patellofemoral DOFs cannot reliably measured when undertaking standard optical motion capture without the capabilities of biplane x-ray equipment (Leardini *et al.*, 2005) and thereby form the secondary DOFs. Pelvis translations and rotations are also measurable but classified as prescribed DOFs such that their accelerations are prescribed within the multibody model to ensure consistency with observed multibody dynamics.

The joint kinematics of the primary and prescribed DOFs are calculated from the measured motion capture marker trajectories using inverse kinematics. The inverse kinematics routine is formulated as a global optimisation to determine the generalised coordinates of the primary and prescribed DOFs that minimise the sum of squared differences between model marker positions and measured marker positions at each time step (Lu and O'Connor, 1999). To enable this calculation, the secondary DOFs must be also determined, but are unknown at this stage. To account for this, the secondary generalised coordinates are constrained to be functions of the primary generalised coordinates during inverse kinematics optimisation (Gerus *et al.*, 2013). These constraints are then removed when later performing the optimisation for dynamically consistent soft tissue loads and secondary kinematics.

Fundamentally, this approach assumes that the differences between the constrained secondary kinematics and load-dependent secondary kinematics subsequently predicted by COMAK have negligible influence on the primary coordinates calculated by inverse kinematics. For the knee joint, internal rotation, varus-valgus rotation, and all translations can be defined as functions of the knee flexion angle which have been reported in literature (Walker, Rovick and Robertson, 1988). In practice, secondary kinematics are calculated so that they are consistent with the articular geometry of the model by performing passive (i.e., minimal muscle activation) forward simulations where a primary DOF is prescribed to travel through its range of motion and the secondary kinematics evolve because of the contact, ligament, and passive muscle forces (Lenhart et al., 2015).

4.5.6 Simultaneous optimisation

After inverse kinematics, an optimisation problem (COMAK) is solved to simultaneously predict the muscle and soft tissue loading and secondary kinematics required to generate the measured primary accelerations. The optimisation is formulated to solve for the muscle activations and secondary coordinates that minimise an objective function while satisfying overall dynamic constraints. The dynamic constraints require that optimised muscle forces and internal joint loads (ligament and contact forces) resulting from the optimised secondary kinematics generate the measured primary accelerations, while inducing equilibrium (zero accelerations) in the secondary DOFs.

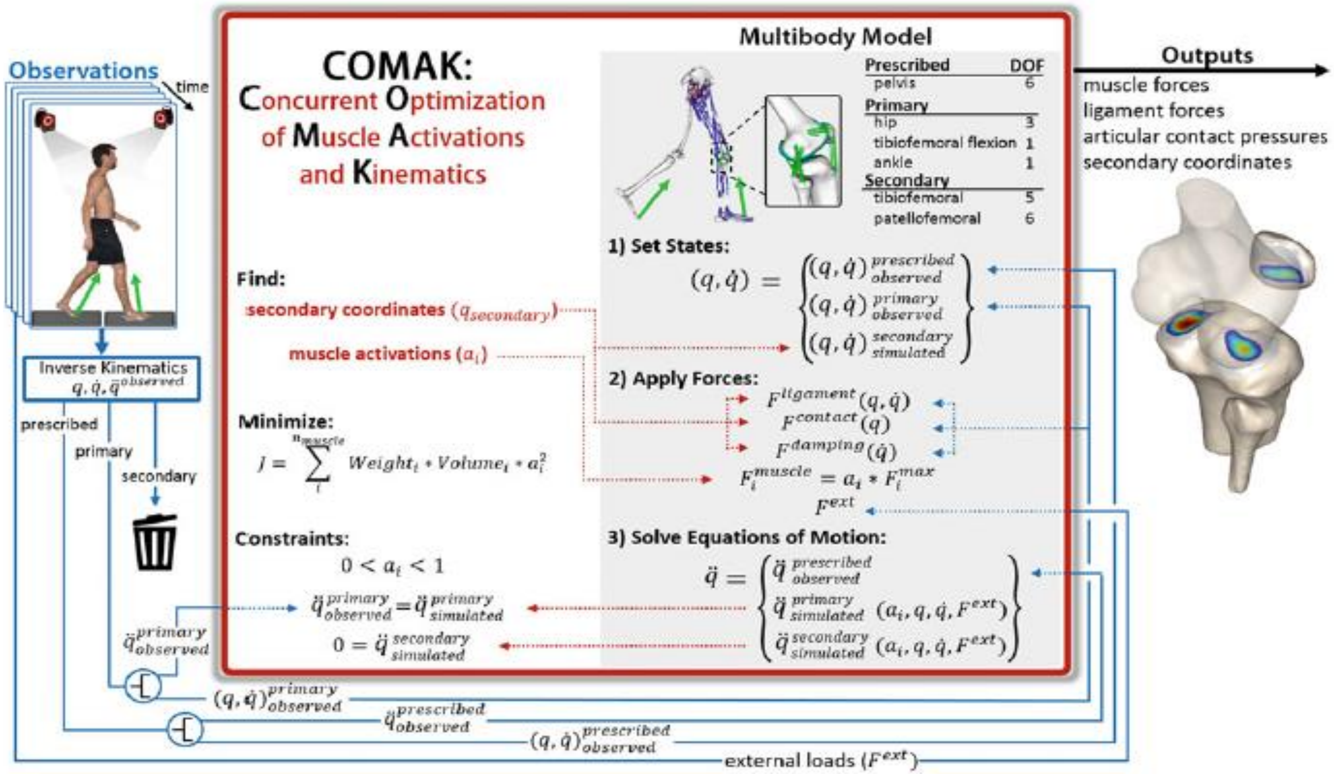


Figure 20 COMAK: Concurrent Optimisation of Muscle Activations and Kinematics framework

The COMAK algorithm is a concurrent simulation method that integrates a multibody musculoskeletal with a detailed knee joint representation and external observations of full body movement to predict soft tissue loading. The optimisation is formulated to solve for the muscle activations and secondary coordinates that minimize an objective function while satisfying overall dynamic constraints. The dynamic constraints require that optimised muscle forces and internal joint loads (ligament and contact forces) resulting from the optimised secondary kinematics generate the measured primary accelerations, while inducing equilibrium (zero accelerations) in the secondary DOFs. Figure extracted from (Brandon et al., 2018)

At the first-time step in COMAK, the prescribed coordinates, speeds, and accelerations $(q,\dot{q},\ddot{q})^{prescribed}$ and primary coordinates and speeds $(q, \dot{q})$ are set to their observed values, and a forward simulation is performed with minimal muscle activations (a_i=0.01) to allow the secondary coordinates, $q^{secondary}$, to settle into an initial pose. At each subsequent time step, $(q, \dot{q})^{prescribed}$ and $(q, \dot{q})^{primary}$ are set to their observed values, while $q^{secondary}$ and a_i are determined by the optimisation.

The secondary speeds, $\dot{q}^{secondary}$, are determined from the difference between $q^{secondary}$ at the current and previous time steps. After setting the states of the model, the generalised forces are computed and applied.

The contact forces are calculated using the elastic foundation model, $F^{contact}(q)$**,** while the ligament forces are computed using the nonlinear spring model, $F^{ligament}(q, \dot{q})$.

Viscous damping forces are applied to each DOF to ensure minimal changes in kinematics between time steps $F^{damping}(\dot{q})$. The muscle forces are computed from the activations $F^{muscle}(ai)$ and the measured external forces, F^{ext}, are applied to their corresponding segments. The $\ddot{q}^{prescribed}$ are then prescribed to their measured values and the equations of motion are solved for $\ddot{q}^{primary}$ and $\ddot{q}^{secondary}$.

Three constraints to the COMAK pipeline, which are detailed below, are muscle activations, the simulated accelerations are consistent with the primary accelerations collected during the motion capture session, and third is that the secondary accelerations are constrained to zero.

First constraint:

Muscle activations are constrained to vary between 0 and 1, to ensure the resulting muscle forces are physiologically reasonable.

$$0 < a_i < 1$$

Second constraint:

Consistency with measured gait dynamics is ensured by satisfying the constraint that the simulated accelerations of the primary DOF matched the observed values.

$$\ddot{q}^{primary}_{j\ observed} = \sum_{i=1}^{n\ muscles} a_i F_i^{max} \ddot{\hat{q}}^{muscle}_{j,i}(q) + \ddot{q}^{other}_{j}(q,\dot{q})$$

Third constraint:

Accelerations of secondary DOFs are constrained to be zero.

$$\ddot{q}_k^{secondary} = 0 = \sum_{i=1}^{n\ muscles} a_i F_i^{max} \ddot{q}_{k,i}^{\wedge muscle}(q) + \ddot{q}_k^{other}(q, \dot{q})$$

In these equations, $\boldsymbol{\ddot{q}_{j,i}^{\wedge muscle}}$the acceleration along coordinate *j* due to a unit force in muscle *i*, **while** $\boldsymbol{\ddot{q}_{j}^{other}}$constitute the accelerations due to all other forces in the multibody system (contact, ligament, damping, external, gravitational, centripetal and Coriolis). The third constraint assumes that inertial effects due to accelerations in the secondary degrees of freedom are negligible. During gait, this assumption is justified given the small mass of the patella and the small magnitudes of rotational and translational excursion in secondary degrees of freedom. The MSK system allows multiple combinations of $q^{secondary}$ and *ai* to fulfil these constraints, thus static optimisation must be performed to minimise an objective function and identify a unique solution.

The objective function (*J*) used by COMAK is generalisable, allowing any user defined quantity to be minimised. Smith et al. (2018) found a common cost function proposed for static optimisation performs well for COMAK in most applications:

$$J = \sum_{i}^{n\ muscles} W_i * V_i * a_i^2$$

where W_i is a weighting term, V_i is muscle volume, and a_i is the muscle activation. The weighting term enables the activation of individual muscles to be penalised within the optimisation. Penalising biarticular muscles (i.e., gastrocnemii and rectus femoris) is necessary to predict tibiofemoral contact forces consistent with measurements from instrumented implants during walking. This redistributes the hip flexor moment and ankle plantar-flexor moments to the uni-articular muscles during late stance, reducing the loading in the muscles crossing the knee and thus the compressive contact force.

Introducing an additional term in the cost function to minimise articular contact energy also produces similar contact force predictions, largely by similarly redistributing the muscle loading to uni-articular muscles (Smith et al. 2016).

However, the errors in the numerical calculation of the gradient of contact energy with respect to the optimised secondary kinematics can cause convergence issues within the optimisation.

4.5.7 Adaptations to the Concurrent Optimisation of Muscle Activations and Kinematics pipeline: Editing the Lenhart model so baseline varus angle better-approximates patients lower limb alignment

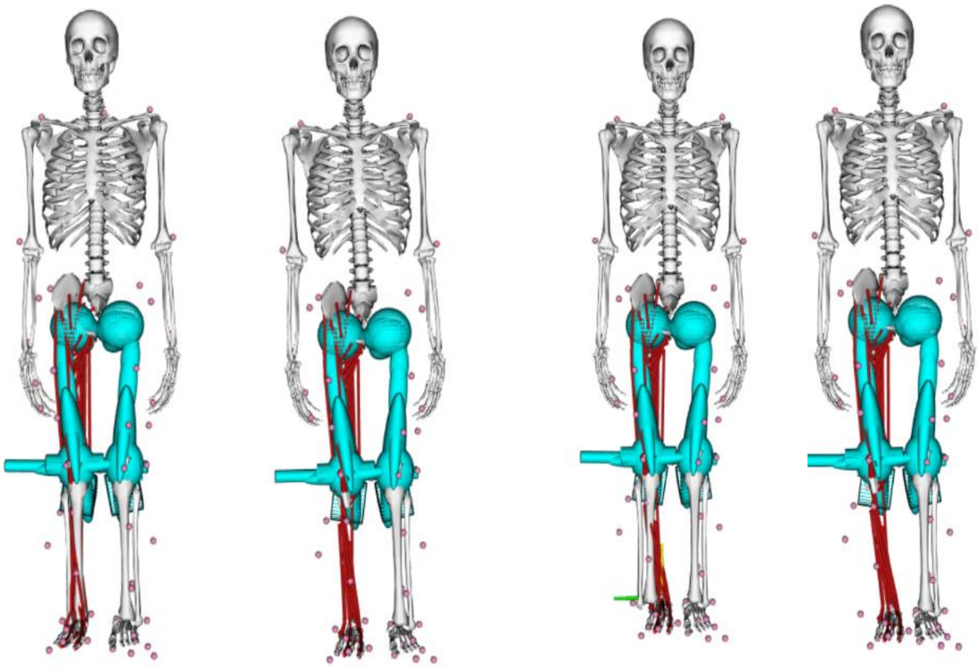
With discussions with Professor Jonkers and Dr van Rossom from KU Leuven, we aimed to incorporate patient-specific lower-limb frontal plane malalignment into each patient's model. We integrated patient-specific mTFA that was recorded by a trained clinician into the lower-limb of each patient's session. The approach taken is addressed below. Upon initial examination, this adaptation did not alter total/medial/lateral tibiofemoral joint contact forces considerably.

When simulating with the malalignment, the location of the foot with respect to the measured GRF application point was medialised. To ensure that for these simulations the application of the GRF to the foot was identical to the reference simulation, the COP (of the GRF) was expressed in the local reference frame of the foot.

Varus alignment was accounted for by using the mTFA angle and implemented within the original COMAK model. When doing this the configuration of the bone file, joint file, contact file, muscle file, and muscle-tendon parameters were accounted for.

The model was altered so that, in a static pose, the HKA angle (frontal plane) better matched the patients lower limb alignment. In simpler models, researchers have implemented a knee adduction angle to OpenSim models for scaling, then locked this DOF for dynamic activities. The reasoning behind this is that bone pin data (Benoit *et al.*, 2006) shows that the knee moves very little in frontal plane during gait.

For the Lenhart model, the tibiofemoral adduction is defined by the geometry of the contact surfaces and so it was not feasible to manually alter this. Therefore, it was decided to adjust the tibial and foot segments, i.e., ankle joint. The ankle is translated to a new location based on the mTFA angle to effectively introduce varus/valgus and corrected for foot alignment, so it lands flat on the floor.



0° angle alteration. 5.6° angle alteration.

Figure 21 Visual implementation of varus angle into the right lower-limb model of a patient participant. Visualised within OpenSim 3.3

The mechanical tibiofemoral angle that is recorded from a long-legged x-ray is implemented within the patient models to adjust for a lower-limb alignment. The tibia was not rotated, the malalignment is implemented through repositioning the ankle joint.

4.5.8 Calculation of discrete metrics in the Concurrent Optimisation of Muscle Activations and Kinematics framework

For each trial, the stance phase was identified as the period in which the GRF exceeded 20N.

Next, the magnitude and timing of the first and second peak (FP and SP) of the resultant tibiofemoral contact force was determined during the first and second half of the stance phase, respectively, as well as the minimum force during single leg support (MS). This approach is the same as the methods used in van Rossom et al. (2019). Like the approach implemented in Visual 3D, the mean ±2 standard deviation of first and second peak was established using a cohort of pre-HTO and control participants that had a clear double peak waveforms in total tibiofemoral contact force during unaltered level gait. Due to time constraints, the altered gait styles were not included in the calculations of first and second peak. Out of 88 participants, 76 participants were used to calculate the windows for first and second peak. First peak total tibiofemoral contact force window was 23-31% of stance, whilst the second peak tibiofemoral contact force window was 69-82% of stance.

The concomitant average and maximum pressure over the contact surface was analysed. Tibiofemoral mean pressure and maximum pressure were determined for the total tibiofemoral joint, as well as for the medial and lateral condyle separately, and were averaged over three trials. Additionally, the joint angles in the hip, knee, and ankle were determined at FP and SP tibiofemoral contact force as well as at midstance (MS). Range of motion (ROM) and the external joint moments in the hip, knee and ankle at FP, SP and MS were analysed.

Furthermore, the point of application of the total knee, medial and lateral contact force expressed in the local reference frame of the tibia at FP, SP and MS were analysed.

External joint moments were scaled to %BW.h, contact forces were scaled to bodyweight (BW) and contact pressures are presented as megapascal (MPa).

4.5.9 Advantages of using the Concurrent Optimisation of Muscle Activations and Kinematics framework

Prior to being awarded the OATech Network+, the author of this work was going to define medial compartment knee joint loading via EKAM and KAAI parameters to align with what has been undertaken in the literature (Simic et al., 2011; Bowd et al., 2019). However, the COMAK pipeline afforded the opportunity to use an enhanced MSM pipeline to predict internal knee joint contact forces for the first time in this cohort.

An understanding of in vivo soft tissue loading is essential for investigating non-surgical and surgical interventions. Directly measuring the loads on in vivo soft tissues (e.g., ligament, tendon, and cartilage) is generally very invasive in nature and impractical for widespread use. Concurrent simulation approaches have been introduced, which enable simultaneous solution of muscle forces and soft tissue mechanics underlying human movement. Concurrent approaches are advantageous to directly probe inherent coupling between muscle coordination, joint kinematics, cartilage contact pressure, and ligamentous behaviour that can arise with soft tissue damage.

4.6 Principal Component Analysis and the Cardiff Classifier

4.6.1 Background

Principal component analysis (PCA) is a technique for reducing the amount of data from biomechanical waveforms to interpretable sizes whilst maintaining the critical valuable information from the data. It does so by creating new ‘variables’ that account for the largest variance between groups. As such, PCA is a useful method for analysing human motion data (Chau, 2001; Deluzio et al., 1999, 1997).

The appeal of this multivariate statistical tool is that it does not rely on a pre-determined selection of specific features of a waveform such as peaks and troughs that are conventionally assessed in the human movement biomechanical literature. PCA provides an objective characterisation of waveform features, helping to identify the main gait patterns within participants in a dataset. Deluzio et al. (1999) paper showed that principal component (PC) scores were sensitive to gait changes associated with KOA and therefore provide a sound basis to identify important gait features that would differ from a control non-pathological group to that of individuals who have mKOA and associated varus deformity.

At the MSKBRF, School of Engineering at Cardiff University, the application of PCA has been combined with a classification method based on a Dempster-Shafer theory of evidence, termed the ‘Cardiff Classifier’. This method has been demonstrated to accurately characterise the biomechanical changes in late-stage KOA subjects and for measuring recovering post TKR (Biggs, 2016). The technique enables decision making in the presence of ignorance, which is particularly pertinent when classifying KOA function due to the quantity of corroborating and conflicting evidence. Additionally, the Cardiff Classifier facilitates the identification of key gait parameters following TKR surgery (Biggs, 2016; Biggs et al., 2019; Jones & Holt, 2008; Metcalfe et al., 2013; Whatling et al., 2008) as well as hip OA (Biggs et al., 2021).

In Chapter 5, PCA and the Cardiff Classifier are used to better understand biomechanical factors affecting varus deformity of the knee in an HTO cohort and this is the first time this method was utilised in this population. Chapter 8 will outline the future work that is planned to follow the findings from this thesis.

PCA is a common approach to assess the variance in a dataset. There will be n number of PCs for n number of data points, where PC1 corresponds to the axis of primary variation within the standardised dataset and PC2 corresponds to the second amount of variance in the dataset (but must be orthogonal to PC1). Each PC is orthogonal to the preceding PC.

When considered during the gait cycle, PCA can be performed on each of the 101 percentage points. PCA is therefore performed on all 101 data points. When attempting to calculate a dataset with many dimensions, it is much easier to undertake such calculations within computer software. The software used to perform eigen decomposition of the correlation matrix PCA within this thesis was MATLAB. Whereby all scripts were previously written and edited by a previous researcher (Dr Paul Biggs).

The reader is directed to Dr Paul Biggs PhD thesis (Biggs, 2016) for a comprehensive introduction and understanding of the theory related to PCA and the Cardiff Classifier. PCA and the Cardiff Classifier will be briefly outlined in this thesis.

4.6.2 Methods

Twenty-two knees (21 participants) scheduled for HTO surgery were included in this study. Participants were assessed pre-operatively and again at a target of 12 months post-operatively. Additionally, twenty healthy volunteers were recruited into the study. The study was approved by the Research Ethics Committee for Wales and Cardiff and Vale University Health Board. Human motion analysis was performed during level gait using a modified Cleveland marker set. Participants walked at their self-selected speed for a minimum of 6 successful trials. The motion capture procedure has been previously outlined in detail in this thesis. Hip, knee and ankle kinematics and kinetics were calculated within Visual 3D (C-Motion) (as shown in Section 4.4 of this chapter).

4.6.2.1.1 Principal Component Analysis of biomechanical waveforms

Typically, biomechanical analysis involves comparing discrete metrics that are taken from biomechanical waveforms. One such example would be to assess the frontal plane knee joint ROM during the stance phase between a healthy and a pathological cohort. However, this approach subjectively discards vital temporal information.

Another approach, which was first used with our research group in Jones (2004) involves treating every percentage point of the gait cycle (*or stance phase*) as an independent variable. The variables are then related to each other via PCA. PCA creates a new axes system that has as many dimensions as the number of data points in the original waveform.

As outlined in the Pataky et al. study, a downside of extracting discrete metrics is the potential introduction of type 1 (false positive) statistical errors if the metrics are not defined a priori to the analysis (Pataky, Vanrenterghem and Robinson, 2016). It is therefore arguable that PCA is an advantageous approach to undertake as there is less chance of type 1 statistical error along with the temporal information being kept.

PCA is not just one fixed technique. The approach that has been used at the MSKBRF is PCA using eigendecomposition of the correlation matrix. This methods section will briefly outline the fundamentals of PCA.

Standardisation

The first step to the PCA approach used in our research group is to standardise the data (Chau, 2001). This can be calculated by removing the mean and dividing it by the standard deviation. The resulting value is referred to as the ‘z-score’. This effectively scales the independent variables by different amounts and normalises each input value, so it has a zero mean, and a standard deviation of 1. This PhD used the ‘z-score’ function within an in-house MATLAB script.

By calculating the z-score, the assumption is that the variance in the data points is not important. This has a subtle knock-on effect on the interpretation of the PCs and is beyond the scope of this PhD to delve into further.

Correlation matrix

The next step in PCA utilised within this thesis was to calculate the correlation coefficient matrix. This is the correlation between all the input variables with each other. With the data being standardised in the previous step, the covariance matrix is equivalent to the correlation matrix (Chau, 2001).

Eigendecomposition

The eigenvalue is the amount of variance in that data explained along that PC and the Eigenvectors are essentially the PCs. In PCA, the PCs are always orthogonal. Once PC1 is defined as being the PC with the most variance, then PC2 can be calculated as the PC with the second most amount of variance as well as being orthogonal to PC1. This then continues until n PCs have been calculated for n amount of data points.

There is no full consensus on how many PCs should be kept for further analysis. However, previous research has suggested to (a) have a cut off percentage point on the number of PCs to keep (b) keeping eigenvalues greater than 1 (c) keeping all the eigenvalues that make up 95% of the variance. The critical factor to consider is how much of the amount of the original data that is decided to be kept. For this thesis, the top 3 PCs for each variable of interest were selected to go forward and inputted into the Cardiff Classifier. A full explanation of eigendecomposition can be found on pages 69-71 of Biggs (2016) PhD thesis.

Transforming data points

Each point can be transformed to the axis system simply by multiplying by the eigenvector. This can be referred to as the PC scores. For this work, the interpretation of the single component reconstruction was used. Figure 22 below shows the original data and the reconstruction of the first three PC values for that data. Figure 23 is an example of PC1 for the frontal plane knee moment from the data collected within this thesis. Figure 23 represents the magnitude difference during the stance phase of gait that is typically seen between mKOA and a control healthy group.

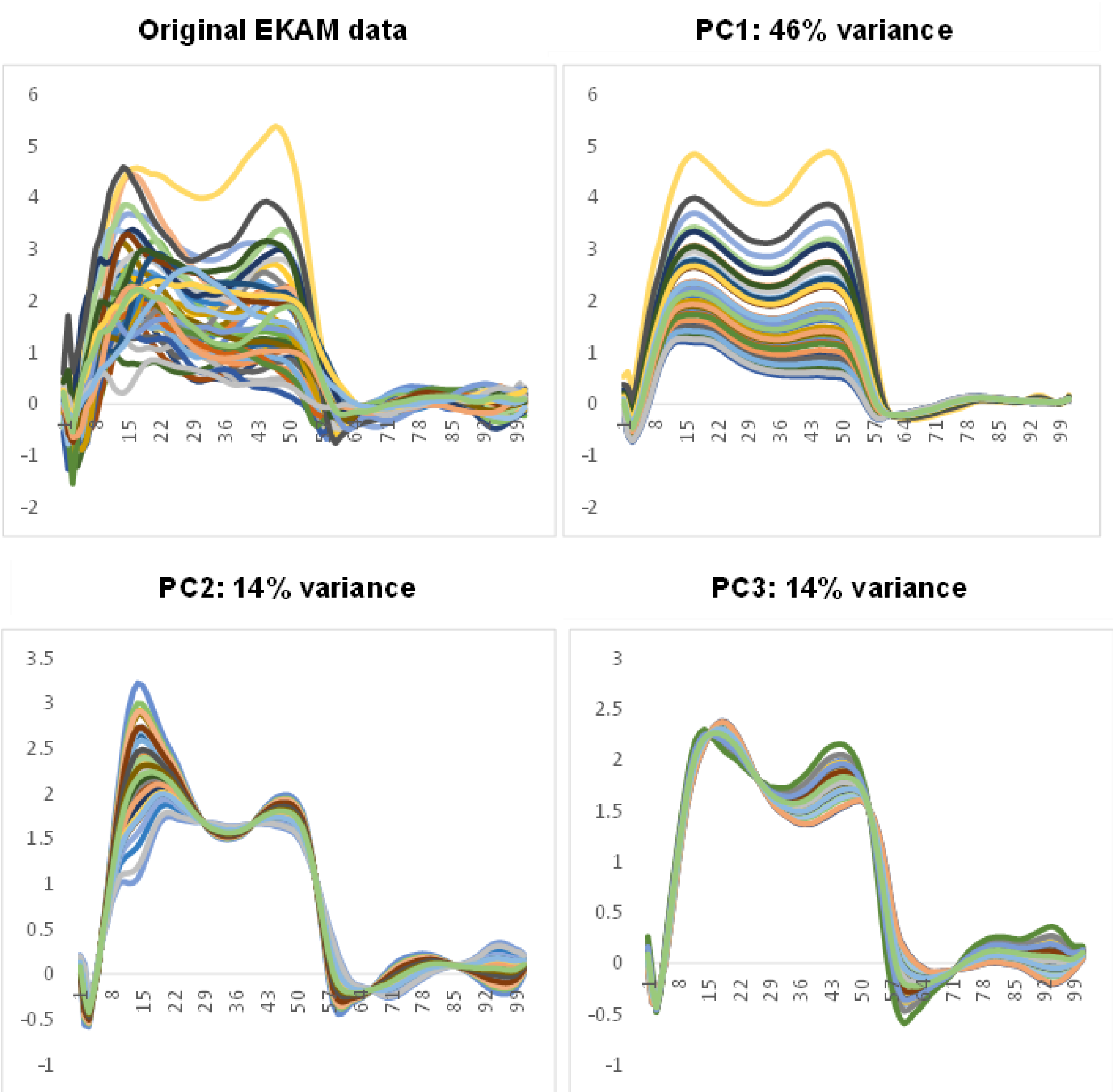


Figure 22 An illustrative demonstration of the application of principal component reconstruction to external knee adduction moment waveforms

The original data (top left) for all subjects is plotted, alongside the reconstruction using PC1 (top right), PC2 (bottom left), and PC3 (bottom right).

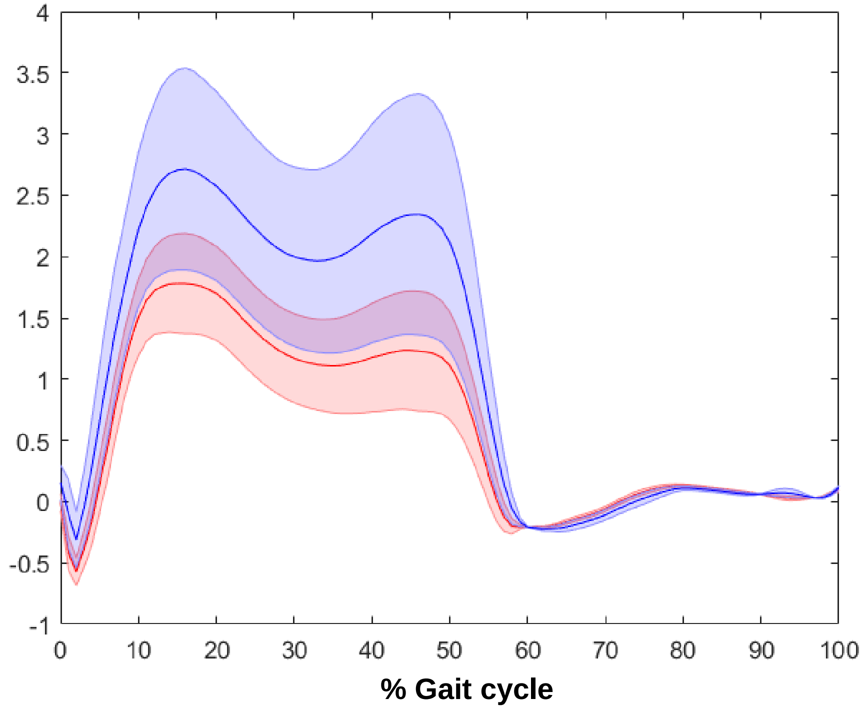


Figure 23 Frontal plane knee moment PC1 example

The external knee adduction moment PC1, in this instance, represents a magnitude difference between the control group (red line) waveforms and the pre-HTO cohort (blue line). This means that the pre-HTO patients had a greater knee adduction moment during stance. The knee adduction moment PC1 explains 46% of the variance in the data.

Software used to perform Principal Component Analysis

This thesis used a MATLAB code called ‘Defining_PCs.m’, written by Dr Paul Biggs. The code calculates the ensemble average per participant for all inputted walking trials per input variable. As such no intra-participant variability between trials data is kept for this analysis. This would be especially important for any individuals who had any walking trials that were not closely aligned to the rest of the trials. The script generates a ‘results’ folder containing as many Excel workbooks as the number of original variables inputted for the analysis. Each Excel workbook includes separate sheets for the original data, eigenvalues, eigenvectors, factor loadings, filtered factor loadings, the first six principal components of the variable in question and the PC values. Each sheet contains the data for each participant.

Retention of principal components

For this thesis, and per the same approach previously taken within this research group (Biggs, 2016) the first three PCs of each input variable were initially selected for each participant. Single-component reconstruction of the PCs was performed alongside representative extremes of each PC to aid interpretation of the biomechanical feature reconstructed by each component (Figure 24).

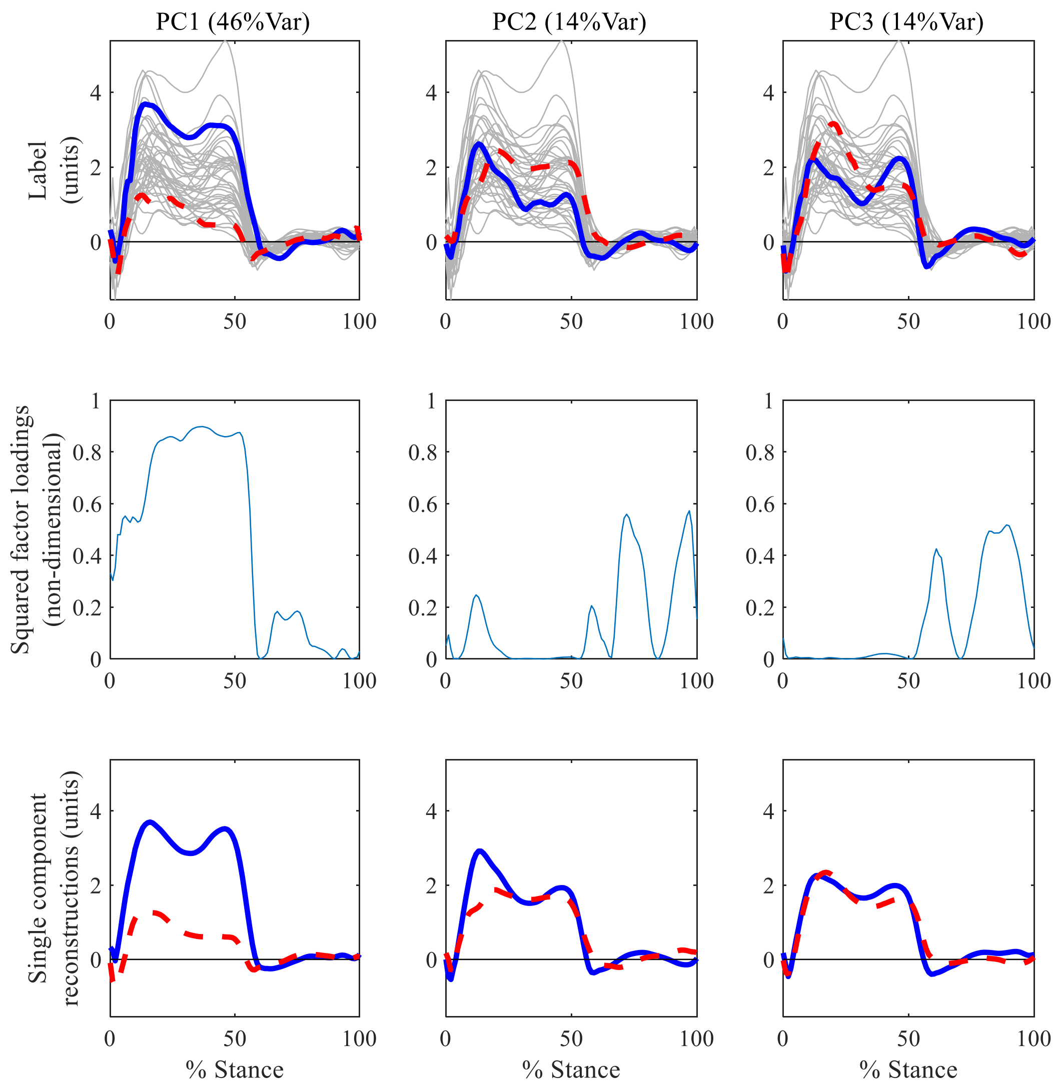


Figure 24 Principal Component Analysis outputs to aide interpretation

4.6.2.1.2 The Cardiff Classifier

The Cardiff Classifier is based on the Dempster-Shafer theory which has the advantage to deal with uncertainty. The method behind the Cardiff Classifier is comprehensively described in Dr Biggs' PhD thesis (Biggs, 2016) and recently published in peer-reviewed papers (Biggs et al., 2019; Biggs et al., 2019; Biggs et al., 2021). This thesis will therefore only briefly outline the Cardiff Classifier and how it was utilised to undertake an exploratory study into classifying mKOA in varus deformed individuals pre- and post-HTO.

Time-normalised hip, knee, and ankle kinematics, kinetics, and powers in all three planes, alongside the three components of the ground reaction force were extracted for each participant from Visual 3D (C-Motion) and retained for further analysis. This means that there were 36 variables for each participant (non-pathological control individual and patients with mKOA and varus deformity included in the study). Firstly, PCA was performed (as outlined in the previous section) on the abovementioned gait variables of each participant. The first three PCs of each variable were selected for the next stage of the analysis, resulting in 108 discrete variables per participant.

Data reduction, ranking, and selection of input features

The present study implements the same approach utilised by Biggs et al. (2021). This approach made improvements on previously published feature-selection methods before the application of the Cardiff Classifier (Metcalfe *et al.*, 2017). These changes aimed to reduce the risk of over-fitting the model.

The training data were split into two equal groups and the classifier was used to rank the input features within both datasets rendering the top 15 most robustly discriminatory input features for classification. For both training datasets, each of the 108 input features, all the participants were classified using the feature for which a classification accuracy was obtained. The average classification accuracy across the two sets was used to rank the input features. The top 15 most robustly discriminatory input features are shown in Chapter 5 (

Parameters		**Accuracy (%)**	**Variance represented (%)**	**Low PC Interpretation**
Kinematics – operative limb				

Hip	Hip flexion	PC2	74	13%	Reduced hip ROM during stance phase.
	Hip adduction	PC1	74	60%	Reduced hip magnitude throughout the gait cycle.
Knee	Knee adduction	PC1	74	70%	Increased knee magnitude throughout the gait cycle.
Ankle	Ankle inversion	PC1	74	66%	Reduced ankle magnitude throughout the gait cycle.
Kinetics – operative limb					
Hip	Hip transverse moment	PC2	81	14%	Increase in the swing phase of the gait cycle.
	Hip transverse moment	PC1	71	42%	Increased magnitude in the first 50% of the gait cycle.
Knee	Knee adduction moment	PC1	74	46%	Increased magnitude in the first 60% of the gait cycle.
	Knee flexion moment	PC2	69	19%	Decreased magnitude between 10-20% of the gait cycle, increased magnitude between 20-50% of the gait cycle.
	Knee internal rotation moment	PC1	69	41%	Increased magnitude of internal rotation moment during the stance phase of gait.
Ankle	Ankle sagittal power	PC2	71	14%	Reduced power output between 40-60% of the gait cycle.
	Ankle transverse power	PC1	71	16%	Difficult to give interpretation.
		PC3	69	11%	Phase shift.
	Ankle adduction power	PC1	69	27%	First 50% of gait cycle.
	Ankle sagittal moment	PC2	69	22%	Difficult to give interpretation.
Ground reaction force					
	Medial-lateral GRF	PC3	69	11%	Phase shift.

Table 5-16). The Classifiers robustness was validated using a 'leave-one-out cross-validation approach whereby n−1 participants are used to train the classifier. This robustness approach is repeated n times until each participant has been left out.

Figure 25 outlines the training process which defines the control parameters of the sigmoid curve (known as the Confidence Function). This process converts each PC value for each individual into a value between 0 and 1 for which then converts into a body of evidence of 'Belief of Non-Pathologic', 'Belief of OA', and 'Uncertainty'. This classifier implemented a threshold into defining/attributing to the belief system of Belief of OA ≥0.5 and determined dominant Belief of OA. Finally, a single individual value is given using Dempster's rule of combination which results in a single combined body of evidence for each of the participants. The change in Belief of OA is denoted as the change in overall gait function by calculating the difference between the pre-and post-operative Belief of OA values.

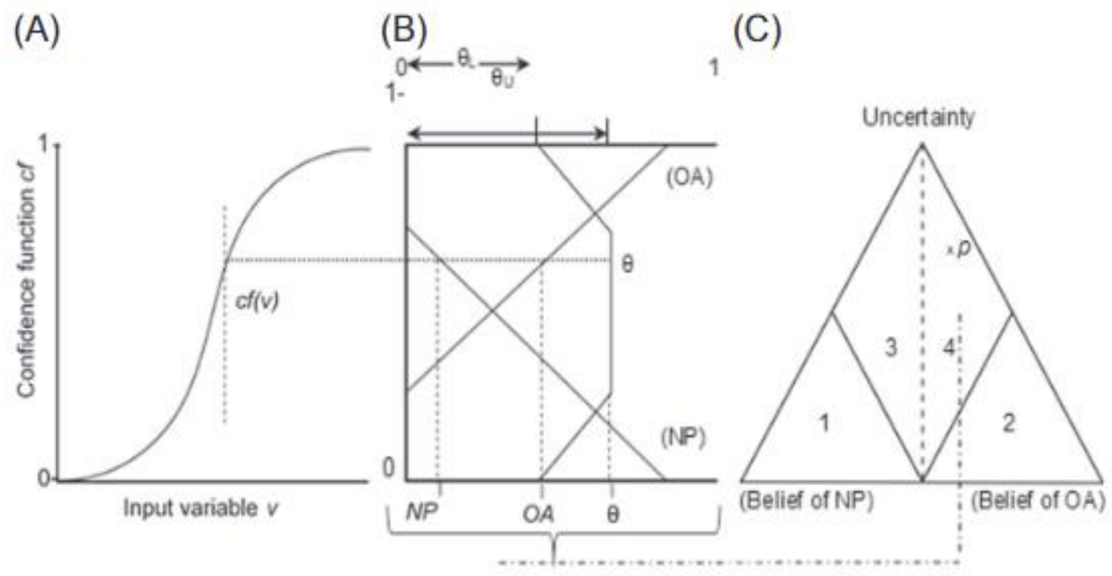


Figure 25 The classification method showing the interaction of its three main stages. This figure is extracted from Biggs et al. (2021)

In the figure above, the conversion of input variable, v, into confidence factor cf(v) using the sigmoid confidence function is performed (A). Then, a conversion of confidence factor into body of evidence (BOE) (B). Then (C) is a visualisation of the BOE within a simplex plot, denoted by the point p. The three belief values are plotted as a distance towards the corresponding vertex (NP), (OA), and θ. The simplex plot is divided into four regions: 1 denotes the dominant NP classification region, where B(NP)≥0.5; 2 denotes the dominant OA classification region where B(OA)≥0.5; 3 denotes the nondominant NP classification region, where 0.5>B(NP)>B(OA), and 4 denotes the nondominant OA classification regions, where 0.5>B(OA)>B(NP). The dotted vertical line, along which B(OA) = B(NP), is the decision boundary between a classification of NP and OA gait characteristics.

A Pearson product-moment correlation coefficient was computed using SPSS 25 to assess the relationship between the change in B(OA) to the change of OKS, as well as the change in mTFA pre-to-post HTO.

4.7 Statistical analysis

Unless stated elsewhere in this thesis, the following statistical analysis was undertaken. Paired samples t-test was performed in MATLAB (MATLAB 2020a, The Math Works, Inc., Natick, Massachusetts, USA) to identify significant differences associated with HTO surgery or undergoing the altered gait styles to the pathological groups unaltered level gait. Where parametric assumptions were not met, a Wilcoxon signed-rank test was used.

Independent t tests were used to determine significant differences in the pre- and post-HTO measurements as well as the altered gait styles compared to the control group. Where parametric assumptions were not met, a Mann–Whitney U test was performed. Significance was determined when $p < 0.05$ for all statistical tests. Irrespective of the analysis undertaken, a full inspection for any outliers was undertaken.

CHAPTER 5: BIOMECHANICAL DIFFERENCES BETWEEN A CONTROL COHORT, AND PRE- & POST-HTO

5.1 Chapter background

mKOA and associated varus alignment has been proposed to alter knee joint loading by increasing medial compartment loading. HTO surgery aims to correct the varus malalignment and unload the medial compartment by lateralising the weight bearing line. Osteotomy around the knee is increasingly employed by surgeons looking to offer joint preserving surgery for younger patients. These younger patients occupy a treatment gap when mKOA is mild or moderate. These patients have often been told to wait until they are candidates for TKA without receiving symptomatic relief. A well-executed HTO can delay disease progression and treat mKOA with pain relief and durable restoration of function. However, when compared to uni condylar and TKA, HTO has been criticised for exposure to the risks of failure and revision surgery; a critique that fails to observe the greater functional benefit from retaining native knee kinematics and ignores that primary arthroplasties are also threatened by revision. A total of 1,776 cases of osteotomy surgery were registered in the United Kingdom Knee Osteotomy Registry between 1 December 2014 and 1 December 2017. A total of 1,652 patients were entered into the Registry in this period, suggesting several bilateral or revision cases. Of these, 621 patients have undergone surgery (34.97%) and 1,155 (65.03%) are either waiting for surgery or have no operative data entered on the registry (Palmer *et al.*, 2018).

Internal loads cannot be measured in-vivo, and so joint loading is either inferred by joint moments or by computational models that estimate the joint contact forces. The overarching purpose of this chapter is to quantify biomechanical differences between individuals with varus deformity and mKOA pre-surgery compared to a healthy cohort, and then to establish the effects of HTO by comparing pre-HTO and post-HTO as well as comparing the healthy cohort to post-HTO.

This chapter will firstly use a pipeline developed in Visual 3D (C-motion, USA) to answer the following:

1. Compare spatial temporal parameters between a healthy cohort, individual's pre-HTO, and 12 months post-HTO.
2. Compare hip, knee and ankle external moments and kinematics between a healthy cohort, individual's pre-HTO, and 12 months post-HTO.

After which, this chapter then describes the use of a MSK model in the form of the COMAK (Lenhart *et al.*, 2015; Smith *et al.*, 2018) simulation framework to answer the following:

3. Compare the magnitude and location of joint loading between a healthy cohort, individual's pre-HTO, and 12 months post-HTO.

4. Compare tibiofemoral joint contact area between a healthy cohort, individual's pre-HTO, and 12 months post-HTO.

The final analysis in Chapter 5 reports the findings for a sub-cohort of individuals undergoing HTO using the Cardiff Classifier technique to objectively determine the biomechanical changes following HTO to provide insight into surgical efficacy and relationship to patient reported outcome measures. The reason as to why the results are on a sub-cohort is that these findings were presented at the Orthopaedic Research Society (ORS) at the mid-point of this PhD journey. Future work aims to use the Cardiff Classifier on the full HTO cohort presented in this thesis. The Cardiff Classifier reports on the following:

5. The first objective was to use the Cardiff Classifier technique to identify the strongest discriminating features of mKOA (pre-HTO) vs non-pathological gait. This study also aimed to understand the relationship between change in the belief of mKOA (B(OA)) and the change in the Oxford Knee Score, as well as the change in the mechanical tibio-femoral angle pre-to-post HTO.

This chapter will then conclude with a summary of the key findings from the 3 different analysis and will provide a short commentary on the effectiveness of HTO from a biomechanical perspective.

5.1.1 Group demographics

Table 5-1 shows participant demographics and clinical measures for the three groups. The control group was significantly younger (38 (± 11) years) than both the pre-HTO cohort (51 (± 9) years)) and the post-HTO cohort (52 (± 9) years). The control group's mass (72 (±16) kg) was significantly less than the two patient groups ((pre-HTO (91 (±20) kg) and post-HTO (90 (±20) kg)). As indicated by a change in the mTFA from 8° (±4) to 1° (3) varus, surgery successfully realigned the lower limbs.

Table 5-1 Pre- to Post-HTO: Demographic and Clinical Characteristics

	Controls	Pre-HTO	Post-HTO	Controls vs pre-HTO	Controls vs post-HTO	Pre vs post-HTO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Number of knees	28	30	30			
Gender (M/F)	13/15	25/5	25/5			
Age, years	38.14 (11.09)	50.70 (8.71)	51.83 (8.79)	**.000†††**	**.000†††**	
Height, m	1.70 (.08)	1.75 (.11)	1.75 (.11)	**.018***	**.017**	
Mass, kg	71.84 (15.74)	90.57 (20.17)	89.92 (19.98)	**.000†††**	**.000†††**	.242
BMI, kg/m²	24.96 (4.36)	29.27 (5.04)	29.08 (4.93)	**.001†††**	**.001†††**	.302
KL Grade	n/a	6 KL2; 19 KL3;5 KL4	n/a			
mTFA (°)	n/a	7.75 (3.72) varus	.92 (2.82) varus (n=27)			**.000*****

Significant difference (p < 0.01) indicated by ** where parametric or †† where non-parametric tests used. mTFA = varus alignment calculated as the mechanical tibiofemoral angle (mTFA) from long leg weight bearing radiographs. Positive value for mTFA = varus. One patient did not return for their post-HTO x-ray.

5.2 Visual 3D

5.2.1 Spatial-temporal parameters

Table 5-2 outlines the spatial-temporal parameters for the control cohort, pre-HTO, and post-HTO groups. Compared to the control group, pre-surgery patients walked significantly slower, spent longer in stance, and adopted a significantly wider stance. Patients 12 months post-HTO remained significantly slower in gait speed, longer cycle time, stance time and step time as well as a wider stride width. This indicates that even though some spatiotemporal parameters improve as a result of surgery, they do not normalise to that of the control cohort.

Table 5-2 Pre- to Post-HTO: Spatial-Temporal Parameters

	NP Group	Pre-HTO NL Group	Post-HTO NL Group	Pre-HTO vs NP	Post-HTO vs NP	Pre-HTO vs post-HTO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Cycle time (s)	1.08 (0.08)	1.17 (0.15)	1.15 (0.11)	**0.003††**	**0.003****	0.280
Stance time (s)	0.65 (0.06)	0.73 (0.12)	0.71 (0.08)	**0.000††**	**0.001****	0.221
FPA at HS (+) = toe out (°)	15.69 (5.68)	16.28 (7.44)	16.99 (7.43)	0.542	0.463	0.315
Step length (m)	0.64 (0.07)	0.60 (0.10)	0.63 (0.07)	0.062	0.556	**0.001††**
Step time (s)	0.54 (0.04)	0.59 (0.07)	0.57 (0.05)	**0.003††**	**0.008****	0.174
Stride length (m)	1.29 (0.13)	1.22 (0.19)	1.26 (0.14)	0.087	0.351	**0.014***
Swing time (s)	0.43 (0.03)	0.44 (0.04)	0.44 (0.03)	0.297	0.301	0.838
Speed (m/s)	1.21 (0.16)	1.06 (0.23)	1.10 (0.16)	**0.008††**	**0.018***	**0.026†**
Stride width (m)	0.14 (0.03)	0.16 (0.04)	0.17 (0.04)	**0.048†**	**0.007††**	**0.034***

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. s = seconds; ° = degrees; m/s = metres per second; m = metres; ROM = range of motion. HS = heel strike.

5.2.2 Knee loading parameters

Table 5-3 and Table 5-4 below outline the external knee moments and knee adduction angular impulse (KAAI) parameters, respectively. The external knee adduction moment (EKAM) is a surrogate of medial compartment knee joint loading (Prodromos, Andriacchi and Galante, 1985) whilst KAAI has been proposed as a surrogate measure of medial compartment knee dynamic knee loading (Bhatnagar and Jenkyn, 2010). High measures for medial knee joint loading that were present pre-HTO were reduced following HTO.

High EKAM and KAAI measures pre-surgery were significantly reduced following HTO and the majority were normalised to that of the control cohort. Noticeably EKAM1 and EKAM2 were reduced from 3.1 %BW.h (1.12) pre-HTO to 2.1 %BW.h (0.88) 12 months post-HTO and 2.48 %BW.h (1.1) to 1.55 %BW.h (0.83), respectively. Figure 26 gives a visual outline of the mean group waveforms for the knee external moments for the three groups.

Table 5-3 Pre- to Post-HTO: Knee Loading - External Moments

External knee moments, %BW.h	**NP Group**	**Pre-HTO Group**	**Post-HTO NL Group**	**Pre-HTO vs NP**	**Post-HTO vs NP**	**Pre-HTO vs post-HTO**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) moment						
Maximum	2.11 (0.81)	3.19 (1.14)	2.03 (0.99)	**0.000****	0.737	**0.000††**
1st peak (1st half stance)	2.27 (0.65)	3.10 (1.12)	2.10 (0.88)	**0.001****	0.412	**0.000††**
2nd peak (2nd half stance)	1.50 (0.67)	2.48 (1.10)	1.55 (0.83)	**0.000****	0.789	**0.000****
Midstance	1.15 (0.49)	2.15 (0.83)	1.32 (0.64)	**0.000****	0.259	**0.000††**
Flexion (+) moment peak	3.62 (1.65)	2.87 (1.56)	2.66 (1.23)	0.079	**0.014****	0.404
Extension (−) moment peak	-2.45 (0.85)	-1.94 (0.83)	-1.84 (0.72)	**0.043†**	**0.001†††**	0.388
Internal (+) rotation moment peak	0.60 (0.37)	1.01 (0.48)	0.64 (0.36)	**0.001****	0.632	**0.000††**
External (−) rotation moment peak	-0.16 (0.08)	-0.14 (0.13)	-0.13 (0.08)	**0.050†**	0.122	0.614

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. %BW.h = % of body weight multiplied by height.

Table 5-4 Pre- to Post-HTO: Dynamic Knee Loading - Knee Adduction Angular Impulse

	NP Group	Pre-HTO Group	Post-HTO NL Group	Pre-HTO vs NP	Post-HTO vs NP	Pre-HTO vs post-HTO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) angular impulse, %BW.h.s						
Stance	0.74 (0.28)	1.33 (0.51)	0.81 (0.39)	**0.000****	0.398	**.000****
1st half stance	0.43 (0.14)	0.73 (0.25)	0.46 (0.20)	**0.000****	0.502	**.000****
2nd half stance	0.31 (0.16)	0.60 (0.27)	0.35 (0.21)	**0.000****	0.346	**.000****
0–16% stance	0.06 (0.03)	0.10 (0.06)	0.06 (0.04)	**0.000††**	0.699	**.000****
17%–midstance	0.36 (0.11)	0.61 (0.20)	0.39 (0.17)	**0.000****	0.466	**.000****
Midstance–83% stance	0.26 (0.13)	0.50 (0.22)	0.30 (0.17)	**0.000****	0.444	**.000****
84%-100% stance	0.04 (0.02)	0.07 (0.04)	0.04 (0.03)	**0.002****	0.787	**.000****
Abduction (−) angular impulse						
1st half stance	-0.02 (0.01)	-0.01 (0.01)	-0.02 (0.01)	**0.001††**	0.236	**0.004††**
2nd half stance	-0.02 (0.02)	-0.01 (0.01)	-0.01 (0.01)	**0.021†**	0.063	0.289

Significant difference (p < 0.01) indicated by ** where parametric or †† where non-parametric tests used. %BW.h.s = % of body weight multiplied by height per second.

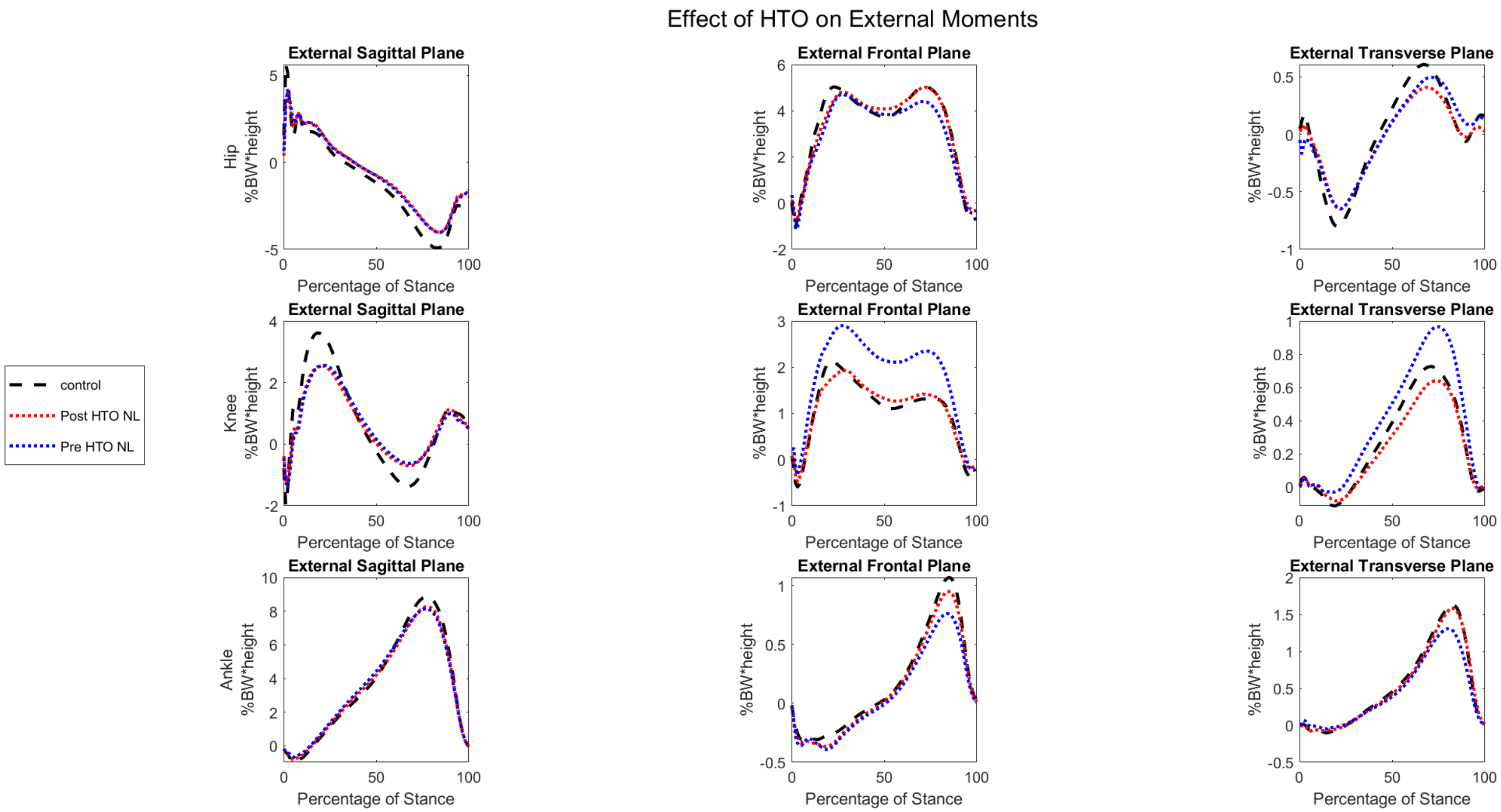


Figure 26 Visual 3D: Pre-and post-HTO group average external moments

Positive values represent external moments for knee flexion, adduction, and internal rotation moments.

5.2.3 Knee joint kinematics

Table 5-5 outlines the key discrete knee kinematics for the control group, pre-HTO, and post-HTO. Figure 27 gives a visual outline of the mean group waveforms for the knee kinematics for the three groups.

Pre-surgery, sagittal plane knee ROM was smaller than controls pre-HTO ($p<0.000$) and remained smaller post-HTO. Maximum knee flexion angle was significantly smaller than the control group ($p = 0.006$), which increased post-HTO and normalised to the control cohort. Maximum knee adduction angle was significantly larger in the pre-HTO group compared to the control group. Post-HTO, the maximum knee adduction angle was restored to the level of the controls ($p>0.000$).

Table 5-5 Pre- to Post-HTO: Knee Kinematic Parameters

	NP Group	Pre-HTO Group	Post-HTO NL Group	Pre-HTO vs NP	Post-HTO vs NP	Pre-HTO vs post-HTO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Knee sagittal plane ROM (°)	64.34 (3.50)	57.95 (6.05)	59.62 (5.48)	**0.000†††**	**0.000*****	0.131
Knee frontal plane ROM (°)	11.32 (3.76)	11.60 (4.53)	12.16 (3.59)	0.932	0.390	0.510
Knee transverse plane ROM (°)	16.46 (4.64)	16.05 (4.33)	16.97 (4.29)	0.731	0.665	0.303
Maximum knee flexion angle (°)	66.78 (4.79)	62.48 (6.44)	64.82 (5.12)	**0.006****	0.139	**0.018***
Maximum knee adduction angle (°)	5.79 (5.15)	10.03 (4.34)	6.54 (5.9)	**0.001****	0.609	**0.003****
Maximum knee internal angle (°)	2.63 (3.75)	2.19 (4.80)	1.91 (4.72)	0.702	0.526	0.741

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. ROM = range of motion; ° = degrees.

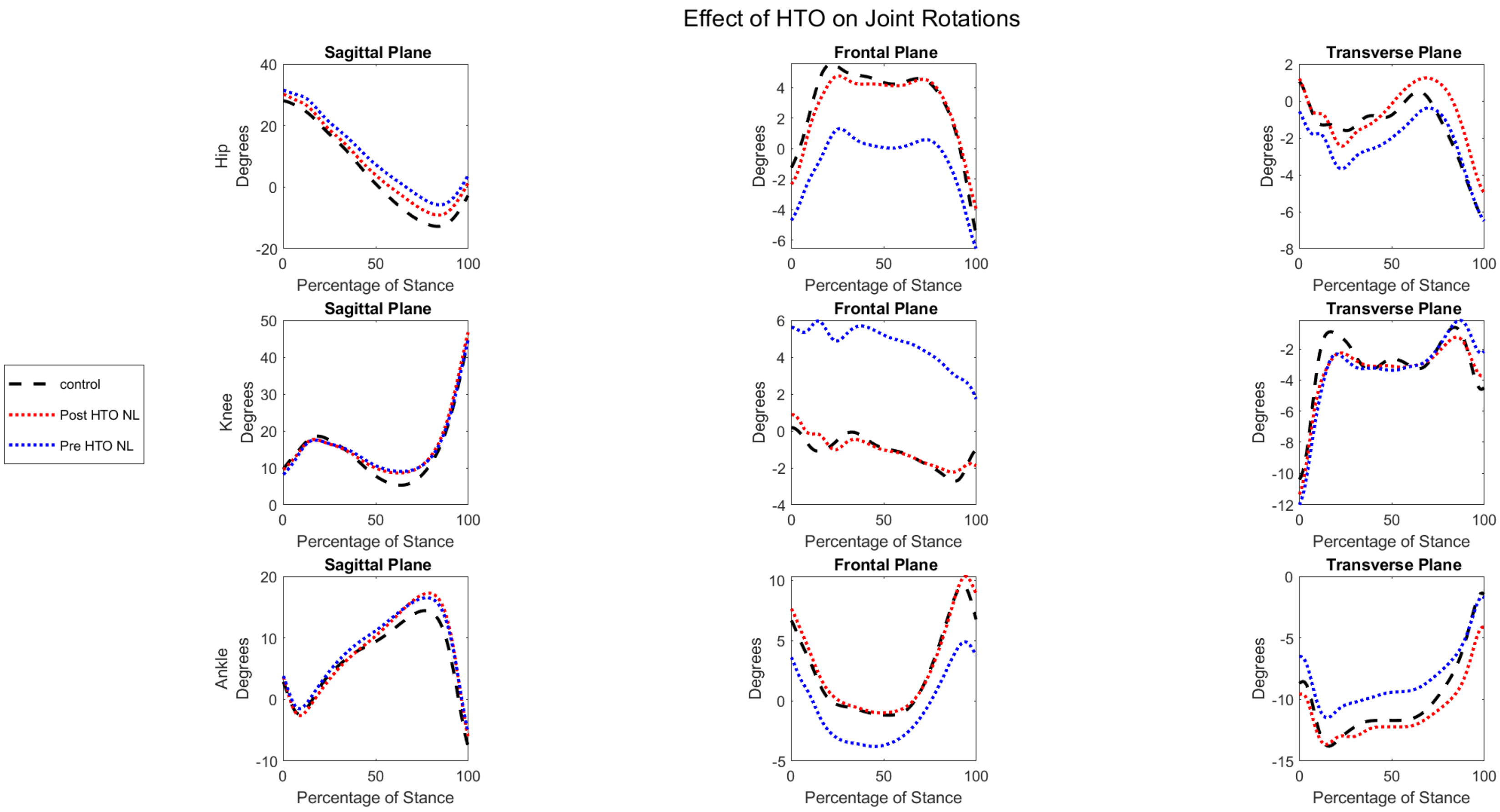


Figure 27 Visual 3D: Pre-and post-HTO group average joint rotations

Positive values represent knee flexion, adduction, and internal rotations.

5.2.4 External ankle moments parameters

Table 5-6 provides the external ankle moments during the whole of stance (heel-strike to toe-off), first half of stance (heel-strike to midstance), and the second half of stance (midstance to toe-off). Figure 26 gives a visual outline of the mean group waveforms for the ankle external moments for the three groups.

During the first half of stance, the control group had a higher peak plantarflexion moment compared to the pre-HTO group (1.03 %BW.h (0.28) vs 0.74 %BW.h (0.38), p = 0.002) and remained significantly lower than the control group 12-months post-HTO (1.03 %BW.h (0.28) vs 0.87 %BW.h (0.29), p = 0.048).

During the second half of stance the control group had a higher peak dorsiflexion (8.94 %BW.h (0.74) vs 8.25 %BW.h (1.22), p = 0.004), peak inversion and peak internal rotation moments compared to the pre-HTO group (Table 5-6). HTO resulted in a normalisation of the peak inversion and peak internal rotation moments; peak dorsiflexion remained significantly lower (8.94 %BW.h (0.74) vs 8.35 %BW.h (0.77), p = 0.004).

Table 5-6 Pre- to Post-HTO: External Ankle Moments Parameters

%BW.h	**NP Group**	**Pre-HTO Group**	**Post-HTO Group**	**Pre-HTO vs NP**	**Post-HTO vs NP**	**Pre-HTO vs post-HTO**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak dorsi-flexion (+) moment	6.78 (3.93)	8.25 (1.22)	8.01 (1.42)	0.594	0.501	0.781
Peak plantar-flexion (−) moment	-0.97 (0.36)	-0.74 (0.37)	-0.81 (0.36)	**0.025***	0.112	0.298
Peak inversion (+) moment	0.82 (0.54)	0.84 (0.48)	0.95 (0.53)	0.787	0.787	0.268
Peak eversion (-) moment	-0.41 (0.22)	-0.53 (0.51)	-0.42 (0.25)	0.638	0.890	0.245
Peak internal rotation (+) moment	1.25 (0.85)	1.37 (0.63)	1.57 (0.83)	0.763	0.740	0.096
Peak external rotation (−) moment	-0.17 (0.08)	-0.14 (0.10)	-0.15 (0.09)	0.239	0.261	0.572
0-50% (HS to midstance)						
Peak dorsi-flexion (+) moment	4.08 (0.99)	4.42 (0.76)	4.22 (0.93)	0.143	0.589	0.054
Peak plantar-flexion (−) moment	-1.03 (0.28)	-0.74 (0.38)	-0.87 (0.29)	**0.002****	**0.048***	**0.024†**
Peak inversion (+) moment	0.14 (0.14)	0.17 (0.21)	0.15 (0.20)	0.694	0.775	0.813
Peak eversion (-) moment	-0.43 (0.20)	-0.53 (0.51)	-0.49 (0.24)	0.994	0.282	0.992
Peak internal rotation (+) moment	0.47 (0.26)	0.42 (0.23)	0.44 (0.22)	0.417	0.552	0.650
Peak external rotation (−) moment	-0.18 (0.08)	-0.13 (0.11)	-0.15 (0.09)	0.100	0.248	0.390
50-100% (Midstance to toe-off)						
Peak dorsi-flexion (+) moment	8.94 (0.74)	8.25 (1.22)	8.35 (0.77)	**0.004††**	**0.004****	0.245
Peak plantar-flexion (−) moment	-0.03 (0.12)	-0.04 (0.10)	-0.06 (0.10)	0.734	0.264	0.095
Peak inversion (+) moment	1.11 (0.34)	0.84 (0.48)	1.00 (0.52)	**0.019***	0.091	0.098
Peak eversion (-) moment	-0.11 (0.14)	-0.19 (0.29)	-0.14 (0.13)	0.521	0.098	0.877
Peak internal rotation (+) moment	1.67 (0.53)	1.37 (0.63)	1.65 (0.81)	**0.006††**	0.206	**0.014***
Peak external rotation (−) moment	0.00 (0.06)	-0.01 (0.06)	0.01 (0.04)	0.969	0.426	0.094

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. %BW.h = % of body weight multiplied by height. HS = heel strike.

5.2.5 Ankle kinematics parameters

The pre-surgery group had significant differences in peak ankle inversion angle (10.97° (4.55) vs (7.62° (5.23), $p = 0.012$) and eversion angle (2.99° (4.27) vs (5.01° (2.79), $p = 0.012$). These differences were normalised 12 months post-HTO. The control group had a significantly larger peak ankle inversion angle whilst having a significantly smaller peak ankle eversion angle compared to the pre-HTO cohort ($p = 0.012$ for both). Again, these differences were normalised post-HTO.

Additionally, ankle ROM in the transverse plane was significantly higher for the control group compared to both the pre-HTO group ($p = 0.010$) and post-HTO ($p = 0.004$).

Table 5-7 Pre- to Post-HTO: Ankle Kinematics Parameters

	NP Group	Pre-HTO Group	Post-HTO Group	Pre-HTO vs NP	Post-HTO vs NP	Pre-HTO vs post-HTO
	Mean (std)	Mean (std)	Mean (std)	P value	P value	P value
Peak ankle dorsiflexion (+) angle (°)	14.97 (3.59)	16.95 (4.00)	17.67 (3.41)	0.054	**0.005****	0.240
Peak ankle plantarflexion (-) angle (°)	-10.31 (4.24)	-9.34 (7.64)	-9.92 (6.21)	0.549	0.782	0.861
Ankle sagittal ROM (°)	25.32 (4.63)	26.24 (7.20)	27.60 (6.59)	0.957	0.206	0.085
Peak ankle inversion (+) angle (°)	10.97 (4.55)	7.62 (5.23)	12.08 (5.84)	**0.012***	0.823	**0.000††**
Peak ankle eversion (-) angle (°)	-2.99 (4.27)	-5.01 (2.79)	-2.00 (2.83)	**0.012†**	0.391	**0.000****
Ankle frontal ROM (°)	13.98 (3.12)	12.62 (4.71)	14.11 (5.80)	0.075	0.492	0.417
Peak ankle internal rotation (+) angle (°)	-0.79 (6.30)	-0.83 (7.34)	-3.49 (7.08)	0.982	0.132	**0.011***
Peak ankle external rotation (-) angle (°)	-15.42 (6.28)	-12.84 (7.01)	-15.29 (6.53)	0.146	0.938	**0.010†**
Ankle transverse ROM (°)	14.64 (4.10)	11.97 (3.68)	11.84 (2.86)	**0.010††**	**0.004****	0.783

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. ROM = range of motion; ° = degrees.

5.2.6 External hip moments parameters

Table 5-8 and Figure 26 outline hip external moments. During the first half of stance, there were no significant differences between the three groups.

During the second half of stance, surgery resulted in significant reductions in both peak external hip abduction moment (0.86 %BW.h (0.55) vs 0.46 %BW.h (0.54), $p = 0.007$) and peak external hip internal rotation moment (0.64 %BW.h (0.24) vs 0.46 %BW.h (0.28), $p = 0.009$) when comparing the surgery group to the control group. When comparing pre-HTO to post-HTO, peak external hip adduction moment was increased (4.64 %BW.h (1.23) vs 5.16 %BW.h (0.84), $p = 0.003$), and peak external hip internal rotation moment was reduced (0.57 %BW.h (0.34) vs 0.46 %BW.h (0.28), $p = 0.026$).

Table 5-8 Pre- to Post-HTO: External Hip Moments Parameters

%BW.h	NP Group	Pre-HTO Group	Post-HTO Group	Pre-HTO vs NP	Post-HTO vs NP	Pre-HTO vs post-HTO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak external hip flexion (+) moment	6.31 (2.64)	4.95 (1.73)	4.70 (1.91)	**0.025***	**0.010****	0.586
Peak external hip extension (−) moment	-4.36 (1.70)	-4.31 (1.56)	-4.10 (1.12)	0.898	0.492	0.427
Peak external hip adduction (+) moment	5.09 (1.41)	5.20 (1.16)	5.23 (0.85)	0.742	0.639	0.829
Peak external hip abduction (-) moment	-1.65 (1.03)	-1.60 (1.08)	-1.18 (0.61)	0.811	**0.047†**	**0.032†**
Peak external hip internal rotation (+) moment	0.54 (0.30)	0.57 (0.33)	0.48 (0.22)	0.691	0.562	0.085
Peak external hip rotation (−) moment	-0.80 (0.43)	-0.73 (0.42)	-0.69 (0.29)	0.546	0.247	0.504
0-50% (HS to midstance)						
Peak external hip extension (−) moment	-1.25 (0.92)	-1.02 (1.03)	-0.96 (0.72)	0.389	0.142	0.861
Peak external hip adduction (+) moment	5.27 (1.03)	5.05 (1.14)	5.11 (0.81)	0.453	0.513	0.802
Peak external hip abduction (-) moment	-1.28 (1.29)	-1.37 (1.25)	-1.02 (0.80)	0.660	0.361	0.094
Peak external hip internal rotation (+) moment	0.34 (0.20)	0.25 (0.24)	0.27 (0.15)	0.058	0.180	0.669
Peak external hip rotation (−) moment	-0.86 (0.42)	-0.73 (0.42)	-0.71 (0.27)	0.250	0.116	0.658
50-100% (HS to midstance)						
Peak external hip adduction (+) moment	5.20 (1.17)	4.64 (1.23)	5.16 (0.84)	0.082	0.886	**0.003††**
Peak external hip abduction (-) moment	-0.86 (0.55)	-0.59 (0.50)	-0.46 (0.54)	0.055	**0.007****	0.127
Peak external hip internal rotation (+) moment	0.64 (0.24)	0.57 (0.34)	0.46 (0.28)	0.329	**0.009****	**0.026†**
Peak external hip rotation (−) moment	-0.11 (0.12)	-0.09 (0.19)	-0.11 (0.15)	0.303	0.573	0.658

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. %BW.h = % of body weight multiplied by height. HS = heel strike.

5.2.7 Hip kinematic parameters

Peak hip extension angle was significantly larger in the control group compared to the pre-HTO group (12.89° (5.64) vs 5.91° (8.32), $p < 0.000$), and was normalised post-HTO. Additionally, the control group had a significantly higher hip sagittal plane ROM (45.38° (5.06) vs 39.9° (6.19), $p = 0.001$), which also remained post-HTO, albeit hip sagittal plane ROM did increase when comparing pre-HTO to post-HTO, just not to that of the control group.

Peak hip adduction angle was significantly larger in the control group compared to the pre-HTO group (7.03° (3.41) vs 2.66° (3.66), $p < 0.000$); this was corrected post-HTO when compared to the control group ($p = 0.365$). Additionally, the control group had a significantly higher hip frontal plane ROM (15.69° (3.26) vs 12.68° (2.44), $p < 0.000$) which remained post-HTO.

Table 5-9 Pre- to Post-HTO: Hip Kinematics Parameters

	NP Group	Pre-HTO Group	Post-HTO Group	Pre-HTO vs NP	Post-HTO vs NP	Pre-HTO vs post-HTO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Peak hip flexion (+) angle (°)	32.47 (6.47)	34.01 (7.32)	32.69 (6.60)	0.401	0.900	0.393
Peak hip extension (-) angle (°)	-12.89 (5.64)	-5.91 (8.32)	-9.21 (8.56)	**0.000****	0.058	0.058
Hip sagittal ROM (°)	45.38 (5.06)	39.90 (6.19)	41.87 (5.31)	**0.001****	**0.002††**	**0.011***
Maximum hip adduction (= +) angle (°)	7.03 (3.41)	2.66 (3.66)	6.29 (4.25)	**0.000****	0.365	**0.000****
Maximum hip abduction (= -) angle (°)	-8.65 (3.63)	-10.01 (3.72)	-7.35 (4.64)	0.167	0.242	**0.020†**
Hip frontal ROM (°)	15.69 (3.26)	12.68 (2.44)	13.63 (3.72)	**0.000****	**0.029***	0.153
Maximum hip transverse (internal = +) angle (°)	4.71 (8.79)	3.72 (7.28)	5.77 (8.34)	0.640	0.640	0.169
Minimum hip transverse (+) angle (°)	-8.92 (8.76)	-9.36 (7.74)	-7.89 (8.21)	0.840	0.645	0.308
Hip transverse ROM (°)	13.63 (3.66)	13.10 (3.63)	13.64 (3.85)	0.578	0.999	0.445

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. ROM = range of motion; ° = degrees.

5.3 Concurrent Optimisation of Muscle and Secondary Kinematics

5.3.1 Section background

This section outlines the predictive internal joint loading pre- and post-HTO using the Concurrent Optimisation of Muscle and Secondary Kinematics (COMAK) framework.

5.3.1.1 Understanding tibiofemoral joint loading at first and second peak total knee contact force

The magnitude and timing of the first and second peak (FP and SP) of the resultant total tibiofemoral contact force was determined during the first and second half of the stance phase, respectively, as well as the minimum force during single leg support (MS). Each variable was determined for the total knee as well as for the medial and lateral condyles separately and were averaged over three trials. Furthermore, the point of application of the total knee, medial and lateral contact force expressed in the local reference frame of the tibia as well as the contact area at FP, SP and MS were analysed. Figure 28 outlines knee kinematics, moments, and internal contact forces for pre-, post-HTO and the control group.

5.3.2 Spatial-temporal parameters

Table 5-10 shows the gait speed measured within the COMAK analysis. Post-surgery, patients walked with a faster gait speed compared to pre-surgery (p = 0.020) but remained slower than the control group. Gait speed findings using the COMAK framework agree with the work outlined in the Visual 3D analysis.

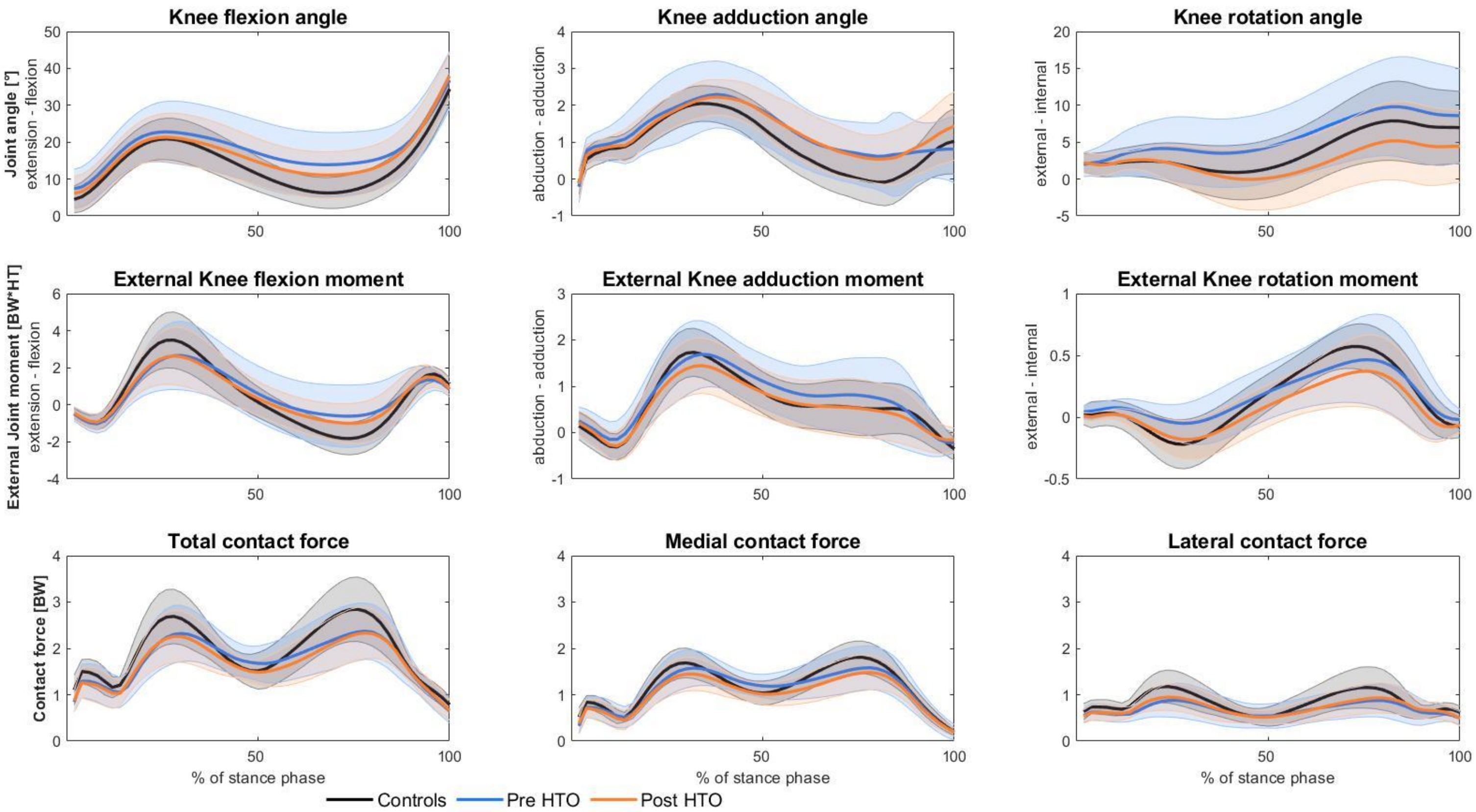


Figure 28 Pre- and post-HTO COMAK: Knee kinematics, moments, & contact forces

Table 5-10 Pre- to Post-HTO: COMAK Gait Speed

	Controls	Pre-HTO	Post-HTO	Controls vs pre-HTO	Controls vs post-HTO	Pre vs Post HTO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Gait speed (m/s)	1.26 (0.17)	1.10 (0.24)	1.15 (0.17)	**0.006****	**0.020***	**0.020†**

Significant difference (p < 0.01) indicated by ** where parametric or †† where non-parametric tests used. m/s = metres per second.

5.3.3 Knee loading

Medial knee compartment contact force at FP significantly reduced because of surgery (pre-HTO 1.61BW (0.36)) and post-HTO (1.46BW (0.36)). This decrease in medial compartment loading was met with no significant changes in FP lateral compartment contact force pre- and post-HTO. Mean pressure and maximum pressure for total, medial, lateral compartments at FP were not significantly different between the groups.

At MS, total knee contact forces decreased post-HTO and was normalised to the control group (1.25BW (0.24) control group vs 1.23BW (0.18) post-HTO, p = 0.859). Medial compartment contact forces significantly reduced pre- vs post-HTO (1.36BW (0.25) vs 1.23BW (0.18)) and normalised to the control group. Although total and medial compartment knee mean pressure and maximum pressure significantly reduced due to surgery, both metrics remained significantly higher compared to the control group (Table 5-11). There were no significant differences between all group comparisons for MS lateral compartment contact force, mean pressure and maximum pressure.

At SP, differences remained post-surgery when compared to the control group for total knee contact force (2.9BW (0.7) control group vs 2.43BW (0.54) post-HTO, p = 0.006). Total knee mean pressure differences that were present pre-HTO (5.98MPa (1.08)) when compared to the control group (5.32MPA (0.64)) were normalised post-HTO (5.77 MPa (1.11)). Medial compartment contact force whilst not significantly different between pre- and post-HTO, showed a significant difference when comparing the control group to the post-HTO group (1.83BW (0.35) control group vs 1.51BW (0.37) post-HTO). Post-HTO lateral compartment contact force increased and was normalised to the control group (1.2BW (0.44) control group vs 1.01BW (0.3) post-HTO, $p = 0.071$). Mean and maximum pressure were both significantly higher in the patient group pre- and post-HTO when compared to the control group and were not significantly changed due to surgery (Table 5-11).

Table 5-11 Pre- to Post-HTO: Internal Knee Joint Loading

	Controls	Pre-HTO	Post-HTO	Controls vs pre-HTO	Controls vs post-HTO	Pre vs Post HTO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total knee						
Contact force [BW]	2.74 (0.57)	2.41 (0.57)	2.33 (0.52)	**0.033***	**0.006****	0.418
Mean pressure [MPa]	5.63 (1.25)	6.11 (1.63)	5.77 (1.13)	0.349	0.671	0.144
Max pressure	12.92 (3.32)	14.19 (4.13)	13.22 (2.46)	0.248	0.501	0.094
Medial knee						
Contact force [BW]	1.69 (0.31)	1.61 (0.36)	1.46 (0.36)	0.362	**0.011***	**0.017***
Mean pressure [MPa]	5.76 (1.12)	6.44 (1.67)	5.95 (1.12)	0.073	0.526	**0.046***
Max pressure	12.21 (2.52)	13.36 (3.72)	12.69 (2.45)	0.171	0.469	0.184
Lateral knee						
Contact force [BW]	1.13 (0.37)	0.87 (0.38)	0.94 (0.27)	**0.011***	**0.027***	0.332
Mean pressure [MPa]	5.37 (1.62)	5.38 (2.12)	5.45 (1.43)	0.969	0.883	0.837
Max pressure	11.57 (3.76)	11.48 (4.71)	11.78 (3.12)	0.908	0.671	0.677
Midstance						
Total knee						
Contact force [BW]	1.25 (0.24)	1.36 (0.25)	1.23 (0.18)	**0.037***	0.859	**0.001††**
Mean pressure [MPa]	3.40 (0.36)	4.36 (0.81)	3.94 (0.58)	**0.000††**	**0.000****	**0.001****
Max pressure	7.62 (1.12)	9.97 (2.03)	8.93 (1.59)	**0.000††**	**0.001****	**0.000****
Medial knee						
Contact force [BW]	0.92 (0.22)	1.03 (0.26)	0.89 (0.20)	0.084	0.644	**0.001****
Mean pressure [MPa]	3.81 (0.58)	4.79 (0.91)	4.31 (0.70)	**0.000****	**0.004****	**0.001****
Max pressure	7.43 (1.22)	9.73 (2.07)	8.59 (1.51)	**0.000****	**0.002****	**0.001****
Lateral knee						
Contact force [BW]	0.36 (0.14)	0.36 (0.21)	0.37 (0.15)	0.920	0.660	0.911
Mean pressure [MPa]	2.59 (0.54)	3.03 (1.37)	3.04 (1.06)	0.255	0.070	0.530
Max pressure	5.56 (1.07)	6.29 (2.85)	6.42 (2.17)	0.333	0.065	0.772
Second peak						
Total knee						
Contact force [BW]	2.90 (0.70)	2.48 (0.62)	2.43 (0.54)	**0.018***	**0.006****	0.621

Mean pressure [MPa]	5.32 (0.64)	5.98 (1.08)	5.77 (1.11)	**0.022†**	0.147	0.201
Max pressure	12.70 (1.76)	14.12 (3.05)	13.34 (3.05)	0.126	0.811	0.058
Medial knee						
Contact force [BW]	1.83 (0.35)	1.63 (0.47)	1.51 (0.37)	0.075	**0.001****	0.100
Mean pressure [MPa]	5.91 (0.8)	6.42 (1.36)	6.14 (1.23)	0.084	0.386	0.263
Max pressure	12.61 (1.80)	13.14 (2.83)	12.65 (2.76)	0.393	0.605	0.333
Lateral knee						
Contact force [BW]	1.20 (0.44)	0.94 (0.37)	1.01 (0.30)	**0.021***	0.071	0.148
Mean pressure [MPa]	4.53 (0.73)	5.20 (1.48)	5.23 (1.39)	**0.045†**	**0.014†**	0.889
Max pressure	9.43 (1.48)	11.21 (3.61)	11.20 (3.30)	**0.021†**	**0.003††**	0.991

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. BW = body weight; MPa = megapascal.

5.3.3.1 Medial and lateral contact forces as a ratio of total contact force

When considering the medial compartment as a ratio of the total compartment (Table 5-12), there were significant reductions at FP (0.68 (0.01) vs 0.63 (0.08), p = 0.004) and at SP (0.65 (0.11) vs 0.62 (0.08), p = 0.022). As well as this, there was an increased lateral to total ratio at FP (0.35 (0.11) vs 0.4 (0.08), p = 0.007) and SP (0.38 (0.12) vs 0.41 (0.08), p = 0.047) of stance.

Table 5-12 Pre- to Post-HTO: Medial and Lateral Contact Force Ratios

	Controls	**Pre-HTO**	**Post-HTO**	**Controls vs pre-HTO**	**Controls vs post-HTO**	**Pre vs post-HTO**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
MED / TOTAL	0.63 (0.07)	0.68 (0.10)	0.63 (0.08)	**0.030***	0.920	**0.004††**
LAT / TOTAL	0.41 (0.07)	0.35 (0.11)	0.40 (0.08)	**0.030***	0.879	**0.007††**
Midstance						
MED / TOTAL	0.73 (0.09)	0.76 (0.14)	0.72 (0.12)	0.333	0.705	0.101
LAT / TOTAL	0.29 (0.10)	0.27 (0.15)	0.30 (0.12)	0.357	0.763	0.149
Second peak						
MED / TOTAL	0.64 (0.07)	0.65 (0.11)	0.62 (0.08)	0.573	0.302	**0.022†**
LAT / TOTAL	0.40 (0.07)	0.38 (0.12)	0.41 (0.08)	0.472	0.463	**0.047†**

Significant difference (p < 0.01) indicated by ** where parametric or †† where non-parametric tests used. MED = medial. LAT = lateral.

5.3.4 Knee loading location

A positive value in the medial-lateral direction indicates lateral point of contact from the centre of the tibial plateau. A positive value in the anterior-posterior direction indicates an anterior point of application from the centre of the tibial plateau.

At FP medial compartment knee, pre-HTO cohort COP was more lateral compared to the control group by a mean group difference of ~1mm (-17.19mm (1.29) control vs -18.11mm (1.5) pre-HTO, $p = 0.015$). Post-HTO, FP medial compartment knee point of contact was normalised when compared to the control cohort ($p = 0.845$). For the FP lateral compartment knee loading location, surgery resulted in a lateral shift of the centre point of application by a group mean difference of ~1.26mm ($p = 0.000$).

At MS, total knee centre point of application shifted more lateral when comparing pre- and post-HTO by a group mean difference of ~2.2mm ($p = 0.046$). The lateralisation in the point of contact was also observed in the medial compartment of the knee ($p = 0.036$).

At SP total knee compartment, HTO resulted in a more lateralised point of application compared to pre-surgery by a mean group difference of ~2mm ($p = 0.007$). There were no alterations due to surgery in the medial compartment; however, the point of application was lateralised in the lateral knee compartment.

Table 5-13 Pre- to Post-HTO: Point of Application of the Contact Forces

	Controls	Pre-HTO	Post-HTO	Controls vs Pre-HTO	Controls vs Post-HTO	Pre vs Post HTO
Position, mm	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total knee						
Anterior (+) / posterior (-)	-2.46 (2.08)	-2.10 (2.93)	-2.56 (2.20)	0.799	0.728	0.186
Lateral (+) / medial (-)	-2.27 (3.27)	-5.20 (4.40)	-2.44 (3.43)	**0.006****	0.845	**0.000††**
Medial knee						
Anterior (+) / posterior (-)	-1.01 (2.25)	-0.38 (3.24)	-1.18 (2.12)	0.492	0.811	0.075
Lateral (+) / medial (-)	-17.19 (1.29)	-18.11 (1.50)	-17.44 (1.53)	**0.015***	0.493	**0.016†**
Lateral knee						
Anterior (+) / posterior (-)	-4.67 (1.83)	-5.11 (2.46)	-4.65 (2.27)	0.170	0.847	0.223
Lateral (+) / medial (-)	20.59 (2.34)	19.63 (2.12)	20.89 (2.08)	0.107	0.614	**0.000****
Midstance						
Total knee						
Anterior (+) / posterior (-)	4.66 (1.86)	4.12 (2.24)	4.64 (1.96)	0.325	0.974	0.278
Lateral (+) / medial (-)	-5.84 (4.72)	-7.70 (7.04)	-5.47 (5.84)	0.246	0.791	**0.046***
Medial knee						
Anterior (+) / posterior (-)	6.63 (2.77)	6.41 (2.63)	6.34 (2.74)	0.755	0.688	0.899
Lateral (+) / medial (-)	-16.33 (1.53)	-17.10 (2.09)	-16.65 (1.82)	**0.036**	0.453	0.188
Lateral knee						
Anterior (+) / posterior (-)	-0.42 (1.55)	-2.16 (2.17)	0.32 (1.74)	**0.001****	0.090	**0.000††**
Lateral (+) / medial (-)	20.34 (2.43)	18.91 (3.49)	20.88 (3.22)	0.077	0.471	**0.003****
Second peak						
Total knee						
Anterior (+) / posterior (-)	5.70 (1.64)	4.45 (3.31)	4.94 (3.22)	0.349	0.717	0.436
Lateral (+) / medial (-)	-1.39 (2.90)	-2.69 (5.07)	-0.68 (3.24)	0.065	0.417	**0.007††**
Medial knee						
Anterior (+) / posterior (-)	9.56 (1.83)	8.13 (3.34)	8.26 (3.68)	0.180	0.201	0.847
Lateral (+) / medial (-)	-14.59 (1.31)	-15.85 (1.74)	-15.46 (1.27)	**0.003****	**0.013***	0.140
Lateral knee						
Anterior (+) / posterior (-)	-0.89 (2.43)	-2.59 (3.34)	-0.39 (2.73)	**0.030†**	0.466	**0.000****

Lateral (+) / medial (-)	19.74 (2.84)	19.49 (3.34)	21.55 (2.90)	0.761	**0.020***	**0.000****

Significant difference (p < 0.01) indicated by ** where parametric or †† where non-parametric tests used. mm = millimetres. +X = anterior; +Z = lateral. COP = centre of pressure.

5.3.5 Knee contact area

The only significant changes between pre- and post-HTO contact area was at the FP where medial compartment significantly decreased, and the lateral compartment contact area significantly increased due to surgery (Table 5-14).

Table 5-14 Pre- to Post-HTO: Tibiofemoral Contact Area

	Controls	**Pre-HTO**	**Post-HTO**	**Controls vs pre-HTO**	**Controls vs post-HTO**	**Pre vs Post-HTO**
mm^2	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total	352.35 (39.00)	362.26 (49.94)	362.97 (49.26)	0.357	0.453	0.923
Medial	206.80 (22.63)	223.16 (29.08)	213.55 (32.49)	**0.012†**	0.472	**0.030***
Lateral	145.55 (20.95)	139.10 (29.50)	149.43 (21.45)	0.638	0.242	**0.030†**
Midstance						
Total	262.26 (34.10)	285.17 (55.54)	282.31 (46.27)	0.111	0.080	0.718
Medial	167.14 (24.46)	188.99 (36.33)	180.35 (29.87)	**0.010***	0.108	0.091
Lateral	95.12 (19.62)	96.18 (35.19)	101.96 (27.28)	0.847	0.111	0.345
Second peak						
Total	399.23 (90.44)	381.58 (90.11)	387.36 (90.55)	0.444	0.620	0.702
Medial	219.07 (41.52)	222.87 (49.90)	217.24 (46.55)	0.755	0.763	0.515
Lateral	180.16 (52.60)	158.71 (47.48)	170.12 (46.05)	0.108	0.442	0.132

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. mm^2 = millimetres squared.

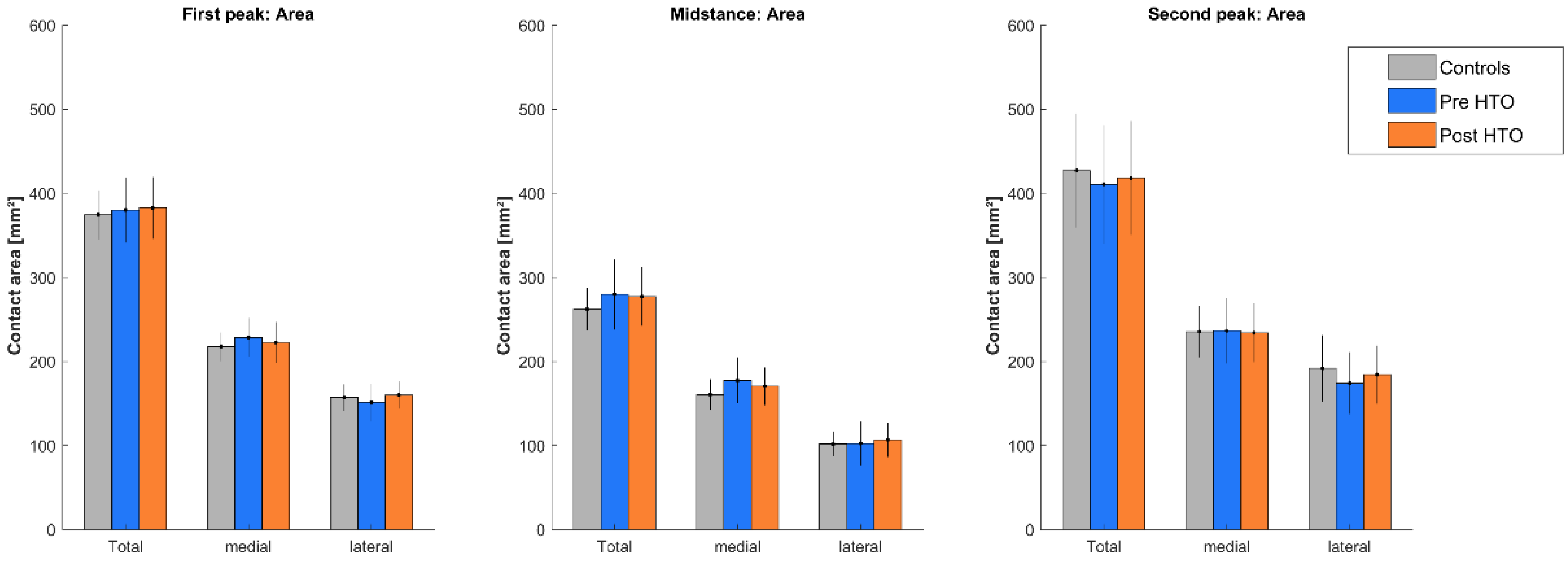


Figure 29 Knee contact area

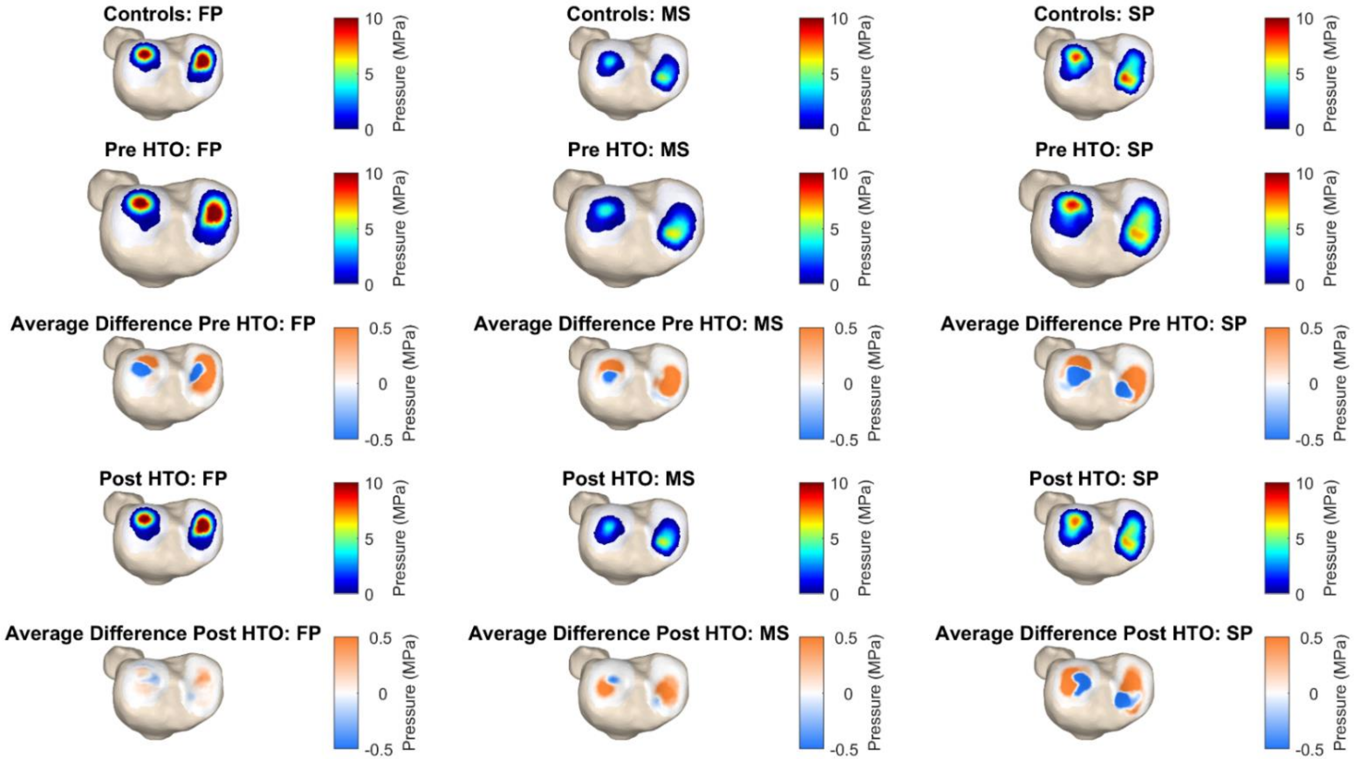


Figure 30 Pre-and post-HTO contact pressure distribution on the tibia

Average contact pressure patterns at first peak, midstance and second peak for the control group and the patients pre- and post-HTO. Furthermore, the average difference between the pressure pattern in patients and the healthy control pressure pattern is shown. Orange indicates more loading in the patient on that specific location, blue indicates decreased loading compared to the controls.

5.4 Waveform analysis using Principal Component Analysis and the Cardiff Classifier to better understand biomechanical factors affecting varus deformity of the knee

5.4.1 Key findings

Healthy controls and patients' pre- and post-HTO demographic, anthropometric data and OKS are reported in Table 5-15. The HTO cohort was significantly older and had a higher mass than the control cohort participants. Gait velocity increased significantly following surgery but remained significantly lower than the control group following HTO. The Cardiff Classifier was able to correctly classify between the control cohort and OA gait biomechanics for 34 out of the 42 cases (81% accuracy).

Table 5-15 Group Demographics for the Cardiff Classifier Study

Group	**Healthy Controls (n=20)**	**Pre-operative patients (n=22)**	**Post-operative patients (n=22)**	**P-value**	
	Mean (std)	**Mean (std)**	**Mean (std)**	**Pre-Post**	**Pre-NP**
Age	33.60 (10.80)	51.50 (6.40)	52.60 (7.50)	**<0.001**	**<0.001**
Height (m)	1.70 (0.80)	1.80 (0.10)	1.80 (0.10)	ns	**<0.01**
Body mass (kg)	69.88 (13.40)	92.50 (16.20)	91.30 (16.70)	ns	**<0.05**
K/L score	n/a	3 (0.60)	n/a	n/a	n/a
mTFA (°)	n/a	7.70 (3.60) varus	0.80 (2.60) varus	**<0.001**	n/a
Oxford knee score	47.74 (.93)	26.10 (9.10)	37.60 (6.60)	**<0.001**	**<0.001**
Gait speed (m/s)	1.19 (0.14)	1.04 (0.24)	1.09 (0.18)	**<0.05**	**<0.001**

mTFA = varus alignment calculated as the mechanical tibiofemoral angle (mTFA) from long leg weight bearing radiographs.

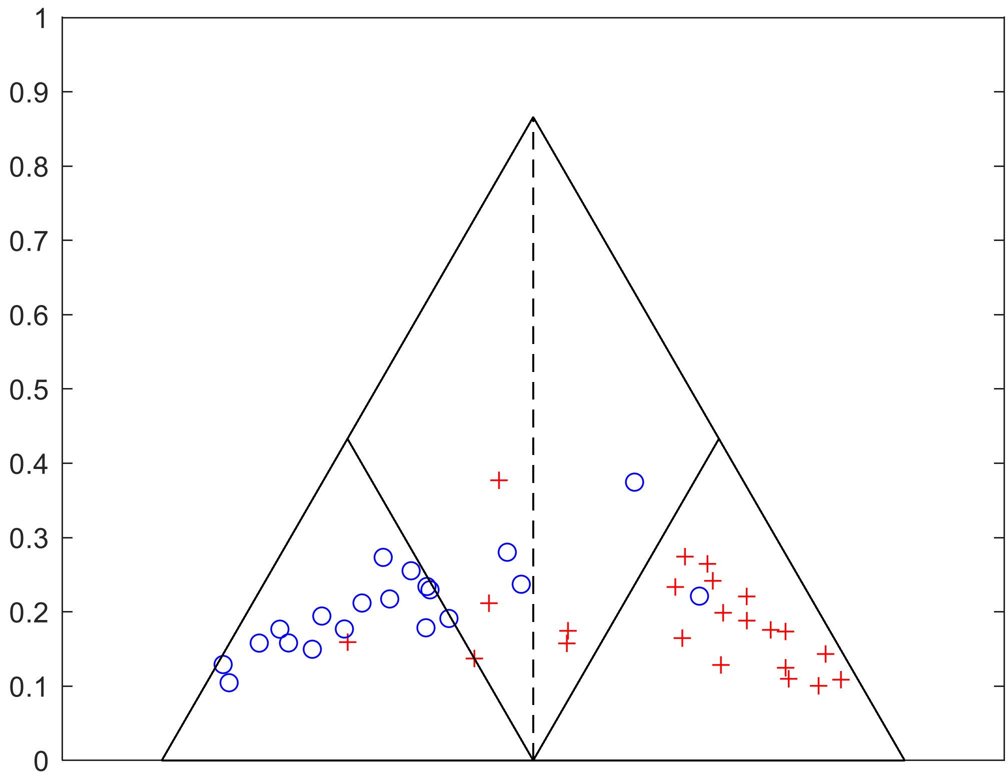


Figure 31 Simplex plot of the classification

Simplex plot of the classification of the 20 control participants (blue circle) and 22 pre-HTO (red cross) participants who were used to train the Cardiff Classifier. The three vertices represent the points where belief of non-pathological function B(NP), belief of osteoarthritic function B(OA) and uncertainty, U is equal to 1 (or 100%). The decision boundary where B(OA) = B(NP) is shown as a dashed line. The boundaries where B(OA) = 0.5 and B(NP)= 0.5 are shown as interior solid lines.

15 PCs were retained for analysis, the accuracy in discriminating between controls and pre-HTO, the percentage of variance represented, and the interpretation of each reconstructed PC are reported in

Parameters			**Accuracy (%)**	**Variance represented (%)**	**Low PC Interpretation**
Kinematics – operative limb					
Hip	Hip flexion	PC2	74	13%	Reduced hip ROM during stance phase.

	Hip adduction	PC1	74	60%	Reduced hip magnitude throughout the gait cycle.
Knee	Knee adduction	PC1	74	70%	Increased knee magnitude throughout the gait cycle.
Ankle	Ankle inversion	PC1	74	66%	Reduced ankle magnitude throughout the gait cycle.
Kinetics – operative limb					
Hip	Hip transverse moment	PC2	81	14%	Increase in the swing phase of the gait cycle.
	Hip transverse moment	PC1	71	42%	Increased magnitude in the first 50% of the gait cycle.
Knee	Knee adduction moment	PC1	74	46%	Increased magnitude in the first 60% of the gait cycle.
	Knee flexion moment	PC2	69	19%	Decreased magnitude between 10-20% of the gait cycle, increased magnitude between 20-50% of the gait cycle.
	Knee internal rotation moment	PC1	69	41%	Increased magnitude of internal rotation moment during the stance phase of gait.
Ankle	Ankle sagittal power	PC2	71	14%	Reduced power output between 40-60% of the gait cycle.
	Ankle transverse power	PC1	71	16%	Difficult to give interpretation.
		PC3	69	11%	Phase shift.
	Ankle adduction power	PC1	69	27%	First 50% of gait cycle.
	Ankle sagittal moment	PC2	69	22%	Difficult to give interpretation.
Ground reaction force					
	Medial-lateral GRF	PC3	69	11%	Phase shift.

Table 5-16. Nine of the top 15 ranked discriminating features were parameters of the hip and ankle.

Of concern to this analysis is the classification of all the participants involved. 34 out of the 42 included participants were correctly classified. Four controls and four OA patients were misclassified. These misclassifications are explored in further detail at the end of this section.

There was no significant correlation between the change in B(OA) and the change in OKS [r = -.41, n = 20, p = .075]. There was also no significant correlation between the change in B(OA) and the change in mTFA [r = .18, n = 19, p = .632]. There was a correlation between the change in OKS and the change in mTFA pre-to-post HTO [r = -.669, n = 17, p = .003].

Parameters			**Accuracy (%)**	**Variance represented (%)**	**Low PC Interpretation**
Kinematics – operative limb					
Hip	Hip flexion	PC2	74	13%	Reduced hip ROM during stance phase.
	Hip adduction	PC1	74	60%	Reduced hip magnitude throughout the gait cycle.
Knee	Knee adduction	PC1	74	70%	Increased knee magnitude throughout the gait cycle.
Ankle	Ankle inversion	PC1	74	66%	Reduced ankle magnitude throughout the gait cycle.
Kinetics – operative limb					
Hip	Hip transverse moment	PC2	81	14%	Increase in the swing phase of the gait cycle.
	Hip transverse moment	PC1	71	42%	Increased magnitude in the first 50% of the gait cycle.
Knee	Knee adduction moment	PC1	74	46%	Increased magnitude in the first 60% of the gait cycle.
	Knee flexion moment	PC2	69	19%	Decreased magnitude between 10-20% of the gait cycle, increased magnitude between 20-50% of the gait cycle.
	Knee internal rotation moment	PC1	69	41%	Increased magnitude of internal rotation moment during the stance phase of gait.
Ankle	Ankle sagittal power	PC2	71	14%	Reduced power output between 40-60% of the gait cycle.
	Ankle transverse power	PC1	71	16%	Difficult to give interpretation.
		PC3	69	11%	Phase shift.

	Ankle adduction power	PC1	69	27%	First 50% of gait cycle.
	Ankle sagittal moment	PC2	69	22%	Difficult to give interpretation.
Ground reaction force					
	Medial-lateral GRF	PC3	69	11%	Phase shift.

Table 5-16 The Interpretation of the Top 15 Biomechanical Features Inputted into the Cardiff Classifier

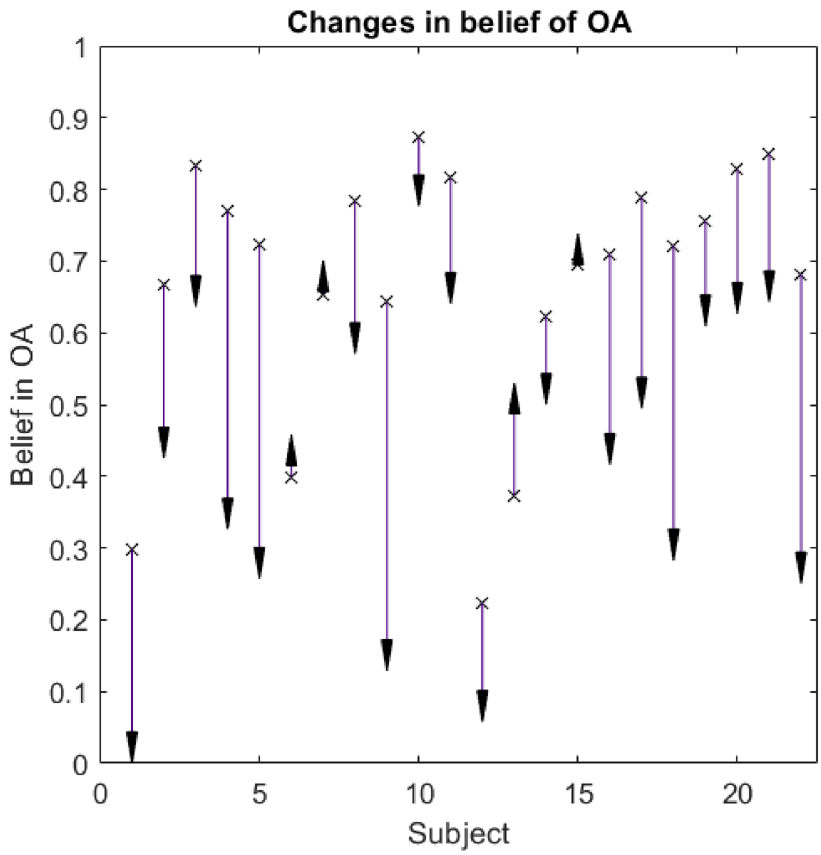


Figure 32 Change in classification of the 22 HTO participant between pre- and post-HTO

The belief system ranged from 1 being B(OA) and 0 meaning B(NP).

The most discriminatory biomechanical gait features of varus patients with mKOA appear to be in the transverse and sagittal plane of the hip. Post-operative patients with the largest correction change in mTFA also reported the greatest improvement in OKS, however this did not correlate with improvements in biomechanical function [reduction in B(OA)].

5.4.1.1 Misclassified participants

The four misclassified patients pre-HTO that were classified as 'healthy' all had a fast gait speed for pre-HTO functionality (mean gait speeds of 1 x 1.5 m/s, 2 x 1.2 m/s, and 1.3 m/s). This consideration may indicate that there are some patients pre-HTO which can walk at a faster gait speed to their healthy counterparts. This would show that the pre-HTO cohort is a heterogenic group and vary considerably in function pre-surgery. This is in comparison of a pre-HTO group mean value of 1.04 m/s. These four patients had a range of OKS results (43, 15, 28 and 23 out of 48).

The non-pathological healthy group had 4 individuals who were misclassified. These four participants had the following gait speeds: 1 m/s, 1.1 m/s, 1.2 m/s and 1.1 m/s. Apart from the participant with the mean gait speed of 1.2 m/s, these results would suggest that these 'healthy control' participants had lower gait speed than that of the control cohort which was 1.2 m/s.

5.4.1.2 Section summary

Post-operative patients who underwent the largest correction change in mTFA also reported the greatest improvement in OKS, however this did not correlate with improvements in biomechanical function [reduction in B(OA)]. HTO surgery reduced the belief in OA in 20 out of 22 of the included patients, indicating biomechanical improvement occurs due to realignment surgery.

The biomechanical significance of the highest scoring discriminating variable of an increased hip transverse moment during the swing phase of the gait cycle should be ignored for these findings due to the lack of clinical relevance and significance. Future work that develops on from this thesis should only focus on joint moments during the stance phase of the gait when the joints are under the greatest loads, and not during the swing phase.

Interestingly, after discarding the first discriminatory PC, the findings from this exploratory study are also observed in Whatling et al. (2019) whereby patients pre-HTO compared to the control group have a reduced hip flexion rotation ROM during stance phase, a reduced hip adduction rotation magnitude throughout the gait cycle, an increased knee adduction rotation magnitude throughout the gait cycle, a reduced ankle inversion rotation magnitude throughout the gait cycle and an increased knee adduction moment magnitude in the first 60% of the gait cycle.

In this thesis, HTO reduced the belief in OA in 20 out of 22 of the included patients, indicating biomechanical improvement occurs due to realignment surgery. This study has introduced waveform analysis, in the form of PCA and classification of changes using the Dempster–Shafer theory (DST) in trying to better understand the biomechanical changes pre-to-post HTO which can have direct influences on clinical decision making.

5.5 Kinematic and moment comparisons in hip, knee, ankle joints between Visual 3D modelling and Concurrent Optimisation of Muscle Activations and Kinematics framework

This thesis has used two different biomechanical models to obtain kinematic and joint moment data. The COMAK model then goes a step further in predicting internal tibiofemoral joint contact forces and pressures. This section outlines a visual comparison between the two approaches for joint kinematics and moments and aims to provide a descriptive analysis to outline the possible reasons for these visual differences.

5.5.1 Kinematics

5.5.1.1 Visual 3D kinematic approach

Visual 3D models are based on a linked set of rigid segments. The traditional Visual 3D model that was used in this thesis was a 6 DOF model that assumes that segments were implicitly linked by the motion capture data and the joints were modelled with 6 DOF, e.g., all segments were treated as if they were independent. The mapping of motion capture markers to 6 DOF segments is a matter of tracking a set of markers that are linked rigidly to the segment. This least square solution requires the specification of the segment coordinate system and the tracking of the pose (position and orientation) of a segment. Essentially this is a straightforward pattern recognition; the pattern (configuration) of the tracking markers is specified in a standing trial, and this pattern is fitted to the marker configuration in each frame of motion capture data.

5.5.1.2 Concurrent Optimisation of Muscle Activations and Kinematics framework kinematic approach

An alternative to the 6 DOF solution is to define joints, e.g., explicitly state which segments are connected by a joint, and to specify the properties of all joints. The targets used to track the segments are often subject to measurement error and soft tissue artefact, motion about some of the degrees of freedom may be much larger than the motion that would be realistically possible. Lu and O'Connor (1999) described a global optimisation process where physically realistic joint constraints can be added to the model to minimise the effect of the soft tissue and measurement error. Lu and O'Connor termed this process global optimisation while others inside the biomechanics community prefer the term inverse kinematics.

5.5.1.3 Summarising key differences between the two approaches

The difference between the traditional 6 DOF model and the inverse kinematics model is that constraints can be added between segments that restrict the relative motion between the segments. This is accomplished by creating one or more inverse kinematic chains. inverse kinematics is the process of determining the parameters of a jointed flexible object (a kinematic chain) to achieve a desired pose. An inverse kinematics solution is dependent on the choice of hierarchical model because the task is to identify an articulated figure consisting of a set of rigid segments connected with joints. Varying angles of the joints yield an indefinite number of configurations, so in the general case there is no analytic solution.

The author of this thesis created a simple MATLAB script to input waveform data of the stance phase of gait to visualise the two approaches on 16 patients pre-HTO (Figure 33) and then on 14 health controls (Figure 34). The reason on graphing the two different groups is to establish whether patients with varus deformity have a larger visual disparity between the two methods.

The main observations from these comparisons are in the magnitude differences in knee frontal plane kinematics. As expected, the Visual 3D pipeline, with using a 6 DOF kinematic approach, has a larger range of degrees compared to the inverse kinematic approach in the COMAK pipeline. This is not surprising as the sole purpose of an inverse kinematic approach that applies constraints is to limit the frontal plane movement to within 'realistic' ranges.

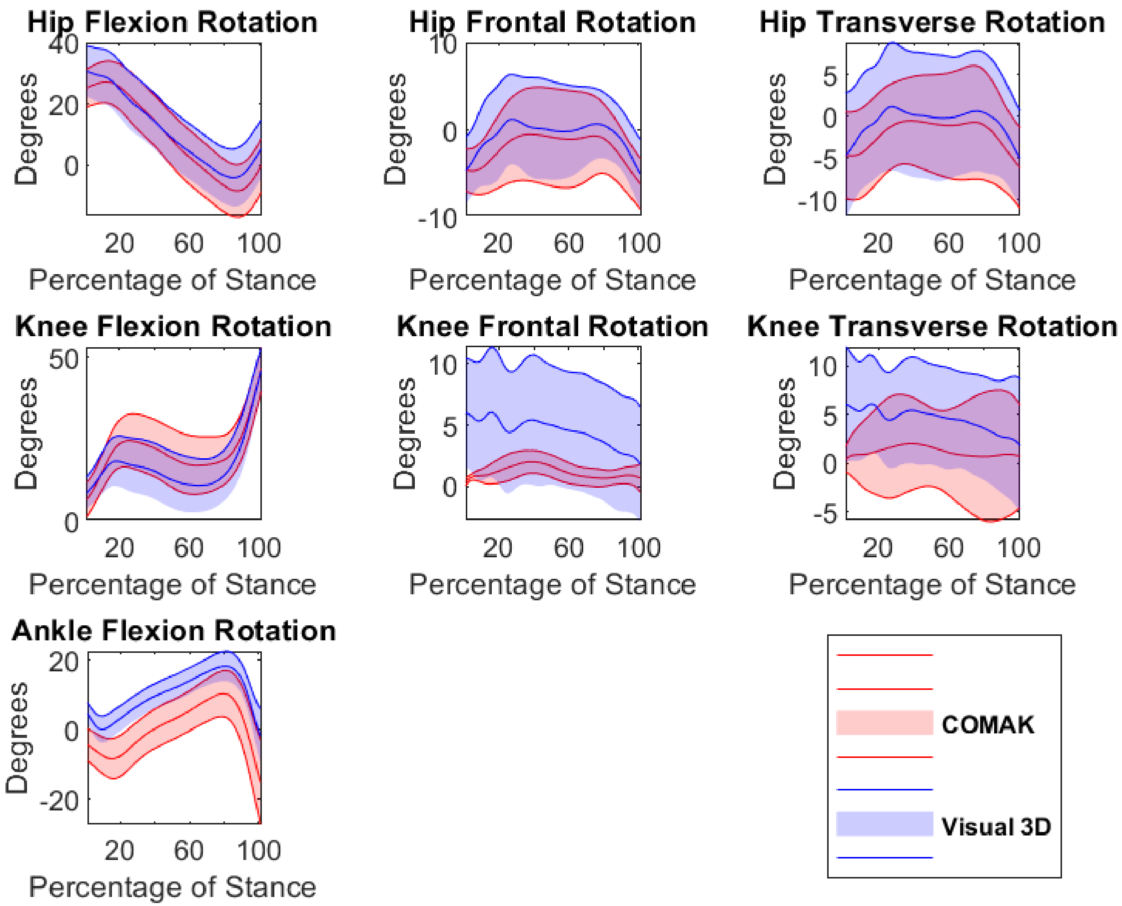


Figure 33 Comparing kinematics pre-HTO: Visual 3D pipeline vs COMAK

Positive values represent knee flexion, adduction, and internal rotations.

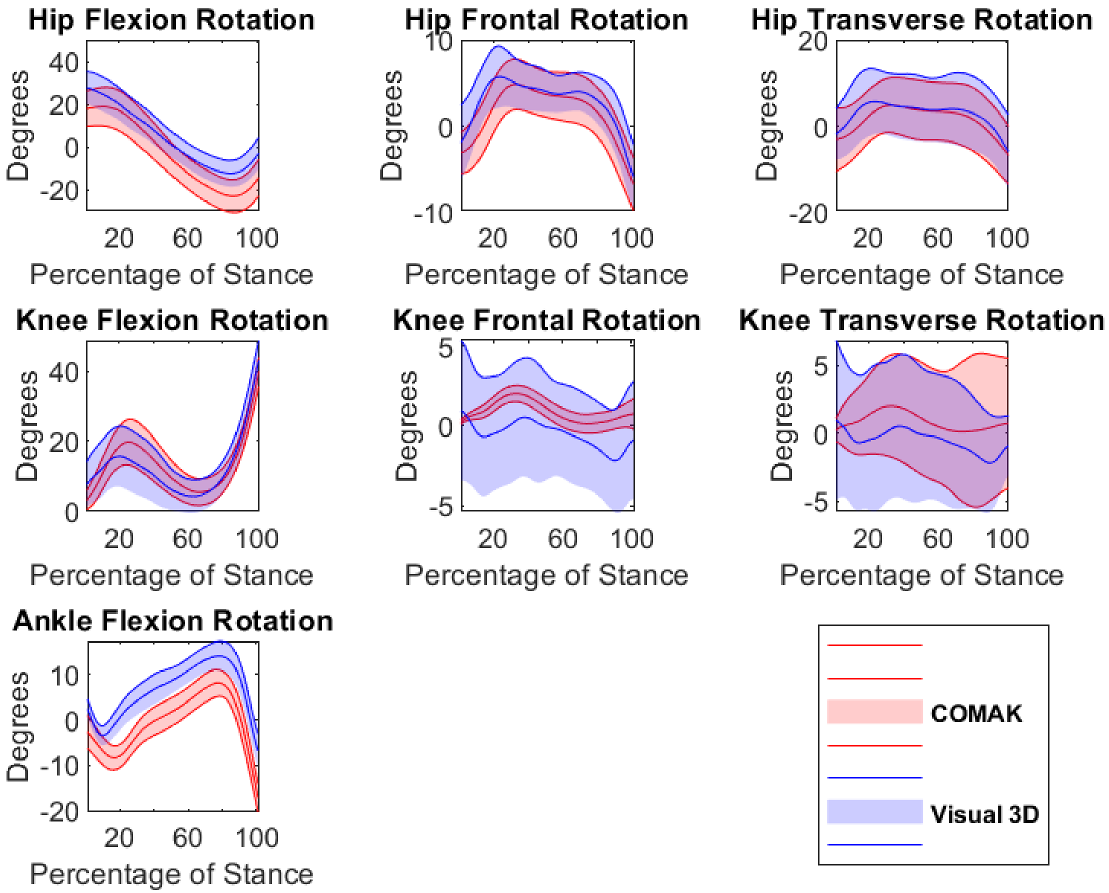


Figure 34 Comparing kinematics for a healthy cohort: Visual 3D pipeline vs COMAK

Positive values represent knee flexion, adduction, and internal rotations.

The differences between the two approaches for the kinematic data had a direct influence on the differences in joint moments. The same comparison but with joint moments are shown in Figure 35 and Figure 36 below. The generic COMAK model assumes a fixed HKA alignment, and the Visual 3D model allows knee adduction DOF, so given the same input marker data the model's knee could be located more laterally in Visual 3D versus COMAK (*inverse kinematics*). Thus, the higher DOF models, e.g., Visual 3D, allows for a larger frontal plane lever arm of the GRF vs knee joint centre, which yields a larger EKAM. It is therefore important for the reader of this thesis to remember to keep the conclusions made by both approaches separate. The graphs presented here should give the reader an appreciation that the two approaches that have been undertaken should be considered independently.

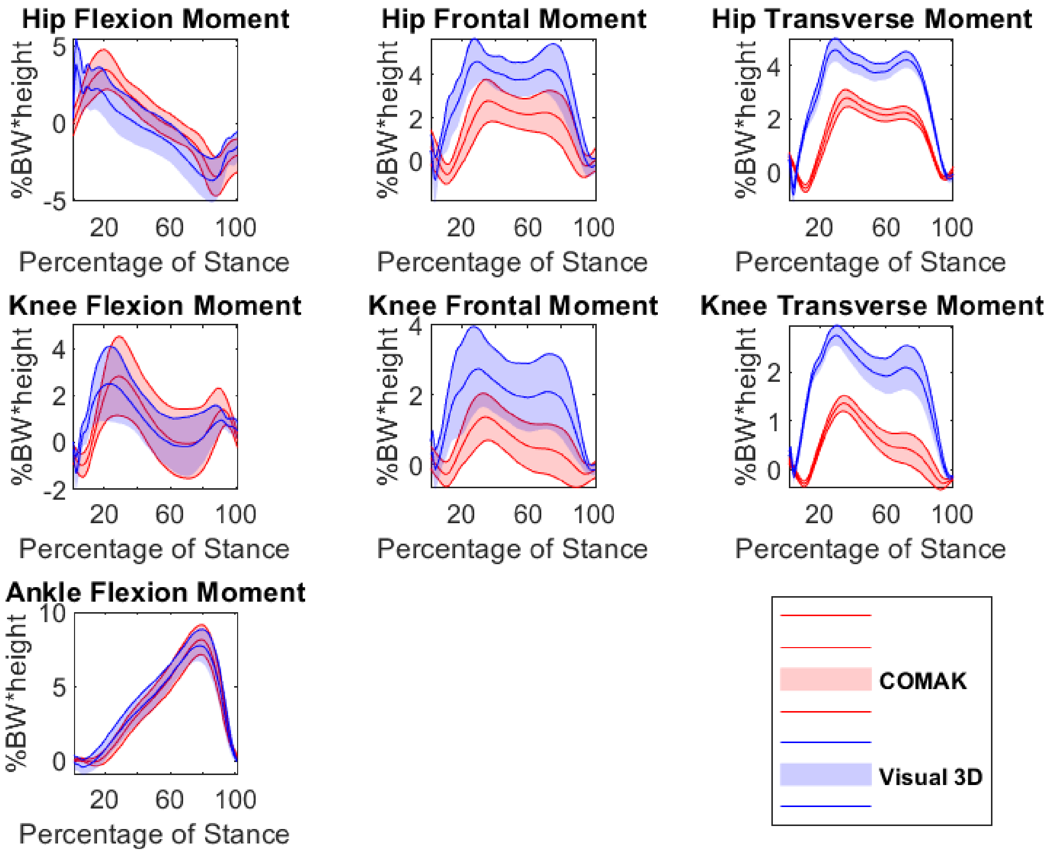


Figure 35 Comparing joint moments pre-HTO: Visual 3D pipeline vs COMAK

Positive values represent external moments for knee flexion, adduction, and internal rotation moments.

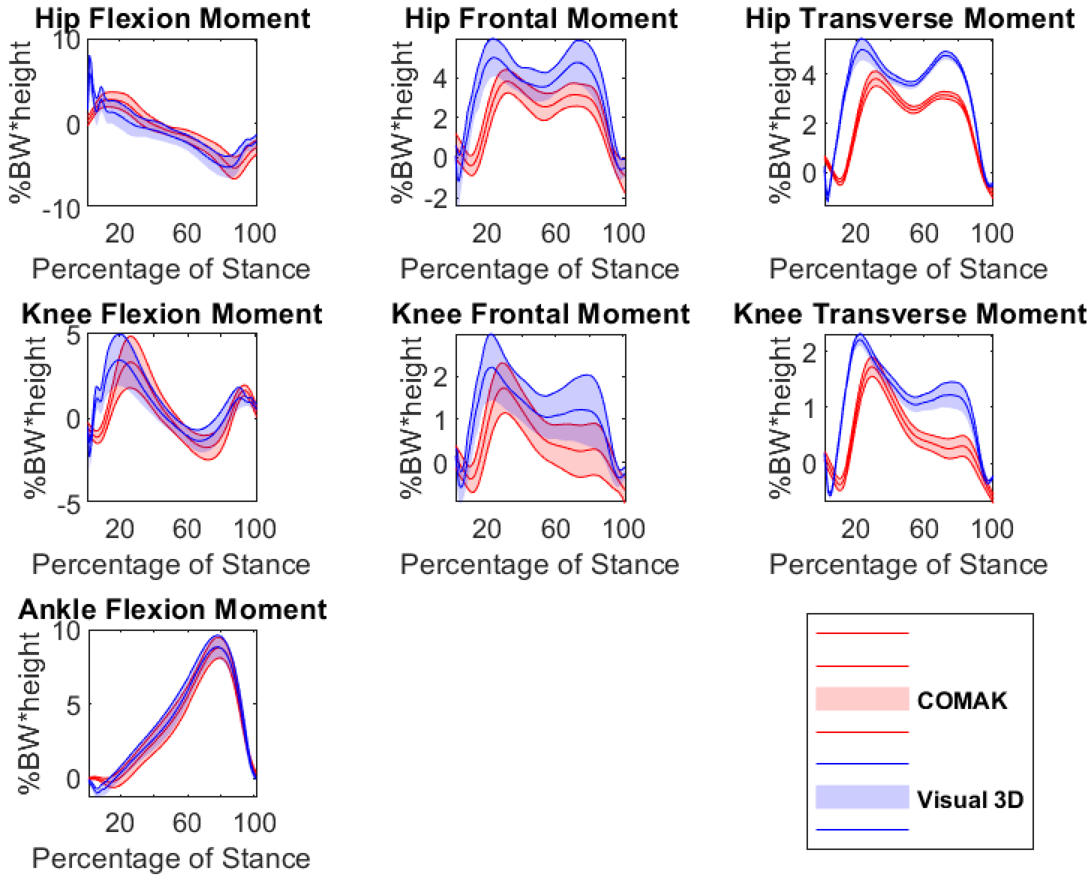


Figure 36 Comparing joint moments for a healthy cohort: Visual 3D pipeline vs COMAK

Positive values represent external moments for knee flexion, adduction, and internal rotation moments.

5.6 Chapter 5 summary

5.6.1 Visual 3D modelling summary

HTO altered many spatiotemporal parameters, including gait speed. Fundamentally, HTO reduced medial compartment knee joint loading. More specifically, EKAM1 and EKAM2, surrogate measurements of medial compartment knee loading, were significantly reduced at 12 months post-HTO. As well as EKAM, KAAI (a measurement of dynamic loading) were also normalised to the control group 12 months post HTO.

During the first half of stance, the control group had a higher peak external plantarflexion moment compared to the pre-HTO group and remained significantly higher when comparing the control group to post-HTO. The clinical significance of this difference cannot be stated within the current body of evidence; however, it does appear to show that surgery brings this metric towards that of the control group. This would be a desirable outcome of surgery. When comparing the control group to post-HTO, the peak external inversion and peak external internal rotation moments were normalised. The interesting take-home message here is that the frontal and transverse planes are ‘corrected’ 12-months post-HTO and that in the sagittal plane, corrective surgery directs discrete metrics towards that of the control group without reaching statistical significance.

When comparing pre-HTO to post-HTO, peak external hip adduction moment was increased, and peak external hip internal rotation moment was reduced. It is difficult to determine if the identified biomechanical changes were involved in the development of the disease, a response to degenerative changes in the joint and soft tissue, or a compensatory mechanism to the disease process (Astephen *et al.*, 2008).

5.6.2 Concurrent Optimisation of Muscle Activations and Kinematics modelling summary

The most significant finding from this chapter is that HTO reduces medial knee contact force at FP and MS whilst not significantly increasing the lateral knee compartment contact forces. At SP, surgery resulted in a significant decrease in medial compartment contact force when compared to the control group whilst no significant increase in the lateral compartment contact force. This reduction is observed following surgery despite a significant increase in walking speed ($p = 0.020$) which is typically known to increase joint loading.

In addition to this, surgery significantly lateralises the COP of the total knee at FP, SP and at MS. At MS, mean pressure and maximum pressure for total knee and medial compartment knee were significantly reduced due to remained elevated compared to the control group.Click or tap here to enter text.Click or tap here to enter text.Click or tap here to enter text.Click or tap here to enter text.Click or tap here to enter text. Findings from this thesis support the use of HTO to slow down cartilage degeneration/OA progression.

Surgery successfully reduced joint loading parameters, changed the point of force application, and normalised the contact area on the medial compartment tibia surface at FP and MS to that of the control group. For the first time, this study quantifies the estimated effects of HTO using a generic model that incorporates a detailed knee model to better understand tibiofemoral contact loading.

5.6.3 Principal Component Analysis and The Cardiff Classifier

Post-operative patients who underwent with the largest correction change in mTFA also reported the greatest improvement in OKS, however this did not correlate with improvements in biomechanical function [reduction in B(OA)]. HTO surgery reduced the belief in OA in 20 out of 22 of the included patients, indicating biomechanical improvement occurs due to realignment surgery.

The Cardiff Classifier was able to correctly classify between the control cohort and OA gait biomechanics for 34 out of the 42 cases (~80% of cases). The classifier also established that 18 out of the 22 patients decreased their OA belief having undergone lower limb realignment surgery.

CHAPTER 6: BIOMECHANICAL DIFFERENCES BETWEEN A CONTROL COHORT, PRE-HTO UNALTERED LEVEL GAIT AND PRE-HTO ALTERED GAIT STYLES

6.1 Chapter background

mKOA is becoming more frequently diagnosed in younger patients, however, surgeons are reluctant to replace joints for this cohort of individuals due to the replacements limited life span. Younger patients with moderate mKOA can find themselves in a treatment gap since they are not candidates for a replacement joint. In this case, orthopaedic surgeons may decide to wait for the patients' symptoms to progress, or they may be offered an HTO. As mentioned previously, between 1 December 2014 and 1 December 2017, 65% of individuals on the UKKOR were either waiting for surgery or had no operative data entered on the registry (United Kingdom Knee Osteotomy Registry: The First Annual Report 2018). Importantly, at the time of writing this thesis, surgeries were being delayed further due to COVID-19, resulting in even more patients not receiving symptomatic relief to their condition.

This treatment gap further justifies the research into altering an individual’s gait to reduce medial compartment knee loading, and consequently may reduce pain. This is like offloading braces; however, it is altering the whole body as opposed to the affected leg only. As outlined in the Literature Review Chapter, braces are bulky and the adherence to wearing them is questionable. Therefore, a simpler and easier intervention would be advantageous. In addition, offloading the diseased side of the knee may arrest further development of the OA, associated tissue damage, inflammation, and pain. Therefore, gait retraining has, in theory, has the potential to provide symptomatic relief leading up to surgery by distributing knee loading towards the healthy side of the joint.

Gait retraining is not currently part of clinical practice for this specific patient population who have mKOA and varus deformity, making this line of research novel. Recent systematic reviews have highlighted the need for further research to be undertaken to establish the effectiveness of gait retraining as a viable clinical recommendation. Bowd et al. (2019) (the systematic review published from this thesis) has indicated that limited research has been undertaken to understand the consequences of a gait retraining intervention on the hip and ankle joints.

The purpose of this chapter was to establish whether altering an individual's gait before undergoing a HTO offloads the damaged medial compartment of the knee, and if it does, whether this also effects the hip and ankle joints moments. This chapter addressed medial compartment knee joint loading with two approaches. The first being the more 'traditional' and widely reported method of measuring medial knee joint loading in the form of the external knee adduction moment (EKAM) and the knee adduction angular impulse (KAAI). The second approach was a MSK modelling technique, in the form of the COMAK framework ((Lenhart *et al.*, 2015)), to predict internal joint loading. Each of the three gait retraining styles were assessed in isolation to better understand each style independently and to produce recommendations that will inform future gait retraining programmes. At the end of this chapter, there is a chapter summary to address the potential benefits of altering gait pre-HTO.

6.2 Group demographics

29 patients (30 knees) were recruited from the Cardiff and Vale Orthopaedic Centre. Pre-HTO group demographics are presented in the previous chapter (Chapter 5). The table below (Table 6-1) outlines the final numbers that were used in each analysis within this chapter.

Table 6-1 Pre-HTO Altered Gait Styles: Participant Numbers Per Analysis

Gait style	**Controls**	**Pre-HTO: Visual 3D**	**Pre-HTO: COMAK**
Unaltered level gait	28	30	29
Toe out gait	x	30	29
Wide stance gait	x	29	29
Medial thrust gait	x	20	19

Note: Due to data compatibility issues with one participant, that participant was not included in the COMAK analysis. 10 participants did not achieve a reduction in the maximum knee adduction angle during the first half of stance and therefore discounted as not performing an effective medial thrust gait style.

6.3 Toe out gait

6.3.1 Quantifying toe out gait

All patient's pre-surgery were able to successfully adopt a toe out altered gait style. Adopting a toe out gait resulted in a FPA mean increase of 12° from baseline (Table 6-2).

Table 6-2 Pre-HTO Toe Out Gait: Establishing Altered Gait Style

	Controls	Pre-HTO NL	Pre-HTO TO	Controls vs pre-HTO NL	Controls vs pre-HTO TO	Pre NL vs Pre TO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Foot progression angle (°)	**15.69 (5.68)**	**16.28 (7.44)**	**28.00 (8.14)**	**0.542**	**0.000††**	**0.000****

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. A positive foot progression angle (°) indicates a toe out foot progression angle. std = standard deviation. ° = degree.

6.3.2 Spatial-temporal parameters

Apart from a significantly reduced operative limb stride length when adopting a toe out gait compared to an unaltered level gait pre-HTO (1.22m (0.19) vs 1.2m (0.19), p = 0.024), there were no significant changes in the other spatial-temporal parameters in Table 6-3.

Table 6-3 Pre-HTO Toe Out Gait Spatial Temporal Parameters

	Controls	**Pre-HTO NL**	**Pre-HTO TO**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO TO**	**Pre NL vs Pre TO**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Operative limb cycle time (s)	1.08 (0.08)	1.17 (0.15)	1.16 (0.15)	**0.003††**	**0.003††**	0.256
Operative limb stance time (s)	0.65 (0.06)	0.73 (0.12)	0.72 (0.12)	**0.000††**	**0.001††**	0.096
Operative limb step length (m)	0.64 (0.07)	0.60 (0.10)	0.60 (0.10)	0.068	**0.040***	0.315
Operative limb step time (s)	0.54 (0.04)	0.59 (0.07)	0.58 (0.07)	**0.003****	**0.002††**	0.665
Operative limb stride length	1.29 (0.13)	1.22 (0.19)	1.20 (0.19)	0.087	**0.027***	**0.024****
Swing time (s)	0.43 (0.03)	0.44 (0.04)	0.44 (0.03)	0.297	0.224	0.972
Speed (m/s)	1.21 (0.16)	1.06 (0.23)	1.05 (0.23)	**0.008****	**0.003****	0.182

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. std = standard deviation. s = seconds; m = metre; m/s = metre/second.

6.3.3 Knee joint loading: External moments

Table 6-4 and Table 6-5 outline the discrete knee moments and KAAI, whilst Figure 37 and Figure 38 give a visual outline of the alterations a pre-HTO toe out gait has on hip, knee, and ankle rotations and moments. Adopting a toe out gait did not significantly change EKAM1, but it did significantly decrease EKAM2 when compared to an unaltered level gait pre-surgery (2.48 %BW.h (1.1) vs 2.18 %BW.h (0.94), p = 0.000). However, a toe out altered gait style EKAM2 remained significantly higher compared to a control group (1.5 %BW.h (0.67) vs 2.18 %BW.h (0.94), p = 0.003).

Adopting a toe out gait significantly increased peak flexion moment when compared to pre-HTO unaltered level gait (2.87 %BW.h (1.56) vs 3.11 %BW.h (1.51), p = 0.041). Additionally, adopting a toe out gait significantly reduced peak extension moment when compared to a pre-HTO unaltered level gait (1.94 %BW.h (0.83) vs 1.84 %BW.h (0.84), p = 0.011).

In terms of the peak transverse plane knee moment changes, adopting a toe out gait compared to unaltered level gait significantly reduced the peak internal rotation moment (1.01 %BW.h (0.48) vs 0.86 %BW.h (0.36), p = 0.000) and significantly increased the peak external rotation moment (0.14 %BW.h (0.13) vs 0.16 %BW.h (0.11), p = 0.001).

6.3.4 Knee joint loading: Impulses

Adopting a toe out gait significantly increased KAAI between heel strike and 16% of stance when compared to pre-HTO unaltered level gait (0.10 %BW.h.s (0.06) vs 0.12 %BW.h.s (0.07), p = 0.000) (Table 6-5). There were no significant differences between pre-HTO unaltered level gait and adopting a toe out gait when comparing the first half of stance.

However, KAAI metrics during the second of stance were significantly reduced when adopting a toe out gait pre-HTO compared to pre-HTO unaltered level gait (0.60 %BW.h.s (0.27) vs 0.52 %BW.h.s (0.24), p = 0.000). When compared to pre-HTO unaltered level gait, a toe out gait significantly increased second half of stance abduction angular impulse (p = 0.001).

Table 6-4 Pre-HTO Toe Out Gait External Knee Moments

	Controls	Pre-HTO NL	Pre-HTO TO	Controls vs Pre-HTO NL	Controls vs Pre-HTO TO	Pre NL vs Pre HTO
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) moment						
Maximum	2.11 (0.81)	3.19 (1.14)	3.22 (1.15)	**0.000****	**0.000****	0.318
1st peak (1st half stance)	2.27 (0.65)	3.10 (1.12)	3.18 (1.17)	**0.001****	**0.001****	0.102
2nd peak (2nd half stance)	1.50 (0.67)	2.48 (1.10)	2.18 (0.94)	**0.000****	**0.003****	**0.000****
Midstance	1.15 (0.49)	2.15 (0.83)	2.13 (0.82)	**0.000****	**0.000****	0.641
Flexion (+) moment peak	3.62 (1.65)	2.87 (1.56)	3.11 (1.51)	0.079	0.221	**0.041†**
Extension (−) moment peak	-2.45 (0.85)	-1.94 (0.83)	-1.84 (0.84)	**0.043†**	0.018	**0.011†**
Internal (+) rotation moment	0.60 (0.37)	1.01 (0.48)	0.86 (0.36)	**0.001****	**0.009****	**0.000****
External (−) rotation moment	-0.16 (0.08)	-0.14 (0.13)	-0.16 (0.11)	0.050	0.311	**0.001†**

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. %BW.h = % of body weight multiplied by height.

Table 6-5 Pre-HTO Toe Out Gait External Knee Angular Impulse

	Controls	Pre-HTO NL	Pre-HTO TO	Controls vs pre-HTO NL	Controls vs pre-HTO TO	Pre NL vs Pre TO
%BW.h.s	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) angular impulse						
Stance	0.74 (0.28)	1.33 (0.51)	1.27 (0.48)	**0.000****	**0.000****	**0.004††**
1st half stance	0.43 (0.14)	0.73 (0.25)	0.75 (0.26)	**0.000****	**0.000****	0.102
2nd half stance	0.31 (0.16)	0.60 (0.27)	0.52 (0.24)	**0.000****	**0.000****	**0.000††**
0–16% stance	0.06 (0.03)	0.10 (0.06)	0.12 (0.07)	**0.000††**	**0.000****	**0.000††**
17%–midstance	0.36 (0.11)	0.61 (0.20)	0.61 (0.20)	**0.000****	**0.000****	0.688
Midstance–83% stance	0.26 (0.13)	0.50 (0.22)	0.45 (0.20)	**0.000****	**0.000****	**0.000††**
84%-100% stance	0.04 (0.02)	0.07 (0.04)	0.05 (0.04)	**0.002****	0.154	**0.000****
Abduction (−) angular impulse in Stance						
1st half stance	-0.02 (0.01)	-0.01 (0.01)	-0.01 (0.01)	**0.001††**	**0.000††**	0.064
2nd half stance	-0.02 (0.02)	-0.01 (0.01)	-0.02 (0.02)	**0.021†**	0.775	**0.001††**

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. %BW.h.s = % of body weight multiplied by height per second.

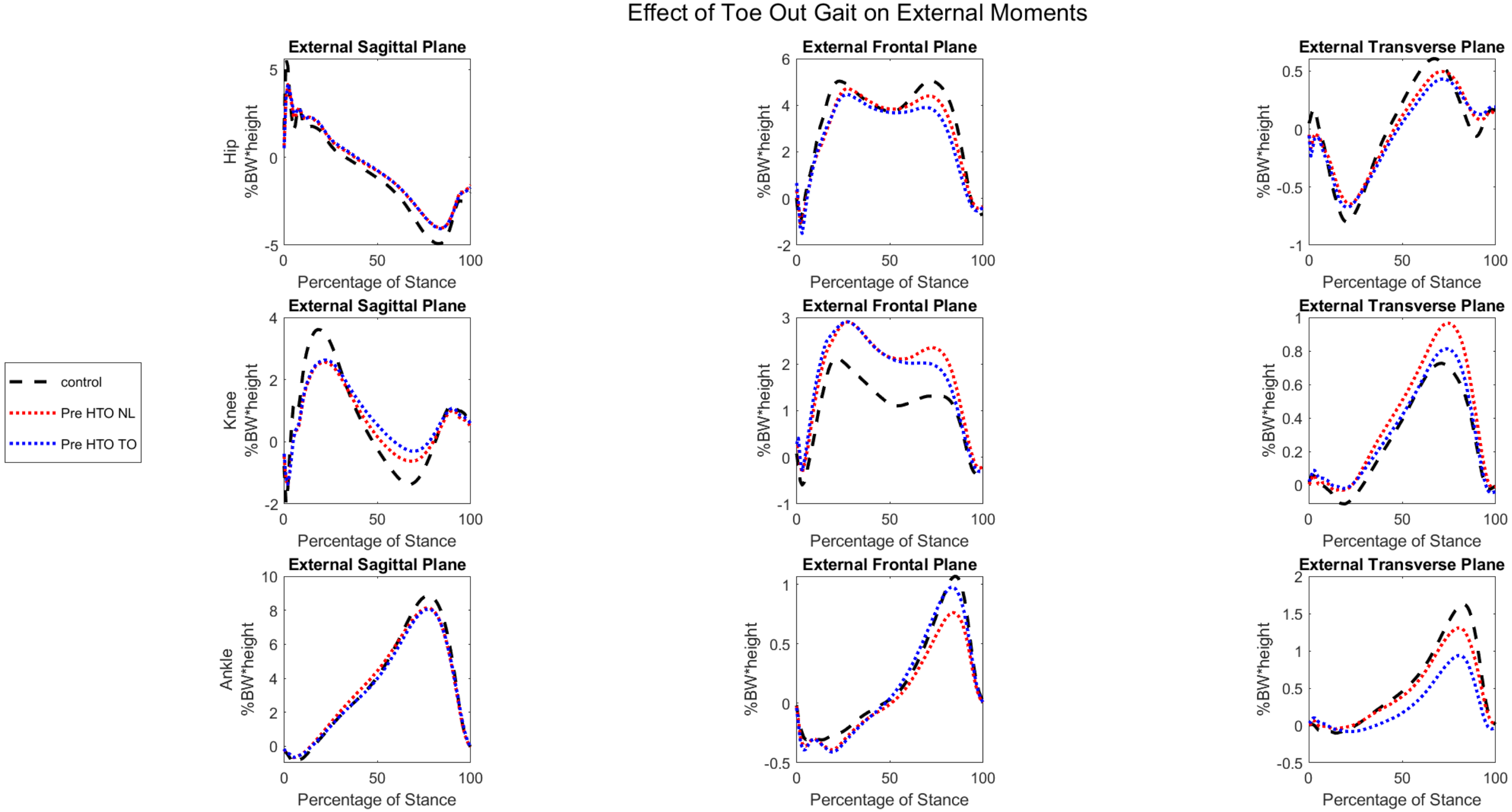


Figure 37 Visual 3D: Pre-HTO toe out gait group average joint external moments

Positive values represent external moments for knee flexion, adduction, and internal rotation

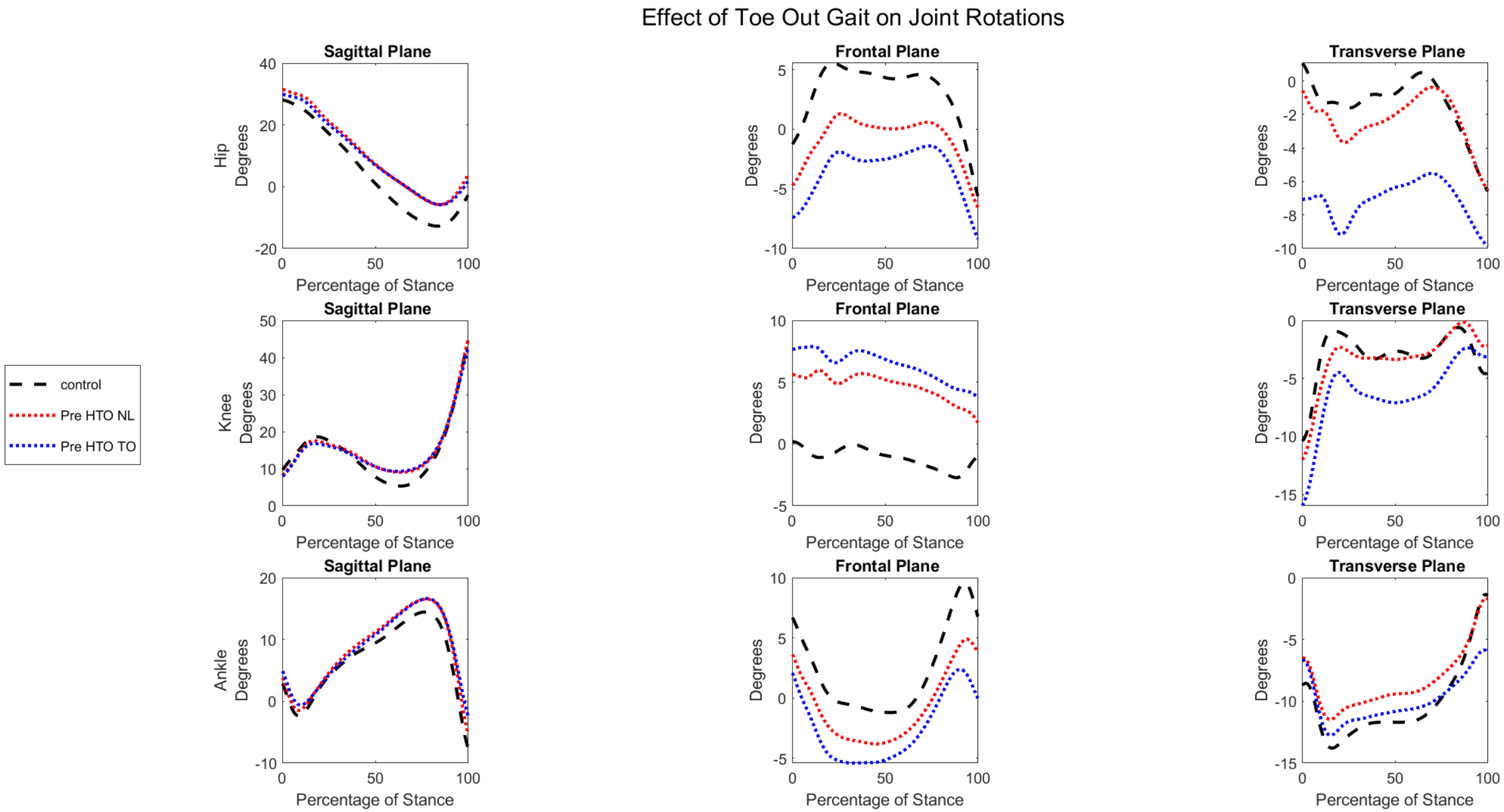


Figure 38 Visual 3D: Pre-HTO toe out gait group average joint kinematics

Positive values represent knee flexion, adduction, and internal rotations.

6.3.5 External ankle moments

Table 6-6 presents ankle moments for adopting a toe out gait style pre-surgery. During the first half of stance, peak dorsiflexion moment significantly reduced when adopting a toe out gait style (4.42 %BW.h (0.76) vs 3.99 %BW.h (0.73), p = 0.000). In the transverse plane, adopting a toe out gait resulted in a significantly reduced peak internal rotation moment (0.42 %BW.h (0.23) vs 0.38 %BW.h (0.21), p = 0.032) and a significantly increased peak external rotation moment compared to a pre-HTO unaltered level gait (0.13 %BW.h (0.11) vs 0.28 %BW.h (0.63), p = 0.012).

During the second half of stance, peak plantarflexion moment significantly reduced when adopting a toe out gait style (0.04 %BW.h (0.1) vs 0.00 %BW.h (0.11), p = 0.007). In the frontal plane, peak inversion moment was significantly increased when adopting a toe out gait when compared to a pre-HTO unaltered level gait (0.84 %BW.h (0.48) vs 1.06 %BW.h (0.58), p = 0.002). In the transverse plane, adopting a toe out gait resulted in a significantly increased peak external rotation moment compared to a pre-HTO unaltered level gait (0.01 %BW.h (0.06) vs 0.31 %BW.h (1.18), p = 0.000).

Table 6-6 Pre-HTO Toe Out Gait: External Ankle Moments Parameters

	Controls	**Pre-HTO NL**	**Pre-HTO TO**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO TO**	**Pre NL vs Pre TO**
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak dorsi-flexion (+) moment	6.78 (3.93)	8.25 (1.22)	8.19 (1.28)	0.594	0.583	0.213
Peak plantar-flexion (−) moment	-0.97 (0.36)	-0.74 (0.37)	-0.73 (0.40)	**0.025***	**0.023***	0.861
Peak inversion (+) moment	0.82 (0.54)	0.84 (0.48)	1.06 (0.57)	0.787	0.268	**0.001†**
Peak eversion (-) moment	-0.41 (0.22)	-0.53 (0.51)	-0.55 (0.42)	0.638	0.195	0.111
Peak internal rotation (+) moment	1.25 (0.85)	1.37 (0.63)	1.26 (0.68)	0.763	0.649	0.178
Peak external rotation (−) moment	-0.17 (0.08)	-0.14 (0.10)	-0.42 (1.16)	0.239	0.835	**0.001††**
0-50% (HS to midstance)						
Peak dorsi-flexion (+) moment	4.08 (0.99)	4.42 (0.76)	3.99 (0.73)	0.143	0.702	**0.000****
Peak plantar-flexion (−) moment	-1.03 (0.28)	-0.74 (0.38)	-0.73 (0.40)	**0.002****	**0.002****	0.813
Peak inversion (+) moment	0.14 (0.14)	0.17 (0.21)	0.18 (0.23)	0.694	0.649	0.862
Peak eversion (-) moment	-0.43 (0.20)	-0.53 (0.51)	-0.55 (0.43)	0.994	0.382	0.116
Peak internal rotation (+) moment	0.47 (0.26)	0.42 (0.23)	0.38 (0.21)	0.417	0.115	**0.032††**
Peak external rotation (−) moment	-0.18 (0.08)	-0.13 (0.11)	-0.28 (0.63)	0.100	0.511	**0.012†**
50-100% (Midstance to toe-off)						
Peak dorsi-flexion (+) moment	8.94 (0.74)	8.25 (1.22)	8.16 (1.32)	**0.004††**	**0.003††**	0.206
Peak plantar-flexion (−) moment	-0.03 (0.12)	-0.04 (0.10)	0.00 (0.11)	0.734	0.377	**0.007****
Peak inversion (+) moment	1.11 (0.34)	0.84 (0.48)	1.06 (0.58)	**0.019****	0.677	**0.002††**
Peak eversion (-) moment	-0.11 (0.14)	-0.19 (0.29)	-0.15 (0.26)	0.521	0.435	0.111
Peak internal rotation (+) moment	1.67 (0.53)	1.37 (0.63)	1.21 (0.76)	**0.006**	**0.010***	0.271
Peak external rotation (−) moment	0.00 (0.06)	-0.01 (0.06)	-0.31 (1.18)	0.969	**0.001**	**0.000**

Significant difference (p < 0.01) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. %BW.h = % of body weight multiplied by height. HS = heel strike.

6.3.6 External hip moments

During the first half of stance, peak external hip adduction moment significantly reduced when adopting a toe out gait style (5.05 %BW.h (1.14) vs 4.85 %BW.h (1.1), p = 0.017) (Table 6-7). A toe out altered gait significantly increased peak external hip abduction moment compared to pre-HTO unaltered level gait (1.37 %BW.h (1.25) vs 1.73 %BW.h (1.27), p = 0.001). In the transverse plane, adopting a toe out gait resulted in a significantly reduced peak external hip internal rotation moment when compared to a pre-HTO unaltered level gait (0.25 %BW.h (0.24) vs 0.2 %BW.h (0.23), p = 0.005).

During the second half of stance, peak external hip adduction moment significantly reduced when adopting a toe out gait style compared to pre-HTO unaltered level gait (4.64 %BW.h (1.23) vs 4.17 %BW.h (1.19), p = 0.000). In the transverse plane, adopting a toe out gait resulted in a significantly reduced peak external hip internal rotation moment to a pre-HTO unaltered level gait (0.57 %BW.h (0.34) vs 0.51 %BW.h (0.25), p = 0.049).

Table 6-7 Pre-HTO Toe Out Gait External Hip Moments

	Controls	Pre-HTO NL	Pre-HTO TO	Controls vs pre-HTO NL	Controls vs pre-HTO TO	Pre NL vs Pre TO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak external hip flexion (+) moment	6.31 (2.64)	4.95 (1.73)	5.09 (1.92)	**0.025***	**0.047***	0.629
Peak external hip extension (−) moment	-4.36 (1.70)	-4.31 (1.56)	-4.41 (1.97)	0.898	0.932	0.572
Peak external hip adduction (+) moment	5.09 (1.41)	5.20 (1.16)	4.96 (1.14)	0.742	0.712	**0.004****
Peak external hip abduction (-) moment	-1.65 (1.03)	-1.60 (1.08)	-1.93 (1.09)	0.811	0.212	**0.001††**
Peak external hip internal rotation (+) moment	0.54 (0.30)	0.57 (0.33)	0.52 (0.25)	0.691	0.750	**0.028***
Peak external hip rotation (−) moment	-0.80 (0.43)	-0.73 (0.42)	-0.77 (0.38)	0.546	0.790	0.115
0-50% (HS to midstance)						
Peak external hip extension (−) moment	-1.25 (0.92)	-1.02 (1.03)	-0.99 (1.11)	0.389	0.353	0.742
Peak external hip adduction (+) moment	5.27 (1.03)	5.05 (1.14)	4.85 (1.10)	0.453	0.145	**0.017***
Peak external hip abduction (-) moment	-1.28 (1.29)	-1.37 (1.25)	-1.73 (1.27)	0.660	0.189	**0.001††**
Peak external hip internal rotation (+) moment	0.34 (0.20)	0.25 (0.24)	0.20 (0.23)	0.058	**0.003††**	**0.005††**
Peak external hip rotation (−) moment	-0.86 (0.42)	-0.73 (0.42)	-0.77 (0.38)	0.250	0.400	0.115
50-100% (midstance to toe off)						
Peak external hip adduction (+) moment	5.20 (1.17)	4.64 (1.23)	4.17 (1.19)	0.082	**0.000††**	**0.000****
Peak external hip abduction (-) moment	-0.86 (0.55)	-0.59 (0.50)	-0.71 (0.55)	0.055	0.306	0.096
Peak external hip internal rotation (+) moment	0.64 (0.24)	0.57 (0.34)	0.51 (0.25)	0.329	0.052	**0.049***
Peak external hip rotation (−) moment	-0.11 (0.12)	-0.09 (0.19)	-0.10 (0.22)	0.303	0.190	0.504

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. %BW.h = % of body weight multiplied by height. HS = heel strike.

6.3.7 Concurrent Optimisation of Muscle and Secondary Kinematics framework

6.3.7.1 Internal knee joint loading

Table 6-8 presents internal knee joint contact force and pressures when adopting a toe out gait style pre-HTO, whilst Figure 39 outlines the knee kinematics, moments, and internal contact forces.

At FP, total knee, and medial compartment knee contact force, mean pressure, and maximum pressure all increased because of adopting a toe out gait pre-HTO compared to unaltered level gait pre-HTO. Medial knee compartment contact force at FP significantly increased because of adopting a toe out gait compared to an unaltered level (1.59BW (0.34)) vs (1.7BW (0.32), p = 0.000). This increase of medial compartment loading was met with no significant changes in FP lateral compartment contact force unaltered level gait compared to toe out gait. Additionally, FP lateral compartment knee maximum pressure significantly increased because of adopting a toe out gait style pre-operatively (11.67 MPa (4.69)) vs (12.11 MPa (4.37), p = 0.048).

At MS, total, medial, and lateral compartment contact forces did not alter when adopting a toe out gait style compared to an unaltered level gait pre-HTO. Total and medial compartment knee mean, and maximum pressures also did not significantly alter when adopting the toe out gait. Compared to pre-HTO unaltered level gait, adopting a toe out gait resulted in a significant increase in lateral knee compartment mean pressure (3.08 MPa (1.37)) vs (3.37 MPa (1.49), p = 0.027) and maximum pressure (6.4 MPa (2.85)) vs 7.01 MPa (3.12), p = 0.037).

At SP, total knee contact forces were significantly decreased when adopting a toe out gait compared to pre-HTO unaltered level gait (2.49BW (0.63)) vs (2.31BW (0.49), p = 0.006). The decreased in total knee contact forces was also observed in the medial compartment (1.63BW (0.48)) vs (1.44BW (0.37), p = 0.000). Additionally, when adopting a toe out gait compared to pre-HTO unaltered level gait, medial compartment mean pressure decreased (6.44 MPa (1.37)) vs 6.11 MPa (1.24), p = 0.006), as well as maximal pressure (13.18 MPa (2.87)) vs 12.36 MPa (2.68), p = 0.004). Adopting the toe out gait did not alter lateral compartment joint loading and remained elevated compared to the control group.

6.3.7.1.1 Internal knee joint loading ratios

Table 6-9 presents internal knee joint contact force ratios when adopting a toe out gait style pre-HTO. At FP and MS, adopting a toe out gait did not significantly alter either medial to total, or lateral to total, contact forces when compared to a pre-HTO unaltered level gait. When comparing the pre-surgery group to the control group, a toe out gait resulted in a significant increase in first peak medial to total contact force (0.63 (0.07) vs 0.68 (0.08), $p = 0.018$) and a significant decrease in lateral to total contact force (0.41 (0.07) vs 0.35 (0.08), $p = 0.015$). No significant changes were observed between the groups at MS.

At SP, when comparing pre-HTO unaltered gait to adopting a toe out gait, the medial to total contact force ratio significantly reduced (0.65 (0.12) vs 0.62 (0.11), $p = 0.002$), whilst there was a significant increase in lateral to total contact force ratio (0.38 (0.12) vs 0.41 (0.11), $p = 0.005$).

Table 6-8 Pre-HTO Toe Out Gait: Internal Knee Joint Loading

	Controls	**Pre-HTO NL**	**Pre-HTO TO**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO TO**	**Pre NL vs Pre TO**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total knee						
Contact force [BW]	2.74 (0.57)	2.40 (0.58)	2.54 (0.55)	**0.032***	0.199	**0.002****
Mean pressure [MPa]	5.63 (1.25)	6.10 (1.66)	6.38 (1.74)	0.396	0.092	**0.001****
Max pressure	12.92 (3.32)	14.14 (4.20)	14.68 (4.51)	0.313	0.174	**0.042***
Medial knee						
Contact force [BW]	1.69 (0.31)	1.59 (0.34)	1.70 (0.32)	0.233	0.970	**0.000****
Mean pressure [MPa]	5.76 (1.12)	6.40 (1.69)	6.79 (1.84)	0.097	**0.037†**	**0.000****
Max pressure	12.21 (2.52)	13.28 (3.76)	14.31 (4.11)	0.210	**0.024***	**0.000****
Lateral knee						
Contact force [BW]	1.13 (0.37)	0.89 (0.38)	0.92 (0.34)	**0.017***	**0.031***	0.163
Mean pressure [MPa]	5.37 (1.62)	5.47 (2.12)	5.63 (1.98)	0.880	0.686	0.158
Max pressure	11.57 (3.76)	11.67 (4.69)	12.11 (4.37)	0.931	0.640	**0.048†**
Midstance						
Total knee						
Contact force [BW]	1.25 (0.24)	1.36 (0.25)	1.43 (0.36)	**0.031†**	**0.035†**	0.198
Mean pressure [MPa]	3.40 (0.36)	4.39 (0.82)	4.46 (0.96)	**0.000****	**0.000****	0.284
Max pressure	7.62 (1.12)	10.01 (2.05)	10.04 (2.35)	**0.000****	**0.000****	0.627
Medial knee						
Contact force [BW]	0.92 (0.22)	1.03 (0.26)	1.05 (0.26)	0.093	**0.046***	0.448
Mean pressure [MPa]	3.81 (0.58)	4.81 (0.92)	4.85 (1.02)	**0.000****	**0.000††**	0.593
Max pressure	7.43 (1.22)	9.76 (2.10)	9.64 (2.13)	**0.000****	**0.000****	0.494
Lateral knee						
Contact force [BW]	0.36 (0.14)	0.37 (0.21)	0.43 (0.27)	0.918	0.607	0.058
Mean pressure [MPa]	2.59 (0.54)	3.08 (1.37)	3.37 (1.49)	0.164	0.063	**0.027†**
Max pressure	5.56 (1.07)	6.40 (2.85)	7.01 (3.12)	0.224	0.120	**0.037†**
Second peak						
Total knee						
Contact force [BW]	2.90 (0.70)	2.49 (0.63)	2.31 (0.49)	**0.023***	**0.001****	**0.006****
Mean pressure [MPa]	5.32 (0.64)	6.02 (1.07)	5.85 (0.98)	**0.004****	**0.020***	0.071

Max pressure	12.70 (1.76)	14.19 (3.06)	13.70 (3.06)	0.102	0.405	0.226
Medial knee						
Contact force [BW]	1.83 (0.35)	1.63 (0.48)	1.44 (0.37)	0.083	**0.000****	**0.000**††
Mean pressure [MPa]	5.91 (0.80)	6.44 (1.37)	6.11 (1.24)	0.077	0.452	**0.006****
Max pressure	12.61 (1.80)	13.18 (2.87)	12.36 (2.68)	0.376	0.682	**0.004****
Lateral knee						
Contact force [BW]	1.20 (0.44)	0.95 (0.37)	0.95 (0.34)	**0.028***	**0.023***	0.905
Mean pressure [MPa]	4.53 (0.73)	5.25 (1.46)	5.37 (1.54)	**0.025**†	**0.007**††	0.133
Max pressure	9.43 (1.48)	11.33 (3.58)	11.54 (3.66)	0.011	**0.002**††	0.230

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. BW = body weight; MPa = megapascals.

Table 6-9 Pre-HTO Toe Out Gait Contact Force Ratios

	Controls	**Pre-HTO NL**	**Pre-HTO TO**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO TO**	**Pre NL vs Pre TO**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
MED / TOTAL	0.63 (0.07)	0.67 (0.10)	0.68 (0.08)	0.051	**0.018***	0.708
LAT / TOTAL	0.41 (0.07)	0.36 (0.10)	0.35 (0.08)	0.051	**0.015***	0.643
Midstance						
MED / TOTAL	0.73 (0.09)	0.76 (0.14)	0.74 (0.13)	0.451	0.794	0.149
LAT / TOTAL	0.29 (0.10)	0.27 (0.15)	0.29 (0.14)	0.471	0.893	0.153
Second peak						
MED / TOTAL	0.64 (0.07)	0.65 (0.12)	0.62 (0.11)	0.643	0.511	**0.002****
LAT / TOTAL	0.40 (0.07)	0.38 (0.12)	0.41 (0.11)	0.542	0.695	**0.005****

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. Significant difference (p < 0.01) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. BW = body weight; MPa = megapascals. MED = medial compartment contact force; LAT = lateral compartment contact force; TOTAL = total tibiofemoral contact force.

6.3.7.2 Point of application

At FP, there were no significant changes in point of application in the medial-lateral direction between pre-HTO unaltered level gait and pre-HTO toe out gait.

At MS medial compartment knee point of application, the pre-surgery unaltered gait was significantly more lateral than the control group (mean group difference of <1mm, $p = 0.045$). This difference was no longer present when adopting a toe out gait.

At SP, medial knee compartment point of application was more lateral for both the pre-HTO unaltered gait and toe out gait compared to the control cohort (Table 6-10).

Table 6-10 Pre-HTO Toe Out Gait Point of Application

	Controls	**Pre-HTO NL**	**Pre-HTO TO**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO TO**	**Pre NL vs Pre TO**
mm	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total knee						
Anterior (+) / posterior (-)	-2.46 (2.08)	-2.07 (2.98)	-2.45 (2.98)	0.757	0.868	**0.040***
Lateral (+) / medial (-)	-2.27 (3.27)	-4.91 (4.19)	-5.00 (3.42)	**0.011***	**0.003****	0.782
Medial knee						
Anterior (+) / posterior (-)	-1.01 (2.25)	-0.30 (3.27)	-0.97 (3.30)	0.387	0.968	**0.030†**
Lateral (+) / medial (-)	-17.19 (1.29)	-18.08 (1.52)	-18.14 (1.65)	**0.021***	**0.019***	0.170
Lateral knee						
Anterior (+) / posterior (-)	-4.67 (1.83)	-5.12 (2.50)	-5.19 (2.41)	0.169	0.083	0.623
Lateral (+) / medial (-)	20.59 (2.34)	19.69 (2.14)	19.89 (2.11)	0.133	0.238	0.245
Midstance						
Total knee						
Anterior (+) / posterior (-)	4.66 (1.86)	4.07 (2.26)	3.82 (3.07)	0.291	0.220	0.530
Lateral (+) / medial (-)	-5.84 (4.72)	-7.53 (7.14)	-6.98 (5.97)	0.300	0.429	0.277
Medial knee						
Anterior (+) / posterior (-)	6.63 (2.77)	6.42 (2.69)	6.09 (3.07)	0.765	0.487	0.820
Lateral (+) / medial (-)	-16.33 (1.53)	-17.09 (2.14)	-17.13 (1.75)	**0.045†**	0.052	0.794
Lateral knee						
Anterior (+) / posterior (-)	-0.42 (1.55)	-2.18 (2.21)	-2.39 (2.36)	**0.001****	**0.000****	0.336
Lateral (+) / medial (-)	20.34 (2.43)	18.91 (3.55)	19.09 (3.21)	0.081	0.103	0.462
Second peak						
Total knee						
Anterior (+) / posterior (-)	5.70 (1.64)	4.45 (3.36)	3.68 (4.12)	0.370	0.132	0.078
Lateral (+) / medial (-)	-1.39 (2.90)	-2.59 (5.13)	-1.43 (4.70)	0.086	0.596	**0.002****
Medial knee						
Anterior (+) / posterior (-)	9.56 (1.83)	8.15 (3.38)	7.53 (4.37)	0.218	0.195	0.275
Lateral (+) / medial (-)	-14.59 (1.31)	-15.83 (1.76)	-15.88 (1.65)	**0.004****	**0.002****	0.738
Lateral knee						
Anterior (+) / posterior (-)	-0.89 (2.43)	-2.59 (3.39)	-2.72 (3.37)	**0.034***	**0.022***	0.417
Lateral (+) / medial (-)	19.74 (2.84)	19.48 (3.40)	20.37 (3.03)	0.755	0.426	**0.007****

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. std = standard deviation. mm = millimetres. X = anterior; Z = lateral. COP = centre of pressure

6.3.7.3 Contact area

At FP, there was a significant increase in contact area in the medial compartment of the knee when adopting a toe out gait compared to a pre-surgery unaltered level gait (223.63 mm^2 (29.53) vs 226.39 mm^2 (29.74), $p = 0.008$). At MS, there was no significant differences in contact area between the pre-HTO unaltered level gait and a toe out gait. At SP, total and medial contact area significantly decreased when adopting a toe out gait compared to an unaltered level gait; medial compartment contact area reduced from 223.86 mm^2 (50.52) with unaltered level gait to 210.67 mm^2 (43.85) when adopting a toe out gait, $p = 0.004$.

Table 6-11 Pre-HTO Toe Out Gait Knee Contact Area

	Controls	**Pre-HTO NL**	**Pre-HTO TO**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO TO**	**Pre NL vs Pre TO**
mm^2	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total	352.35 (39.00)	364.30 (49.61)	369.79 (46.44)	0.249	0.078	0.055
Medial	206.80 (22.63)	223.63 (29.53)	226.39 (29.74)	**0.012†**	**0.003††**	**0.008††**
Lateral	145.55 (20.95)	140.68 (28.69)	143.40 (25.71)	0.794	0.943	0.194
Midstance						
Total	262.26 (34.10)	286.65 (55.88)	295.92 (59.84)	0.080	**0.022†**	0.102
Medial	167.14 (24.46)	189.50 (36.97)	192.73 (36.07)	**0.010***	**0.003††**	0.417
Lateral	95.12 (19.62)	97.16 (35.50)	103.20 (34.21)	0.698	0.195	0.095
Second peak						
Total	399.23 (90.44)	383.65 (91.11)	367.38 (76.97)	0.500	0.136	**0.018***
Medial	219.07 (41.52)	223.86 (50.52)	210.67 (43.85)	0.698	0.291	**0.004****
Lateral	180.16 (52.60)	159.80 (48.03)	156.72 (39.22)	0.132	0.061	0.329

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. std = standard deviation. mm^2 = millimetres squared

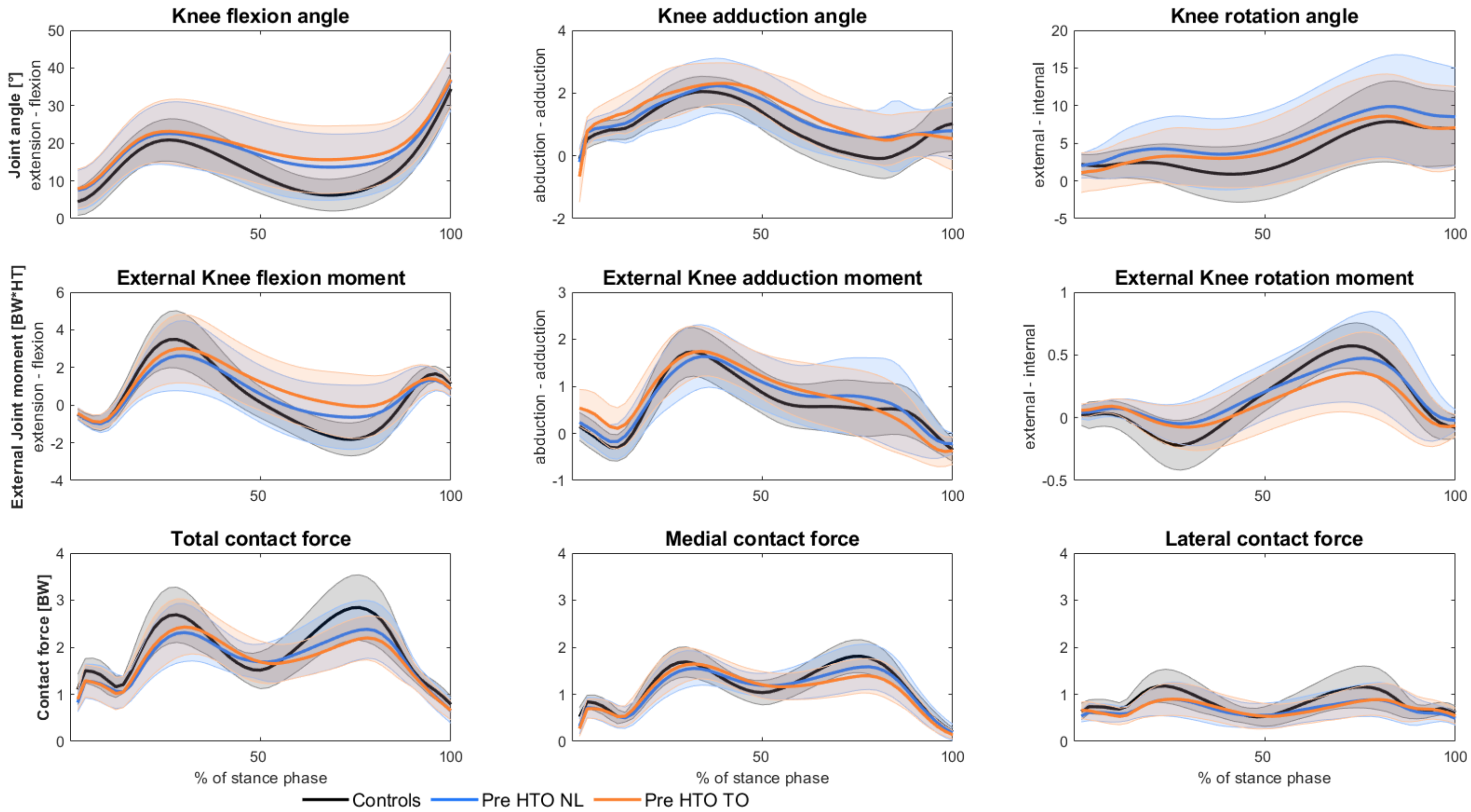


Figure 39 Pre-HTO toe out gait knee joint kinematics, external moments, contact forces

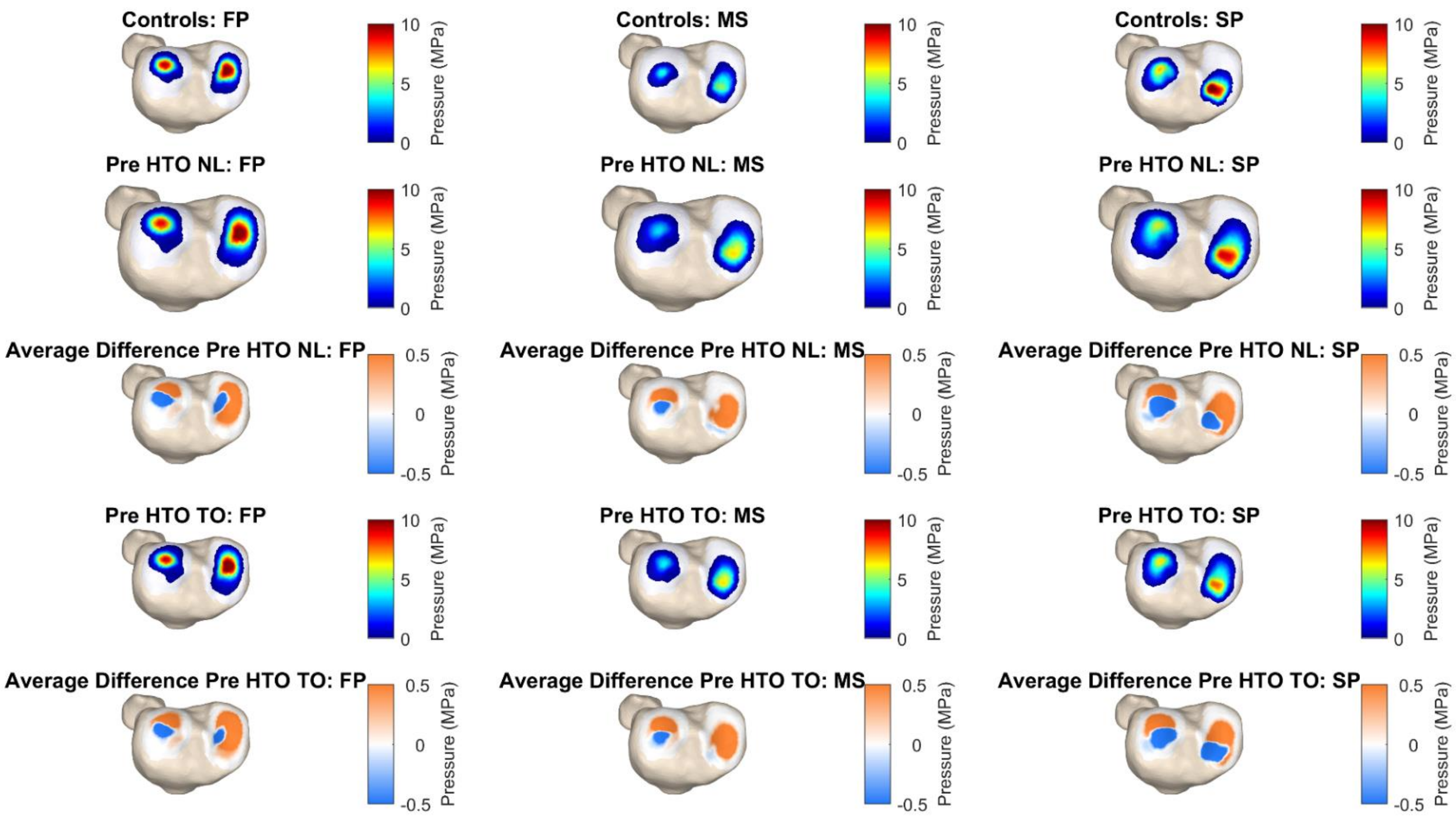


Figure 40 Pre-HTO toe out contact pressure distribution on the tibia

Average contact pressure patterns at first peak, midstance and second peak for the control group and the patients pre- and post-HTO. Furthermore, the average difference between the pressure pattern in patients and the healthy control pressure pattern is shown. Orange indicates more loading in the patient on that specific location, blue indicates decreased loading compared to the controls.

6.4 Wide stance gait

6.4.1 Quantifying wide stance gait

All patient participants were able to successfully adopt a wide stance gait style pre-HTO. Adopting a wide stance gait resulted in a stride width of ~0.25m compared a pre-HTO unaltered level gait stride width of ~0.16m (p = 0.000).

6.4.2 Spatial-temporal parameters

There were no significant changes in any spatiotemporal parameters when adopting a wide stance gait compared to an unaltered level gait pre-HTO.

Table 6-12 Pre-HTO Wide Stance Gait Quantifying Gait Style

	Controls	**Pre-HTO NL**	**Pre-HTO WS**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO WS**	**Pre NL vs Pre WS**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Stride width (m)	0.14 (0.03)	0.16 (0.04)	0.25 (0.07)	0.058	**0.000††**	**0.000****

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. m = metre.

Table 6-13 Pre-HTO Wide Stance Gait Spatial Temporal Parameters

	Controls	**Pre-HTO NL**	**Pre-HTO WS**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO WS**	**Pre NL vs Pre WS**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Operative limb cycle time (s)	1.08 (0.08)	1.17 (0.15)	1.18 (0.17)	**0.002††**	**0.002††**	0.338
Operative limb stance time (s)	0.65 (0.06)	0.73 (0.12)	0.73 (0.14)	**0.000††**	**0.001††**	0.604
Operative limb step length (m)	0.64 (0.07)	0.6 (0.10)	0.61 (0.11)	0.072	0.182	0.118
Operative limb step time (s)	0.54 (0.04)	0.59 (0.07)	0.59 (0.08)	**0.004****	**0.006††**	0.312
Operative limb stride length (m)	1.29 (0.13)	1.22 (0.19)	1.23 (0.21)	0.083	0.164	0.180
Swing time (s)	0.43 (0.03)	0.44 (0.04)	0.45 (0.05)	0.295	0.282	0.095
Speed (m/s)	1.21 (0.16)	1.06 (0.23)	1.07 (0.24)	**0.007****	**0.012***	0.367

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. s = seconds; m = metre; m/s = metre/second.

6.4.3 Knee joint loading: External moments

Adopting a wide stance gait did not significantly alter EKAM1 but it did significantly decrease EKAM2 when compared to an unaltered level gait pre-surgery (2.41 %BW.h (1.05) vs 2.19 %BW.h (1.04), p = 0.000). However, EKAM2 remained significantly higher than the control group when adopting a wide stance altered gait style EKAM2 (1.5 %BW.h (0.67) vs 2.19 %BW.h (1.04), p = 0.000).

Adopting a wide stance gait pre-HTO significantly increased peak external knee flexion moment when compared to pre-HTO unaltered level gait (2.87 %BW.h (1.59) vs 3.2 %BW.h (1.8), p = 0.003).

In terms of the peak transverse plane external knee moment changes, adopting a wide stance gait compared to unaltered level gait significantly reduced peak internal rotation moment (0.99 %BW.h (0.48) vs 0.9 %BW.h (0.46), p = 0.000) and significantly increased peak external rotation moment (0.13 %BW.h (0.13) vs 0.17 %BW.h (0.16), p = 0.002).

Table 6-14 Pre-HTO Wide Stance Gait External Knee Moments

	Controls	**Pre-HTO NL**	**Pre-HTO WS**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO WS**	**Pre NL vs Pre WS**
% BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) moment						
Maximum	2.11 (0.81)	3.08 (1.01)	3.01 (1.08)	**0.000****	**0.001****	0.135
1st peak (1st half stance)	2.27 (0.65)	3.00 (0.97)	2.94 (1.07)	**0.002****	**0.006****	0.236
2nd peak (2nd half stance)	1.50 (0.67)	2.41 (1.05)	2.19 (1.04)	**0.000****	**0.004****	**0.000****
Midstance	1.15 (0.49)	2.09 (0.79)	2.09 (0.88)	**0.000****	**0.000****	0.947
Flexion (+) moment peak	3.62 (1.65)	2.87 (1.59)	3.20 (1.80)	0.087	0.363	**0.003****
Extension (−) moment peak	-2.45 (0.85)	-1.96 (0.84)	-1.83 (0.91)	**0.065†**	0.010††	0.088
Internal (+) rotation moment peak	0.60 (0.37)	0.99 (0.48)	0.90 (0.46)	**0.001****	**0.008****	**0.000****
External (−) rotation moment peak	-0.16 (0.08)	-0.13 (0.13)	-0.17 (0.16)	0.026†	0.337	**0.002††**

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. % BW.h = percentage of body weight multiplied by height.

6.4.4 Knee joint loading: Impulses

Adopting a wide stance gait significantly reduced KAAI over the whole of stance (1.29 %BW.h.s (0.47) vs 1.21 %BW.h.s (0.5), p = 0.000). However, when splitting stance into first and second half, there were no significant differences for KAAI between wide stance gait and unaltered level gait pre-HTO. Between midstance and 83% of stance there was a significant reduction in KAAI when adopting a wide stance gait compared to unaltered level gait pre-surgery (0.49 %BW.h.s (0.21) vs 0.45 %BW.h.s (0.23), p = 0.000). This reduction was also seen between 84% and toe-off (0.07 %BW.h.s (0.04) vs 0.04 %BW.h.s (0.03), p = 0.000). When compared to pre-HTO unaltered level gait, a wide stance gait significantly increased first and second half of stance abduction angular impulse (p = 0.000).

Table 6-15 Pre-HTO Wide Stance Gait Knee Angular Impulse

	Controls	**Pre-HTO NL**	**Pre-HTO WS**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO WS**	**Pre NL vs Pre WS**
%BW.h.s	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) angular impulse						
Stance	0.74 (0.28)	1.29 (0.47)	1.21 (0.5)	**0.000****	**0.000****	**0.000††**
1st half stance	0.43 (0.14)	0.70 (0.22)	0.69 (0.25)	**0.000****	**0.000****	0.208
2nd half stance	0.31 (0.16)	0.58 (0.26)	0.52 (0.26)	**0.000****	**0.001††**	0.642
0–16% stance	0.06 (0.03)	0.10 (0.05)	0.10 (0.06)	**0.000****	**0.000****	**0.000††**
17%–midstance	0.36 (0.11)	0.59 (0.18)	0.58 (0.21)	**0.000****	**0.000****	0.130
Midstance–83% stance	0.26 (0.13)	0.49 (0.21)	0.45 (0.23)	**0.000****	**0.000****	**0.001****
84%-100% stance	0.04 (0.02)	0.07 (0.04)	0.04 (0.03)	**0.004****	0.472	**0.000****
Abduction (−) angular impulse in Stance						
1st half stance	-0.02 (0.01)	-0.01 (0.01)	-0.02 (0.01)	**0.001††**	0.224	**0.000****
2nd half stance	-0.02 (0.02)	-0.01 (0.01)	-0.02 (0.02)	**0.023†**	0.931	**0.000††**

Significant difference (p < 0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. % BW.h.s = % of body weight multiplied by height per second.

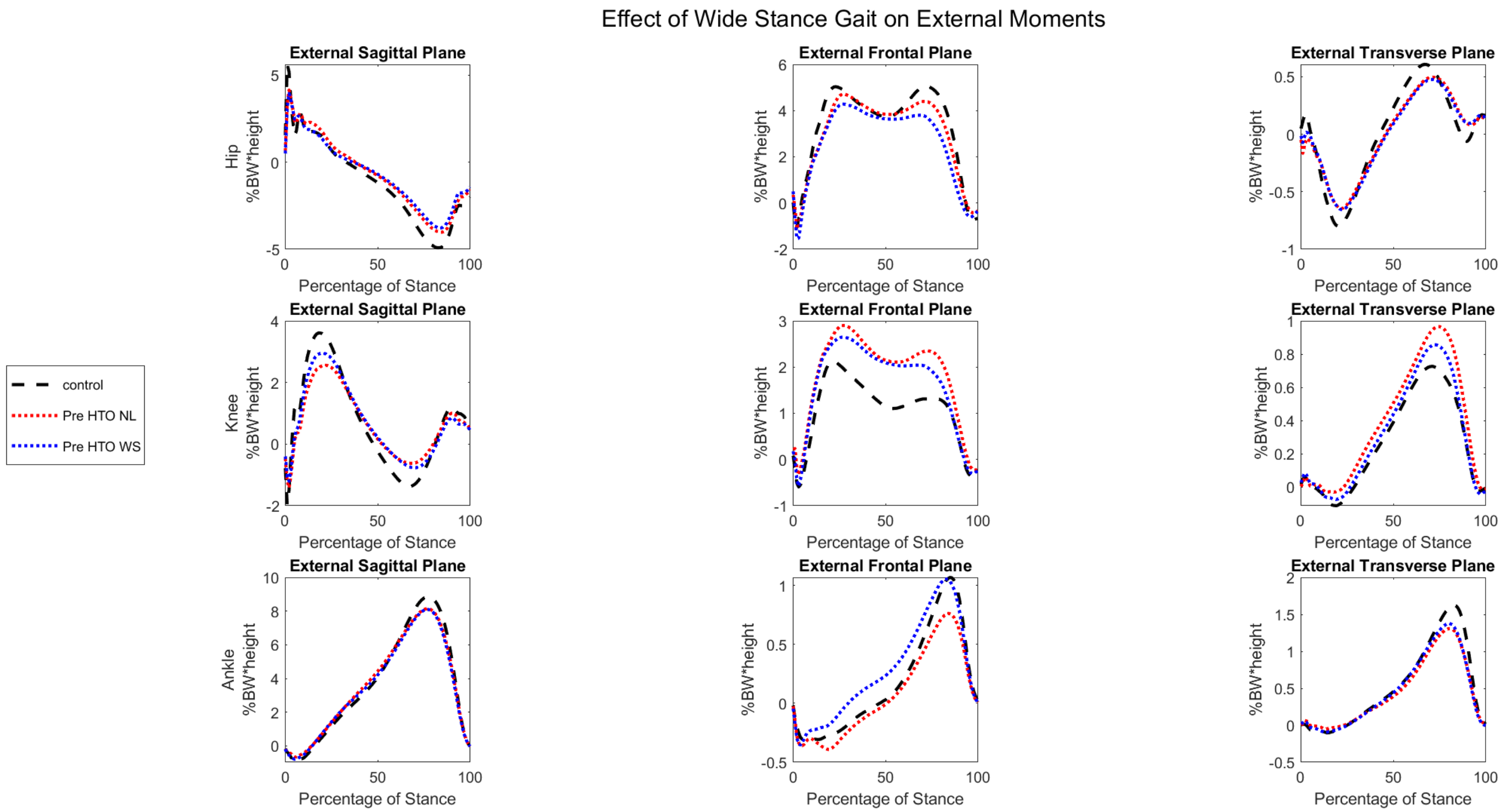


Figure 41 Visual 3D: Pre-HTO wide stance gait group average joint external moments

Positive values represent external moments for knee flexion, adduction, and internal rotation moments.

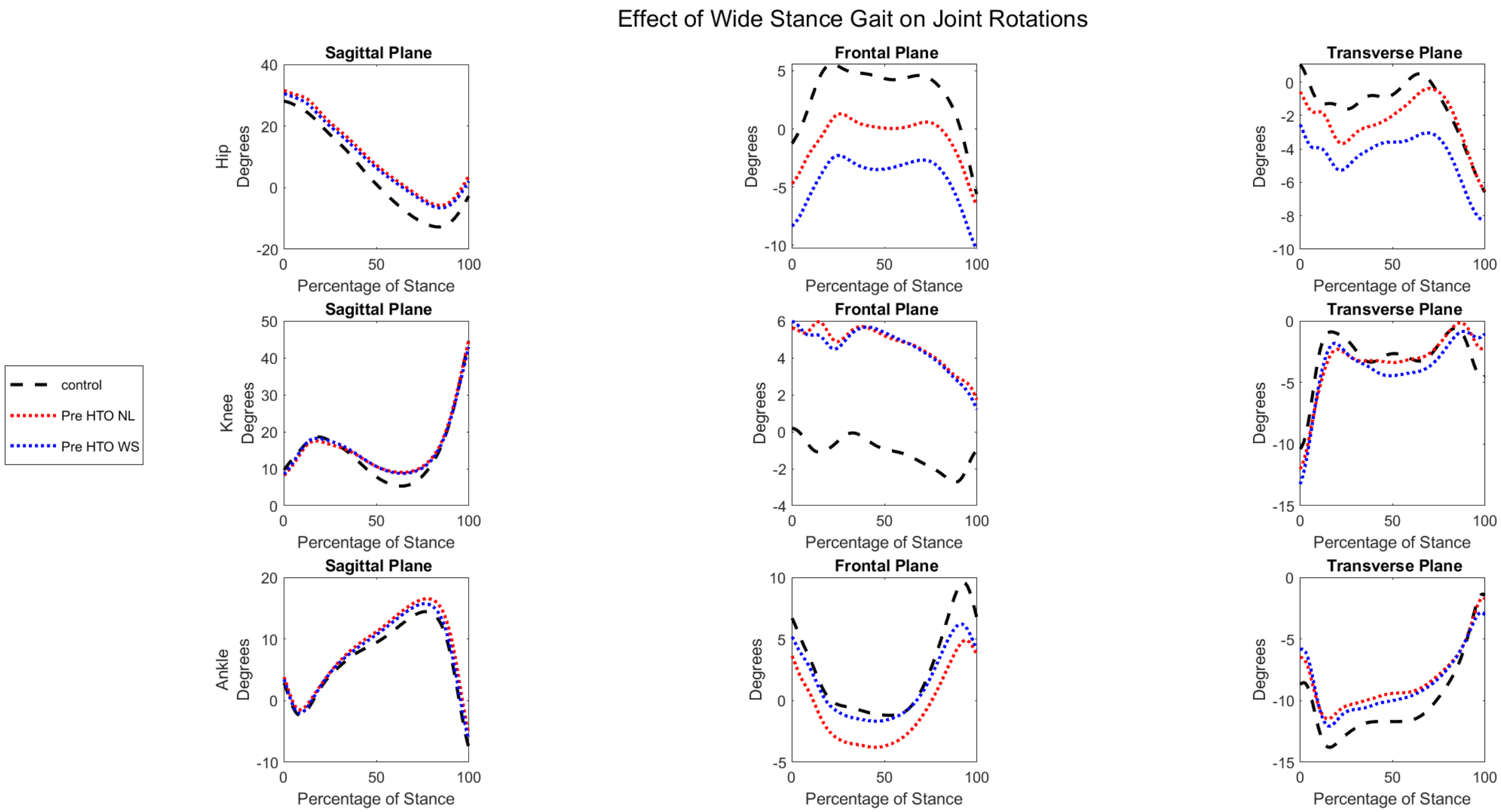


Figure 42 Visual 3D: Pre-HTO wide stance gait group average joint kinematics

Positive values represent knee flexion, adduction, and internal rotations.

6.4.5 External ankle moments

During the whole of stance, adopting a wide stance gait significantly increased peak external plantarflexion moment compared to pre-HTO unaltered level gait (0.74 %BW.h (0.38) vs 0.87 %BW.h (0.4), p = 0.000). Wide stance gait also increased peak inversion moment compared to pre-HTO unaltered level gait (0.83 %BW.h (0.49) vs 1.13 %BW.h (0.62), p = 0.000). Adopting a wide stance gait significantly reduced peak eversion moment when compared to an unaltered level gait pre-HTO (0.54 (0.51) vs 0.45 (0.51), p = 0.001). In addition to this, adopting a wide stance gait style also significantly increased the peak external rotation moment (0.14 %BW.h (0.1) vs 0.18 %BW.h (0.14), p = 0.001).

During the first half of stance, peak dorsiflexion moment significantly reduced when adopting a wide stance gait style (4.4 %BW.h (0.77) vs 4.23 %BW.h (0.87), p = 0.031), as well as a significant increase in peak plantarflexion moment (0.74 %BW.h (0.39) vs 0.86 %BW.h (0.42), p = 0.000). In the frontal plane there was a significant increase in peak inversion moment when adopting a wide stance gait compared to a pre-HTO unaltered level gait (0.17 %BW.h (0.21) vs 0.34 %BW.h (0.31), p = 0.000), as well as significantly reducing peak eversion moment (0.54 %BW.h (0.51) vs 0.45 %BW.h (0.51), p = 0.001). In the transverse plane, adopting a wide stance gait resulted in a significant increase in peak external rotation moment when compared to a pre-HTO unaltered level gait (0.13 %BW.h (0.11) vs 0.16 %BW.h (0.15), p = 0.001).

During the second half of stance, the frontal plane peak inversion moment is significantly increased when adopting a wide stance gait when compared to a pre-HTO unaltered level gait (0.83 %BW.h (0.49) vs 1.13 %BW.h (0.62), p = 0.000), as well as significantly reducing peak eversion moment (0.19 %BW.h (0.3) vs 0.09 %BW.h (0.24), p = 0.000). In the transverse plane, adopting a wide stance gait resulted in a significantly increased peak external rotation moment compared to a pre-HTO unaltered level gait (0.01 %BW.h (0.06) vs 0.03 %BW.h (0.05), p = 0.002).

Table 6-16 Pre-HTO Wide Stance Gait External Ankle Moments

	Controls	Pre-HTO NL	Pre-HTO WS	Controls vs pre-HTO NL	Controls vs pre-HTO WS	Pre NL vs Pre WS
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak dorsi-flexion (+) moment	6.78 (3.93)	8.21 (1.22)	8.22 (1.41)	0.511	0.686	0.956
Peak plantar-flexion (−) moment	-0.97 (0.36)	-0.74 (0.38)	-0.87 (0.40)	**0.028***	0.358	**0.000****
Peak inversion (+) moment	0.82 (0.54)	0.83 (0.49)	1.13 (0.62)	0.745	0.136	**0.000****
Peak eversion (-) moment	-0.41 (0.22)	-0.54 (0.51)	-0.45 (0.51)	0.542	0.442	**0.001††**
Peak internal rotation (+) moment	1.25 (0.85)	1.36 (0.64)	1.43 (0.65)	0.733	0.733	0.355
Peak external rotation (−) moment	-0.17 (0.08)	-0.14 (0.10)	-0.18 (0.14)	0.180	0.542	**0.001††**
0-50% (HS to midstance)						
Peak dorsi-flexion (+) moment	4.08 (0.99)	4.40 (0.77)	4.23 (0.87)	0.169	0.547	**0.031***
Peak plantar-flexion (−) moment	-1.03 (0.28)	-0.74 (0.39)	-0.86 (0.42)	**0.002****	0.085	**0.000****
Peak inversion (+) moment	0.14 (0.14)	0.17 (0.21)	0.34 (0.31)	0.757	**0.009††**	**0.000††**
Peak eversion (-) moment	-0.43 (0.20)	-0.54 (0.51)	-0.45 (0.51)	0.893	0.200	**0.001††**
Peak internal rotation (+) moment	0.47 (0.26)	0.43 (0.22)	0.47 (0.23)	0.521	1.000	0.581
Peak external rotation (−) moment	-0.18 (0.08)	-0.13 (0.11)	-0.16 (0.15)	0.070	0.313	**0.001††**
50-100% (Midstance to toe-off)						
Peak dorsi-flexion (+) moment	8.94 (0.74)	8.21 (1.22)	8.22 (1.41)	**0.002††**	**0.006††**	0.956
Peak plantar-flexion (−) moment	-0.03 (0.12)	-0.04 (0.10)	-0.03 (0.12)	0.723	0.928	0.631
Peak inversion (+) moment	1.11 (0.34)	0.83 (0.49)	1.13 (0.62)	**0.017***	0.857	**0.000****
Peak eversion (-) moment	-0.11 (0.14)	-0.19 (0.30)	-0.09 (0.24)	0.596	0.078	**0.000††**
Peak internal rotation (+) moment	1.67 (0.53)	1.36 (0.64)	1.43 (0.65)	**0.006††**	0.130	0.361
Peak external rotation (−) moment	0.00 (0.06)	-0.01 (0.06)	-0.03 (0.05)	0.943	0.065	**0.002††**

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. % BW.h = percentage of body weight multiplied by height. HS = heel strike.

6.4.6 External hip moments

During the whole of stance, adopting a wide stance gait resulted in a significantly reduced peak external hip extension moment compared to unaltered level gait pre-HTO (4.27 %BW.h (1.57) vs 4.03 %BW.h (1.72), p = 0.023). Adopting a wide stance gait significantly reduced peak external hip adduction moment compared to pre-HTO unaltered level gait (5.16 %BW.h (1.16) vs 4.76 %BW.h (1.28), p = 0.000). A wide stance altered gait significantly increased peak external hip abduction moment compared to pre-HTO unaltered level gait (1.64 %BW.h (1.08) vs 1.99 %BW.h (1.27), p = 0.000).

During the first half of stance, peak external hip adduction moment significantly reduced when adopting a wide stance gait style compared to an unaltered level gait pre-HTO (5.01 %BW.h (1.14) vs 4.67 %BW.h (1.27), p = 0.000). A wide stance altered gait significantly increased peak external hip abduction moment compared to pre-HTO unaltered level gait (1.41 %BW.h (1.26) vs 1.83 %BW.h (1.42), p = 0.000).

During the second half of stance, peak external hip adduction moment significantly reduced when adopting a wide stance gait style (4.59 %BW.h (1.22) vs 4.08 %BW.h (1.31), p = 0.000). During the second half of stance, peak external hip abduction moment significantly increased when adopting a wide stance gait style (0.6 %BW.h (0.51) vs 0.77 %BW.h (0.53), p = 0.001).

Table 6-17 Pre-HTO Wide Stance Gait External Hip Moments

	Controls	Pre-HTO NL	Pre-HTO WS	Controls vs pre-HTO NL	Controls vs pre-HTO WS	Pre NL vs Pre WS
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak external hip flexion (+) moment	6.31 (2.64)	4.92 (1.75)	4.85 (1.51)	**0.024***	**0.014***	0.618
Peak external hip extension (−) moment	-4.36 (1.70)	-4.27 (1.57)	-4.03 (1.72)	0.838	0.552	**0.023***
Peak external hip adduction (+) moment	5.09 (1.41)	5.16 (1.16)	4.76 (1.28)	0.829	0.365	**0.000****
Peak external hip abduction (-) moment	-1.65 (1.03)	-1.64 (1.08)	-1.99 (1.27)	0.968	0.224	**0.000††**
Peak external hip internal rotation (+) moment	0.54 (0.30)	0.57 (0.33)	0.55 (0.33)	0.715	0.868	0.352
Peak external hip rotation (−) moment	-0.80 (0.43)	-0.70 (0.38)	-0.74 (0.36)	0.357	0.537	0.141
0-50% (HS to midstance)						
Peak external hip extension (−) moment	-1.25 (0.92)	-1.05 (1.04)	-1.02 (1.11)	0.448	0.411	0.763
Peak external hip adduction (+) moment	5.27 (1.03)	5.01 (1.14)	4.67 (1.27)	0.329	**0.030†**	**0.000****
Peak external hip abduction (-) moment	-1.28 (1.29)	-1.41 (1.26)	-1.83 (1.42)	0.552	0.116	**0.000††**
Peak external hip internal rotation (+) moment	0.34 (0.20)	0.26 (0.24)	0.26 (0.23)	0.086	0.063	0.907
Peak external hip rotation (−) moment	-0.86 (0.42)	-0.70 (0.38)	-0.74 (0.36)	0.137	0.232	0.141
50-100% (Midstance to toe-off)						
Peak external hip adduction (+) moment	5.20 (1.17)	4.59 (1.22)	4.08 (1.31)	0.061	**0.001****	**0.000****
Peak external hip abduction (-) moment	-0.86 (0.55)	-0.60 (0.51)	-0.77 (0.53)	0.068	0.532	**0.001††**
Peak external hip internal rotation (+) moment	0.64 (0.24)	0.56 (0.34)	0.54 (0.33)	0.322	0.204	0.267
Peak external hip rotation (−) moment	-0.11 (0.12)	-0.10 (0.19)	-0.08 (0.20)	0.353	0.136	0.470

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. % BW.h = percentage of body weight multiplied by height. HS = heel strike.

6.4.7 Concurrent Optimisation of Muscle and Secondary Kinematics

6.4.7.1 Internal knee joint loading

At FP, total and lateral compartment contact force, mean and maximum pressure significantly increased when adopting a wide stance gait compared to an unaltered level gait. Medial compartment maximum pressure significantly increased because of adopting a wide stance gait style compared to pre-HTO unaltered level gait (13.28 MPa (3.76) vs 13.85 MPa (4.1), $p = 0.028$).

At MS, the only significant differences between a wide stance gait and an unaltered level gait was medial compartment maximum pressure, for which a wide stance gait reduced maximum pressure (9.76 MPa (2.1) vs 9.46 MPa (1.96), $p = 0.014$).

At SP, medial compartment knee contact forces, mean and maximum pressure were significantly decreased when adopting a wide stance gait compared to unaltered level gait. Medial compartment contact force reduced from 1.63BW (0.48) to 1.52BW (0.53) ($p = 0.005$) when adopting a wide stance gait pre-HTO compared to an unaltered level gait pre-HTO. Additionally, lateral compartment contact force, mean and maximal pressure were significantly increased when adopting the wide stance gait style. Lateral compartment contact force went from 0.95BW (0.37) to 1.08BW (0.46) ($p = 0.006$).

6.4.7.2 Internal knee joint loading ratios

At FP, adopting a wide stance gait compared to an unaltered level gait pre-HTO resulted in a significant increase in medial to total contact force ratio (0.67 (0.01) vs 0.64 (0.1), $p = 0.010$), and a significant reduced lateral to total contact force ratio (0.36 (0.1) vs 0.39 (0.1), $p = 0.011$). At SP, the medial to total contact force ratio significantly reduced (0.65 (0.12) vs 0.61 (0.13), $p = 0.000$), whilst there was a significant increase in lateral to total contact force ratio (0.38 (0.12) vs 0.43 (0.14), $p = 0.000$).

6.4.7.3 Point of application

At FP, total knee point of application shifted laterally when adopting a wide stance gait compared to an unaltered level gait pre-HTO by a group mean of ~1.3mm ($p = 0.031$). As a result of adopting a wide stance gait, COP was normalised to that of the control group ($p = 0.180$). In the medial knee compartment, the control group's point of application was more medial compared to the unaltered gait and the wide stance gait (group mean differences of ~1mm, p = 0.021).

At MS, the medial knee compartment point of application was significantly more medial compared to the control cohort by a group mean difference of ~0.8mm unaltered gait and ~1mm when adopting a wide stance gait.

At SP, total knee and lateral knee compartment medial-lateral point of direction changed significantly between the pre-HTO unaltered level gait and pre-HTO wide stance gait style; with the point of application occurring more laterally when adopting a wide stance gait (group mean difference of ~2mm across the total knee (p = 0.000)). Additionally, at the medial compartment, both the unaltered and the wide stance gait point of application was more lateral compared to the control cohort.

6.4.7.4 Contact area

At FP, there was a significant increase in total knee contact area when adopting a wide stance when compared to a pre-surgery unaltered level gait (364.3 mm^2 (49.61) vs 372.04 mm^2 (47.95), p = 0.040). There was a significant increase in lateral compartment contact area when adopting a wide stance when compared to a pre-surgery unaltered level gait (140.68 mm^2 (28.69) vs 148.35 mm^2 (28.96), p = 0.007). At MS and SP, there were no significant differences between the pre-HTO unaltered level gait and a wide stance gait.

Table 6-18 Pre-HTO Wide Stance Gait Internal Knee Joint Loading

	Controls	Pre-HTO NL	Pre-HTO WS	Controls vs pre-HTO NL	Controls vs pre-HTO WS	Pre NL vs Pre WS
	Mean (std)	Mean (std)	Mean (std)	P value	P value	P value
First peak						
Total knee						
Contact force [BW]	2.74 (0.57)	2.40 (0.58)	2.61 (0.73)	**0.032***	0.470	**0.002††**
Mean pressure [MPa]	5.63 (1.25)	6.10 (1.66)	6.50 (2.09)	0.396	0.120	**0.003††**
Max pressure	12.92 (3.32)	14.14 (4.20)	15.11 (5.69)	0.313	0.124	**0.024†**
Medial knee						
Contact force [BW]	1.69 (0.31)	1.59 (0.34)	1.63 (0.36)	0.233	0.485	0.198
Mean pressure [MPa]	5.76 (1.12)	6.40 (1.69)	6.56 (1.80)	0.097	**0.048***	0.102
Max pressure	12.21 (2.52)	13.28 (3.76)	13.85 (4.10)	0.210	0.074	**0.028***
Lateral knee						
Contact force [BW]	1.13 (0.37)	0.89 (0.38)	1.06 (0.52)	**0.017***	0.353	**0.001††**
Mean pressure [MPa]	5.37 (1.62)	5.47 (2.12)	6.23 (2.76)	0.880	0.361	**0.001††**
Max pressure	11.57 (3.76)	11.67 (4.69)	13.52 (6.37)	0.931	0.370	**0.001††**
Midstance						
Total knee						
Contact force [BW]	1.25 (0.24)	1.36 (0.25)	1.41 (0.38)	**0.031†**	0.089	0.642
Mean pressure [MPa]	3.40 (0.36)	4.39 (0.82)	4.49 (0.98)	**0.000****	**0.000****	0.974
Max pressure	7.62 (1.12)	10.01 (2.05)	10.23 (2.80)	**0.000****	**0.000††**	0.294
Medial knee						
Contact force [BW]	0.92 (0.22)	1.03 (0.26)	1.02 (0.29)	0.093	0.123	0.256
Mean pressure [MPa]	3.81 (0.58)	4.81 (0.92)	4.74 (0.88)	**0.000****	**0.000****	0.117
Max pressure	7.43 (1.22)	9.76 (2.10)	9.46 (1.96)	**0.000****	**0.000****	**0.014†**
Lateral knee						
Contact force [BW]	0.36 (0.14)	0.37 (0.21)	0.42 (0.35)	0.918	0.968	0.552
Mean pressure [MPa]	2.59 (0.54)	3.08 (1.37)	3.22 (1.91)	0.164	0.270	0.991
Max pressure	5.56 (1.07)	6.40 (2.85)	6.78 (4.24)	0.224	0.353	0.837
Second peak						
Total knee						
Contact force [BW]	2.90 (0.70)	2.49 (0.63)	2.51 (0.68)	**0.023***	**0.040***	0.417
Mean pressure [MPa]	5.32 (0.64)	6.02 (1.07)	6.06 (1.19)	**0.004****	**0.012†**	0.417

Max pressure	12.70 (1.76)	14.19 (3.06)	14.33 (4.05)	0.102	0.184	0.469
Medial knee						
Contact force [BW]	1.83 (0.35)	1.63 (0.48)	1.52 (0.53)	0.083	**0.016***	**0.005†**
Mean pressure [MPa]	5.91 (0.80)	6.44 (1.37)	6.15 (1.37)	0.077	0.408	**0.002***
Max pressure	12.61 (1.80)	13.18 (2.87)	12.58 (2.82)	0.376	0.960	**0.003***
Lateral knee						
Contact force [BW]	1.20 (0.44)	0.95 (0.37)	1.08 (0.46)	**0.028***	0.184	**0.006††**
Mean pressure [MPa]	4.53 (0.73)	5.25 (1.46)	5.69 (1.83)	**0.025†**	**0.002††**	**0.008††**
Max pressure	9.43 (1.48)	11.33 (3.58)	12.17 (4.54)	**0.011†**	**0.001††**	**0.013†**

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. BW = body weight; MPa = megapascals.

Table 6-19 Pre-HTO Wide Stance Gait Contact Force Ratios

	Controls	**Pre-HTO NL**	**Pre-HTO WS**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO WS**	**Pre NL vs Pre WS**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
MED / TOTAL	0.63 (0.07)	0.67 (0.10)	0.64 (0.10)	0.051	0.556	**0.010****
LAT / TOTAL	0.41 (0.07)	0.36 (0.10)	0.39 (0.10)	0.051	0.545	**0.011****
Midstance						
MED / TOTAL	0.73 (0.09)	0.76 (0.14)	0.74 (0.17)	0.451	0.698	0.325
LAT / TOTAL	0.29 (0.10)	0.27 (0.15)	0.28 (0.18)	0.471	0.698	0.443
Second peak						
MED / TOTAL	0.64 (0.07)	0.65 (0.12)	0.61 (0.13)	0.643	0.221	**0.000††**
LAT / TOTAL	0.40 (0.07)	0.38 (0.12)	0.43 (0.14)	0.542	0.303	**0.000††**

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. MED = medial compartment contact force; LAT = lateral compartment contact force; TOTAL = total tibiofemoral contact force.

Table 6-20 Pre-HTO Wide Stance Gait Point of Application

	Controls	Pre-HTO NL	Pre-HTO WS	Controls vs pre-HTO NL	Controls vs pre-HTO WS	Pre NL vs Pre WS
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total knee						
Anterior (+) / posterior (-)	-2.46 (2.08)	-2.07 (2.98)	-2.43 (3.26)	0.757	0.686	0.064
Lateral (+) / medial (-)	-2.27 (3.27)	-4.91 (4.19)	-3.67 (4.42)	**0.011†**	0.180	**0.031***
Medial knee						
Anterior (+) / posterior (-)	-1.01 (2.25)	-0.30 (3.27)	-0.61 (3.46)	0.387	0.722	0.403
Lateral (+) / medial (-)	-17.19 (1.29)	-18.08 (1.52)	-18.15 (1.74)	**0.021***	**0.021***	0.565
Lateral knee						
Anterior (+) / posterior (-)	-4.67 (1.83)	-5.12 (2.50)	-5.39 (2.76)	0.169	0.253	0.103
Lateral (+) / medial (-)	20.59 (2.34)	19.69 (2.14)	19.80 (2.29)	0.133	0.204	0.637
Midstance						
Total knee						
Anterior (+) / posterior (-)	4.66 (1.86)	4.07 (2.26)	3.69 (3.05)	0.291	0.156	0.754
Lateral (+) / medial (-)	-5.84 (4.72)	-7.53 (7.14)	-7.28 (7.98)	0.300	0.410	0.496
Medial knee						
Anterior (+) / posterior (-)	6.63 (2.77)	6.42 (2.69)	5.83 (3.02)	0.765	0.299	0.174
Lateral (+) / medial (-)	-16.33 (1.53)	-17.09 (2.14)	-17.29 (2.15)	**0.045†**	**0.022†**	0.300
Lateral knee						
Anterior (+) / posterior (-)	-0.42 (1.55)	-2.18 (2.21)	-1.83 (2.53)	**0.001****	**0.014***	0.256
Lateral (+) / medial (-)	20.34 (2.43)	18.91 (3.55)	18.41 (5.42)	0.081	0.145	0.673
Second peak						
Total knee						
Anterior (+) / posterior (-)	5.70 (1.64)	4.45 (3.36)	3.89 (3.99)	0.370	0.361	0.090
Lateral (+) / medial (-)	-1.39 (2.90)	-2.59 (5.13)	-0.62 (5.84)	0.086	0.956	**0.000††**
Medial knee						
Anterior (+) / posterior (-)	9.56 (1.83)	8.15 (3.38)	7.53 (4.16)	0.218	0.218	0.084
Lateral (+) / medial (-)	-14.59 (1.31)	-15.83 (1.76)	-15.68 (1.93)	**0.004****	**0.015†**	0.524
Lateral knee						
Anterior (+) / posterior (-)	-0.89 (2.43)	-2.59 (3.39)	-1.97 (3.73)	**0.034***	0.201	**0.045†**
Lateral (+) / medial (-)	19.74 (2.84)	19.48 (3.40)	20.26 (3.38)	0.755	0.532	**0.016****

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. mm = millimetres. WS = wide stance gait; NL = unaltered level gait. X = anterior; Z = lateral. COP = centre of pressure.

Table 6-21 Pre-HTO Wide Stance Gait Knee Contact Area

	Controls	Pre-HTO NL	Pre-HTO WS	Controls vs pre-HTO NL	Controls vs pre-HTO WS	Pre NL vs Pre WS
mm^2	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total	352.35 (39.00)	364.30 (49.61)	372.04 (47.95)	0.249	0.063	**0.040***
Medial	206.80 (22.63)	223.63 (29.53)	223.69 (29.87)	**0.012†**	**0.015†**	0.978
Lateral	145.55 (20.95)	140.68 (28.69)	148.35 (28.96)	0.794	0.563	**0.007***
Midstance						
Total	262.26 (34.10)	286.65 (55.88)	287.79 (62.65)	0.080	0.102	0.837
Medial	167.14 (24.46)	189.50 (36.97)	189.90 (35.87)	**0.010***	**0.009†**	0.642
Lateral	95.12 (19.62)	97.16 (35.50)	97.90 (45.34)	0.698	0.698	0.957
Second peak						
Total	399.23 (90.44)	383.65 (91.11)	384.70 (87.65)	0.500	0.521	0.905
Medial	219.07 (41.52)	223.86 (50.52)	216.92 (50.99)	0.698	0.863	0.071
Lateral	180.16 (52.60)	159.80 (48.03)	167.77 (45.71)	0.132	0.387	0.112

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. mm^2 = millimetres squared.

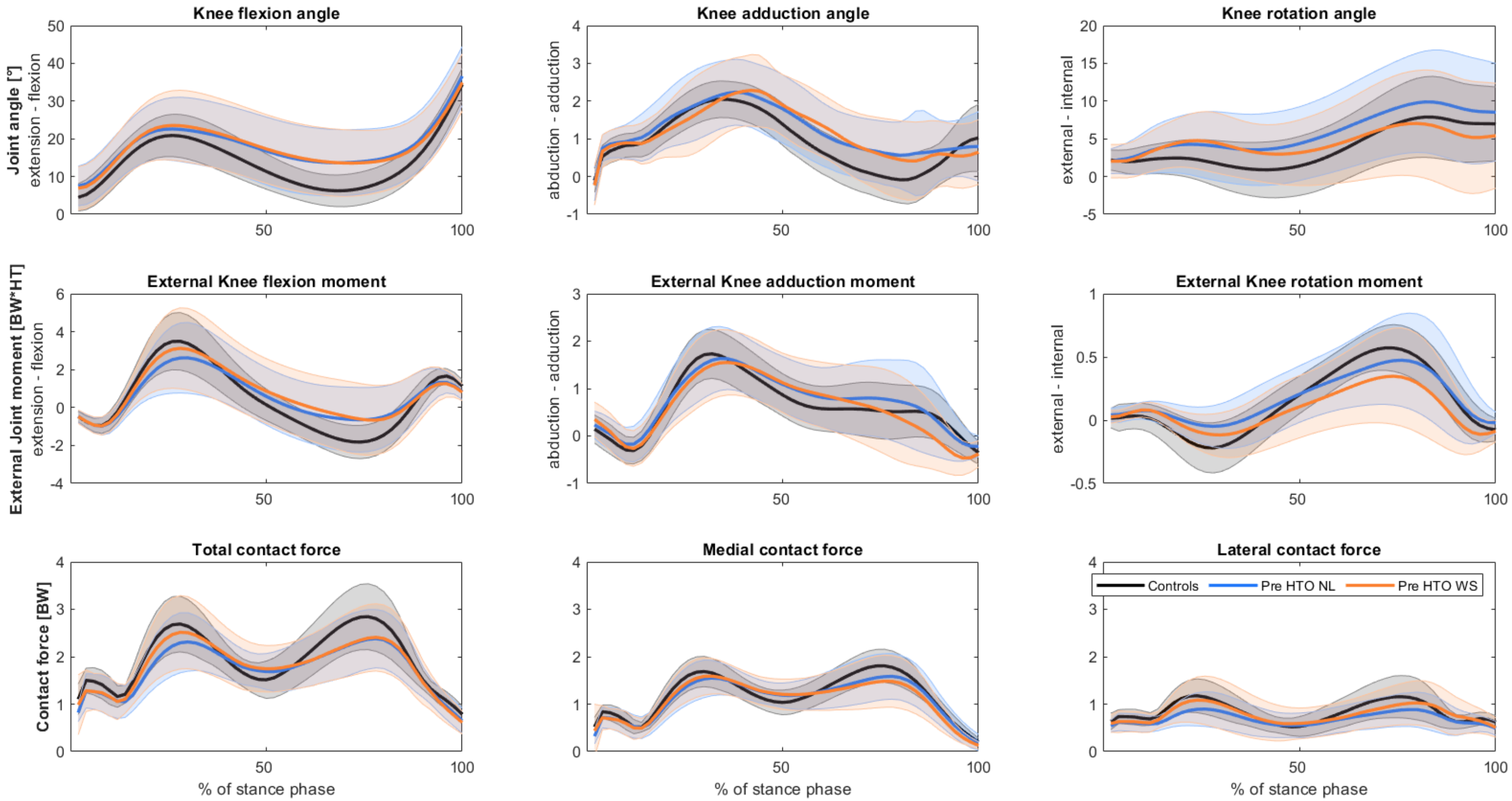


Figure 43 Pre-HTO wide stance gait knee joint kinematics, external moments, contact forces

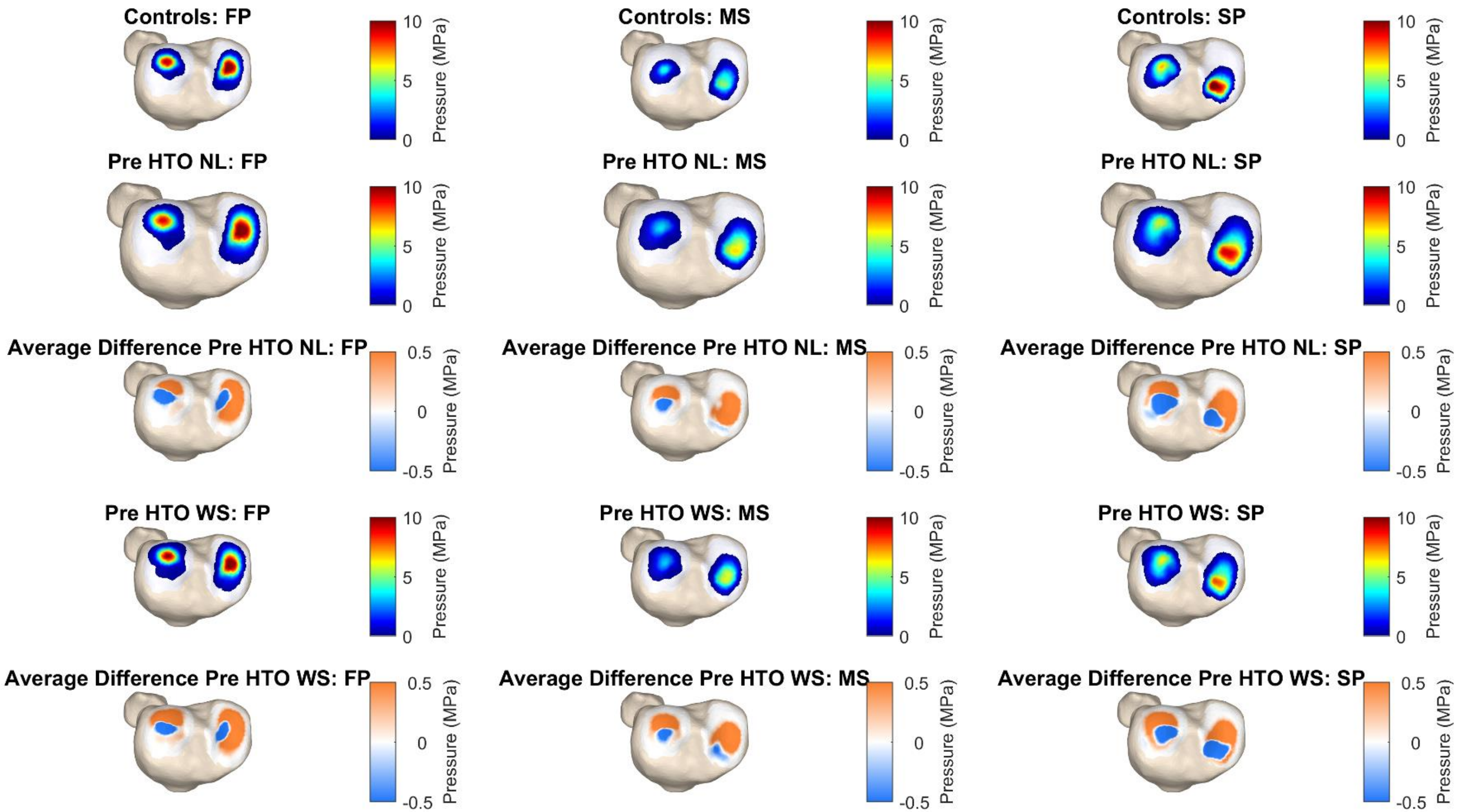


Figure 44 Pre-HTO wide stance gait contact pressure distribution on the tibia

6.5 Medial thrust gait

6.5.1 Quantifying medial thrust gait

The metric used to determine whether an individual successfully adapted their gait to a medial thrust gait was whether an individual could decrease their knee adduction angle during the first half of stance compared to when walking with an unaltered level gait. Only 20 patients were able to successfully adopt a medial thrust altered gait style pre-surgery. Adopting a medial thrust gait resulted in a maximum knee adduction angle in the first half of stance of ~6° compared a pre-HTO unaltered level gait maximum knee adduction angle in the first half of stance of ~8° (p = 0.000).

Table 6-22 Pre-HTO Medial Thrust Gait Quantifying Gait Style

	Controls	**Pre-HTO NL**	**Pre-HTO MT**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO MT**	**Pre NL vs Pre MT**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Maximum knee adduction angle in first half of stance (°)	1.53 (3.87)	7.73 (4.52)	5.86 (3.97)	**0.000††**	**0.000****	**0.000††**

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. std = standard deviation. m = metre.

6.5.2 Spatial-temporal parameters

Adopting a medial thrust gait resulted in many spatiotemporal changes including a reduction in stride length and an increased stance time.

Table 6-23 Pre-HTO Medial Thrust Gait Spatial Temporal Parameters

	Controls	**Pre-HTO NL**	**Pre-HTO MT**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO MT**	**Pre NL vs Pre MT**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Operative limb cycle time (s)	1.08 (0.08)	1.16 (0.17)	1.25 (0.24)	**0.021†**	**0.000††**	**0.001††**
Operative limb stance time (s)	0.65 (0.06)	0.73 (0.14)	0.80 (0.20)	**0.002††**	**0.000††**	**0.005††**
Operative limb step length (m)	0.64 (0.07)	0.59 (0.11)	0.57 (0.12)	**0.046***	**0.024***	**0.018***
Operative limb step time (s)	0.54 (0.04)	0.58 (0.08)	0.62 (0.10)	**0.035***	**0.000††**	**0.000††**
Operative limb stride length (m)	1.29 (0.13)	1.19 (0.21)	1.15 (0.23)	**0.048***	**0.019***	**0.017***
Swing time (s)	0.43 (0.03)	0.43 (0.04)	0.45 (0.05)	0.815	0.111	**0.003††**
Speed (m/s)	1.21 (0.16)	1.06 (0.26)	0.96 (0.25)	**0.025***	**0.001****	**0.001††**

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. std = standard deviation. s = seconds; m = metre; m/s = metre/second.

6.5.3 Knee joint loading: External moments

Adopting a medial thrust gait significantly reduced maximum EKAM when compared to an unaltered level gait pre-surgery (3.21 %BW.h (1.09) vs 2.86 %BW.h (1.05), p = 0.000).

Adopting a medial thrust gait significantly reduced EKAM1 (3.17 %BW.h (1.09) vs 2.68 %BW.h (0.86), p = 0.001) and EKAM2 (2.49 %BW.h (0.98) vs 2.3 %BW.h (1.07), p = 0.041) when compared to an unaltered level gait pre-surgery.

Adopting a medial thrust gait significantly increased peak flexion moment when compared to pre-HTO unaltered level gait (3.04 %BW.h (1.57) vs 4.05 %BW.h (1.88), p = 0.000), as well as significantly reducing peak extension moment (1.84 %BW.h (0.81) vs 1.37 %BW.h (0.83), p = 0.001).

In terms of the peak transverse plane knee moment changes, adopting a medial thrust gait compared to unaltered level gait significantly reduced peak internal rotation moment (0.96 %BW.h (0.39) vs 0.84 %BW.h (0.33), p = 0.037) and significantly increased peak external rotation moment (0.13 %BW.h (0.09) vs 0.18 %BW.h (0.13), p = 0.036).

Table 6-24 Pre-HTO Medial Thrust Gait External Knee Moments

	Controls	Pre-HTO NL	Pre-HTO MT	Controls vs pre-HTO NL	Controls vs pre-HTO MT	Pre NL vs Pre MT
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) moment						
Maximum	2.11 (0.81)	3.21 (1.09)	2.86 (1.05)	**0.000****	**0.008****	**0.000****
1st peak (1st half stance)	2.27 (0.65)	3.17 (1.09)	2.68 (0.86)	**0.003****	0.067	**0.001††**
2nd peak (2nd half stance)	1.50 (0.67)	2.49 (0.98)	2.30 (1.07)	**0.000****	**0.003****	**0.041***
Midstance	1.15 (0.49)	2.20 (0.78)	2.08 (1.07)	**0.000****	**0.001††**	0.093
Flexion (+) moment peak	3.62 (1.65)	3.04 (1.57)	4.05 (1.88)	0.227	0.406	**0.000****
Extension (−) moment peak	-2.45 (0.85)	-1.84 (0.81)	-1.37 (0.83)	**0.031†**	**0.000††**	**0.001††**
Internal (+) rotation moment peak	0.60 (0.37)	0.96 (0.39)	0.84 (0.33)	**0.002****	**0.023****	**0.037†**
External (−) rotation moment peak	-0.16 (0.08)	-0.13 (0.09)	-0.18 (0.13)	0.187	0.611	**0.036***

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. % BW.h = percentage of body weight multiplied by height.

6.5.4 Knee joint loading: Impulses

Adopting a medial thrust gait significantly reduced KAAI over the first half of stance when compared to pre-surgery unaltered level gait (0.74 %BW.h.s (0.23) vs 0.70 %BW.h.s (0.35), $p = 0.028$). Additionally, between heel strike and 16% of stance, adopting a medial thrust gait significantly reduced KAAI (0.11 %BW.h.s (0.06) vs 0.09 %BW.h.s (0.05), $p = 0.003$).

When compared to pre-HTO unaltered level gait, a medial thrust gait significantly increased first half of stance abduction angular impulse ($p = 0.002$).

Table 6-25 Pre-HTO Medial Thrust Gait Knee Angular Impulse

	Controls	Pre-HTO NL	Pre-HTO MT	Controls vs pre-HTO NL	Controls vs pre-HTO MT	Pre NL vs Pre MT
%BW.h.s	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) angular impulse						
Stance	0.74 (0.28)	1.34 (0.45)	1.34 (0.85)	**0.000****	**0.000††**	0.067
1st half stance	0.43 (0.14)	0.74 (0.23)	0.70 (0.35)	**0.000****	**0.000††**	**0.028†**
2nd half stance	0.31 (0.16)	0.60 (0.23)	0.64 (0.51)	**0.000****	**0.000††**	0.313
0–16% stance	0.06 (0.03)	0.11 (0.06)	0.09 (0.05)	**0.001††**	**0.025***	**0.003††**
17%–midstance	0.36 (0.11)	0.62 (0.17)	0.61 (0.31)	**0.000****	**0.000††**	0.108
Midstance–83% stance	0.26 (0.13)	0.51 (0.19)	0.54 (0.41)	**0.000****	**0.000††**	0.411
84%-100% stance	0.04 (0.02)	0.06 (0.04)	0.07 (0.08)	**0.014***	**0.039†**	0.709
Abduction (−) angular impulse in Stance						
1st half stance	-0.02 (0.01)	-0.01 (0.01)	-0.01 (0.01)	**0.002††**	0.097	**0.002††**
2nd half stance	-0.02 (0.02)	-0.01 (0.01)	-0.01 (0.01)	0.058	**0.017†**	0.126

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. % BW.h.s = % of body weight multiplied by height per second.

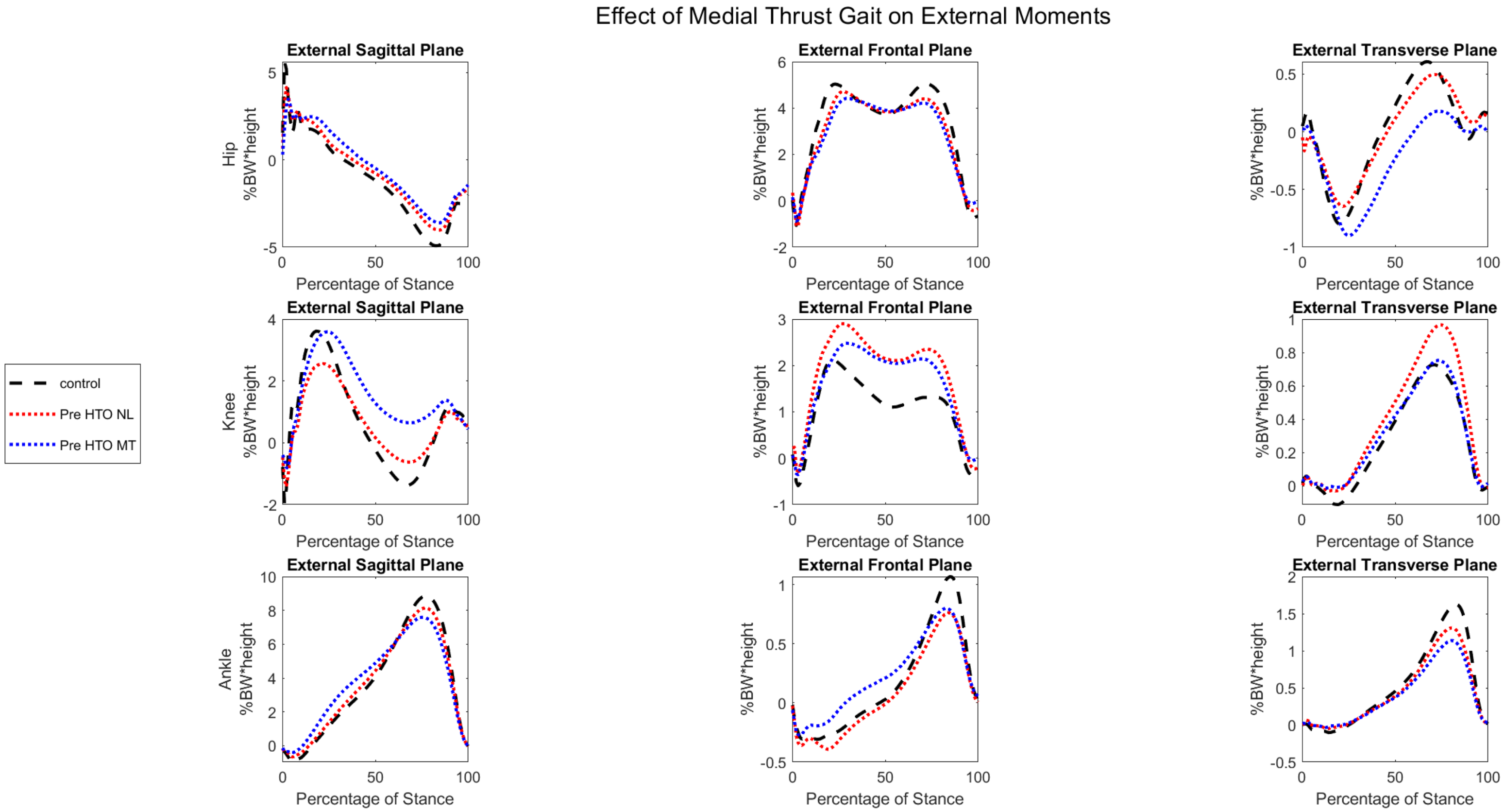


Figure 45 Visual 3D: Pre-HTO medial thrust gait group average external moments

Positive values represent external moments for knee flexion, adduction, and internal rotation moments.

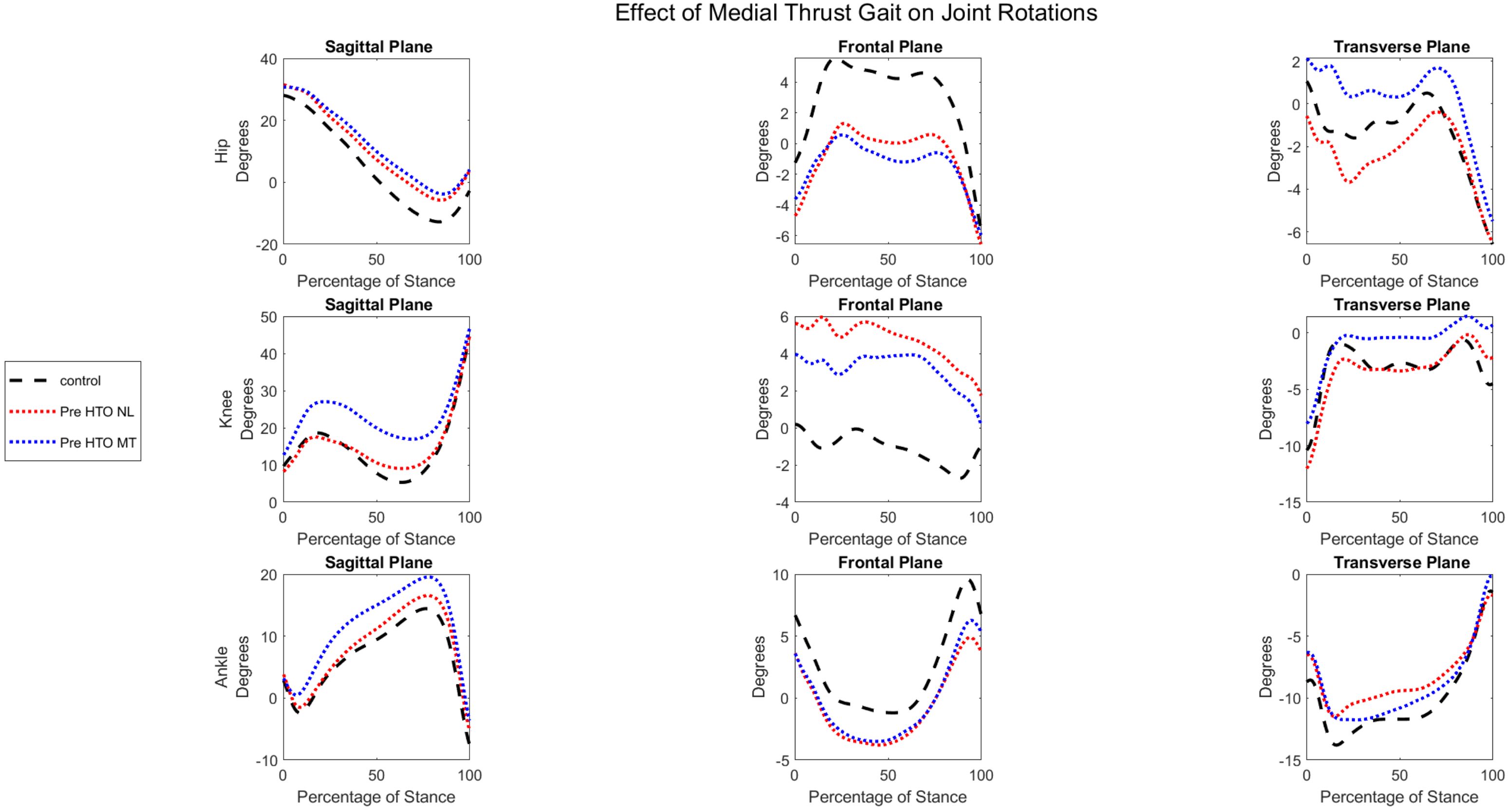


Figure 46 Visual 3D: Pre-HTO medial thrust gait group average joint kinematics

Positive values represent knee flexion, adduction, and internal rotations.

6.5.5 External ankle moments

When adopting a medial thrust gait compared to an unaltered level gait, most changes that occurred with external ankle moments occurred in the first half of stance.

During the first half of stance, peak external dorsiflexion moment significantly increased when adopting a medial thrust gait style (4.2 %BW.h (0.74) vs 5.00 %BW.h (1.03), p = 0.009), as well as a significant decrease in peak external plantarflexion moment (0.75 %BW.h (0.41) vs 0.58 %BW.h (0.48), p = 0.030).

In the frontal plane there was a significant increase in peak external inversion moment when adopting a medial thrust gait compared to a pre-HTO unaltered level gait (0.15 %BW.h (0.21) vs 0.31 %BW.h (0.27), p = 0.004), as well as significantly reducing peak external eversion moment (0.49 %BW.h (0.3) vs 0.38 %BW.h (0.27), p = 0.006). In the transverse plane, adopting a medial thrust gait resulted in a significant increase in peak external rotation moment compared to a pre-HTO unaltered level gait (p = 0.043).

During the second half of stance, the only significant difference between pre-HTO unaltered level gait and adopting a medial thrust gait was in the frontal plane peak external eversion moment which significantly decreased when adopting a medial thrust gait (0.17 %BW.h (0.19) vs 0.07 %BW.h (0.11), p = 0.032).

Table 6-26 Pre-HTO Medial Thrust Gait External Ankle Moments

	Controls	Pre-HTO NL	Pre-HTO MT	Controls vs pre-HTO NL	Controls vs pre-HTO MT	Pre NL vs Pre MT
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak dorsi-flexion (+) moment	6.78 (3.93)	8.05 (1.33)	7.81 (1.28)	0.294	0.097	0.145
Peak plantar-flexion (−) moment	-0.97 (0.36)	-0.75 (0.41)	-0.61 (0.43)	0.065	**0.003****	**0.033†**
Peak inversion (+) moment	0.82 (0.54)	0.88 (0.52)	0.90 (0.53)	1.000	0.992	0.762
Peak eversion (-) moment	-0.41 (0.22)	-0.49 (0.30)	-0.38 (0.26)	0.287	0.366	**0.007††**
Peak internal rotation (+) moment	1.25 (0.85)	1.31 (0.69)	1.23 (0.80)	0.583	0.449	0.167
Peak external rotation (−) moment	-0.17 (0.08)	-0.13 (0.09)	-0.16 (0.15)	0.091	0.612	0.156
0-50% (HS to midstance)						
Peak dorsi-flexion (+) moment	4.08 (0.99)	4.42 (0.74)	5.00 (1.03)	0.196	**0.003****	**0.009****
Peak plantar-flexion (−) moment	-1.03 (0.28)	-0.75 (0.41)	-0.58 (0.48)	**0.009****	**0.001****	**0.030†**
Peak inversion (+) moment	0.14 (0.14)	0.15 (0.21)	0.31 (0.27)	0.812	**0.018†**	**0.004****
Peak eversion (-) moment	-0.43 (0.20)	-0.49 (0.30)	-0.38 (0.27)	0.764	0.147	**0.006††**
Peak internal rotation (+) moment	0.47 (0.26)	0.39 (0.18)	0.44 (0.31)	0.224	0.671	0.765
Peak external rotation (−) moment	-0.18 (0.08)	-0.12 (0.10)	-0.15 (0.12)	**0.026***	0.374	**0.043***
50-100% (Midstance to toe-off)						
Peak dorsi-flexion (+) moment	8.94 (0.74)	8.05 (1.33)	7.80 (1.29)	**0.000††**	**0.000††**	0.145
Peak plantar-flexion (−) moment	-0.03 (0.12)	-0.04 (0.09)	-0.02 (0.11)	0.643	0.758	0.210
Peak inversion (+) moment	1.11 (0.34)	0.88 (0.52)	0.89 (0.53)	0.069	**0.024†**	0.818
Peak eversion (-) moment	-0.11 (0.14)	-0.17 (0.19)	-0.07 (0.11)	0.275	0.284	**0.032***
Peak internal rotation (+) moment	1.67 (0.53)	1.31 (0.69)	1.22 (0.82)	**0.004††**	**0.001††**	0.156
Peak external rotation (−) moment	0.00 (0.06)	0.00 (0.05)	-0.02 (0.16)	0.656	0.284	0.351

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. % BW.h = percentage of body weight multiplied by height. HS = heel strike.

6.5.6 External hip moments

When adopting a medial thrust gait compared to an unaltered level gait, most changes that occurred with external hip moments occurred in the second half of stance.

During the first half of stance, peak external hip adduction moment significantly reduced when adopting a medial thrust gait style (5.1 %BW.h (1.19) vs 4.89 %BW.h (1.19), p = 0.038). A medial thrust altered gait significantly increased peak external hip rotation moment compared to pre-HTO unaltered level gait (0.77 %BW.h (0.45) vs 0.99 %BW.h (0.47), p = 0.001).

During the second half of stance, peak external hip adduction moment significantly reduced when adopting a medial thrust gait style (4.62 %BW.h (1.28) vs 4.54 %BW.h (1.52), p = 0.048). Peak external hip abduction moment significantly decreased when adopting a medial thrust gait style (0.58 %BW.h (0.5) vs 0.24 %BW.h (0.62), p = 0.009). Peak external hip internal rotation moment significantly decreased when adopting a medial thrust gait style (0.5 %BW.h (0.32) vs 0.3 %BW.h (0.23), p = 0.006), whilst peak external hip external moment significantly increased when adopting a medial thrust gait style (0.09 %BW.h (0.2) vs 0.29 %BW.h (0.32), p = 0.005).

Table 6-27 Pre-HTO Medial Thrust Gait External Hip Moments

	Controls	**Pre-HTO NL**	**Pre-HTO MT**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO MT**	**Pre NL vs Pre MT**
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak external hip flexion (+) moment	6.31 (2.64)	4.87 (1.83)	4.91 (2.04)	**0.040***	**0.037†**	0.765
Peak external hip extension (−) moment	-4.36 (1.70)	-4.34 (1.54)	-3.95 (1.83)	0.968	0.422	**0.006††**
Peak external hip adduction (+) moment	5.09 (1.41)	5.24 (1.19)	5.10 (1.28)	0.893	0.963	**0.037†**
Peak external hip abduction (-) moment	-1.65 (1.03)	-1.52 (0.89)	-1.23 (0.92)	0.780	0.070	**0.017†**
Peak external hip internal rotation (+) moment	0.54 (0.30)	0.51 (0.32)	0.34 (0.20)	0.717	**0.014***	**0.012†**
Peak external hip rotation (−) moment	-0.80 (0.43)	-0.77 (0.45)	-0.99 (0.47)	0.816	0.150	**0.001****
0-50% (HS to midstance)						
Peak external hip extension (−) moment	-1.25 (0.92)	-1.13 (0.86)	-0.92 (0.95)	0.437	0.242	0.135
Peak external hip adduction (+) moment	5.27 (1.03)	5.10 (1.19)	4.89 (1.19)	0.617	0.248	**0.038***
Peak external hip abduction (-) moment	-1.28 (1.29)	-1.22 (1.11)	-1.11 (1.00)	0.959	0.641	0.526
Peak external hip internal rotation (+) moment	0.34 (0.20)	0.21 (0.23)	0.15 (0.13)	**0.009††**	**0.001****	0.243
Peak external hip rotation (−) moment	-0.86 (0.42)	-0.77 (0.45)	-0.99 (0.47)	0.481	0.316	**0.001****
50-100% (Midstance to toe-off)						
Peak external hip adduction (+) moment	5.20 (1.17)	4.62 (1.28)	4.54 (1.52)	0.111	0.097	**0.048†**
Peak external hip abduction (-) moment	-0.86 (0.55)	-0.58 (0.50)	-0.24 (0.62)	0.084	**0.001††**	**0.009††**
Peak external hip internal rotation (+) moment	0.64 (0.24)	0.50 (0.32)	0.30 (0.23)	**0.050***	**0.000****	**0.006††**
Peak external hip rotation (−) moment	-0.11 (0.12)	-0.09 (0.20)	-0.29 (0.32)	0.334	**0.039†**	**0.005††**

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. % BW.h = percentage of body weight multiplied by height. HS = heel strike.

6.5.7 Concurrent Optimisation of Muscle and Secondary Kinematics

6.5.7.1 Internal knee joint loading

At FP, total, medial compartment, and lateral compartment contact force, mean and maximum pressure significantly increased because of adopting a medial thrust gait compared to an unaltered level. When adopting a medial thrust gait compared to an unaltered level gait, medial compartment knee contact force increased from 1.56 BW (0.39) to 1.67 BW (0.46), $p = 0.018$, whilst the lateral compartment of the knee increased from 0.94 BW (0.38) to 1.22 BW (0.47), $p = 0.000$.

At MS, total and lateral compartment contact force, mean and maximum pressure significantly increased when adopting a medial thrust gait compared to an unaltered level gait. Lateral compartment of the knee increased from 0.4 BW (0.22) to 0.51 BW (0.31), $p = 0.026$. In the medial compartment, maximum pressure increased because of adopting a medial thrust gait style compared to pre-HTO unaltered level gait (9.61 MPa (2.13) vs 10.12 MPa (2.33), $p = 0.040$).

At SP, the only significant change that occurred due to adopting a medial thrust gait was that the medial compartment knee maximum pressure significantly decreased compared to an unaltered level gait (12.99 MPa (3.14) vs 12.36 MPa (3.83), $p = 0.049$).

6.5.7.2 Internal knee joint loading ratios

At FP, adopting a medial thrust gait significantly decreased the medial to total contact force ratio (0.65 (0.09) vs 0.6 (0.09), $p = 0.005$), and significantly increased the lateral to total contact force ratio (0.38 (0.09) vs 0.43 (0.1), $p = 0.008$) when compared to a pre-HTO unaltered level gait. No significant differences occurred at MS or SP.

6.5.7.3 Point of application

At FP, point of application shifted laterally when adopting a medial thrust gait compared to an unaltered gait (group mean change of ~2mm, $p = 0.011$). In the medial knee compartment, point of application was located significantly more medially for the patient cohort compared to the control group.

At SP, medial compartment and lateral knee compartment significantly moved laterally when adopting a medial thrust gait compared to pre-HTO unaltered level gait. However, both patient walks were overall more medial compared to the control cohort.

6.5.7.4 Contact area

At FP, there was a significant increase in lateral compartment contact area when adopting a medial thrust gait when compared to a pre-surgery unaltered level gait (145.02 mm^2 (20.95) vs 154.13 mm^2 (27.91), $p = 0.032$).

At MS and SP, there were no significant differences between the pre-HTO unaltered level gait and a medial thrust gait.

Table 6-28 Pre-HTO Medial Thrust Gait Internal Knee Joint Loading

	Controls	**Pre-HTO NL**	**Pre-HTO MT**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO MT**	**Pre NL vs Pre MT**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total knee						
Contact force [BW]	2.74 (0.57)	2.42 (0.65)	2.80 (0.75)	0.082	0.739	**0.000****
Mean pressure [MPa]	5.63 (1.25)	6.13 (1.81)	7.01 (2.14)	0.471	**0.024†**	**0.000****
Max pressure	12.92 (3.32)	14.16 (4.65)	16.23 (5.29)	0.420	**0.036†**	**0.001****
Medial knee						
Contact force [BW]	1.69 (0.31)	1.56 (0.39)	1.67 (0.46)	0.195	0.569	**0.018***
Mean pressure [MPa]	5.76 (1.12)	6.32 (1.79)	6.90 (2.11)	0.234	0.133	**0.001††**
Max pressure	12.21 (2.52)	13.03 (3.98)	14.25 (4.27)	0.432	0.286	**0.002****
Lateral knee						
Contact force [BW]	1.13 (0.37)	0.94 (0.38)	1.22 (0.47)	0.089	0.438	**0.000****
Mean pressure [MPa]	5.37 (1.62)	5.75 (2.21)	7.04 (2.59)	0.526	**0.008††**	**0.000****
Max pressure	11.57 (3.76)	12.34 (4.87)	15.07 (5.64)	0.540	**0.009††**	**0.000****
Midstance						
Total knee						
Contact force [BW]	1.25 (0.24)	1.36 (0.29)	1.49 (0.38)	0.133	**0.040†**	**0.046***
Mean pressure [MPa]	3.40 (0.36)	4.37 (0.83)	4.74 (1.05)	**0.000****	**0.000****	**0.004****
Max pressure	7.62 (1.12)	9.87 (2.10)	10.99 (2.54)	**0.000****	**0.000****	**0.003††**
Medial knee						
Contact force [BW]	0.92 (0.22)	0.99 (0.28)	1.03 (0.27)	0.289	0.132	0.411
Mean pressure [MPa]	3.81 (0.58)	4.74 (0.95)	4.95 (1.09)	**0.001****	**0.000††**	0.060
Max pressure	7.43 (1.22)	9.61 (2.13)	10.12 (2.33)	**0.000****	**0.000††**	**0.040***
Lateral knee						
Contact force [BW]	0.36 (0.14)	0.40 (0.22)	0.51 (0.31)	0.584	0.102	**0.026***
Mean pressure [MPa]	2.59 (0.54)	3.27 (1.37)	3.86 (1.86)	0.081	**0.012†**	**0.019***
Max pressure	5.56 (1.07)	6.87 (2.84)	8.05 (3.96)	0.070	**0.028†**	**0.024***
Second peak						
Total knee						
Contact force [BW]	2.90 (0.70)	2.42 (0.61)	2.29 (0.56)	**0.020***	**0.003****	0.271

Mean pressure [MPa]	5.32 (0.64)	5.95 (1.19)	5.90 (1.62)	0.102	0.420	0.147
Max pressure	12.70 (1.76)	14.03 (3.46)	14.05 (4.91)	0.372	0.889	0.260
Medial knee						
Contact force [BW]	1.83 (0.35)	1.57 (0.47)	1.45 (0.37)	**0.040***	**0.001****	0.084
Mean pressure [MPa]	5.91 (0.8)	6.34 (1.53)	6.06 (1.69)	0.271	0.756	0.070
Max pressure	12.61 (1.80)	12.99 (3.14)	12.36 (3.83)	0.637	0.267	**0.049**†
Lateral knee						
Contact force [BW]	1.20 (0.44)	0.94 (0.37)	0.93 (0.45)	**0.047****	**0.050***	0.546
Mean pressure [MPa]	4.53 (0.73)	5.24 (1.47)	5.44 (2.34)	0.057	0.177	0.809
Max pressure	9.43 (1.48)	11.38 (3.71)	11.67 (5.40)	**0.017**†	0.157	0.841

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. BW = body weight; MPa = megapascals. Max = maximum.

Table 6-29 Pre-HTO Medial Thrust Gait Contact Force Ratios

	Controls	**Pre-HTO NL**	**Pre-HTO MT**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO MT**	**Pre NL vs Pre MT**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
MED / TOTAL	0.63 (0.07)	0.65 (0.09)	0.60 (0.09)	0.239	0.328	**0.005****
LAT / TOTAL	0.41 (0.07)	0.38 (0.09)	0.43 (0.10)	0.259	0.346	**0.008****
Midstance						
MED / TOTAL	0.73 (0.09)	0.73 (0.14)	0.70 (0.15)	0.974	0.240	0.097
LAT / TOTAL	0.29 (0.10)	0.30 (0.15)	0.33 (0.16)	0.974	0.258	0.131
Second peak						
MED / TOTAL	0.64 (0.07)	0.64 (0.12)	0.64 (0.12)	0.890	0.929	0.920
LAT / TOTAL	0.40 (0.07)	0.39 (0.12)	0.39 (0.13)	0.792	0.844	0.910

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. MED = medial compartment contact force; LAT = lateral compartment contact force; TOTAL = total tibiofemoral contact force.

Table 6-30 Pre-HTO Medial Thrust Gait Point of Application

	Controls	Pre-HTO NL	Pre-HTO MT	Controls vs pre-HTO NL	Controls vs pre-HTO MT	Pre NL vs Pre MT
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total knee						
Anterior (+) / posterior (-)	-2.46 (2.08)	-1.96 (3.21)	-3.78 (3.08)	0.629	**0.036†**	**0.000****
Lateral (+) / medial (-)	-2.27 (3.27)	-4.24 (3.49)	-2.23 (3.44)	0.055	0.964	**0.011***
Medial knee						
Anterior (+) / posterior (-)	-1.01 (2.25)	0.17 (3.58)	-1.66 (3.09)	0.258	0.396	**0.001††**
Lateral (+) / medial (-)	-17.19 (1.29)	-18.07 (1.51)	-18.37 (1.90)	**0.037***	**0.016†**	0.141
Lateral knee						
Anterior (+) / posterior (-)	-4.67 (1.83)	-5.52 (2.25)	-6.67 (2.52)	0.162	**0.003****	**0.007****
Lateral (+) / medial (-)	20.59 (2.34)	19.69 (1.91)	20.35 (2.07)	0.169	0.719	0.104
Midstance						
Total knee						
Anterior (+) / posterior (-)	4.66 (1.86)	3.92 (2.28)	3.72 (3.97)	0.234	0.350	0.763
Lateral (+) / medial (-)	-5.84 (4.72)	-6.46 (7.04)	-4.89 (6.69)	0.720	0.571	0.125
Medial knee						
Anterior (+) / posterior (-)	6.63 (2.77)	6.57 (2.58)	6.49 (3.33)	0.935	0.868	0.899
Lateral (+) / medial (-)	-16.33 (1.53)	-16.88 (2.11)	-16.80 (1.88)	0.089	0.360	0.752
Lateral knee						
Anterior (+) / posterior (-)	-0.42 (1.55)	-2.51 (1.98)	-2.55 (3.55)	**0.000****	**0.022***	0.955
Lateral (+) / medial (-)	20.34 (2.43)	19.39 (3.17)	19.91 (4.70)	0.252	0.822	0.212
Second peak						
Total knee						
Anterior (+) / posterior (-)	5.70 (1.64)	4.14 (3.53)	2.50 (4.43)	0.258	**0.004††**	**0.035***
Lateral (+) / medial (-)	-1.39 (2.90)	-2.23 (5.42)	-2.72 (4.87)	0.258	0.215	0.419
Medial knee						
Anterior (+) / posterior (-)	9.56 (1.83)	7.85 (3.53)	6.22 (3.86)	0.063	**0.002****	**0.029***
Lateral (+) / medial (-)	-14.59 (1.31)	-15.90 (2.08)	-16.48 (2.11)	**0.021***	**0.001††**	**0.025***
Lateral knee						
Anterior (+) / posterior (-)	-0.89 (2.43)	-2.77 (3.39)	-4.25 (3.17)	**0.032***	**0.000****	**0.040†**

Lateral (+) / medial (-)	19.74 (2.84)	19.71 (3.55)	19.51 (2.89)	0.970	0.788	0.711

Significant difference (p < 0.01) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. mm = millimetres. X = anterior; Z = lateral. COP = centre of pressure.

Table 6-31 Pre-HTO Medial Thrust Gait Knee Contact Area

	Controls	**Pre-HTO NL**	**Pre-HTO MT**	**Controls vs pre-HTO NL**	**Controls vs pre-HTO MT**	**Pre NL vs Pre MT**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total	352.35 (39.00)	367.47 (44.76)	371.84 (44.59)	0.177	0.117	0.198
Medial	206.80 (22.63)	222.46 (22.73)	217.70 (20.86)	**0.012†**	**0.042††**	0.184
Lateral	145.55 (20.95)	145.02 (26.64)	154.13 (27.91)	0.923	0.249	**0.032***
Midstance						
Total	262.26 (34.10)	288.62 (54.06)	291.61 (59.81)	**0.046***	**0.038***	0.744
Medial	167.14 (24.46)	186.14 (33.77)	185.14 (33.83)	**0.030***	**0.040***	0.856
Lateral	95.12 (19.62)	102.47 (35.22)	106.46 (36.11)	0.296	0.117	0.459
Second peak						
Total	399.23 (90.44)	380.12 (86.16)	367.57 (75.74)	0.473	0.216	0.272
Medial	219.07 (41.52)	220.82 (49.23)	217.95 (36.90)	0.896	0.889	0.674
Lateral	180.16 (52.60)	159.30 (45.85)	149.62 (44.11)	0.167	**0.043***	0.097

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. mm^2 = millimetres squared.

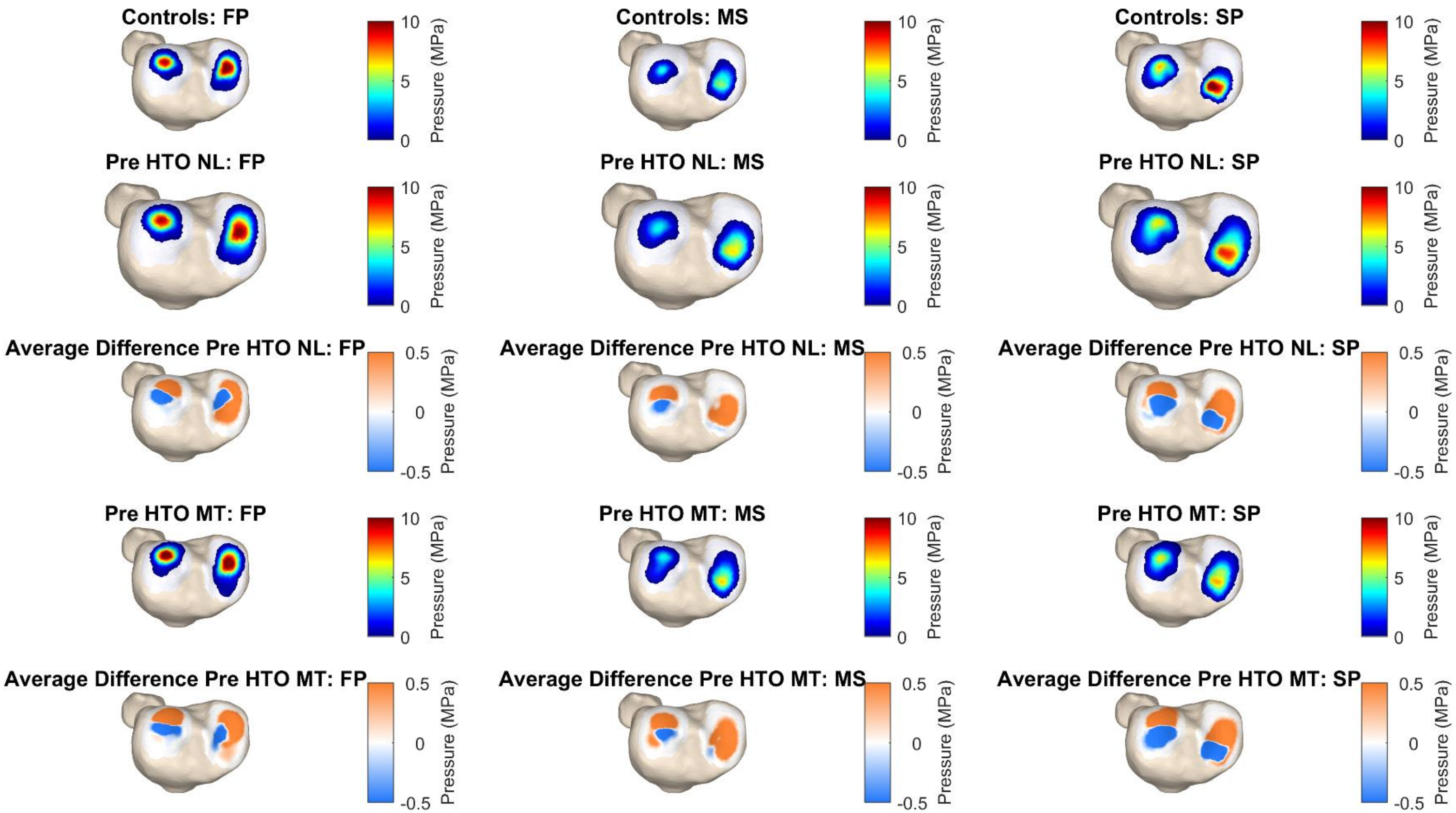


Figure 47 Pre-HTO medial thrust gait contact pressure distribution on the tibia

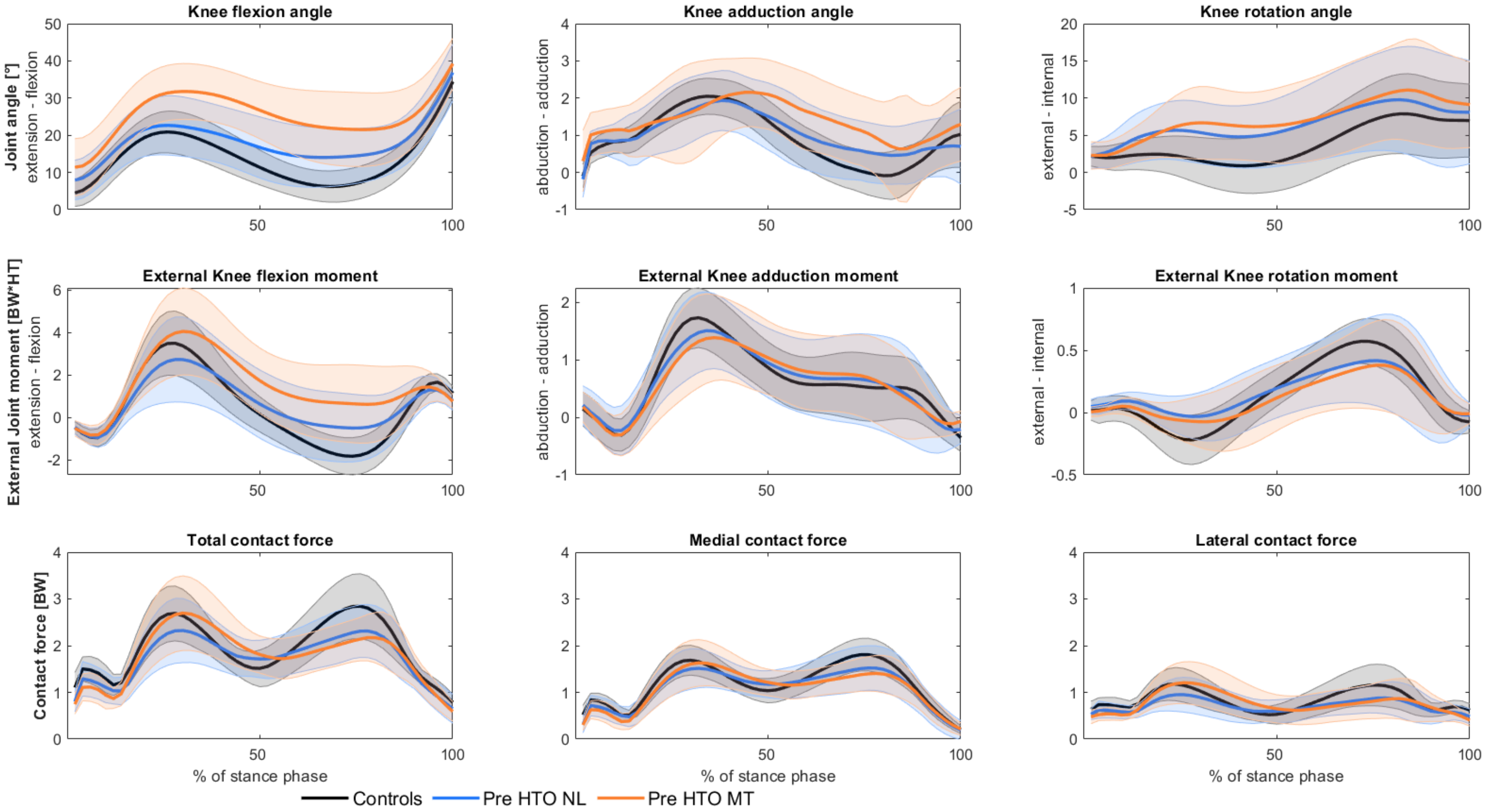


Figure 48 Pre-HTO medial thrust gait knee joint kinematics, external moments, contact forces

Average contact pressure patterns at first peak, midstance and second peak for the control group and the patients pre- and post-HTO. Furthermore, the average difference between the pressure pattern in patients and the healthy control pressure pattern is shown. Orange indicates more loading in the patient on that specific location, blue indicates decreased loading compared to the controls.

6.6 Conclusion

The objective of this chapter was to quantify the biomechanical differences of knee joint loading between a control group, pre-HTO unaltered level gait, and 3 pre-HTO altered gait styles. Accordingly, this chapter addressed lower limb biomechanical differences between a control cohort and pre-HTO unaltered level gait, and biomechanical changes resulting from pre-HTO altered gait styles to assess whether altering gait could offload the medial compartment of the knee as well as understanding the biomechanical consequences of doing so to the hip, knee, and ankle joints.

6.6.1 Toe out gait

6.6.1.1 Pre- High Tibial Osteotomy toe out gait recommendations

Adopting a toe out gait style pre-HTO did not significantly change any spatiotemporal parameters except for a reduced stride length. Importantly gait speed was not affected by altering to a toe out gait style. Adopting a toe out gait style pre-HTO did not alter EKAM1 but did significantly reduce EKAM2 magnitude when compared to an unaltered level gait (2.48 % BW.h (1.1) vs 2.18 % BW.h (0.94), $p = 0.000$). The changes that occurred with EKAM parameters were also reflected when assessing KAAI parameters in the first and second half of stance. Interestingly, even though there was not a significant difference at EKAM1 between pre-HTO unaltered level gait and pre-HTO toe out gait, the COMAK simulation indicates that medial knee compartment contact force at FP significantly increased because of adopting a toe out gait compared to an unaltered level gait. Mean pressure and maximum pressure for total, medial, lateral compartments at FP were increased when adopting a toe out gait except for the lateral compartment mean pressure.

The reduction in EKAM2 was met with a significantly increased peak flexion moment when compared to pre-HTO unaltered level gait and a significantly reduced peak extension moment. As expected, a toe out gait also resulted in a significantly reduced peak internal rotation moment and significantly increased peak external rotation moment.

In collaboration with the EKAM2 changes that occurred, the internal joint loading simulation suggests that at SP, total knee and medial compartment knee contact forces were significantly decreased when adopting a toe out gait. Additionally, medial compartment mean pressure and maximal pressure were significantly decreased when adopting the toe out gait style. Adopting the toe out gait did not alter lateral compartment joint loading and remained elevated compared to the control group. Additionally, total knee and lateral knee compartment medial-lateral point of direction changed significantly between the pre-HTO unaltered level gait and pre-HTO toe out gait style with the point of application occurring more laterally when adopting a toe out gait.

For the first time, internal joint loading has been assessed in individuals with mKOA and varus aligned lower limbs. The findings would suggest that although toe out gait decreases medial tibiofemoral joint loading during the second half of stance, there may be adverse consequences during the first half of stance. These changes would not have been apparent if the only metrics of concern were external moments at the knee. It is therefore imperative to better understand the internal joint alterations when adopting this gait style as an intervention pre-HTO before recommending it as an effective intervention.

6.6.2 Wide stance gait

6.6.2.1 Pre- High Tibial Osteotomy wide stance gait recommendations

All patient’s pre-surgery were able to successfully adopt a wide stance altered gait style. Adopting a wide stance gait resulted in a stride width of ~0.25m compared a pre-HTO unaltered level gait stride width of ~0.16m. There were no significant changes in any spatiotemporal parameters when adopting a wide stance gait compared to an unaltered level pre-HTO.

Like adopting a toe out gait style, a wide stance gait style did not significantly alter EKAM1 but did significantly decrease EKAM2 when compared to an unaltered level gait pre-surgery. However, a wide stance altered gait style EKAM2 remained significantly higher compared to a control group. Adopting a wide stance gait significantly increased peak flexion moment when compared to pre-HTO unaltered level gait. Compared to an unaltered gait, adopting a wide stance gait reduced peak internal rotation moment and significantly increased peak external rotation moment.

At FP, total and lateral compartment contact force, mean and maximum pressure increased due to adopting a wide stance gait compared to unaltered level gait. Medial compartment maximum pressure significantly increased because of adopting a wide stance gait style compared to pre-HTO unaltered level gait. At SP, medial compartment knee contact forces, mean and maximum pressure were significantly decreased when adopting a wide stance gait compared to unaltered level gait. Additionally, lateral compartment contact force, mean and maximal pressure were significantly increased when adopting the wide stance gait style.

At FP, total knee point of application shifted laterally when adopting a wide stance gait compared to an unaltered level gait pre-HTO. At SP, total knee and lateral knee compartment medial-lateral point of direction changed significantly between the pre-HTO unaltered level gait and pre-HTO wide stance gait style with the point of application occurring more laterally when adopting a wide stance gait.

At FP, there was a significant increase in lateral compartment contact area when adopting a wide stance when compared to a pre-surgery unaltered level gait (140.68 mm^2 (28.69) vs 148.35 mm^2 (28.96), $p = 0.007$). At MS and SP, there were no significant differences between the pre-HTO unaltered level gait and a wide stance gait.

Like adopting a toe out gait pre-HTO, for the first time, this thesis has assessed internal joint loading alterations when adopting to a wide stance gait in individuals with mKOA and varus aligned lower limbs. The findings would suggest that although wide stance gait decreases medial tibiofemoral joint loading during the second half of stance, there may be adverse consequences during the first half of stance. In the first half of stance, there is increased total knee contact force as well as an increased medial increase in contact pressure. These changes would not have been apparent if the only metrics of concern were external moments at the knee. Again, like adopting a toe out gait pre-surgery, it is imperative to better understand the internal joint alterations when adopting this gait style as an intervention pre-HTO before recommending it as an effective intervention.

6.6.3 Medial thrust gait

6.6.3.1 Pre- High Tibial Osteotomy medial thrust gait recommendations

Adopting a medial thrust gait style resulted in a significant reduction in gait speed. Any reductions seen in EKAM1, EKAM2 should be viewed considering the reduction in gait speed. Adopting a medial thrust gait significantly reduced EKAM1 and EKAM2 when compared to an unaltered level gait pre-surgery. Adopting a medial thrust gait significantly reduced KAAI over the first half of stance when compared to -pre-surgery unaltered level gait. Additionally, between heel strike and 16% of stance, adopting a medial thrust gait significantly reduced KAAI. Adopting a medial thrust gait significantly increased peak flexion moment when compared to pre-HTO unaltered level gait, as well as significantly reducing peak extension moment. In terms of the peak transverse plane knee moment changes, adopting a medial thrust gait compared to unaltered level gait significantly reduced peak internal rotation moment and significantly increased peak external rotation moment.

At FP, total, medial compartment, and lateral compartment contact force, mean and maximum pressure significantly increased because of adopting a medial thrust gait compared to an unaltered level. At MS, total and lateral compartment contact force, mean and maximum pressure significantly increased when adopting a medial thrust gait compared to an unaltered level gait. In the medial compartment, maximum pressure increased because of adopting a medial thrust gait style compared to pre-HTO unaltered level gait. At SP, the only significant change that occurred due to adopting a medial thrust gait was that the medial compartment knee maximum pressure significantly compared to unaltered level gait; no other changes occurred.

It is clear from the novel findings of this work that adopting a medial thrust gait pre-HTO may not be advantageous for this cohort. The gait speed was significantly reduced, medial tibiofemoral contact force was increased during the first half of stance. Only the medial compartment knee maximum pressure was reduced during the second half of stance. The findings from this work indicate that much more learning of this style needs to occur prior to implementing into a clinical setting.

6.7 Introducing the next chapter

Apart from a stand-alone treatment, gait retraining has been suggested to be beneficial to individuals post-HTO to prolong the benefits of surgery and to preserve the knee joint. However, there has not been any research specifically on altering gait for a cohort of individuals after HTO surgery. The next chapter assesses whether an altered gait style could compliment surgery, but also to improve outcomes of surgery alongside future novel surgical planning, by removing uncertainty with abnormal movement patterns within the body that are not influenced by the surgery alone.

CHAPTER 7: BIOMECHANICAL DIFFERENCES BETWEEN A CONTROL COHORT, POST-HTO UNALTERED LEVEL GAIT AND POST-HTO ALTERED GAIT STYLES

7.1 Chapter background

The purpose of this chapter was to establish whether altering an individual's gait 12 months post- HTO offloads the damaged medial compartment of the knee, and if it does, whether this also effected the hip and ankle joints moments.

This chapter addressed medial compartment knee joint loading with two approaches. The first being a more 'traditional' way using the external knee adduction moment to infer medial compartment knee joint loading. The second was predicting internal joint loading using an enhanced musculoskeletal simulation technique. This chapter ends with a summary to address the potential of altering gait post-HTO.

The rationale for this chapter is that recent literature has suggested that gait retraining has the potential of reducing medial knee joint loading. However, this work has not focused on individuals with varus deformity as well as not appreciating what the biomechanical consequences are at the ankle and hip joints. The rationale for this chapter is that whether altering an individual's gait could have the potential to compliment surgery to prolong the benefits of HTO and to slow down the progression of mKOA. Chapter seven fulfils objective 4 of this PhD which aims to quantify the biomechanical differences of knee joint loading between post-HTO unaltered level gait and post-HTO altered gait styles. First, lower limb biomechanical differences were identified between the control group, post-HTO unaltered level gait, and post-HTO altered gait styles in the form of a toe out gait, wide stance gait and a medial thrust gait. Visual 3D (C-Motion) was used to extract discrete metrics to understand knee joint loading in the form of external moments. Additionally, hip and ankle external moments and rotations were assessed to determine the consequences of altering gait on the adjacent joints to the knee. Second, an enhanced musculoskeletal model was used to predict internal joint loading differences between the control group, post-HTO unaltered level gait, and post-HTO altered gait styles in the form of a toe out gait, wide stance gait and a medial thrust gait.

This chapter, as with chapters 5 and 6 performed paired samples t-tests in MATLAB (MATLAB 2020a, The Math Works, Inc., Natick, Massachusetts, USA) to identify significant differences associated with undergoing the altered gait styles to the pathological groups unaltered level gait post-HTO. Where parametric assumptions were not met, a Wilcoxon signed-rank test was used. Independent t tests were used to determine significant differences for the altered gait styles compared to the control group. Where parametric assumptions were not met, a Mann–Whitney U test was performed. Significance was determined when $p < 0.05$ for all statistical tests. Irrespective of the analysis undertaken, a full inspection for any outliers was undertaken.

7.2 Group demographics

29 patients (30 knees) were recruited from the Cardiff and Vale Orthopaedic Centre. Post-HTO group demographics are presented in Chapter 5. Table 7-1 outlines the final numbers that were used in each analysis in this chapter.

Table 7-1 Post-HTO Altered Gait Styles: Participant Numbers Per Analysis

Gait style	**Controls**	**Post-HTO: Visual 3D**	**Post-HTO: COMAK**
Unaltered level gait	28	30	30
Toe out gait	x	30	30
Wide stance gait	x	30	30
Medial thrust gait	x	14	14

Note: 2 people did not perform any medial thrust trials. 14 participants did not achieve a reduction in the maximum knee adduction angle during the first half of stance and were therefore discounted as not performing an effective medial thrust gait style.

7.3 Toe out gait

7.3.1 Quantifying toe out gait

All patients were able to successfully adopt a toe out altered gait style. Adopting a toe out gait resulted in a FPA of ~25° compared a post-HTO unaltered level gait foot progression angle of ~17° (p = 0.000), as shown in Table 7-2.

Table 7-2 Post-HTO Toe Out Gait Establishing Altered Gait Style

	Controls	Post-HTO NL	Post-HTO TO	Controls vs post-HTO NL	Controls vs Post-HTO TO	Post NL vs Post TO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Foot progression angle (°)	15.69 (5.68)	16.99 (7.43)	25.35 (19.04)	0.463	**0.000††**	**0.000††**

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. A positive foot progression angle (°) indicates a toe out foot progression angle. std = standard deviation. ° = degree.

7.3.2 Spatial-temporal parameters

Adopting a toe out gait during the 12-month post-HTO motion analysis visit did not significantly alter any spatial temporal parameters compared to their unaltered level gait, as reported in Table 7-3.

Table 7-3 Post-HTO Toe Out Gait Spatial Temporal Parameters

	Controls	Post-HTO NL	Post-HTO TO	Controls vs Post-HTO NL	Controls vs Post-HTO TO	Post NL vs Post TO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Operative limb cycle time (s)	**1.08 (0.08)**	**1.15 (0.11)**	**1.15 (0.12)**	**0.003****	**0.007****	**0.715**
Operative limb stance time (s)	**0.65 (0.06)**	**0.71 (0.08)**	**0.71 (0.09)**	**0.001****	**0.006††**	**0.188**
Operative limb step length (m)	**0.64 (0.07)**	**0.63 (0.07)**	**0.63 (0.08)**	**0.556**	**0.500**	**0.621**
Operative limb step time (s)	**0.54 (0.04)**	**0.57 (0.05)**	**0.57 (0.06)**	**0.008****	**0.024***	**0.496**
Operative limb stride length (m)	**1.29 (0.13)**	**1.26 (0.14)**	**1.25 (0.15)**	**0.351**	**0.288**	**0.431**
Swing time (s)	**0.43 (0.03)**	**0.44 (0.03)**	**0.44 (0.04)**	**0.301**	**0.171**	**0.221**
Speed (m/s)	**1.21 (0.16)**	**1.10 (0.16)**	**1.10 (0.18)**	**0.018***	**0.023***	**0.994**

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. std = standard deviation. s = seconds; m = metre; m/s = metre/second.

7.3.3 Knee joint loading: External moments

Adopting a toe out gait 12-months post-HTO did not significantly alter EKAM1 but did significantly decrease EKAM2 when compared to an unaltered level gait 12-months post-surgery (1.55 % BW.h (0.83) vs 1.38 % BW.h (1.00), $p = 0.002$). Although there were no significant differences between post-HTO unaltered level gait and adopting a toe out gait, when compared to the control group, adopting a toe out gait resulted in a normalised peak external flexion moment. Patients, whether adopting a toe out gait or with an unaltered level gait had a significantly reduced peak external extension moment compared to the control group.

In terms of the peak transverse plane knee moment changes, adopting a toe out gait compared to unaltered level gait post-HTO significantly reduced peak internal rotation moment ($p = 0.018$) and significantly increased peak external rotation moment ($p = 0.000$).

Table 7-4 Post-HTO Toe Out Gait External Knee Moments

	Controls	Post-HTO NL	Post-HTO TO	Controls vs Post-HTO NL	Controls vs Post-HTO TO	Post NL vs Post TO
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) moment						
Maximum	**2.11 (0.81)**	**2.03 (0.99)**	**2.21 (0.96)**	**0.737**	**0.660**	**0.102**
1st peak (1st half stance)	**2.27 (0.65)**	**2.10 (0.88)**	**2.14 (0.89)**	**0.412**	**0.543**	**0.766**
2nd peak (2nd half stance)	**1.50 (0.67)**	**1.55 (0.83)**	**1.38 (1.00)**	**0.789**	**0.147**	**0.002††**
Midstance	**1.15 (0.49)**	**1.32 (0.64)**	**1.34 (0.73)**	**0.259**	**0.492**	**0.318**
Flexion (+) moment peak	**3.62 (1.65)**	**2.66 (1.23)**	**2.88 (1.27)**	**0.014***	**0.059**	**0.122**
Extension (−) moment peak	**-2.45 (0.85)**	**-1.84 (0.72)**	**-1.78 (0.66)**	**0.001††**	**0.002††**	**0.813**
Internal (+) rotation moment	**0.60 (0.37)**	**0.64 (0.36)**	**0.56 (0.4)**	**0.632**	**0.365**	**0.018†**
External (−) rotation moment	**-0.16 (0.08)**	**-0.13 (0.08)**	**-0.18 (0.09)**	**0.122**	**0.626**	**0.000****

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. % BW.h = percentage of body weight multiplied by height.

7.3.4 Knee joint loading: Impulses

When assessing KAAI during each portion of stance, unaltered level gait post-HTO was not statistically different to the control cohort.

However, when comparing adopting a toe out gait and unaltered level gait post-HTO, adopting a toe out gait significantly decreased KAAI over the whole of stance, during the second half of stance, midstance to 83% of stance, and 84% of stance to toe-off; indicating a decreased second half of stance dynamic joint loading. Adopting a toe out gait post-HTO did not significantly change first half of stance KAAI parameters apart from a slight significant increase in KAAI between heel strike and 16% of stance.

When compared to post-HTO unaltered level gait, a toe out gait significantly decreased first half of stance abduction angular impulse ($p = 0.037$) and significantly increased second half of stance abduction angular impulse ($p = 0.000$).

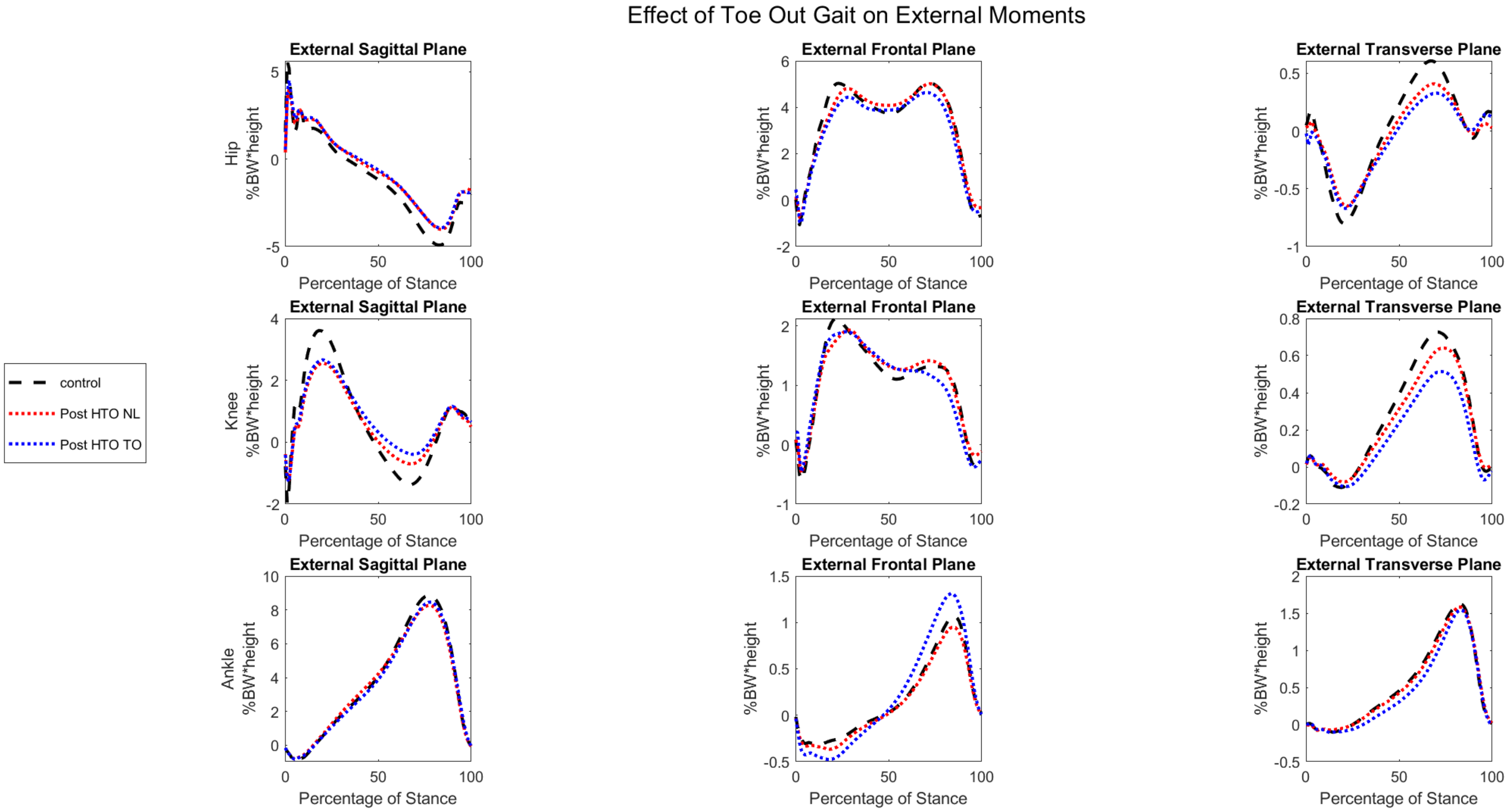


Figure 49 Visual 3D: Post-HTO toe out gait group average joint external moments

Positive values represent external moments for knee flexion, adduction, and internal

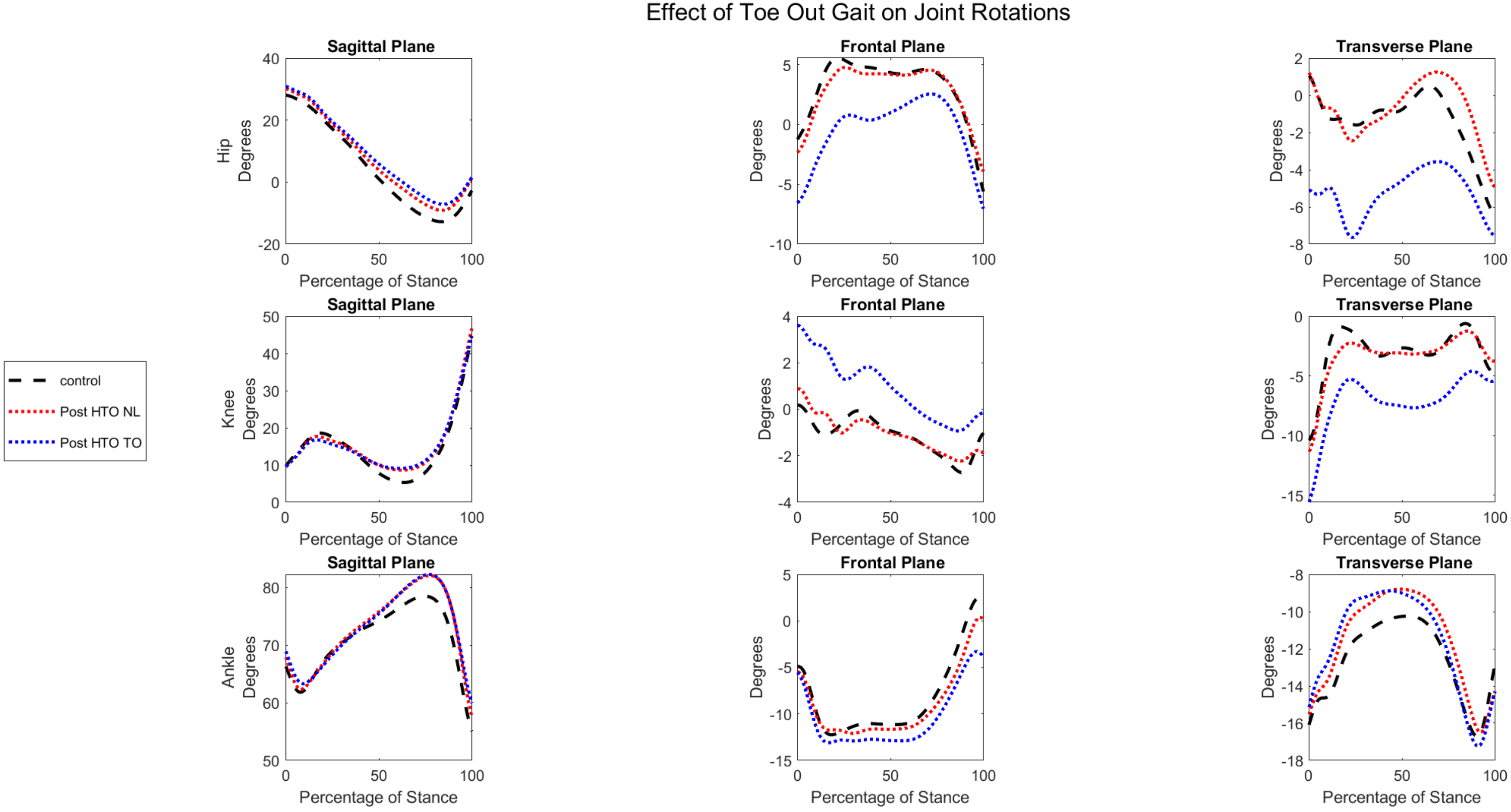


Figure 50 Visual 3D: Post-HTO toe out gait group average joint kinematics

Positive values represent knee flexion, adduction, and internal rotations.

Table 7-5 Post-HTO Toe Out Gait Knee Angular Impulse

	Controls	**Post-HTO NL**	**Post-HTO TO**	**Controls vs Post-HTO NL**	**Controls vs Post-HTO TO**	**Post NL vs Post TO**
%BW.h.s	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) angular impulse						
Stance	0.74 (0.28)	0.81 (0.39)	0.77 (0.42)	0.398	0.896	**0.017†**
1st half stance	0.43 (0.14)	0.46 (0.20)	0.47 (0.20)	0.502	0.325	0.491
2nd half stance	0.31 (0.16)	0.35 (0.21)	0.30 (0.24)	0.346	0.408	**0.001††**
0–16% stance	0.06 (0.03)	0.06 (0.04)	0.07 (0.04)	0.699	0.113	**0.006****
17%–midstance	0.36 (0.11)	0.39 (0.17)	0.39 (0.16)	0.466	0.423	0.441
Midstance–83% stance	0.26 (0.13)	0.30 (0.17)	0.26 (0.20)	0.444	0.682	**0.002††**
84%-100% stance	0.04 (0.02)	0.04 (0.03)	0.02 (0.03)	0.787	**0.016†**	**0.000****
Abduction (–) angular impulse in Stance						
1st half stance	-0.02 (0.01)	-0.02 (0.01)	-0.01 (0.01)	0.236	**0.047†**	**0.037***
2nd half stance	-0.02 (0.02)	-0.01 (0.01)	-0.03 (0.02)	0.063	**0.013†**	**0.000††**

Significant difference p<0.01) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. %BW.h.s = % of body weight multiplied by height per second.

7.3.5 External ankle moments

During the whole of stance, adopting a toe out gait compared to unaltered level gait 12-months post-HTO significantly increased peak external inversion moment (0.95 %BW.h (0.53) vs 1.35 %BW.h (0.54), p = 0.000) and a significantly increased peak external eversion moment (0.42 %BW.h (0.25) vs 0.6 %BW.h (0.23), p = 0.000). In addition to this, adopting a toe out gait style also significantly increased the peak external rotation moment (0.15 %BW.h (0.09) vs 0.2 %BW.h (0.15), p = 0.010).

During the first half of stance, peak external dorsiflexion moment significantly reduced when adopting a toe out gait style (4.22 %BW.h (0.93) vs 3.89 %BW.h (1.05), p = 0.019). In the frontal plane, peak external eversion moment significantly increased when adopting a toe out gait compared to post-HTO unaltered level gait (0.49 %BW.h (0.24) vs 0.6 %BW.h (0.23), p = 0.000). In the transverse plane, adopting a toe out gait resulted in a significantly reduced peak internal rotation and a significantly increased peak external rotation moment compared to a post-HTO unaltered level gait.

During the second half of stance, peak external plantarflexion moment significantly reduced when adopting a toe out gait compared to an unaltered level gait 12-months post-HTO (0.06 %BW.h (0.1) vs 0.02 %BW.h (0.11), p = 0.000). In the frontal plane, peak external inversion moment is significantly increased when adopting a toe out gait when compared to a post-HTO unaltered level gait (1.0 %BW.h (0.52) vs 1.35 %BW.h (0.54), p = 0.000) and peak external eversion moment is significantly decreased (0.14 %BW.h (0.13) vs 0.1 %BW.h (0.12), p = 0.027). In the transverse plane, adopting a toe out gait resulted in a significantly increased peak external rotation moment compared to a 12-month post-HTO unaltered level gait.

Table 7-6 Post-HTO Toe Out Gait External Ankle Moments Parameters

	Controls	**Post-HTO NL**	**Post-HTO TO**	**Controls vs post-HTO NL**	**Controls vs post-HTO TO**	**Post NL vs Post TO**
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak dorsi-flexion (+) moment	6.78 (3.93)	8.01 (1.42)	8.56 (1.33)	0.501	0.932	0.165
Peak plantar-flexion (−) moment	-0.97 (0.36)	-0.81 (0.36)	-0.90 (0.31)	0.112	0.444	0.057
Peak inversion (+) moment	0.82 (0.54)	0.95 (0.53)	1.35 (0.54)	0.787	**0.001††**	**0.000††**
Peak eversion (-) moment	-0.41 (0.22)	-0.42 (0.25)	-0.60 (0.23)	0.890	**0.002****	**0.000††**
Peak internal rotation (+) moment	1.25 (0.85)	1.57 (0.83)	1.62 (0.79)	0.740	0.296	0.534
Peak external rotation (−) moment	-0.17 (0.08)	-0.15 (0.09)	-0.20 (0.15)	0.261	0.752	**0.010††**
0-50% (HS to midstance)						
Peak dorsi-flexion (+) moment	4.08 (0.99)	4.22 (0.93)	3.89 (1.05)	0.589	0.479	**0.019***
Peak plantar-flexion (−) moment	-1.03 (0.28)	-0.87 (0.29)	-0.90 (0.31)	**0.048***	0.103	0.474
Peak inversion (+) moment	0.14 (0.14)	0.15 (0.20)	0.16 (0.20)	0.775	0.982	0.796
Peak eversion (-) moment	-0.43 (0.20)	-0.49 (0.24)	-0.60 (0.23)	0.282	**0.005††**	**0.000††**
Peak internal rotation (+) moment	0.47 (0.26)	0.44 (0.22)	0.34 (0.18)	0.552	**0.020†**	**0.001****
Peak external rotation (−) moment	-0.18 (0.08)	-0.15 (0.09)	-0.18 (0.11)	0.248	0.823	**0.027†**
50-100% (Midstance to toe-off)						
Peak dorsi-flexion (+) moment	8.94 (0.74)	8.35 (0.77)	8.56 (1.33)	**0.004****	**0.022†**	0.766
Peak plantar-flexion (−) moment	-0.03 (0.12)	-0.06 (0.10)	-0.02 (0.11)	0.264	0.815	**0.000****
Peak inversion (+) moment	1.11 (0.34)	1.00 (0.52)	1.35 (0.54)	0.091	0.091	**0.000††**
Peak eversion (-) moment	-0.11 (0.14)	-0.14 (0.13)	-0.10 (0.12)	0.098	0.920	**0.027***

Peak internal rotation (+) moment	1.67 (0.53)	1.65 (0.81)	1.62 (0.79)	0.206	0.326	0.718
Peak external rotation (−) moment	0.00 (0.06)	0.01 (0.04)	-0.06 (0.15)	0.426	**0.043†**	**0.000††**

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. %BW.h = % of body weight multiplied by height. HS = heel strike.

7.3.6 External hip moments

During the whole of stance, adopting a toe out altered gait significantly increased peak hip flexion moment compared to post-HTO unaltered level gait (4.7 %BW.h (1.91) vs 5.47 %BW.h (1.84), $p = 0.002$). A toe out altered gait significantly increased peak hip abduction moment compared to post-HTO unaltered level gait (1.18 %BW.h (0.61) vs 1.52 %BW.h (0.7), $p = 0.001$).

During the first half of stance, peak external hip adduction moment significantly reduced when adopting a toe out gait style (5.11 %BW.h (0.81) vs 4.78 %BW.h (0.93), $p = 0.000$). A toe out altered gait significantly increased peak external hip abduction moment compared to post-HTO unaltered level gait (1.02 %BW.h (0.8) vs 1.29 %BW.h (0.91), $p = 0.007$). In the transverse plane, adopting a toe out gait resulted in a significantly reduced peak external hip internal rotation moment when compared to a post-HTO unaltered level gait (0.27 (0.15) vs 0.19 (0.14), $p = 0.000$).

During the second half of stance, peak external hip adduction moment significantly reduced when adopting a toe out gait style compared to post-HTO unaltered level gait (5.16 %BW.h (0.84) vs 4.85 %BW.h (1.1), $p = 0.004$), whilst significantly increasing peak external hip abduction moment (0.46 %BW.h (0.28) vs 0.64 %BW.h (0.58), $p = 0.002$).

Table 7-7 Post-HTO Toe Out Gait External Hip Moments

	Controls	Post-HTO NL	Post-HTO TO	Controls vs post-HTO NL	Controls vs post-HTO TO	Post NL vs Post TO
	Mean (std)	Mean (std)	Mean (std)	P value	P value	P value
0-100% Stance						
Peak external hip flexion (+) moment	6.31 (2.64)	4.70 (1.91)	5.47 (1.84)	**0.010****	0.161	**0.002††**
Peak external hip extension (−) moment	-4.36 (1.70)	-4.10 (1.12)	-4.19 (1.30)	0.492	0.665	0.600
Peak external hip adduction (+) moment	5.09 (1.41)	5.23 (0.85)	5.14 (1.10)	0.639	0.864	0.153
Peak external hip abduction (-) moment	-1.65 (1.03)	-1.18 (0.61)	-1.52 (0.70)	**0.047†**	0.994	**0.001††**
Peak external hip internal rotation (+) moment	0.54 (0.30)	0.48 (0.22)	0.42 (0.22)	0.562	0.195	0.129
Peak external hip rotation (−) moment	-0.80 (0.43)	-0.69 (0.29)	-0.74 (0.31)	0.247	0.550	0.209
0-50% (HS to midstance)						
Peak external hip extension (−) moment	-1.25 (0.92)	-0.96 (0.72)	-0.86 (0.68)	0.142	0.077	0.203
Peak external hip adduction (+) moment	5.27 (1.03)	5.11 (0.81)	4.78 (0.93)	0.513	0.064	**0.000††**
Peak external hip abduction (-) moment	-1.28 (1.29)	-1.02 (0.8)	-1.29 (0.91)	0.361	0.983	**0.007††**
Peak external hip internal rotation (+) moment	0.34 (0.20)	0.27 (0.15)	0.19 (0.14)	0.180	**0.001****	**0.000††**
Peak external hip rotation (−) moment	-0.86 (0.42)	-0.71 (0.27)	-0.74 (0.31)	0.116	0.218	0.434
50-100% (Midstance to toe-off)						
Peak external hip adduction (+) moment	5.20 (1.17)	5.16 (0.84)	4.85 (1.10)	0.886	0.256	**0.004††**
Peak external hip abduction (-) moment	-0.86 (0.55)	-0.46 (0.54)	-0.64 (0.58)	**0.007****	0.147	**0.002****
Peak external hip internal rotation (+) moment	0.64 (0.24)	0.46 (0.28)	0.41 (0.23)	**0.009****	**0.000††**	0.058
Peak external hip rotation (−) moment	-0.11 (0.12)	-0.11 (0.15)	-0.11 (0.17)	0.573	0.671	0.976

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. % BW.h = percentage of body weight multiplied by height. HS = heel strike.

7.3.7 Concurrent Optimisation of Muscle and Secondary Kinematics

Post-surgery toe out gait did not alter gait speed or stance time when compared to post-HTO unaltered level gait.

Table 7-8 Post-HTO Toe Out Gait COMAK Gait Speed

	Controls	Post-HTO NL	Post-HTO TO	Controls vs Post-HTO NL	Controls vs Post-HTO TO	Post-HTO NL vs Post-HTO TO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Gait speed (m/s)	1.26 (0.17)	1.15 (0.17)	1.15 (0.18)	**0.020****	**0.019****	0.726

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. std= standard deviation. m/s = metres per second.

7.3.8 Internal knee joint loading

At FP, total knee compartment contact force significantly increased because of adopting a toe out gait compared to an unaltered level (2.33BW (0.52) vs 2.53BW (0.62), p = 0.03), as well as significant increases in mean pressure and maximum pressure. Lateral compartment contact force increased when adopting a toe out gait compared to unaltered level gait (0.94BW (0.27)) vs 1.14BW (0.36), p = 0.000) as well as a significant increase in mean pressure and maximum pressures. At SP, the only significant change between the post-HTO unaltered level gait and post-HTO toe out gait was a reduced medial compartment knee contact force when adopting a toe out gait (1.51BW (0.37)) vs 1.41BW (0.40), p = 0.045).

Internal knee joint loading ratios

At FP adopting a toe out gait significantly increased medial to total compartment total contact forces when compared to a post-HTO unaltered level gait (0.63 (0.08) vs 0.59 (0.09), p = 0.000). At FP, adopting a toe out gait significantly decreased lateral to total compartment total contact forces when compared to a post-HTO unaltered level gait (0.4 (0.08) vs 0.45 (0.09), p = 0.000).

At SP, adopting a toe out gait significantly increased medial to total compartment total contact forces when compared to a post-HTO unaltered level gait (0.62 (0.08) vs 0.59 (0.09), $p = 0.000$). At SP, adopting a toe out gait significantly decreased lateral to total compartment total contact forces when compared to a post-HTO unaltered level gait (0.41 (0.08) vs 0.44 (0.09), $p = 0.000$).

7.3.9 Point of application

At FP, post-HTO toe out gait shifted lateral knee compartment slightly medially compared to post-HTO unaltered level gait (20.89 (2.08) vs 20.65 (2.05), $p = 0.041$).

At SP, total knee and lateral knee compartment medial-lateral point of direction changed significantly between the post-HTO unaltered level gait and post-HTO toe out gait style with the point of application occurring more laterally when adopting a toe out gait.

7.3.10 Contact area

At FP, there was a significant increase in medial compartment contact area when adopting a toe out compared to a post-surgery unaltered level gait (213.55 mm^2 (32.49) vs 217.74 mm^2 (29.95), $p = 0.030$).

At MS, there was a significant increase in lateral compartment contact area when adopting a toe out when compared to a post-surgery unaltered level gait (101.96mm^2 (27.28) vs 109.13 mm^2 (24.34), $p = 0.044$).

At SP, total and medial contact area significantly decreased when adopting a toe out gait compared to an unaltered level gait; medial compartment contact area reduced from 217.24 mm^2 (46.55) with unaltered level gait to 202.7 mm^2 (39.93) when adopting a toe out gait style ($p = 0.005$).

Table 7-9 Post-HTO Toe Out Gait Internal Knee Joint Loading

	Controls	**Post-HTO NL**	**Post-HTO TO**	**Controls vs Post-HTO NL**	**Controls vs Post-HTO TO**	**Post-HTO NL vs Post-HTO TO**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total knee						
Contact force [BW]	2.74 (0.57)	2.33 (0.52)	2.53 (0.62)	**0.006****	0.195	**0.003****
Mean pressure [MPa]	5.63 (1.25)	5.77 (1.13)	6.13 (1.20)	0.671	0.054	**0.018†**
Max pressure	12.92 (3.32)	13.22 (2.46)	14.63 (3.15)	0.501	**0.032***	**0.004††**
Medial knee						
Contact force [BW]	1.69 (0.31)	1.46 (0.36)	1.48 (0.39)	**0.011***	**0.023***	0.662
Mean pressure [MPa]	5.76 (1.12)	5.95 (1.12)	5.95 (1.18)	0.526	0.531	0.974
Max pressure	12.21 (2.52)	12.69 (2.45)	12.72 (2.45)	0.469	0.442	0.864
Lateral knee						
Contact force [BW]	1.13 (0.37)	0.94 (0.27)	1.14 (0.36)	**0.027***	0.932	**0.000††**
Mean pressure [MPa]	5.37 (1.62)	5.45 (1.43)	6.29 (1.65)	0.883	**0.042†**	**0.000††**
Max pressure	11.57 (3.76)	11.78 (3.12)	13.84 (3.81)	0.671	**0.021†**	**0.000††**
Midstance						
Total knee						
Contact force [BW]	1.25 (0.24)	1.23 (0.18)	1.21 (0.18)	0.859	0.705	0.075
Mean pressure [MPa]	3.40 (0.36)	3.94 (0.58)	3.97 (0.68)	**0.000****	**0.000****	0.750
Max pressure	7.62 (1.12)	8.93 (1.59)	8.89 (1.71)	**0.001****	**0.001****	0.405
Medial knee						
Contact force [BW]	0.92 (0.22)	0.89 (0.20)	0.86 (0.22)	0.644	0.367	**0.043†**
Mean pressure [MPa]	3.81 (0.58)	4.31 (0.70)	4.28 (0.79)	**0.004****	**0.013***	0.329
Max pressure	7.43 (1.22)	8.59 (1.51)	8.54 (1.56)	**0.002****	**0.004****	0.697
Lateral knee						
Contact force [BW]	0.36 (0.14)	0.37 (0.15)	0.37 (0.19)	0.660	0.787	0.544
Mean pressure [MPa]	2.59 (0.54)	3.04 (1.06)	3.05 (1.22)	0.070	0.138	0.894

Max pressure	5.56 (1.07)	6.42 (2.17)	6.47 (2.49)	0.065	0.130	0.830
Second peak						
Total knee						
Contact force [BW]	2.90 (0.70)	2.43 (0.54)	2.38 (0.62)	**0.006****	**0.004****	0.417
Mean pressure [MPa]	5.32 (0.64)	5.77 (1.11)	5.66 (1.05)	0.147	0.268	0.417
Max pressure	12.70 (1.76)	13.34 (3.05)	13.05 (2.77)	0.811	0.859	0.393
Medial knee						
Contact force [BW]	1.83 (0.35)	1.51 (0.37)	1.41 (0.40)	**0.001****	**0.000****	**0.045***
Mean pressure [MPa]	5.91 (0.80)	6.14 (1.23)	5.86 (1.29)	0.386	0.365	0.054
Max pressure	12.61 (1.80)	12.65 (2.76)	12.07 (2.91)	0.605	0.088	0.178
Lateral knee						
Contact force [BW]	1.20 (0.44)	1.01 (0.30)	1.06 (0.36)	0.071	0.196	0.188
Mean pressure [MPa]	4.53 (0.73)	5.23 (1.39)	5.33 (1.33)	**0.014†**	**0.002††**	0.381
Max pressure	9.43 (1.48)	11.20 (3.30)	11.37 (2.95)	**0.003†**	**0.001††**	0.528

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. std = standard deviation. BW = body weight; MPa = megapascals.

Table 7-10 Post-HTO Toe Out Gait Contact Force Ratios

	Controls	Post-HTO NL	Post-HTO TO	Controls vs Post-HTO NL	Controls vs Post-HTO TO	Post-HTO NL vs Post-HTO TO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
MED / TOTAL	0.63 (0.07)	0.63 (0.08)	0.59 (0.09)	0.920	0.056	**0.000††**
LAT / TOTAL	0.41 (0.07)	0.40 (0.08)	0.45 (0.09)	0.879	0.057	**0.000††**
Midstance						
MED / TOTAL	0.73 (0.09)	0.72 (0.12)	0.72 (0.14)	0.705	0.717	0.877
LAT / TOTAL	0.29 (0.10)	0.30 (0.12)	0.31 (0.15)	0.763	0.682	0.975
Second peak						
MED / TOTAL	0.64 (0.07)	0.62 (0.08)	0.59 (0.09)	0.302	**0.004††**	**0.000††**
LAT / TOTAL	0.40 (0.07)	0.41 (0.08)	0.44 (0.09)	0.463	**0.020†**	**0.000††**

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. std = standard deviation. MED = medial compartment tibiofemoral joint; LAT = lateral compartment tibiofemoral joint; TOTAL = total tibiofemoral joint.

Table 7-11 Post-HTO Toe Out Gait Point of Application

	Controls	Post-HTO NL	Post-HTO TO	Controls vs Post-HTO NL	Controls vs Post-HTO TO	Post-HTO NL vs Post-HTO TO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total knee						
Anterior (+) / posterior (-)	-2.46 (2.08)	-2.56 (2.20)	-2.61 (2.33)	0.728	0.594	0.805
Lateral (+) / medial (-)	-2.27 (3.27)	-2.44 (3.43)	-3.02 (3.96)	0.845	0.440	0.360
Medial knee						
Anterior (+) / posterior (-)	-1.01 (2.25)	-1.18 (2.12)	-1.17 (2.51)	0.811	0.896	0.959
Lateral (+) / medial (-)	-17.19 (1.29)	-17.44 (1.53)	-17.59 (1.53)	0.493	0.284	0.134
Lateral knee						
Anterior (+) / posterior (-)	-4.67 (1.83)	-4.65 (2.27)	-4.86 (2.01)	0.847	0.700	0.180
Lateral (+) / medial (-)	20.59 (2.34)	20.89 (2.08)	20.65 (2.05)	0.614	0.923	**0.041†**
Midstance						
Total knee						
Anterior (+) / posterior (-)	4.66 (1.86)	4.64 (1.96)	3.69 (2.22)	0.974	0.080	**0.015***
Lateral (+) / medial (-)	-5.84 (4.72)	-5.47 (5.84)	-4.38 (5.76)	0.791	0.297	0.052
Medial knee						
Anterior (+) / posterior (-)	6.63 (2.77)	6.34 (2.74)	5.85 (2.96)	0.688	0.305	0.215
Lateral (+) / medial (-)	-16.33 (1.53)	-16.65 (1.82)	-16.47 (1.75)	0.453	0.303	**0.049†**
Lateral knee						
Anterior (+) / posterior (-)	-0.42 (1.55)	0.32 (1.74)	-0.59 (1.92)	0.090	0.720	**0.006****
Lateral (+) / medial (-)	20.34 (2.43)	20.88 (3.22)	21.04 (2.57)	0.471	0.341	0.861
Second peak						
Total knee						
Anterior (+) / posterior (-)	5.70 (1.64)	4.94 (3.22)	4.63 (3.08)	0.717	0.147	0.316
Lateral (+) / medial (-)	-1.39 (2.90)	-0.68 (3.24)	1.08 (3.60)	0.417	**0.004††**	**0.000****

Medial knee						
Anterior (+) / posterior (-)	9.56 (1.83)	8.26 (3.68)	7.86 (3.59)	0.201	**0.029†**	0.213
Lateral (+) / medial (-)	-14.59 (1.31)	-15.46 (1.27)	-15.44 (1.39)	**0.013***	**0.019***	0.906
Lateral knee						
Anterior (+) / posterior (-)	-0.89 (2.43)	-0.39 (2.73)	0.08 (3.01)	0.466	0.185	0.226
Lateral (+) / medial (-)	19.74 (2.84)	21.55 (2.90)	22.34 (2.96)	**0.020***	**0.001****	**0.020***

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. std = standard deviation. mm = millimetres. X = anterior; Z = lateral. COP = centre of pressure.

Table 7-12 Post-HTO Toe Out Gait Knee Contact Area

mm^2	Controls	Post-HTO NL	Post-HTO TO	Controls vs Post-HTO NL	Controls vs Post-HTO TO	Post-HTO NL vs Post-HTO TO
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total	352.35 (39.00)	362.97 (49.26)	367.11 (48.70)	0.453	0.255	0.069
Medial	206.80 (22.63)	213.55 (32.49)	217.74 (29.95)	0.472	0.118	**0.030***
Lateral	145.55 (20.95)	149.43 (21.45)	149.37 (25.60)	0.242	0.282	0.530
Midstance						
Total	262.26 (34.10)	282.31 (46.27)	289.30 (39.68)	0.080	**0.008****	0.086
Medial	167.14 (24.46)	180.35 (29.87)	180.17 (28.74)	0.108	0.069	0.530
Lateral	95.12 (19.62)	101.96 (27.28)	109.13 (24.34)	0.111	**0.007††**	**0.044***
Second peak						
Total	399.23 (90.44)	387.36 (90.55)	365.87 (73.62)	0.620	0.128	**0.041†**
Medial	219.07 (41.52)	217.24 (46.55)	202.70 (39.27)	0.763	0.160	**0.005††**
Lateral	180.16 (52.60)	170.12 (46.05)	163.17 (36.93)	0.442	0.158	0.141

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. TO = toe out gait; NL = unaltered level gait. std = standard deviation. Measurement = mm^2.

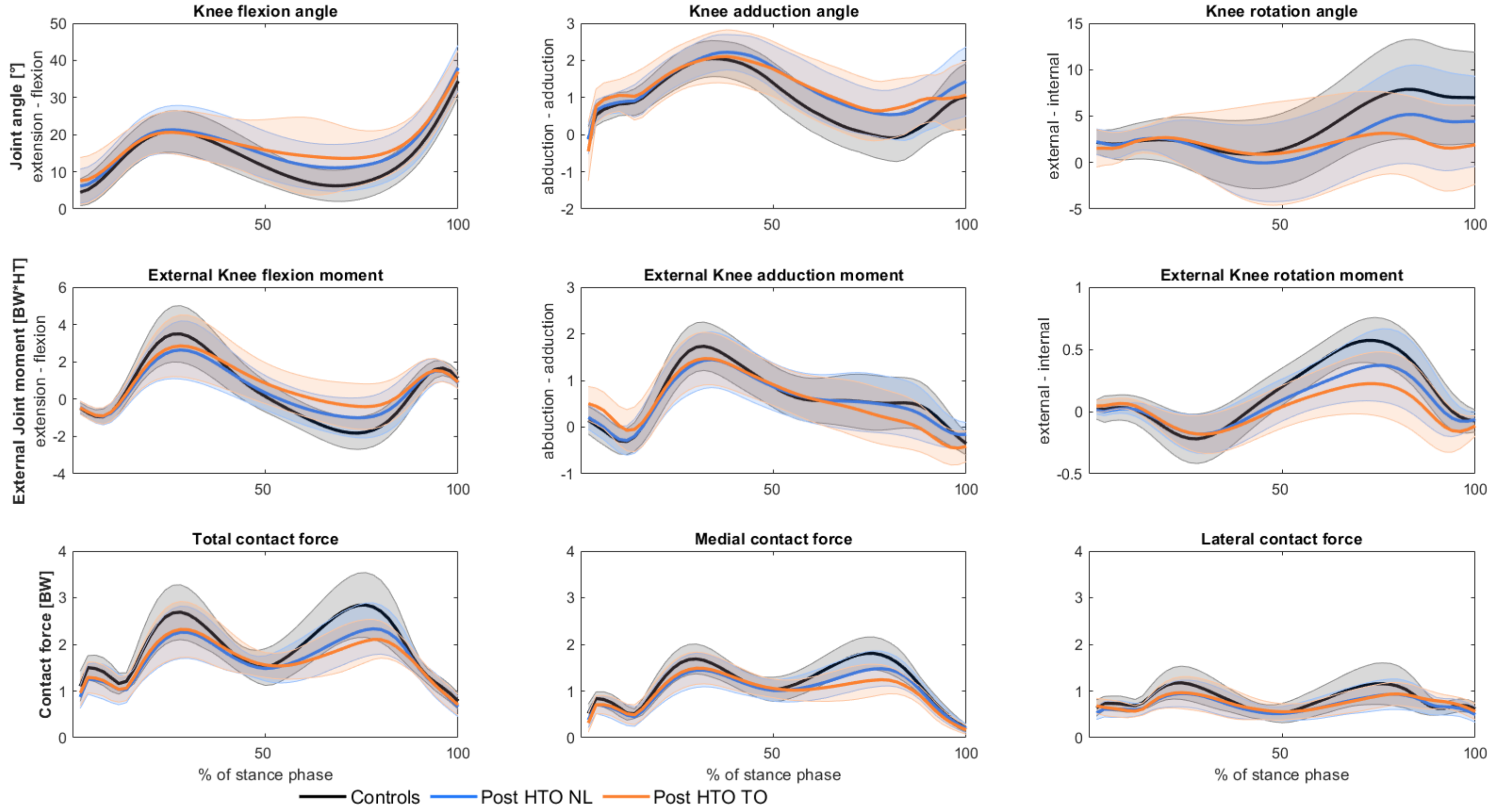


Figure 51 Post-HTO toe out gait knee joint kinematics, external moments, contact forces

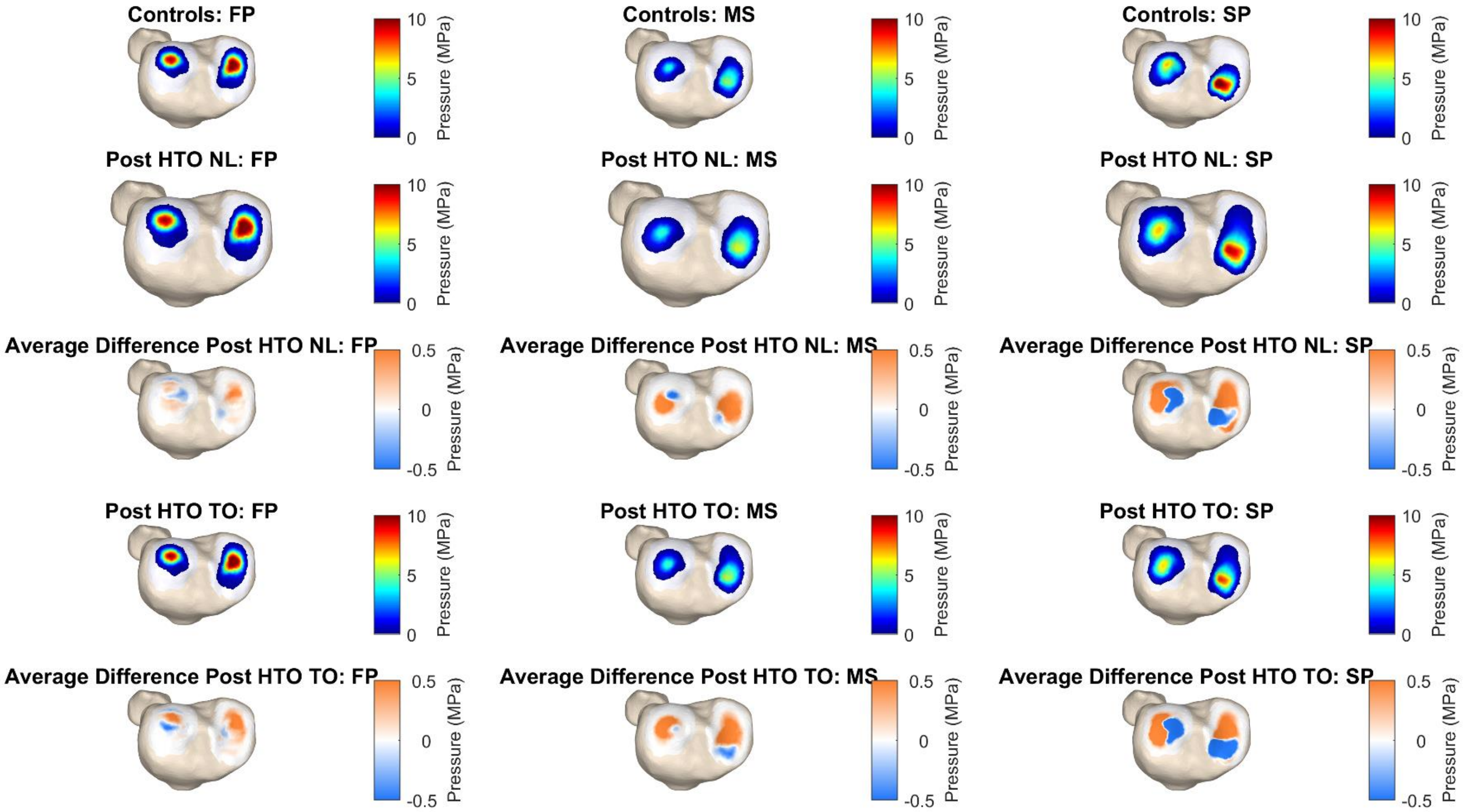


Figure 52 Post-HTO toe out contact pressure distribution on the tibia

7.4 Wide stance gait

7.4.1 Quantifying wide stance gait

All patient's post-surgery were able to successfully adopt a wide stance altered gait style. Adopting a wide stance gait resulted in a stride width of ~0.26m compared a post-HTO unaltered level gait stride width of ~0.17m (p = 0.000).

Table 7-13 Post-HTO Wide Stance Gait Quantifying Gait Style

	Controls	**Post-HTO NL**	**Post-HTO WS**	**Controls vs post-HTO NL**	**Controls vs Post-HTO WS**	**Post NL vs Post WS**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Stride width (m)	0.14 (0.03)	0.17 (0.04)	0.26 (0.07)	**0.007††**	**0.000††**	**0.000††**

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. m = metre.

7.4.2 Spatial-temporal parameters

There were no significant changes in any spatiotemporal parameters when adopting a wide stance gait compared to an unaltered level post-HTO apart from a significant increase in operative limb step length (0.63m (0.07) vs 0.64m (0.08), p = 0.016).

Table 7-14 Post-HTO Wide Stance Gait Spatial Temporal Parameters

	Controls	**Post-HTO NL**	**Post-HTO WS**	**Controls vs post-HTO NL**	**Controls vs post-HTO WS**	**Post NL vs Post WS**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Operative limb cycle time (s)	1.08 (0.08)	1.15 (0.11)	1.15 (0.11)	**0.003****	**0.004****	0.982
Operative limb stance time (s)	0.65 (0.06)	0.71 (0.08)	0.71 (0.08)	**0.001****	**0.002††**	0.484
Operative limb step length (m)	0.64 (0.07)	0.63 (0.07)	0.64 (0.08)	0.556	0.983	**0.016***
Operative limb step time (s)	0.54 (0.04)	0.57 (0.05)	0.58 (0.06)	**0.008****	**0.006****	0.522
Operative limb stride length (m)	1.29 (0.13)	1.26 (0.14)	1.28 (0.15)	0.351	0.672	0.058
Swing time (s)	0.43 (0.03)	0.44 (0.03)	0.44 (0.04)	0.301	0.123	0.074
Speed (m/s)	1.21 (0.16)	1.10 (0.16)	1.12 (0.19)	**0.018***	0.058	0.197

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. s = seconds; m = metre; m/s = metre/second.

7.4.3 Knee joint loading: External moments

Adopting a wide stance gait significantly decreased EKAM1 (2.1 %BW.h (0.88) vs 2.05 %BW.h (0.89), $p = 0.002$) and EKAM2 (1.55 %BW.h (0.83) vs 1.34 %BW.h (0.87), $p = 0.000$) when compared to an unaltered level gait post-surgery.

Adopting a wide stance gait significantly increased peak external flexion moment when compared to post-HTO unaltered level gait (2.66 %BW.h (1.23) vs 3.12 %BW.h (1.27), $p = 0.004$).

In terms of the peak transverse plane knee moment changes, adopting a wide stance gait compared to unaltered level gait significantly increased peak external rotation moment (0.13 %BW.h (0.08) vs 0.17 %BW.h (0.08), $p = 0.000$).

Table 7-15 Post-HTO Wide Stance Gait External Knee Moments

	Controls	**Post-HTO NL**	**Post-HTO WS**	**Controls vs post-HTO NL**	**Controls vs post-HTO WS**	**Post NL vs Post WS**
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) moment						
Maximum	2.11 (0.81)	2.03 (0.99)	2.05 (0.89)	0.737	0.789	0.382
1st peak (1st half stance)	2.27 (0.65)	2.10 (0.88)	1.95 (0.87)	0.412	0.119	**0.002††**
2nd peak (2nd half stance)	1.50 (0.67)	1.55 (0.83)	1.34 (0.78)	0.789	0.282	**0.000****
Midstance	1.15 (0.49)	1.32 (0.64)	1.33 (0.69)	0.259	0.271	0.908
Flexion (+) moment peak	3.62 (1.65)	2.66 (1.23)	3.12 (1.27)	**0.014***	0.201	**0.004††**
Extension (−) moment peak	-2.45 (0.85)	-1.84 (0.72)	-1.85 (0.76)	**0.001††**	**0.002††**	0.849
Internal (+) rotation	0.60 (0.37)	0.64 (0.36)	0.60 (0.34)	0.632	0.823	0.099
External (−) rotation	-0.16 (0.08)	-0.13 (0.08)	-0.17 (0.08)	0.122	0.880	**0.000****

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. % BW.h = percentage of body weight multiplied by height.

7.4.4 Knee joint loading: Impulses

Adopting a wide stance gait significantly reduced KAAI over the whole of stance (0.81 (0.39) vs 0.71 (0.36), $p = 0.002$). Both first and second half of KAAI had significant reductions in KAAI when adopting a post-HTO wide stance gait compared to post-HTO unaltered level gait. Adopting a wide stance gait significantly reduced KAAI between heel strike and 16% of stance (0.06 (0.04) vs 0.05 (0.04), $p = 0.026$).

Between midstance and 83% of stance there was a significant reduction in KAAI when adopting a wide stance gait compared to unaltered level gait post-surgery. This reduction remained true between 84% and toe-off too.

When compared to post-HTO unaltered level gait, a wide stance gait significantly increased first and second half of stance abduction angular impulse ($p = 0.000$).

Table 7-16 Post-HTO Wide Stance Gait Knee Angular Impulse

	Controls	**Post-HTO NL**	**Post-HTO WS**	**Controls vs post-HTO NL**	**Controls vs post-HTO WS**	**Post NL vs Post WS**
%BW.h.s	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) angular impulse						
Stance	0.74 (0.28)	0.81 (0.39)	0.71 (0.36)	0.398	0.718	**0.002††**
1st half stance	0.43 (0.14)	0.46 (0.20)	0.41 (0.19)	0.502	0.750	**0.028†**
2nd half stance	0.31 (0.16)	0.35 (0.21)	0.29 (0.19)	0.346	0.408	**0.000††**
0–16% stance	0.06 (0.03)	0.06 (0.04)	0.05 (0.04)	0.699	0.585	**0.026†**
17%–midstance	0.36 (0.11)	0.39 (0.17)	0.35 (0.16)	0.466	0.834	0.057
Midstance–83% stance	0.26 (0.13)	0.30 (0.17)	0.26 (0.16)	0.444	0.682	**0.004††**
84%-100% stance	0.04 (0.02)	0.04 (0.03)	0.02 (0.02)	0.787	**0.005††**	**0.000****
Abduction (−) angular impulse in Stance						
1st half stance	-0.02 (0.01)	-0.02 (0.01)	-0.03 (0.02)	0.236	0.063	**0.000****
2nd half stance	-0.02 (0.02)	-0.01 (0.01)	-0.02 (0.02)	0.063	0.115	**0.000††**

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. % BW.h.s = % of body weight multiplied by height per second.

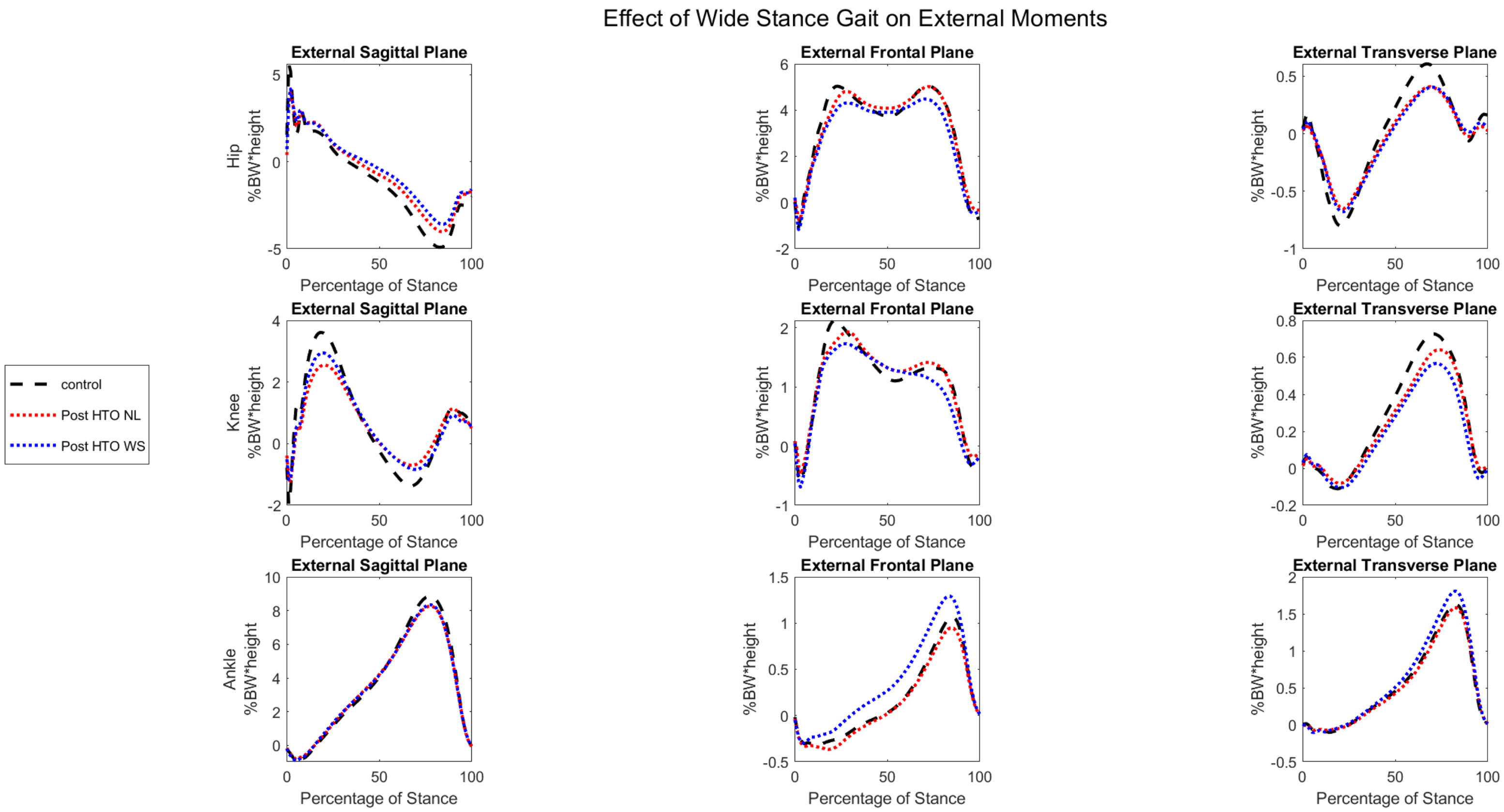


Figure 53 Post-HTO wide stance gait group average joint external moments

Positive values represent external moments for knee flexion, adduction, and internal moments.

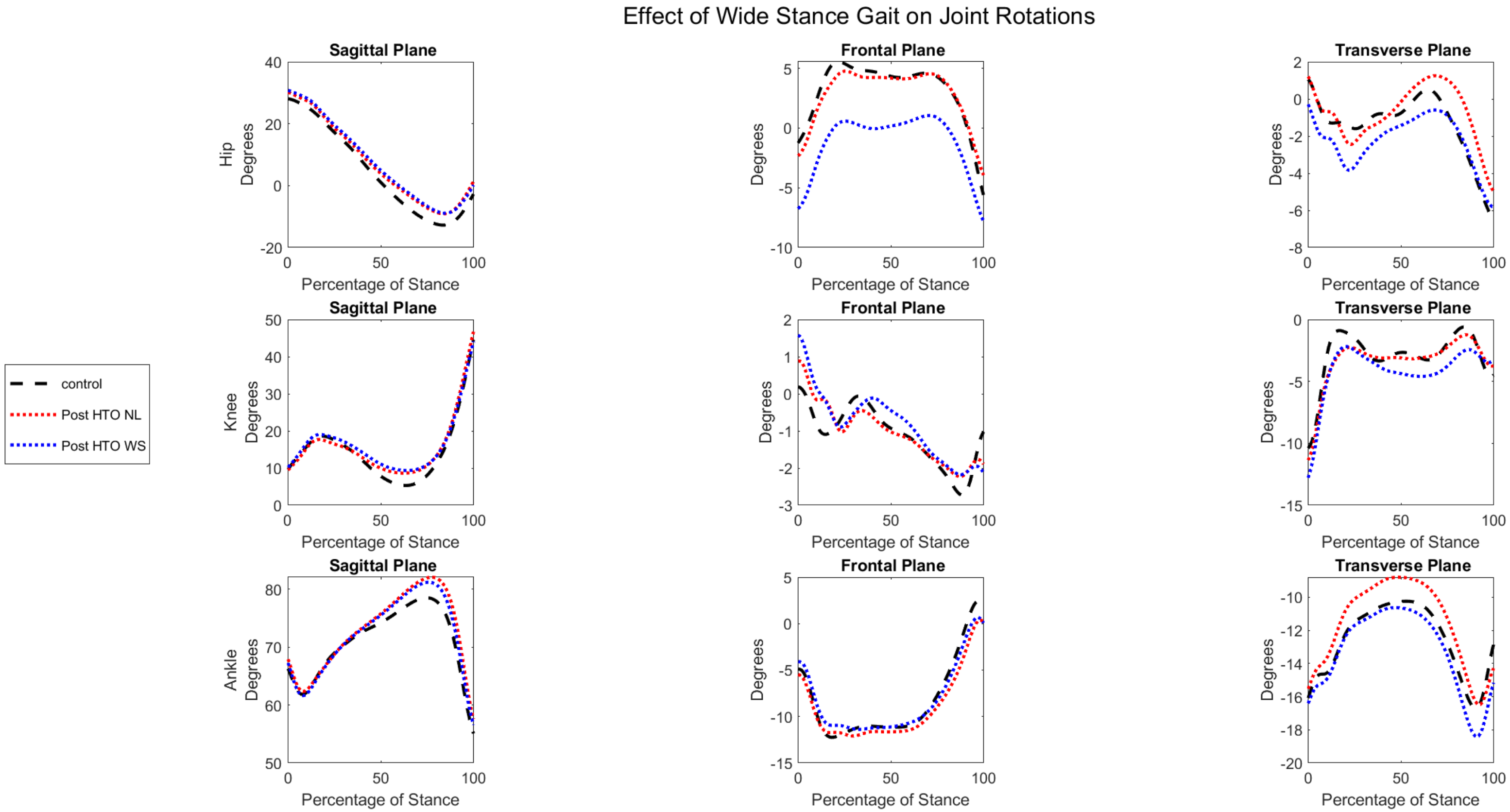


Figure 54 Post-HTO wide stance gait group average joint rotations

Positive values represent knee flexion, adduction, and internal rotations.

7.4.5 External ankle moments

During the whole of stance, adopting a toe out altered gait significantly increased peak external dorsiflexion moment compared to post-HTO unaltered level gait (8.01 %BW.h (1.42) vs 8.44 %BW.h (0.88), $p = 0.024$). Adopting a toe out altered gait significantly increased peak external plantarflexion moment compared to post-HTO unaltered level gait (0.81 %BW.h (0.36) vs 0.96 %BW.h (0.31), $p = 0.000$).

Wide stance gait also increased peak external inversion moment compared to post-HTO unaltered level gait (0.95 %BW.h (0.53) vs 1.34 %BW.h (0.58), $p = 0.000$. In addition to this, adopting a wide stance gait style also significantly increased the peak rotation moment (1.57 %BW.h (0.83) vs 1.87 %BW.h (0.74), $p = 0.000$). In addition to this, adopting a wide stance gait style also significantly increased the peak external rotation moment (0.15 %BW.h (0.09) vs 0.17 %BW.h (0.1), $p = 0.041$).

During the first half of stance, peak external plantarflexion moment significantly reduced when adopting a wide stance gait style (0.87 %BW.h (0.29) vs 0.96 %BW.h (0.31), $p = 0.002$). In the frontal plane there was a significant increase in peak inversion moment when adopting a wide stance gait compared to a post-HTO unaltered level gait (0.15 %BW.h (0.2) vs 0.33 %BW.h (0.26), $p = 0.000$), as well as significantly reducing peak eversion moment (0.49 %BW.h (0.24) vs 0.4 %BW.h (0.21), $p = 0.001$). In the transverse plane, adopting a wide stance gait resulted in a significant increase in peak internal and peak external rotation moment compared to a post-HTO unaltered level gait ($p < 0.05$).

During the second half of stance, peak plantar flexion moment was significantly reduced when adopting a wide stance gait compared to unaltered level gait post-HTO (0.06 %BW.h (0.1) vs 0.03 %BW.h (0.13), $p = 0.010$). In the frontal plane, peak inversion moment is significantly increased when adopting a wide stance gait when compared to a post-HTO unaltered level gait (1.00 %BW.h (0.52) vs 1.34 %BW.h (0.58), $p = 0.000$), as well as significantly reducing peak eversion moment (0.14 %BW.h (0.13) vs 0.04 %BW.h (0.07), $p = 0.000$). In the transverse plane, adopting a wide stance gait resulted in a significantly increased peak internal rotation moment compared to a post-HTO unaltered level gait (1.65 %BW.h (0.81) vs 1.87 %BW.h (0.74), $p = 0.000$).

Table 7-17 Post-HTO Wide Stance Gait External Ankle Moments

	Controls	**Post-HTO NL**	**Post-HTO WS**	**Controls vs post-HTO NL**	**Controls vs post-HTO WS**	**Post NL vs Post WS**
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak dorsi-flexion (+) moment	6.78 (3.93)	8.01 (1.42)	8.44 (0.88)	0.501	0.920	**0.024**†
Peak plantar-flexion (−) moment	-0.97 (0.36)	-0.81 (0.36)	-0.96 (0.31)	0.112	0.946	**0.000**††
Peak inversion (+) moment	0.82 (0.54)	0.95 (0.53)	1.34 (0.58)	0.787	**0.003**††	**0.000**††
Peak eversion (-) moment	-0.41 (0.22)	-0.42 (0.25)	-0.40 (0.21)	0.890	0.921	0.147
Peak internal rotation (+) moment	1.25 (0.85)	1.57 (0.83)	1.87 (0.74)	0.740	**0.037**†	**0.000****
Peak external rotation (−) moment	-0.17 (0.08)	-0.15 (0.09)	-0.17 (0.10)	0.261	0.979	**0.041**†
0-50% (HS to midstance)						
Peak dorsi-flexion (+) moment	4.08 (0.99)	4.22 (0.93)	4.19 (0.86)	0.589	0.660	0.517
Peak plantar-flexion (−) moment	-1.03 (0.28)	-0.87 (0.29)	-0.96 (0.31)	**0.048***	0.396	**0.002****
Peak inversion (+) moment	0.14 (0.14)	0.15 (0.20)	0.33 (0.26)	0.775	**0.001**††	**0.000**††
Peak eversion (-) moment	-0.43 (0.20)	-0.49 (0.24)	-0.40 (0.21)	0.282	0.932	**0.001****
Peak internal rotation (+) moment	0.47 (0.26)	0.44 (0.22)	0.52 (0.24)	0.552	0.275	**0.020***
Peak external rotation (−) moment	-0.18 (0.08)	-0.15 (0.09)	-0.17 (0.10)	0.248	0.812	**0.037***
50-100% (Midstance to toe-off)						
Peak dorsi-flexion (+) moment	8.94 (0.74)	8.35 (0.77)	8.44 (0.88)	**0.004****	**0.023***	0.187
Peak plantar-flexion (−) moment	-0.03 (0.12)	-0.06 (0.10)	-0.03 (0.13)	0.264	0.967	**0.010***
Peak inversion (+) moment	1.11 (0.34)	1.00 (0.52)	1.34 (0.58)	0.091	0.147	**0.000**††
Peak eversion (-) moment	-0.11 (0.14)	-0.14 (0.13)	-0.04 (0.07)	0.098	0.045†	**0.000****
Peak internal rotation (+) moment	1.67 (0.53)	1.65 (0.81)	1.87 (0.74)	0.206	0.616	**0.000****
Peak external rotation (−) moment	0.00 (0.06)	0.01 (0.04)	0.00 (0.05)	0.426	0.883	0.079

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. % BW.h = percentage of body weight multiplied by height. HS = heel strike.

7.4.6 External hip moments

During the whole of stance, adopting a wide stance gait resulted in a significantly increased peak external hip flexion moment compared to unaltered level gait post-HTO (4.7 (1.91) vs 5.25 (1.71), p = 0.024). During the whole of stance, adopting a wide stance gait resulted in a significantly reduced peak external hip extension compared to unaltered level gait post-HTO (4.1 (1.12) vs 3.83 (1.11), p = 0.000). Adopting a wide stance gait significantly reduced peak hip adduction moment compared to an unaltered gait (5.23 %BW.h (0.85) vs 5.04 %BW.h (0.83), p = 0.021) whilst also significantly increasing peak hip abduction moment (1.18 %BW.h (0.61) vs 1.59 %BW.h (0.73), p = 0.000). In the transverse plane, a wide stance altered gait significantly increased peak external hip moment compared to post-HTO unaltered level gait (0.69 %BW.h (0.29) vs 0.74 %BW.h (0.28), p = 0.015).

During the first half of stance, peak external hip extension moment significantly reduced when adopting a wide stance gait style (0.96 %BW.h (0.72) vs 0.69 %BW.h (0.69), p = 0.000). During the first half of stance, peak external hip adduction moment significantly reduced when adopting a wide stance gait style (5.11 %BW.h (0.81) vs 4.74 %BW.h (0.85), p = 0.000) whilst peak external hip abduction moment significantly increased (1.02 %BW.h (0.8) vs 1.4 %BW.h (0.93), p = 0.000). Adopting a wide stance gait significantly increased peak external hip rotation moment compared to post-HTO unaltered level gait (0.71 %BW.h (0.27) vs 0.74 %BW.h (0.28), p = 0.045).

During the second half of stance, peak external hip adduction moment significantly reduced when adopting a wide stance gait style (5.16 %BW.h (0.84) vs 4.72 %BW.h (0.85), p = 0.000). During the second half of stance, peak external hip abduction moment significantly increased when adopting a wide stance gait style (0.46 %BW.h (0.54) vs 0.63 %BW.h (0.61), p = 0.004).

Table 7-18 Post-HTO Wide Stance Gait External Hip Moments

	Controls	**Post-HTO NL**	**Post-HTO WS**	**Controls vs post-HTO NL**	**Controls vs post-HTO WS**	**Post NL vs Post WS**
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak external hip flexion (+) moment	6.31 (2.64)	4.70 (1.91)	5.25 (1.71)	**0.010****	0.076	**0.024†**
Peak external hip extension (−) moment	-4.36 (1.70)	-4.10 (1.12)	-3.83 (1.11)	0.492	0.165	**0.000****
Peak external hip adduction (+) moment	5.09 (1.41)	5.23 (0.85)	5.04 (0.83)	0.639	0.893	**0.021***
Peak external hip abduction (-) moment	-1.65 (1.03)	-1.18 (0.61)	-1.59 (0.73)	**0.047†**	0.763	**0.000****
Peak external hip internal rotation (+) moment	0.54 (0.30)	0.48 (0.22)	0.49 (0.20)	0.562	0.496	0.495
Peak external hip rotation (−) moment	-0.80 (0.43)	-0.69 (0.29)	-0.74 (0.28)	0.247	0.553	**0.015***
0-50% (HS to midstance)						
Peak external hip extension (−) moment	-1.25 (0.92)	-0.96 (0.72)	-0.69 (0.69)	0.142	**0.011***	**0.000††**
Peak external hip adduction (+) moment	5.27 (1.03)	5.11 (0.81)	4.74 (0.85)	0.513	**0.036***	**0.000††**
Peak external hip abduction (-) moment	-1.28 (1.29)	-1.02 (0.80)	-1.40 (0.93)	0.361	0.677	**0.000****
Peak external hip internal rotation (+) moment	0.34 (0.20)	0.27 (0.15)	0.27 (0.14)	0.180	0.130	0.975
Peak external hip rotation (−) moment	-0.86 (0.42)	-0.71 (0.27)	-0.74 (0.28)	0.116	0.212	**0.045†**
50-100% (Midstance to toe-off)						
Peak external hip adduction (+) moment	5.20 (1.17)	5.16 (0.84)	4.72 (0.85)	0.886	0.078	**0.000****
Peak external hip abduction (-) moment	-0.86 (0.55)	-0.46 (0.54)	-0.63 (0.61)	**0.007****	0.134	**0.004****
Peak external hip internal rotation (+) moment	0.64 (0.24)	0.46 (0.28)	0.44 (0.24)	**0.009****	**0.003****	0.992
Peak external hip rotation (−) moment	-0.11 (0.12)	-0.11 (0.15)	-0.08 (0.12)	0.573	0.465	0.181

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. % BW.h = percentage of body weight multiplied by height. HS = heel strike.

7.4.7 Concurrent Optimisation of Muscle and Secondary Kinematics

Table 7-19 shows group gait speed between the three groups. Post-surgery wide stance gait did not alter gait speed nor change stance time when compared to post-HTO unaltered level gait.

Table 7-19 Post-HTO Wide Stance Gait COMAK Gait Speed

Demographics	**Controls**	**Post-HTO NL**	**Post-HTO WS**	**Controls vs post-HTO NL**	**Controls vs post-HTO WS**	**Post NL vs Post WS**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Gait speed (m/s)	1.26 (0.17)	1.15 (0.17)	1.16 (0.20)	**0.020****	0.051	0.351

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std= standard deviation. m/s = metres per second.

7.4.8 Internal knee joint loading

At FP, total and lateral compartment contact force, mean and maximum pressure significantly increased because of adopting a wide stance gait compared to an unaltered level. No significantly changes occurred for FP medial compartment loading.

At MS, the only significant differences between a wide stance gait and an unaltered level gait were for medial compartment contact force, for which a wide stance gait significantly reduced medial compartment contact force.

At SP, medial compartment knee contact forces significantly decreased when adopting a wide stance gait compared to unaltered level gait post-HTO.

7.4.9 Internal knee joint loading ratios

At FP and SP, adopting a wide stance gait significantly reduced medial to total contact force ratios, and significantly increased lateral to total contact force ratios when compared to a post-HTO unaltered level gait ($p = 0.000$).

7.4.10 Point of application

At FP, total knee point of application shifted laterally when adopting a wide stance gait compared to an unaltered level gait post-HTO, as well as shifting significantly more posterior. Lateral compartment point of application at FP was shifted more anterior.

At MS, total knee and lateral compartment knee point of application shifted more posterior when adopting a wide stance gait compared to unaltered level gait post-HTO.

At SP, total knee compartment medial-lateral point of direction changed significantly between the post-HTO unaltered level gait and post-HTO wide stance gait style with the point of application occurring more laterally when adopting a wide stance gait.

7.4.11 Contact area

At FP, there was a significant increase in total knee and lateral compartment contact area when adopting a wide stance when compared to a post-surgery unaltered level gait.

At MS and SP, there were no significant differences between the post-HTO unaltered level gait and a wide stance gait.

Table 7-20 Post-HTO Wide Stance Gait Internal Knee Joint Loading

	Controls	Post-HTO NL	Post-HTO WS	Controls vs post-HTO NL	Controls vs post-HTO WS	Post NL vs Post WS
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total knee						
Contact force [BW]	2.74 (0.57)	2.33 (0.52)	2.53 (0.62)	**0.006****	0.197	**0.003****
Mean pressure [MPa]	5.63 (1.25)	5.77 (1.13)	6.13 (1.20)	0.671	0.054	**0.018†**
Max pressure	12.92 (3.32)	13.22 (2.46)	14.63 (3.15)	0.501	**0.032†**	**0.004**
Medial knee						
Contact force [BW]	1.69 (0.31)	1.46 (0.36)	1.48 (0.39)	**0.011***	**0.023***	0.662
Mean pressure [MPa]	5.76 (1.12)	5.95 (1.12)	5.95 (1.18)	0.526	0.531	0.974
Max pressure	12.21 (2.52)	12.69 (2.45)	12.72 (2.45)	0.469	0.442	0.864
Lateral knee						
Contact force [BW]	1.13 (0.37)	0.94 (0.27)	1.14 (0.36)	**0.027***	0.929	**0.000††**
Mean pressure [MPa]	5.37 (1.62)	5.45 (1.43)	6.29 (1.65)	0.883	**0.042†**	**0.000††**
Max pressure	11.57 (3.76)	11.78 (3.12)	13.84 (3.81)	0.671	**0.021†**	**0.000††**
Midstance						
Total knee						
Contact force [BW]	1.25 (0.24)	1.23 (0.18)	1.21 (0.18)	0.871	0.717	0.075
Mean pressure [MPa]	3.40 (0.36)	3.94 (0.58)	3.97 (0.68)	**0.000****	**0.000****	0.750
Max pressure	7.62 (1.12)	8.93 (1.59)	8.89 (1.71)	**0.001****	**0.001****	0.405
Medial knee						
Contact force [BW]	0.92 (0.22)	0.89 (0.20)	0.86 (0.22)	0.648	0.370	**0.043†**
Mean pressure [MPa]	3.81 (0.58)	4.31 (0.70)	4.28 (0.79)	**0.004****	**0.013***	0.329
Max pressure	7.43 (1.22)	8.59 (1.51)	8.54 (1.56)	**0.002****	**0.004****	0.697
Lateral knee						
Contact force [BW]	0.36 (0.14)	0.37 (0.15)	0.37 (0.19)	0.660	0.787	0.544
Mean pressure [MPa]	2.59 (0.54)	3.04 (1.06)	3.05 (1.22)	0.070	0.138	0.894
Max pressure	5.56 (1.07)	6.42 (2.17)	6.47 (2.49)	0.065	0.130	0.830
Second peak						
Total knee						
Contact force [BW]	2.90 (0.70)	2.43 (0.54)	2.38 (0.62)	**0.006****	**0.004****	0.417
Mean pressure [MPa]	5.32 (0.64)	5.77 (1.11)	5.66 (1.05)	0.147	0.268	0.417

Max pressure	12.70 (1.76)	13.34 (3.05)	13.05 (2.77)	0.811	0.859	0.393
Medial knee						
Contact force [BW]	1.83 (0.35)	1.51 (0.37)	1.41 (0.40)	**0.001****	**0.000****	**0.045***
Mean pressure [MPa]	5.91 (0.80)	6.14 (1.23)	5.86 (1.29)	0.386	0.365	0.054
Max pressure	12.61 (1.80)	12.65 (2.76)	12.07 (2.91)	0.605	0.088	0.178
Lateral knee						
Contact force [BW]	1.20 (0.44)	1.01 (0.30)	1.06 (0.36)	0.072	0.198	0.188
Mean pressure [MPa]	4.53 (0.73)	5.23 (1.39)	5.33 (1.33)	**0.014†**	**0.002††**	0.381
Max pressure	9.43 (1.48)	11.20 (3.30)	11.37 (2.95)	**0.003††**	**0.001††**	0.528

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. BW = body weight; MPa = megapascals. Max = maximum.

Table 7-21 Post-HTO Wide Stance Gait Contact Force Ratios

	Controls		Post-HTO NL	Post-HTO WS	Controls vs post-HTO NL	Controls vs post-HTO WS	Post NL vs Post WS
	Mean (std)		Mean (std)	Mean (std)	P value	P value	P value
First peak							
MED / TOTAL	0.63 (0.07)		0.63 (0.08)	0.59 (0.09)	0.920	0.056	**0.000††**
LAT / TOTAL	0.41 (0.07)		0.40 (0.08)	0.45 (0.09)	0.879	0.057	**0.000††**
Midstance							
MED / TOTAL	0.73 (0.09)		0.72 (0.12)	0.72 (0.14)	0.705	0.717	0.877
LAT / TOTAL	0.29 (0.10)		0.30 (0.12)	0.31 (0.15)	0.763	0.682	0.975
Second peak							
MED / TOTAL	0.64 (0.07)		0.62 (0.08)	0.59 (0.09)	0.302	**0.004††**	**0.000††**
LAT / TOTAL	0.40 (0.07)		0.41 (0.08)	0.44 (0.09)	0.463	**0.020†**	**0.000††**

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. MED = medial compartment tibiofemoral joint; LAT = lateral compartment tibiofemoral joint; TOTAL = total tibiofemoral joint.

Table 7-22 Post-HTO Wide Stance Gait Point of Application

	Controls	Post-HTO NL	Post-HTO TO	Controls vs post-HTO NL	Controls vs post-HTO WS	Post NL vs Post WS
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total knee						
Anterior (+) / posterior (-)	-2.46 (2.08)	-2.56 (2.20)	-3.17 (2.19)	0.728	0.115	**0.045†**
Lateral (+) / medial (-)	-2.27 (3.27)	-2.44 (3.43)	-1.02 (3.54)	0.845	0.168	**0.000††**
Medial knee						
Anterior (+) / posterior (-)	-1.01 (2.25)	-1.18 (2.12)	-1.05 (2.18)	0.811	0.811	0.537
Lateral (+) / medial (-)	-17.19 (1.29)	-17.44 (1.53)	-17.63 (1.62)	0.493	0.255	0.106
Lateral knee						
Anterior (+) / posterior (-)	-4.67 (1.83)	-4.65 (2.27)	-5.73 (2.31)	0.847	**0.026†**	**0.000††**
Lateral (+) / medial (-)	20.59 (2.34)	20.89 (2.08)	20.75 (1.75)	0.614	0.778	0.159
Midstance						
Total knee						
Anterior (+) / posterior (-)	4.66 (1.86)	4.64 (1.96)	4.17 (1.9)	0.974	0.331	**0.041†**
Lateral (+) / medial (-)	-5.84 (4.72)	-5.47 (5.84)	-5.17 (6.91)	0.791	0.663	0.601
Medial knee						
Anterior (+) / posterior (-)	6.63 (2.77)	6.34 (2.74)	6.28 (2.75)	0.688	0.627	0.926
Lateral (+) / medial (-)	-16.33 (1.53)	-16.65 (1.82)	-16.51 (1.94)	0.453	0.501	0.420
Lateral knee						
Anterior (+) / posterior (-)	-0.42 (1.55)	0.32 (1.74)	-0.79 (1.99)	0.090	0.443	**0.001††**
Lateral (+) / medial (-)	20.34 (2.43)	20.88 (3.22)	20.88 (2.83)	0.471	0.365	0.271
Second peak						
Total knee						
Anterior (+) / posterior (-)	5.70 (1.64)	4.94 (3.22)	4.72 (3.04)	0.717	0.242	0.527
Lateral (+) / medial (-)	-1.39 (2.90)	-0.68 (3.24)	0.54 (4.01)	0.417	**0.005††**	**0.002****
Medial knee						
Anterior (+) / posterior (-)	9.56 (1.83)	8.26 (3.68)	7.86 (3.86)	0.201	**0.047†**	0.288
Lateral (+) / medial (-)	-14.59 (1.31)	-15.46 (1.27)	-15.43 (1.43)	**0.013***	**0.023***	0.811
Lateral knee						
Anterior (+) / posterior (-)	-0.89 (2.43)	-0.39 (2.73)	-0.10 (2.85)	0.466	0.264	0.374
Lateral (+) / medial (-)	19.74 (2.84)	21.55 (2.9)	21.82 (3.10)	**0.020***	**0.010***	0.278

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. mm = millimetres. X = anterior; Z = lateral. COP = centre of pressure.

Table 7-23 Post-HTO Wide Stance Gait Knee Contact Area

	Controls	Post-HTO NL	Post-HTO TO	Controls vs post-HTO NL	Controls vs post-HTO WS	Post NL vs Post WS
mm^2	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total	352.35 (39.00)	362.97 (49.26)	370.76 (49.50)	0.453	0.151	**0.040***
Medial	206.80 (22.63)	213.55 (32.49)	214.48 (29.25)	0.472	0.349	0.147
Lateral	145.55 (20.95)	149.43 (21.45)	156.28 (24.74)	0.242	0.050	**0.001****
Midstance						
Total	262.26 (34.10)	282.31 (46.27)	275.45 (38.44)	0.080	0.174	0.141
Medial	167.14 (24.46)	180.35 (29.87)	175.45 (25.96)	0.108	0.215	0.237
Lateral	95.12 (19.62)	101.96 (27.28)	99.99 (29.57)	0.111	0.224	0.673
Second peak						
Total	399.23 (90.44)	387.36 (90.55)	382.63 (87.84)	0.620	0.481	0.499
Medial	219.07 (41.52)	217.24 (46.55)	210.54 (43.89)	0.763	0.451	0.078
Lateral	180.16 (52.60)	170.12 (46.05)	172.09 (47.45)	0.442	0.541	0.601
Lateral	145.55 (20.95)	149.43 (21.45)	156.28 (24.74)	0.242	0.050	**0.001****
Midstance						
Total	262.26 (34.10)	282.31 (46.27)	275.45 (38.44)	0.080	0.174	0.141
Medial	167.14 (24.46)	180.35 (29.87)	175.45 (25.96)	0.108	0.215	0.237
Lateral	95.12 (19.62)	101.96 (27.28)	99.99 (29.57)	0.111	0.224	0.673
Second peak						
Total	399.23 (90.44)	387.36 (90.55)	382.63 (87.84)	0.620	0.481	0.499
Medial	219.07 (41.52)	217.24 (46.55)	210.54 (43.89)	0.763	0.451	0.078
Lateral	180.16 (52.60)	170.12 (46.05)	172.09 (47.45)	0.442	0.541	0.601

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. WS = wide stance gait; NL = unaltered level gait. std = standard deviation. Measurement = mm^2.

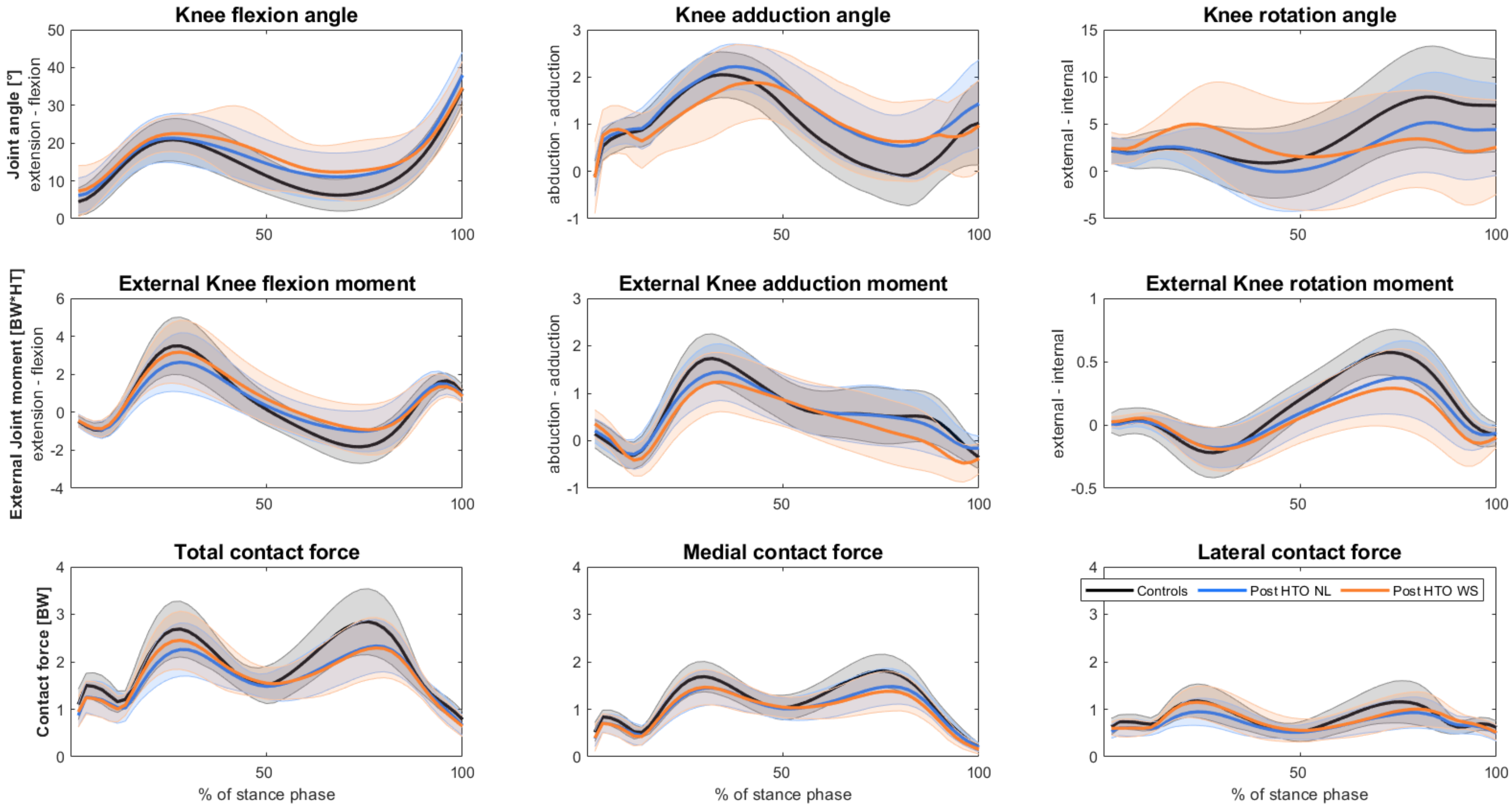


Figure 55 Post-HTO wide stance gait knee joint kinematics, external moments, contact forces

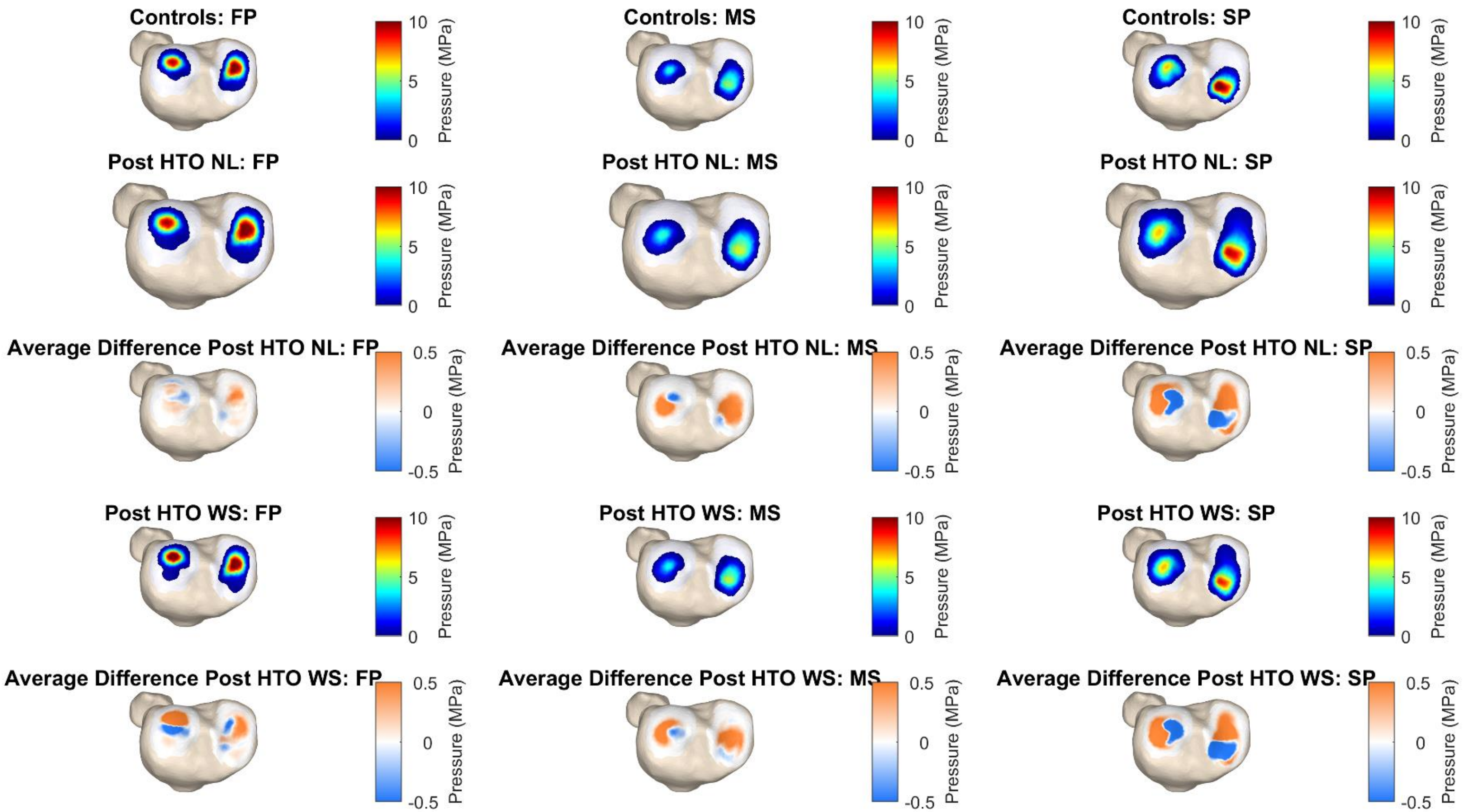


Figure 56 Post-HTO wide stance contact pressure distribution on the tibia

7.5 Medial thrust gait

7.5.1 Quantifying medial thrust gait

The metric used to determine whether an individual successfully adapted their gait to a medial thrust gait was whether an individual could decrease their maximum knee adduction angle during the first half of stance compared to when walking with an unaltered level gait. Only 14 patients were able to successfully adopt a medial thrust altered gait style. Adopting a medial thrust gait resulted in a maximum knee adduction angle in the first half of stance of ~0° knee adduction compared to a post-HTO unaltered level gait peak knee adduction angle during first half of stance being ~1° ($p = 0.000$).

Table 7-24 Post-HTO Medial Thrust Gait Quantifying Gait Style (n = 14)

	Controls	**Post-HTO NL**	**Post-HTO MT**	**Controls vs post-HTO NL**	**Controls vs post-HTO MT**	**Pre NL vs Post MT**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Maximum knee adduction angle (+) in first half of stance (°)	1.53 (3.87)	1.13 (4.71)	-0.37 (5.35)	0.885	0.194	**0.000††**

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. std = standard deviation. m = metre. Only 14 patients were able to successfully perform a medial thrust gait, as determined by reducing their maximum knee adduction angle in first half of stance.

7.5.2 Spatial-temporal parameters

Adopting a medial thrust gait resulted in many spatiotemporal changes, as shown in Table 7-25. Most noticeably, a medial thrust gait reduced gait speed when compared to post-HTO unaltered level gait.

Table 7-25 Post-HTO Medial Thrust Gait Spatial Temporal Parameters

	Controls	**Post-HTO NL**	**Post-HTO MT**	**Controls vs post-HTO NL**	**Controls vs post-HTO MT**	**Pre NL vs Post MT**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Operative limb cycle time (s)	1.08 (0.08)	1.14 (0.07)	1.18 (0.09)	**0.015†**	**0.001††**	**0.044***
Operative limb stance time (s)	0.65 (0.06)	0.70 (0.05)	0.74 (0.07)	**0.003****	**0.000††**	**0.014***
Operative limb step length (m)	0.64 (0.07)	0.66 (0.08)	0.65 (0.09)	0.443	0.803	0.146
Operative limb step time (s)	0.54 (0.04)	0.57 (0.04)	0.59 (0.05)	**0.037†**	**0.001****	**0.040***
Operative limb stride length (m)	1.29 (0.13)	1.32 (0.16)	1.30 (0.19)	0.592	0.886	0.312
Swing time (s)	0.43 (0.03)	0.44 (0.03)	0.44 (0.03)	0.407	0.248	0.609
Speed (m/s)	1.21 (0.16)	1.16 (0.15)	1.11 (0.18)	0.398	0.090	**0.016***

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. std = standard deviation. s = seconds; m = metre; m/s = metre/second.

7.5.3 Knee joint loading: External moments

Adopting a medial thrust gait significantly reduced maximum EKAM when compared to an unaltered level gait post-surgery (2.16%BW.h (0.86) vs 1.74 %BW.h (0.66), $p = 0.001$).

Adopting a medial thrust gait significantly reduced EKAM1 (2.07 %BW.h (0.8) vs 1.57 %BW.h (0.6), $p = 0.001$) and EKAM2 (1.54% BW.h (0.86 vs 1.29 %BW.h (0.73), $p = 0.002$) when compared to an unaltered level gait post-surgery.

Adopting a medial thrust gait significantly increased peak flexion moment when compared to post-HTO unaltered level gait 3.19 %BW.h (1.17) vs 4.72 %BW.h (1.68), $p = 0.000$), as well as significantly reducing peak extension moment (1.79 %BW.h (0.56) vs 1.37 %BW.h (0.52), $p = 0.038$).

In terms of the peak transverse plane knee moment changes, adopting a medial thrust gait compared to unaltered level gait significantly reduced peak internal rotation moment and significantly increased peak external rotation moment.

Table 7-26 Post-HTO Medial Thrust Gait External Knee Moments

	Controls	**Post-HTO**	**Post-HTO**	**Controls vs Post-HTO**	**Controls vs Post-HTO**	**Post NL vs Post**
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) moment						
Maximum	2.11 (0.81)	2.16 (0.86)	1.74 (0.66)	0.864	0.148	**0.001**
1st peak (1st half stance)	2.27 (0.65)	2.07 (0.80)	1.57 (0.60)	0.395	**0.002****	**0.001****
2nd peak (2nd half stance)	1.50 (0.67)	1.54 (0.86)	1.29 (0.73)	0.855	0.364	**0.002****
Midstance	1.15 (0.49)	1.33 (0.68)	1.10 (0.65)	0.340	0.795	**0.008****
Flexion (+) moment peak	3.62 (1.65)	3.19 (1.17)	4.72 (1.68)	0.386	0.107	**0.000††**
Extension (−) moment peak	-2.45 (0.85)	-1.79 (0.56)	-1.37 (0.52)	**0.005††**	**0.000††**	**0.038***
Internal (+) rotation	0.60 (0.37)	0.62 (0.31)	0.52 (0.27)	0.861	0.510	**0.014***
External (−) rotation	-0.16 (0.08)	-0.14 (0.07)	-0.24 (0.12)	0.404	**0.018***	**0.000††**

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. % BW.h = percentage of body weight multiplied by height.

7.5.4 Knee joint loading: Impulses

Adopting a medial thrust gait significantly reduced KAAI over stance when compared to post-surgery unaltered level gait (0.81 %BW.h.s (0.4) vs 0.66 %BW.h.s (0.32), p = 0.001). Adopting a medial thrust gait significantly reduced KAAI over the first half of stance when compared to post-surgery unaltered level gait (0.46 %BW.h.s (0.2) vs 0.35 %BW.h.s (0.16), p = 0.000). Adopting a medial thrust gait significantly reduced KAAI over the second half of stance when compared to post-surgery unaltered level gait (0.35 %BW.h.s (0.21) vs 0.3 %BW.h.s (0.18), p = 0.015).

Between heel strike and 16% of stance, adopting a medial thrust gait significantly reduced KAAI (0.06 %BW.h.s (0.04) vs 0.04 %BW.h.s (0.03), p = 0.000). Between 17% of stance and midstance, adopting a medial thrust gait significantly reduced KAAI (0.39 %BW.h.s (0.16) vs 0.31 %BW.h.s (0.14), p = 0.001). Between 84% of stance and toe-off, adopting a medial thrust gait significantly reduced KAAI (0.3 %BW.h.s (0.18) vs 0.26 %BW.h.s (0.16), p = 0.020). When compared to post-HTO unaltered level gait, a medial thrust gait significantly increased first half of stance abduction angular impulse (p = 0.000).

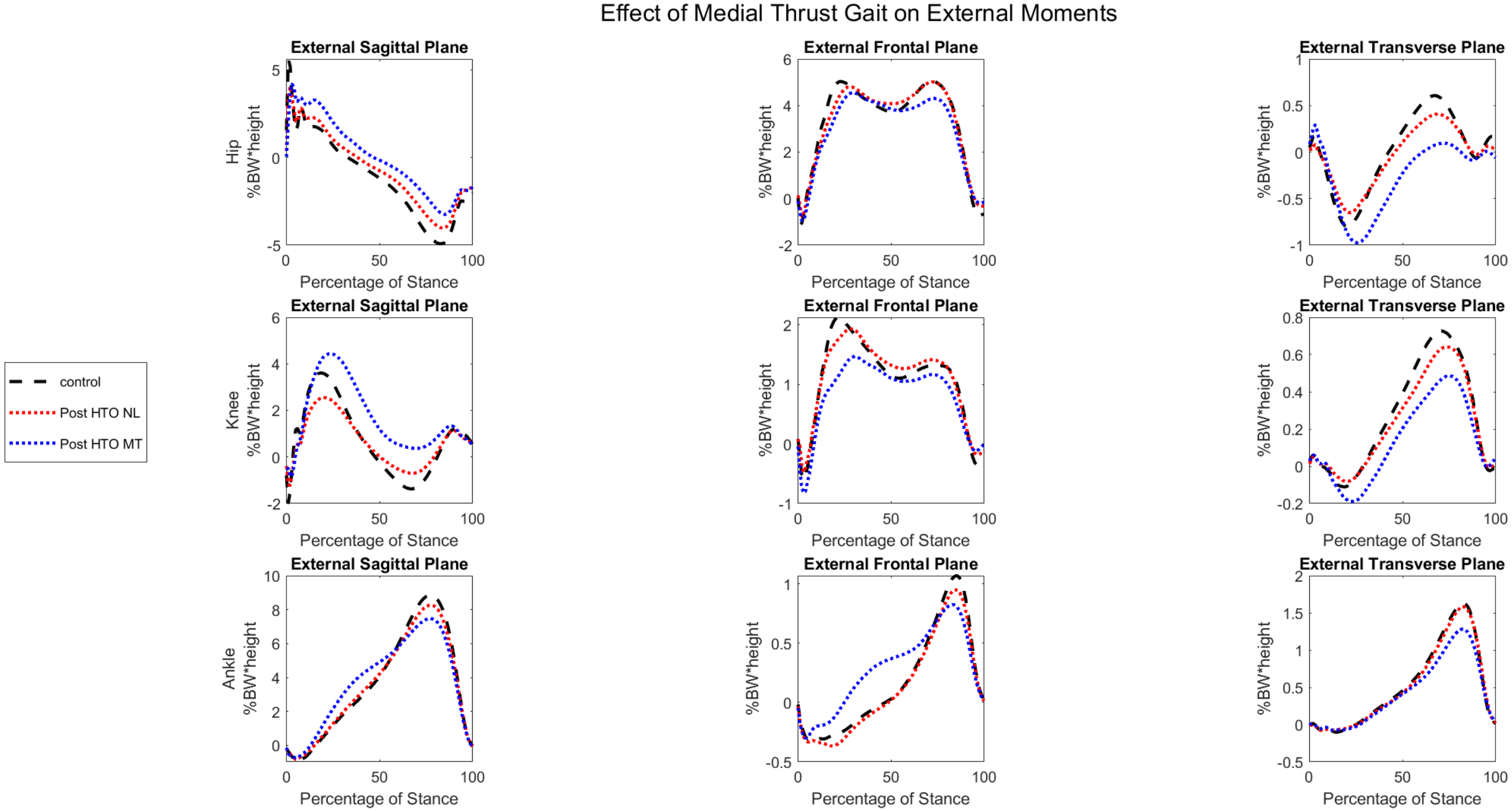


Figure 57 Post-HTO medial thrust gait group average joint external moments

Positive values represent external moments for knee flexion, adduction, and internal moments.

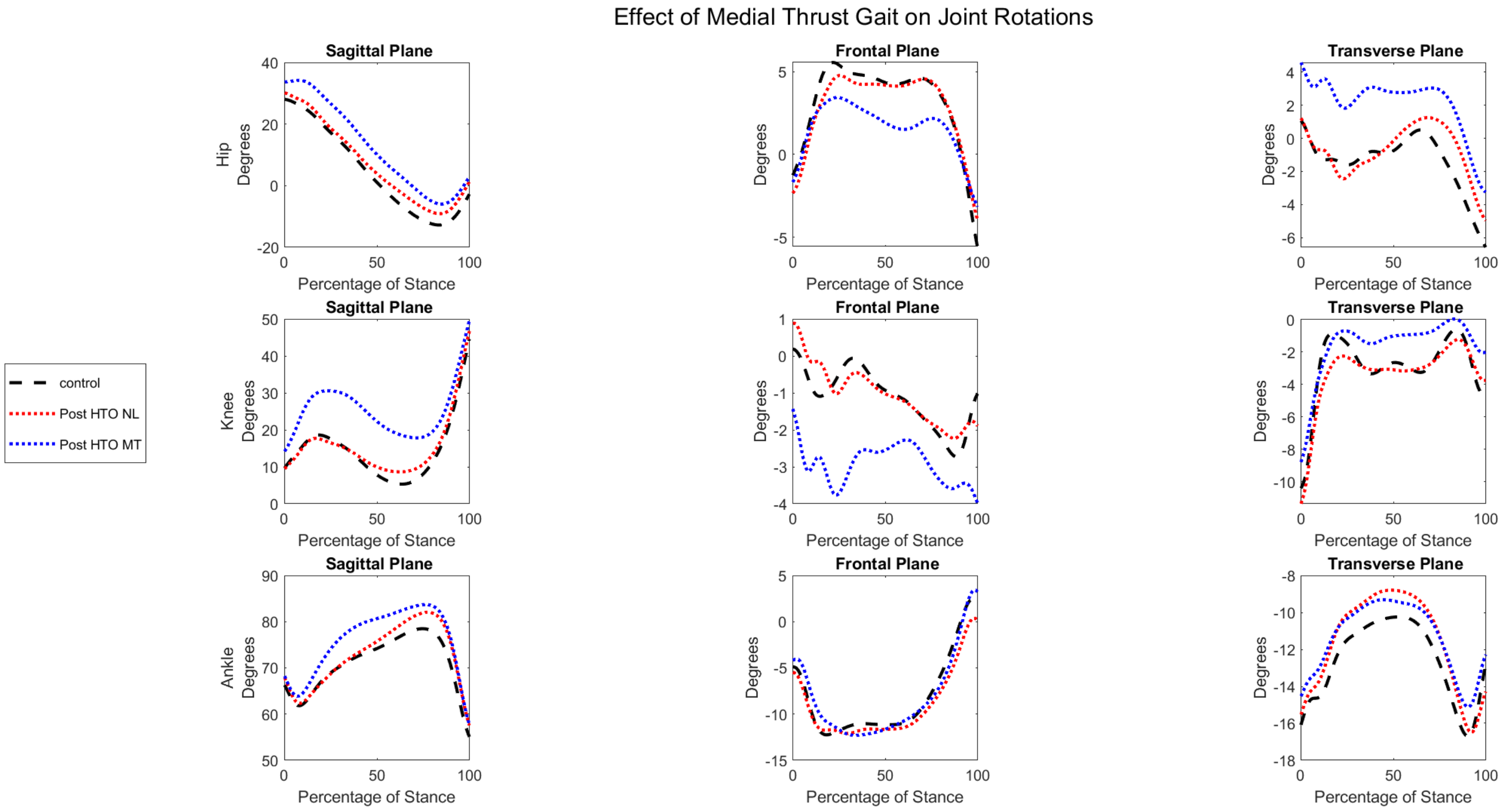


Figure 58 Post-HTO medial thrust gait group average joint rotations

Positive values represent knee flexion, adduction, and internal rotations.

Table 7-27 Post-HTO Medial Thrust Gait Knee Angular Impulse

	Controls	Post-HTO NL	Post-HTO MT	Controls vs Post-HTO NL	Controls vs Post-HTO MT	Post NL vs Post MT
%BW.h.s	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Adduction (+) angular impulse						
Stance	0.74 (0.28)	0.81 (0.40)	0.66 (0.32)	0.487	0.408	**0.001****
1st half stance	0.43 (0.14)	0.46 (0.20)	0.35 (0.16)	0.599	0.129	**0.000** **
2nd half stance	0.31 (0.16)	0.35 (0.21)	0.30 (0.18)	0.472	0.889	**0.015***
0–16% stance	0.06 (0.03)	0.06 (0.04)	0.04 (0.03)	0.613	0.076	**0.000††**
17%–midstance	0.36 (0.11)	0.39 (0.16)	0.31 (0.14)	0.592	0.204	**0.001** **
Midstance–83% stance	0.26 (0.13)	0.30 (0.18)	0.26 (0.16)	0.403	0.968	**0.020***
84%-100% stance	0.04 (0.02)	0.04 (0.03)	0.03 (0.02)	0.975	0.345	0.145
Abduction (−) angular impulse in Stance						
1st half stance	-0.02 (0.01)	-0.02 (0.02)	-0.04 (0.02)	0.571	**0.005** **	**0.000** **
2nd half stance	-0.02 (0.02)	-0.01 (0.01)	-0.02 (0.02)	0.071	0.260	0.194

Significant difference (p < 0.01) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. std = standard deviation. % BW.h.s = % of body weight multiplied by height per second.

7.5.5 External ankle moments

When adopting a medial thrust gait compared to an unaltered level gait, significant changes occurred in both the first and second half of stance.

During the first half of stance, peak dorsiflexion moment significantly increased when adopting a medial thrust gait style (4.22 %BW.h (0.92) vs 4.98 %BW.h (0.98), $p = 0.012$), as well as a significant decrease in peak plantarflexion moment (0.96 %BW.h (0.32) vs 0.81 %BW.h (0.34), $p = 0.011$). In the frontal plane there was a significant increase in peak inversion moment when adopting a medial thrust gait compared to a post-HTO unaltered level gait (0.18 %BW.h (0.23) vs 0.43 %BW.h (0.25), $p = 0.000$), as well as significantly reducing peak eversion moment (0.48 %BW.h (0.2) vs 0.38 %BW.h (0.16), $p = 0.009$).

During the second half of stance, peak dorsiflexion moment significantly decreased when adopting a medial thrust gait style (8.14 %BW.h (0.7) vs 7.67 %BW.h (0.8), $p = 0.002$). Additionally, adopting a medial thrust significantly decreased peak eversion moment (0.13 %BW.h (0.11) vs 0.04 %BW.h (0.08), $p = 0.009$). Finally, adopting a medial thrust significantly decreased peak internal rotation moment (1.46 %BW.h (0.67) vs 1.32 %BW.h (0.74), $p = 0.027$).

Table 7-28 Post-HTO Medial Thrust Gait External Ankle Moments

	Controls	**Post-HTO NL**	**Post-HTO MT**	**Controls vs post-HTO NL**	**Controls vs post-HTO MT**	**Post NL vs Post MT**
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak dorsi-flexion (+) moment	6.78 (3.93)	7.97 (1.15)	7.68 (0.79)	0.308	0.101	**0.049†**
Peak plantar-flexion (−) moment	-0.97 (0.36)	-0.96 (0.32)	-0.81 (0.33)	0.996	0.193	**0.010****
Peak inversion (+) moment	0.82 (0.54)	0.89 (0.41)	0.89 (0.40)	0.864	0.864	0.999
Peak eversion (-) moment	-0.41 (0.22)	-0.48 (0.20)	-0.38 (0.15)	0.328	0.730	**0.017***
Peak internal rotation (+) moment	1.25 (0.85)	1.43 (0.70)	1.32 (0.74)	0.968	0.722	0.176
Peak external rotation (−) moment	-0.17 (0.08)	-0.17 (0.08)	-0.19 (0.09)	0.958	0.334	0.119
0-50% (HS to midstance)						
Peak dorsi-flexion (+) moment	4.08 (0.99)	4.22 (0.92)	4.98 (0.98)	0.669	**0.008****	**0.012****
Peak plantar-flexion (−) moment	-1.03 (0.28)	-0.96 (0.32)	-0.81 (0.34)	0.520	**0.032***	**0.011***
Peak inversion (+) moment	0.14 (0.14)	0.18 (0.23)	0.43 (0.25)	0.742	**0.000††**	**0.000****
Peak eversion (-) moment	-0.43 (0.20)	-0.48 (0.20)	-0.38 (0.16)	0.180	0.553	**0.009****
Peak internal rotation (+) moment	0.47 (0.26)	0.38 (0.17)	0.44 (0.19)	0.249	0.802	0.256
Peak external rotation (−) moment	-0.18 (0.08)	-0.17 (0.08)	-0.19 (0.09)	0.840	0.571	0.100
50-100% (Midstance to toe-off)						
Peak dorsi-flexion (+) moment	8.94 (0.74)	8.14 (0.70)	7.67 (0.80)	**0.002****	**0.000****	**0.002****
Peak plantar-flexion (−) moment	-0.03 (0.12)	-0.10 (0.10)	-0.08 (0.13)	0.097	0.243	0.403
Peak inversion (+) moment	1.11 (0.34)	0.93 (0.39)	0.89 (0.40)	0.128	0.068	0.457
Peak eversion (-) moment	-0.11 (0.14)	-0.13 (0.11)	-0.04 (0.08)	0.107	0.056	**0.009****
Peak internal rotation (+) moment	1.67 (0.53)	1.46 (0.67)	1.32 (0.74)	0.287	**0.016†**	**0.027***
Peak external rotation (−) moment	0.00 (0.06)	0.01 (0.04)	0.02 (0.05)	0.664	0.056	0.182

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. std = standard deviation. % BW.h = percentage of body weight multiplied by height. HS = heel strike.

7.5.6 External hip moments

When adopting a medial thrust gait compared to an unaltered level gait, significant changes at the hip moments occurred in both the first and second half of stance.

During the first half of stance, peak external hip adduction moment significantly reduced when adopting a medial thrust gait style (5.23 %BW.h (0.93) vs 4.78 %BW.h (0.9), $p = 0.007$). A medial thrust altered gait significantly increased peak external hip internal rotation moment compared to post-HTO unaltered level gait (0.27 %BW.h (0.13) vs 0.35 %BW.h (0.2), $p = 0.041$). A medial thrust altered gait significantly increased peak external hip external rotation moment compared to post-HTO unaltered level gait (0.84 %BW.h (0.24) vs 1.05 %BW.h (0.23), $p = 0.000$).

During the second half of stance, peak external hip adduction moment significantly reduced when adopting a medial thrust gait style (5.14 %BW.h (0.94) vs 4.61 %BW.h (1.09), $p = 0.000$). Peak external hip abduction moment significantly decreased when adopting a medial thrust gait style (0.5 %BW.h (0.61) vs 0.33 %BW.h (0.68), $p = 0.044$). Peak external hip internal rotation moment significantly decreased when adopting a medial thrust gait style (0.35 %BW.h (0.23) vs 0.21 %BW.h (0.19), $p = 0.002$), whilst peak external hip external moment significantly increased when adopting a medial thrust gait style (0.16 %BW.h (0.13) vs 0.31 %BW.h (0.17), $p = 0.000$).

Table 7-29 Post-HTO Medial Thrust Gait External Hip Moments

	Controls	Post-HTO NL	Post-HTO MT	Controls vs post-HTO NL	Controls vs post-HTO MT	Post NL vs Post MT
%BW.h	**Mean (std)**	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
0-100% Stance						
Peak external hip flexion (+) moment	6.31 (2.64)	5.29 (1.64)	5.24 (1.52)	0.194	0.104	0.891
Peak external hip extension (−) moment	-4.36 (1.70)	-4.02 (0.94)	-3.55 (0.88)	0.405	**0.047***	**0.003****
Peak external hip adduction (+) moment	5.09 (1.41)	5.39 (0.92)	5.02 (1.01)	0.404	0.879	**0.020***
Peak external hip abduction (-) moment	-1.65 (1.03)	-1.32 (0.73)	-1.38 (0.52)	0.334	0.702	0.620
Peak external hip internal rotation (+) moment	0.54 (0.30)	0.44 (0.17)	0.42 (0.16)	0.468	0.114	0.774
Peak external hip rotation (−) moment	-0.80 (0.43)	-0.83 (0.25)	-1.06 (0.23)	0.802	0.056	**0.000****
0-50% (HS to midstance)						
Peak external hip extension (−) moment	-1.25 (0.92)	-0.71 (0.67)	-0.70 (0.78)	0.061	0.064	0.935
Peak external hip adduction (+) moment	5.27 (1.03)	5.23 (0.93)	4.78 (0.90)	0.906	0.140	**0.007****
Peak external hip abduction (-) moment	-1.28 (1.29)	-1.00 (0.99)	-1.25 (0.67)	0.479	0.939	0.063
Peak external hip internal rotation (+) moment	0.34 (0.20)	0.27 (0.13)	0.35 (0.20)	0.158	0.855	**0.041***
Peak external hip rotation (−) moment	-0.86 (0.42)	-0.84 (0.24)	-1.05 (0.23)	0.857	0.189	**0.000****
50-100% (Midstance to toe-off)						
Peak external hip adduction (+) moment	5.20 (1.17)	5.14 (0.94)	4.61 (1.09)	0.889	0.124	**0.000****
Peak external hip abduction (-) moment	-0.86 (0.55)	-0.50 (0.61)	-0.33 (0.68)	0.062	**0.009****	**0.044***
Peak external hip internal rotation (+) moment	0.64 (0.24)	0.35 (0.23)	0.21 (0.19)	**0.001****	**0.000****	**0.002****
Peak external hip rotation (−) moment	-0.11 (0.12)	-0.16 (0.13)	-0.31 (0.17)	0.240	**0.000††**	**0.000****

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. std = standard deviation. % BW.h = percentage of body weight multiplied by height. HS = heel strike.

7.5.7 Concurrent Optimisation of Muscle and Secondary Kinematics

Table 7-30 shows gait speed between the three groups. Post-surgery medial thrust gait significantly reduced gait speed and increased stance time when compared to post-HTO unaltered level gait.

Table 7-30 Post-HTO Medial Thrust Gait COMAK Gait Speed

Demographics	**Controls**	**Post-HTO NL**	**Post-HTO MT**	**Controls vs post-HTO NL**	**Controls vs post-HTO MT**	**Pre NL vs Post MT**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
Gait speed (m/s)	1.26 (0.17)	1.21 (0.16)	1.16 (0.19)	0.454	0.090	**0.012***

Significant difference (p<0.01) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. std= standard deviation. m/s = metres per second.

7.5.8 Internal knee joint loading

At FP, total and lateral compartment contact force, mean and maximum pressure significantly increased because of adopting a medial thrust gait compared to an unaltered level post-HTO. In addition to this, medial compartment knee mean pressure also significantly increased (6.15 BW (1.22) vs 6.5 BW (1.06), $p = 0.032$).

At MS, lateral compartment contact force, mean and maximum pressure significantly increased when adopting a medial thrust gait compared to an unaltered level gait (lateral knee contact force: 0.33 BW (0.17) vs 0.44 BW (0.24), $p = 0.014$).

At SP, adopting a medial thrust gait resulted in a significant decrease in medial compartment knee contact force and mean pressure compared to unaltered level gait (contact force: 1.4 BW (0.41) vs 1.28 BW (0.35), $p = 0.046$).

7.5.9 Internal knee joint loading ratios

At FP, adopting a medial thrust gait significantly decreased the medial to total contact force ratio (0.63 (0.09) vs 0.55 (0.07), $p = 0.000$), and significantly increased the lateral to total contact force ratio (0.4 (0.1) vs 0.48 (0.08), $p = 0.000$) when compared to a post-HTO unaltered level gait.

At MS, adopting a medial thrust gait significantly decreased the medial to total contact force ratio (0.73 (0.14) vs 0.68 (0.15), $p = 0.004$), and significantly increased the lateral to total contact force ratio (0.29 (0.14) vs 0.35 (0.16), $p = 0.005$) when compared to a post-HTO unaltered level gait.

There were no significant differences between a medial thrust gait and unaltered level gait post-HTO.

7.5.10 Point of application

At FP, total knee point of application shifted laterally and posterior when adopting a medial thrust gait compared to an unaltered level gait post-HTO. FP medial and lateral compartment knee point of application shifted posteriorly when adopting a medial thrust gait compared to an unaltered level gait post-HTO.

At MS, total knee, medial compartment, and lateral knee compartment significantly moved laterally when adopting a medial thrust gait compared to post-HTO unaltered level gait.

7.5.11 Contact area

At FP, there was a significant increase in lateral compartment contact area when adopting a medial thrust gait when compared to a post-surgery unaltered level gait (154.15 mm^2 (24.57) vs 167.93 mm^2 (21.23), $p = 0.001$).

At MS and SP, there were no significant differences between the post-HTO unaltered level gait and a medial thrust gait.

Table 7-31 Post-HTO Medial Thrust Gait Internal Loading Variables

	Controls	Post-HTO NL	Post-HTO MT	Controls vs Post-HTO NL	Controls vs Post-HTO MT	Post-HTO NL vs Post-HTO MT
	Mean (std)	Mean (std)	Mean (std)	P value	P value	P value
First peak						
Total knee						
Contact force [BW]	2.74 (0.57)	2.32 (0.58)	2.79 (0.59)	**0.032***	0.752	**0.001****
Mean pressure [MPa]	5.63 (1.25)	5.97 (1.16)	6.96 (1.08)	0.334	**0.001††**	**0.000****
Max pressure	12.92 (3.32)	13.93 (2.36)	16.64 (3.37)	0.155	**0.000††**	**0.003****
Medial knee						
Contact force [BW]	1.69 (0.31)	1.47 (0.42)	1.55 (0.38)	0.057	0.196	0.147
Mean pressure [MPa]	5.76 (1.12)	6.15 (1.22)	6.50 (1.06)	0.310	**0.045***	**0.032***
Max pressure	12.21 (2.52)	13.23 (2.68)	13.53 (2.32)	0.235	0.110	0.417
Lateral knee						
Contact force [BW]	1.13 (0.37)	0.92 (0.29)	1.33 (0.33)	0.074	0.087	**0.000****
Mean pressure [MPa]	5.37 (1.62)	5.62 (1.52)	7.50 (1.61)	0.607	**0.000††**	**0.000****
Max pressure	11.57 (3.76)	12.14 (3.25)	16.18 (3.78)	0.468	**0.000††**	**0.000****
Midstance						
Total knee						
Contact force [BW]	1.25 (0.24)	1.16 (0.11)	1.25 (0.22)	0.284	0.906	0.268
Mean pressure [MPa]	3.40 (0.36)	4.02 (0.60)	4.34 (1.03)	**0.002****	**0.000††**	0.078
Max pressure	7.62 (1.12)	9.22 (1.68)	10.04 (2.85)	**0.004****	**0.000††**	0.119
Medial knee						
Contact force [BW]	0.92 (0.22)	0.85 (0.19)	0.84 (0.18)	0.332	0.239	0.619
Mean pressure [MPa]	3.81 (0.58)	4.39 (0.68)	4.54 (0.77)	**0.006****	**0.001****	0.304
Max pressure	7.43 (1.22)	8.82 (1.42)	9.20 (1.63)	**0.002****	**0.000****	0.283
Lateral knee						
Contact force [BW]	0.36 (0.14)	0.33 (0.17)	0.44 (0.24)	0.571	0.376	**0.014***
Mean pressure [MPa]	2.59 (0.54)	2.96 (1.32)	3.71 (1.85)	0.362	**0.033†**	**0.007****
Max pressure	5.56 (1.07)	6.13 (2.68)	7.66 (3.90)	0.405	**0.040†**	**0.012***
Second peak						
Total knee						
Contact force [BW]	2.90 (0.70)	2.25 (0.57)	2.09 (0.49)	**0.004****	**0.000****	0.070

Mean pressure [MPa]	5.32 (0.64)	5.76 (1.31)	5.64 (1.59)	0.484	0.947	0.308
Max pressure	12.70 (1.76)	13.09 (3.36)	13.00 (4.58)	0.722	0.321	0.426
Medial knee						
Contact force [BW]	1.83 (0.35)	1.40 (0.41)	1.28 (0.35)	**0.001****	**0.000****	**0.046***
Mean pressure [MPa]	5.91 (0.80)	6.17 (1.36)	5.86 (1.42)	0.513	0.916	**0.020***
Max pressure	12.61 (1.80)	12.41 (3.05)	11.97 (3.35)	0.791	0.155	0.117
Lateral knee						
Contact force [BW]	1.20 (0.44)	0.92 (0.29)	0.88 (0.31)	**0.040***	**0.022***	0.380
Mean pressure [MPa]	4.53 (0.73)	5.19 (1.58)	5.25 (2.12)	0.067	0.189	0.855
Max pressure	9.43 (1.48)	11.16 (3.79)	11.20 (4.88)	**0.018†**	0.171	0.583

Significant difference (p < 0.01) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. std = standard deviation. BW = body weight; MPa = megapascals. Max = maximum.

Table 7-32 Post-HTO Medial Thrust Gait Contact Force Ratios

	Controls	**Post-HTO NL**	**Post-HTO MT**	**Controls vs Post-HTO NL**	**Controls vs Post-HTO MT**	**Post-HTO NL vs Post-HTO MT**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
MED / TOTAL	0.63 (0.07)	0.63 (0.09)	0.55 (0.07)	0.904	**0.003****	**0.000****
LAT / TOTAL	0.41 (0.07)	0.40 (0.10)	0.48 (0.08)	0.847	**0.003****	**0.000****
Midstance						
MED / TOTAL	0.73 (0.09)	0.73 (0.14)	0.68 (0.15)	0.947	0.321	**0.004****
LAT / TOTAL	0.29 (0.10)	0.29 (0.14)	0.35 (0.16)	0.885	0.348	**0.005****
Second peak						
MED / TOTAL	0.64 (0.07)	0.62 (0.08)	0.61 (0.09)	0.455	0.296	0.410
LAT / TOTAL	0.40 (0.07)	0.41 (0.08)	0.42 (0.09)	0.773	0.480	0.288

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. std = standard deviation. MED = medial compartment tibiofemoral joint; LAT = lateral compartment tibiofemoral joint; TOTAL = total tibiofemoral joint.

Table 7-33 Post-HTO Medial Thrust Gait Point of Application

	Controls	Post-HTO NL	Post-HTO TO	Controls vs Post-HTO NL	Controls vs Post-HTO MT	Post-HTO NL vs Post-HTO MT
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total knee						
Anterior (+) / posterior (-)	-2.46 (2.08)	-3.32 (1.49)	-5.06 (1.30)	0.189	**0.000††**	**0.000****
Lateral (+) / medial (-)	-2.27 (3.27)	-2.69 (4.21)	0.43 (3.30)	0.721	**0.016***	**0.000****
Medial knee						
Anterior (+) / posterior (-)	-1.01 (2.25)	-1.85 (1.44)	-3.00 (1.40)	0.272	**0.001††**	**0.003****
Lateral (+) / medial (-)	-17.19 (1.29)	-18.30 (1.46)	-18.36 (1.86)	**0.016***	**0.021***	0.680
Lateral knee						
Anterior (+) / posterior (-)	-4.67 (1.83)	-5.51 (1.69)	-7.27 (1.63)	0.160	**0.000****	**0.000****
Lateral (+) / medial (-)	20.59 (2.34)	21.74 (1.96)	22.15 (1.68)	0.122	**0.032***	0.154
Midstance						
Total knee						
Anterior (+) / posterior (-)	4.66 (1.86)	5.12 (2.38)	4.99 (3.04)	0.496	0.171	0.903
Lateral (+) / medial (-)	-5.84 (4.72)	-6.23 (7.14)	-3.29 (7.80)	0.856	0.274	**0.002****
Medial knee						
Anterior (+) / posterior (-)	6.63 (2.77)	6.84 (3.37)	7.35 (3.55)	0.838	0.147	0.633
Lateral (+) / medial (-)	-16.33 (1.53)	-17.44 (1.89)	-16.62 (2.10)	**0.025†**	0.864	**0.001****
Lateral knee						
Anterior (+) / posterior (-)	-0.42 (1.55)	0.30 (1.52)	-0.62 (3.04)	0.161	0.820	0.542
Lateral (+) / medial (-)	20.34 (2.43)	21.59 (3.80)	22.87 (3.29)	0.090	**0.008****	**0.030†**
Second peak						
Total knee						
Anterior (+) / posterior (-)	5.70 (1.64)	5.17 (3.74)	4.31 (3.74)	0.702	0.218	0.129
Lateral (+) / medial (-)	-1.39 (2.90)	-0.67 (3.40)	-0.24 (4.08)	0.607	0.348	0.442
Medial knee						
Anterior (+) / posterior (-)	9.56 (1.83)	8.69 (4.67)	7.89 (4.72)	0.989	0.518	0.119
Lateral (+) / medial (-)	-14.59 (1.31)	-16.08 (1.19)	-16.01 (1.47)	**0.001****	**0.003****	0.670
Lateral knee						
Anterior (+) / posterior (-)	-0.89 (2.43)	-0.64 (2.63)	-1.65 (2.78)	0.756	0.370	0.183
Lateral (+) / medial (-)	19.74 (2.84)	22.78 (2.88)	22.54 (2.88)	**0.001††**	**0.003††**	0.665

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. std = standard deviation. mm = millimetres. X = anterior; Z = lateral. COP = centre of pressure.

Table 7-34 Post-HTO Medial Thrust Gait Knee Contact Area

	Controls	**Post-HTO NL**	**Post-HTO MT**	**Controls vs Post-HTO NL**	**Controls vs Post-HTO MT**	**Post-HTO NL vs Post-HTO MT**
	Mean (std)	**Mean (std)**	**Mean (std)**	**P value**	**P value**	**P value**
First peak						
Total	352.35 (39.00)	376.11 (54.49)	389.27 (48.83)	0.272	0.007	0.063
Medial	206.8 (22.63)	221.95 (35.37)	221.34 (30.28)	0.180	0.147	0.884
Lateral	145.55 (20.95)	154.15 (24.57)	167.93 (21.23)	0.133	**0.001††**	**0.001****
Midstance						
Total	262.26 (34.10)	280.25 (34.83)	281.71 (43.47)	0.163	0.120	0.882
Medial	167.14 (24.46)	181.93 (25.84)	174.2 (27.12)	0.126	0.400	0.194
Lateral	95.12 (19.62)	98.31 (28.45)	107.51 (30.51)	0.308	0.296	0.127
Second peak						
Total	399.23 (90.44)	386.85 (92.03)	369.72 (73.10)	0.680	0.296	0.124
Medial	219.07 (41.52)	217.55 (50.95)	208.42 (39.11)	0.918	0.429	0.170
Lateral	180.16 (52.60)	169.29 (43.80)	161.31 (39.54)	0.510	0.244	0.141

Significant difference ($p<0.01$) indicated by ** where parametric or †† where non-parametric tests used. MT = medial thrust gait; NL = unaltered level gait. std = standard deviation. Measurement = mm^2.

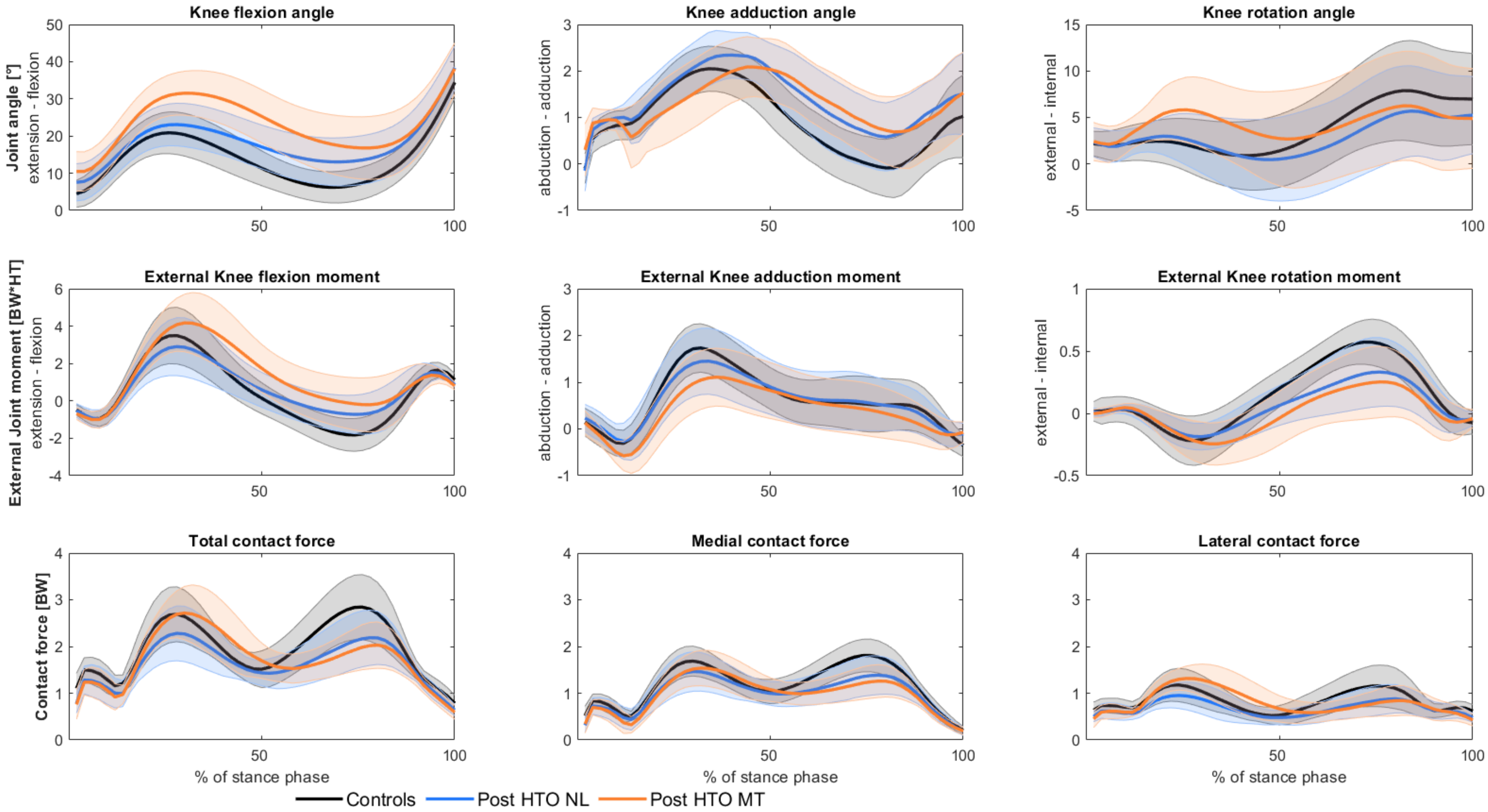


Figure 59 Post-HTO medial thrust gait knee joint kinematics, external moments, contact forces

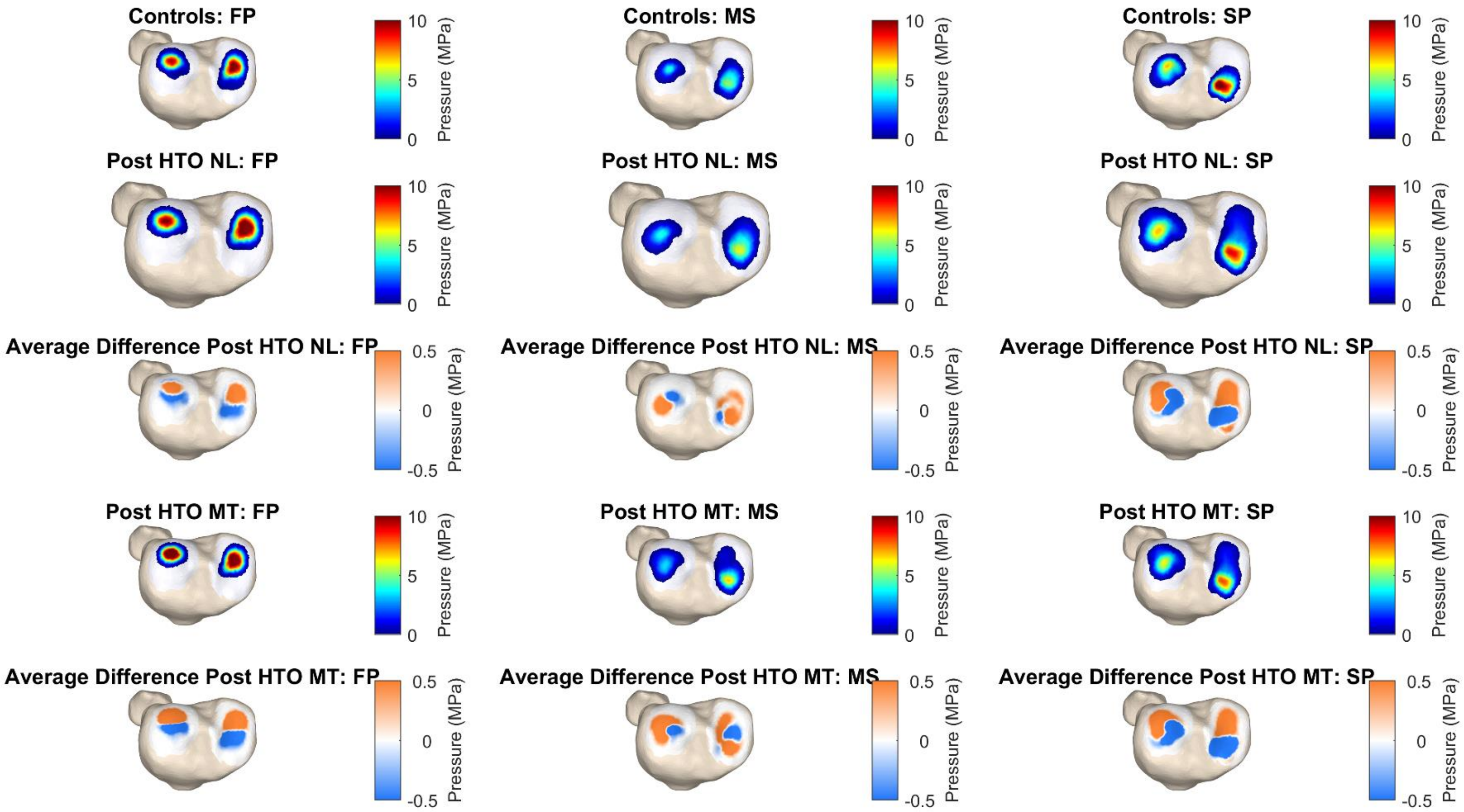


Figure 60 Post-HTO medial thrust contact pressure distribution on the tibia

7.6 Conclusion

This chapter aimed to compare the biomechanical differences of knee joint loading between a control group, post-HTO unaltered level gait, and post-HTO altered gait styles in the form of toe out, wide stance and medial thrust gait styles. If a gait style altered knee joint loading, then analysis is reported to outline the changes that consequently occurred at the hip and ankle joints. Accordingly, this chapter addressed lower limb biomechanical differences between a control group and post-HTO unaltered level gait, and biomechanical changes resulting from post-HTO altered gait styles.

7.6.1 Toe out gait

7.6.1.1 Post- High Tibial Osteotomy toe out gait recommendations

All patients at 12-months post-surgery were able to successfully adopt a toe out altered gait style without significantly altering any spatiotemporal parameters.

Adopting a toe out gait did not significantly alter EKAM1 but did significantly decrease EKAM2 when compared to an unaltered level gait post-surgery. Considering the decrease in EKAM2, adopting a toe out gait significantly decreased the KAAI over the whole of stance and during the second half of stance. These significant differences indicate a decreased second half of stance dynamic joint loading.

Interestingly, the observed non-changes in EKAM1 were met with an increased total knee compartment contact force, mean pressure and maximum pressure at FP when adopting a toe out gait compared to an unaltered level 12 months post-HTO. This increase of total compartment loading was met with significant increases in FP lateral compartment contact force unaltered level gait compared to toe out gait and significant increase in mean pressure and maximum pressure. This may indicate that although net external moments may not change during the first half of stance when adopting a toe out gait, the internal joint loading may be increased on the lateral compartment of the knee. If this is the case, it is important to establish whether there are any detrimental consequences associated with this lateral compartment loading. At SP, there were reductions in medial compartment contact force when adopting a toe out gait style (1.51 BW (0.37)) vs 1.41 BW (0.40)).

As toe out gait may have the potential of reducing medial knee compartment joint loading at SP, it is important to understand what the potential consequences of this gait style are on the hip and ankle joints. During the second half of stance, peak plantarflexion moment significantly reduced when adopting a toe out gait style (0.06 %BW.h (0.1) vs 0.02 %BW.h (0.11), p = 0.000). In the frontal plane, peak external inversion moment is significantly increased when adopting a toe out gait when compared to a post-HTO unaltered level gait (1.0 %BW.h (0.52) vs 1.35 %BW.h (0.54), p = 0.000) and a significant decrease in peak external eversion moment (0.14 %BW.h (0.13) vs 0.1 %BW.h (0.12), p = 0.027). In the transverse plane, adopting a toe out gait resulted in a significantly increased peak external rotation moment compared to a post-HTO unaltered level gait.

During the second half of stance, peak external hip adduction moment significantly reduced when adopting a toe out gait style compared to post-HTO unaltered level gait (5.16 %BW.h (0.84) vs 4.85 %BW.h (1.1), p = 0.004), whilst significantly increasing peak external hip abduction moment (0.46 %BW.h (0.28) vs 0.64 %BW.h (0.58), p = 0.002).

The findings would suggest that although toe out gait decreases medial tibiofemoral joint loading during the second half of stance. However, there may be adverse consequences during the first half of stance where it was reported a significant increase in total knee contact force. These changes would not have been apparent if the only metrics of concern were external moments at the knee. It is therefore imperative to better understand the internal joint alterations when adopting this gait style as an intervention post-HTO before recommending it as an effective intervention.

7.6.2 Wide stance gait

7.6.2.1 Post- High Tibial Osteotomy wide stance gait recommendations

All patient's 12-months post-surgery were able to successfully adopt a wide stance gait style. Adopting a wide stance gait resulted in a stride width of ~0.26m compared a post-HTO unaltered level gait stride width of ~0.17m. There were no significant changes in any spatiotemporal parameters when adopting a wide stance gait compared to an unaltered level gait post-HTO, apart from significant increase in operative limb step length of which the magnitude change is negligible.

Adopting a wide stance gait significantly reduced EKAM1 (2.1 %BW.h (0.88) vs 2.05 %BW.h (0.89), p = 0.002) and EKAM2 (1.55 %BW.h (0.83) vs 1.34 %BW.h (0.87), p = 0.000) when compared to an unaltered level gait post-surgery. Adopting a wide stance gait significantly reduced KAAI over the whole of stance (0.81 %BW.h (0.39) vs 0.71 %BW.h (0.36), p =

0.002), as well as the first and second half of stance when adopting a post-HTO wide stance gait compared to post-HTO unaltered level gait.

At FP, total and lateral compartment contact force, mean pressure and maximum pressure significantly increased because of adopting a wide stance gait compared to an unaltered level gait. No significant changes occurred for FP medial compartment loading. At MS, the only significant differences between a wide stance gait and an unaltered level gait were for medial compartment contact force, for which a wide stance gait significantly reduced medial compartment contact force. At SP, medial compartment knee contact forces significantly decreased when adopting a wide stance gait compared to unaltered level gait post-HTO.

For the first time, this thesis has assessed internal joint loading alterations when adopting to a wide stance gait in individuals with mKOA and varus aligned lower limbs. The findings would suggest that although wide stance gait decreases medial tibiofemoral joint loading during the second half of stance, there may be adverse consequences during the first half of stance. Adopting a wide stance increased total knee contact force during the first half of stance. These changes would not have been apparent if the only metrics of concern were external moments at the knee. It is imperative to better understand the internal joint alterations when adopting this gait style as an intervention post-HTO before recommending it as an effective intervention.

7.6.3 Medial thrust gait

7.6.3.1 Post- High Tibial Osteotomy medial thrust gait recommendations

Adopting a medial thrust gait style resulted in a significant reduction in gait speed. Any reductions seen in EKAM1 and EKAM2 should be viewed in light of the reduction in gait speed.

Adopting a medial thrust gait significantly reduced EKAM1 and EKAM2 when compared to an unaltered level gait post-surgery. Adopting a medial thrust gait significantly reduced KAAI over the first half of stance when compared to -post-surgery unaltered level gait.

At FP, total and lateral compartment contact force, mean and maximum pressure significantly increased because of adopting a medial thrust gait compared to an unaltered level post-HTO. In addition to this, medial compartment knee mean pressure also significantly increased (6.15 BW (1.22) vs 6.5 BW (1.06), $p = 0.032$). At SP, adopting a medial thrust gait resulted in a significant decrease in medial compartment knee contact force and mean pressure compared to unaltered level gait.

At FP, adopting a medial thrust gait significantly decreased the medial to total contact force ratio (0.63 (0.09) vs 0.55 (0.07), $p = 0.000$), and significantly increased the lateral to total contact force ratio (0.4 (0.1) vs 0.48 (0.08), $p = 0.000$) when compared to a post-HTO unaltered level gait. At MS, adopting a medial thrust gait significantly decreased the medial to total contact force ratio (0.73 (0.14) vs 0.68 (0.15), $p = 0.004$), and significantly increased the lateral to total contact force ratio (0.29 (0.14) vs 0.35 (0.16), $p = 0.005$) when compared to a post-HTO unaltered level gait.

Like adopting a medial thrust gait pre-surgery, it is clear from the novel findings of this work that whilst only having one visit to adopt a medial thrust gait, the gait style post-HTO may not be advantageous for this cohort. The gait speed was significantly reduced, and total knee contact force and pressures increased during the first half of stance. During the second half of stance, the medial compartment knee contact force was reduced. The findings from this work indicate that much more learning of this style needs to occur prior to implementing into a clinical setting.

7.7 Comparison of key knee metrics from pre- and post-HTO

Table 7-35 gives a short overview of some of the key knee metrics that have been used in this thesis to establish if altered gait styles pre- and post-HTO reduce medial knee joint loading. Alongside these metrics, gait speed is also included as an indicator as to whether the altered gait styles also alter gait speed.

Table 7-35 Key Knee Metrics Comparison

	Gait speed (m/s)	EKAM1 (%BW.h)	EKAM2 (%BW.h)	Peak EKFM (%BW.h)	MCF1	MCF2
Pre-HTO NL gait	1.06 (0.23)	3.10 (1.12)	2.48 (1.10)	2.87 (1.56)	1.59 (0.34)	1.63 (0.48)
Pre-HTO TO gait	1.05 (0.23)	3.18 (1.17)	2.18 (0.94)*****	3.11 (1.51)*****	1.70 (0.32)*****	1.44 (0.37)*****
Pre-HTO WS gait	1.07 (0.24)	2.94 (1.07)	2.19 (1.04)*****	3.20 (1.80)*****	1.63 (0.36)	1.52 (0.53)*****
Pre-HTO MT gait	0.96 (0.25)*****	2.68 (0.86)*****	2.30 (1.07)*****	4.05 (1.88)*****	1.67 (0.46)*****	1.45 (0.37)
Post-HTO NL gait	1.10 (0.16)	2.10 (0.88)	1.55 (0.83)	2.66 (1.23)	1.46 (0.36)	1.51 (0.37)
Post-HTO TO gait	1.10 (0.18)	2.14 (0.89)	1.38 (1.00)*****	2.88 (1.27)	1.48 (0.39)	1.41 (0.40)*****
Post-HTO WS gait	1.12 (0.19)	1.95 (0.87)*****	1.34 (0.78)*****	3.12 (1.27)*****	1.48 (0.39)	1.41 (0.40)*****
Post-HTO MT gait	1.11 (0.18)*****	1.57 (0.60)*****	1.29 (0.73)*****	4.72 (1.68)*****	1.55 (0.38)	1.28 (0.35)*****
Control cohort	1.21 (0.16)	2.27 (0.65)	1.50 (0.67)	3.62 (1.65)	1.69 (0.31)	1.83 (0.35)

NL = normal level ; TO = toe out ; WS = wide stance ; MT = medial thrust. m/s = metre per second. EKAM1/2 = external knee adduction moment peak 1 and 2 respectively. Peak EKFM = peak external knee flexion moment. MCF1/2 = medial tibiofemoral joint contact force peaks 1 and 2 respectively. * = pre/post-HTO NL is significantly different to the respective pre/post-HTO altered gait style.

CHAPTER 8: DISCUSSION

8.1 Novelty

The research described in this thesis has, for the first time, established the effects of gait retraining on individuals with mKOA and varus deformity, pre- and post-HTO. Previous research has assessed toe out gait, wide stance gait, medial thrust gait on healthy individuals (Schache *et al.*, 2008; Gerbrands, Pisters and Vanwanseele, 2014; Ogaya *et al.*, 2015; Favre *et al.*, 2016; Legrand *et al.*, 2021), those with mKOA but no varus deformity (Fregly, Reinbolt and Chmielewski, 2008; Jenkyn et al., 2008; Reinbolt et al., 2008; Hunt and Takacs, 2014; Charlton et al., 2018; Cheung et al., 2018; Richards et al., 2018) or at best simulated varus deformity (van Rossom *et al.*, 2019). Individuals with varus deformity are an in vivo example of altered knee joint loading. The purpose of intervening with gait retraining pre-surgery is to ascertain whether the knee can be offloaded and preserved prior to undergoing realignment surgery. Whilst the purpose to intervening post-surgery is to ascertain whether there is any merit of gait retraining prolonging the benefits of HTO. Therefore, the current thesis contributes significant knowledge in understanding the role of gait retraining in individuals with mKOA with varus deformity, as well as mKOA with corrected lower limb alignment.

This thesis focused on understanding the influence of altered gait styles at two time points. First pre-HTO and secondly at 12 months post-HTO. In addition to understanding altered gait styles, this thesis aimed to understand the biomechanical consequences of an HTO. The relative importance of findings are as follows. First and foremost are the findings of medial knee joint loading. If an intervention decreased medial knee joint loading parameters, then the findings surrounding the ankle and hip joints become important. Understanding the adjacent joint alterations to the knee are not well documented within the literature. Therefore, this thesis adds novel and clinically meaningful results that need to be considered.

This thesis is the first study to evaluate predictive internal knee joint loading pre- and post-HTO in a cohort of individuals with mKOA and varus deformity. Also, following on from the groups preliminary work and using a larger cohort, a comprehensive analysis has been undertaken to evaluate the effect of altered gait on the biomechanics at the ankle and hip joints pre and post HTO. The importance of understanding the effect of KOA and potential interventions on hip and ankle joints is highlighted in the Ro et al. (2019) study. The study by Ro et al. (2019) showed that changes in gait mechanics in the knee joint have a strong effect on ankle and hip biomechanics. Interestingly, ankle varus moment was 50% higher and associated with an increase of EKAM. Such changes are important as they can be risk factors for the subsequent development of secondary arthritis and result in increased pain (Miyazaki *et al.*, 2002; Mündermann, Dyrby and Andriacchi, 2005). The current thesis also includes results from the first application of PCA and the Cardiff Classifier to the HTO cohort.

By undertaking a novel analysis of the impact of gait retraining, pre- and post-HTO, on knee, hip, and ankle joints; and in predicting internal knee joint loading, this thesis contributes new, novel, and significant knowledge and understanding of the possible effects of gait retraining on a unique cohort of individuals. This PhD focused on assessing the influence of altered gait styles at both pre- and 12 months post-HTO for the following distinct purpose. Introducing an altered gait style pre-HTO aimed to introduce a non-surgical way to dynamically offload the medial compartment of the knee. Whilst introducing altered gait strategies at 12-month post-HTO was to quantify the merit of further offloading the knee joint after having surgery to establish whether there is an extra benefit than surgery alone.

A systematic review (Objective 1) revealed a paucity of research focused on assessing the effectiveness of gait retraining and the consequential alterations at the hip and/or ankle joints (n = 11) in a cohort of individuals who have mKOA (n = 5) and varus deformity (n = 0). Between the time of the systematic review being published and the submission of this thesis (2019 – 2022) only one additional paper (Legrand *et al.*, 2021) met the inclusion criteria of the systematic review. The aim of the Legrand et al.'s (2021) study was to provide insight into the impact of foot progression angle and lateral trunk lean on the sagittal and frontal external moments at the ankle and hip of healthy participants. Legrand's study indicated that whilst not all the gait modifications performed reduced EKAM, they significantly increased the sagittal moment at the ankle and the frontal moment at the hip. Legrand et al. supports the notion of the current thesis that consideration of the biomechanical consequences of gait modifications on the ankle and hip should be understood before a clinical application of gait retraining is feasible.

The biomechanical changes found at the knee, along with associated hip and ankle alterations, following HTO (Objective 2), indicate strong supportive biomechanical evidence for the surgery. Comparing the pre- and post-HTO patient biomechanics with that from a healthy cohort reinforced the groups preliminary work, that post-HTO patients recovered biomechanical function towards that of a healthy cohort in all lower-limb joints. This work has incorporated novel methods to predict internal joint loading on this cohort of individuals for the first time and provides supportive evidence for the operation.

The studies undertaken for Objectives 3 and 4 addressed the systematic review findings (Objective 1), comparing the effects of altered gait pre-HTO and 12 months post-HTO respectively. These studies are the first to investigate the influence of gait alterations on EKAM as well as simulating internal tibiofemoral joint loading for individuals with mKOA and varus deformity pre-HTO, as well as at 12 months post-HTO. The thesis outlines that at no point did adopting an altered gait style normalise medial knee joint loading to that of the control cohort.

The goal of intervention treatment is to decrease joint pain, increase functionality and delay further progression which ultimately leads to improving a patients' quality of life. Therefore, the goal of the interventions outlined in this thesis are not to normalise to the control cohort but to ensure the medial compartment of the knee is offloaded without causing a detriment to the joint functions of the kinetic chain of the lower limbs.

Although the merits of gait retraining pre- and post-HTO are inconclusive; they do however provide preliminary data that supports future development of patient specific gait retraining aimed at clinical translation.

The next section of this chapter will be outlining the impact of the current PhD thesis in relation to the current body of evidence.

8.2 Impact

8.2.1 Impact of High Tibial Osteotomy work

Objective 2 was necessitated to address the lack of research within current literature (Howes *et al.*, 2021). Howes et al. (2021) established that only 3 studies have assessed hip and/or ankle external moments and kinematics pre- and post-HTO (Weidenhielm, Svensson and Broström, 1992, 1995; Kyung *et al.*, 2021). The three studies include two studies that were undertaken over 25 years ago. The remaining study is the most recent which focused only on ankle kinematic alterations; and did not report on ankle moment changes pre- to post-HTO (Kyung et al., 2021).

This study supports the notion that HTO realigns the frontal plane of the knee to near neutral mTFA as well as significantly increasing gait speed which is in line with previous research (Whatling et al., 2019).

These findings are in line with the clinical purpose of HTO which is to unload the medial compartment of the tibiofemoral joint (Black et al., 2018), preventing the degeneration of the joint whilst also increasing gait speed. Previous research has reported a reduction in medial compartment joint loading based on joint moments (Whatling et al., 2019), when systematically simulating a varus knee (van Rossom et al., 2019), and predicting tibiofemoral joint contact forces for a KOA group (Richards et al., 2018). However, no study had previously simulated altered tibiofemoral joint loading in terms of joint contact forces and pressures for this unique cohort before and after HTO.

Within Objective 2, the key ankle biomechanical findings were that the frontal and transverse plane moments are ‘corrected’ towards those of a healthy cohort 12-months post-HTO. However, sagittal plane biomechanics did not reach statistical significance and appeared to remain sub-optimal in comparison to the healthy group. This research contributes significantly to the existing sparse body of evidence and will be used to inform future research.

The findings in Objective 2 relating to ankle biomechanics contrast with the only published research found relating to ankle moments pre- and 12 months post-HTO (Weidenhielm, Svensson and Broström, 1992). Weidenhielm et al. (1992) observed that there were no significant differences in ankle moments pre- and 12 months post-HTO. The contrasting results may be linked to differences in the study design, sample size and normalisation of ankle joint. Weidenhielm et al. (1992) included only 9 individuals with a mean age of ~64 years old, and the moment metrics were not normalised to body weight and height, being presented in Newton-meters. The only other study assessing ankle biomechanics pre- and post-HTO only considered ankle kinematics and did not report on ankle moment alterations (Kyung et al., 2021), thus limiting comparison with the current study.

The findings from the present thesis contrast with the hip moment changes pre- and post-HTO found in the only two published studies assessing hip frontal plane moments pre- and post-HTO (Weidenhielm, Svensson and Broström, 1992, 1995). The current thesis has found that undergoing HTO surgery produces an insignificant change in first peak of the external hip adduction moment with a significant increase in the second peak of the external hip adduction moment. The findings from Weidenhielm et al. (1992 and 1995) found that hip adduction moment was normalised to the control cohort 12 months post-HTO even with an increased gait speed. It is critically important to understand the differences in these findings because these studies can inform the effectiveness of surgery. There are two possible reasons for the disparities in findings. First, Weidenhielm et al. reported moments in Newton-metres (as opposed to normalising them to body weight and height in the current studies). Second, the current thesis split the frontal plane hip moments into the first and second half of stance to extract peak values; in the Weidenhielm studies, peak moment was recorded as the maximum external moment of the hip, knee, and ankle joint in the frontal plane. This approach neglects the biphasic waveform you typically see for hip adduction moment and so the peak value extracted may have been from either the first or second half of stance.

This is the first time the COMAK framework has been used to predict tibiofemoral contact forces and pressures in a cohort of individuals pre- and 12 months post-HTO. The most significant findings from this research are that HTO reduces medial knee contact force with no negative consequences on lateral compartment contact force. In addition to this, surgery significantly lateralised the COP for the total knee at first peak, second peak and at midstance. At midstance, the mean and maximum pressures across the knee as a whole and across the medial compartment were significantly reduced by HTO surgery. These novel findings are in line with the clinical objective of HTO which is to unload the medial compartment of the tibiofemoral joint (Black et al., 2018), to prevent the degeneration of the joint whilst also increasing gait speed. No previous research has undertaken the detailed approach applied in this thesis, and the results therefore go beyond those reported previously: (1) reporting external knee joint moments as a surrogate for medial compartment joint loading (Whatling et al., 2019), (2) systematically simulating a varus knee (van Rossom et al., 2019), and (3) predicting tibiofemoral joint contact forces for a KOA group (Richards et al., 2018). For the first time this thesis has performed predictive simulations to understand joint contact forces and pressures for patients pre- and 12 months post-HTO. The novel findings support HTO as a surgical intervention to offload the medial compartment of the tibiofemoral joint and should be explored further with patient-specific modelling.

This thesis also describes novel findings from the application of the Cardiff Classifier to the HTO cohort for the first time to objectively quantify the positive impact of HTO surgery. The findings in the current thesis contrast with the work by Biggs et al. (2019). The study by Biggs et al. (2019) demonstrated that with patients pre- and post-TKR, there was a significant relationship between the change in OKS scores and the change in objective biomechanical function. The reasons for these disparities may lay in the variables used to undertake PCA, may be due to the misclassifications of the 8 participants or may lay in the differences in the patient cohort. Future research should assess the same input variables as in the Biggs et al. paper to better understand the classification categorisation. The better the Cardiff Classifier is in distinguishing between the control group and the pre-HTO cohort, the more reliable and clinically useful the results. Out of the top 15 Principal Components obtained during the analysis, nine discriminating features were related to the hip and ankle.

Fundamentally supporting the importance in assessing the adjacent joints to the pathological knee joint. The clinical importance of these findings should form the basis for future work and interestingly, the Classifier findings are aligned to the discrete metric work undertaken with this thesis. The biomechanical function of HTO subjects within this study did not return to that found for the non-pathological cohort. The gait deviations that are still present between the control group and the HTO cohort post-surgery could be related to 2 reasons. First, HTO does not fully correct gait to that of a non-pathological individual or, second, because of the age difference between the two groups within this thesis. Future work should look to age-match the control and osteoarthritic cohorts. HTO surgery reduced the Cardiff Classifier belief in KOA in 20 out of 22 patients, indicating biomechanical improvement due to realignment surgery.

Collectively, these studies have produced several clinically important and novel findings and now pave the way for further subject-specific research. HTO is an effective surgery reducing both peak loading and knee angular impulse at the knee. However, there remains substantial biomechanical differences in all three planes at the hip, knee and ankle joints between the control group and the patient cohort. It is important to understand the reasons for the differences so that they can be addressed by targeted rehabilitation in the future.

8.2.2 Impact of gait retraining work

Objective 3 (assessed pre-HTO) and Objective 4 (assessed post-HTO) quantified the biomechanical differences in knee joint loading and the associated alterations at the hip and ankle joints when adopting three different altered gait styles: (1) toe out gait; (2) wide stance; (3) medial thrust. This is the first-time COMAK has been used to predict internal knee joint loading on individuals pre- and post-HTO performing altered gait styles and to reveal the consequences this has on the hip and ankle joint biomechanics. The implications of the findings should be considered before any considerations can be taken to the clinical translation of gait retraining.

The author of this thesis is aware of only one study which has assessed the effectiveness of an altered gait style on a cohort of individuals who were varus aligned, have medial compartment KOA, and underwent HTO surgery (Whelton *et al.*, 2017). The Whelton et al. paper is from the same research group as the author of this thesis and was published prior to the work of the current PhD which included a smaller cohort than is included in this thesis. Whelton et al. did not assess the alterations that occur at the hip and ankle joints when adopting an altered gait style (in this case a toe out gait). Neither did Whelton et al. assess predictive internal tibiofemoral joint loading in the form of musculoskeletal modelling.

Whelton et al. aside, the findings of this PhD will be compared to the gait retraining literature aimed at offloading medial compartment knee joint loading in either a KOA cohort or a healthy participant cohort. The reader of this thesis should bear in mind that a direct comparison is therefore limited due to cohort demographic differences and the inherent mechanical differences that arise when having varus malalignment.

8.2.2.1 Toe out gait

The current thesis findings are novel, clinically relevant to a specific patient cohort and cast new light on gait retraining for individuals with mKOA and varus deformity. They confirm that a toe out gait reduces loading in the second half of the stance, measured as EKAM2, whilst not reducing loading in the first half of the stance, measured as EKAM1, in agreement with recent research (Whelton *et al.*, 2017; Wang *et al.*, 2020).

The systematic review by Wang et al. concluded that for patients with KOA, toe out gait reduced EKAM2 peak and KAAI. The systematic review found that toe out gait did not affect EKAM1 but significantly reduced EKAM2 in both healthy individuals (large effect size) and patients with KOA (Wang *et al.*, 2020). A decrease in EKAM in the latter half of stance can be explained by the reduction of EKAM lever arm, as the COP shifts laterally during late stance. Hunt and Takacs (2014) stated that a ten-week toe out gait retraining programme alleviated knee pain by lowering EKAM2 in patients with KOA.

In this thesis the COMAK framework has been used for the first time to predict internal knee joint contact forces and pressure in individuals with mKOA and varus deformity whilst adopting toe out gait. When adopting a toe out gait, the EKAM1 metric calculated using the Visual 3D inverse dynamic modelling pipeline and the first peak medial knee compartment contact force predicted by the COMAK framework produced contradictory results. When comparing toe out gait to unaltered gait, the COMAK framework predicted an increase at first peak medial compartment knee contact force whilst EKAM1 was not different. The findings from this thesis would therefore indicate that a toe out gait may increase medial compartment joint loading during the first half of stance. These findings also support recent results from Legrand et al. (2021) with toe out gait increasing medial compartment joint loading in the form of EKAM1. Post-HTO, toe out gait resulted in a similar altered pattern compared to adopting a pre-HTO toe out gait pattern. Although these findings can shed new light on the understanding of toe out gait as a gait retraining approach, they differ in their approach to the research study design and clinicians should therefore be cautious when recommending this gait style until systematic, repeated and thus conclusive results are published.

Patients with KOA were reported to naturally walk with FPA from ~2° toe in gait to ~28° toe out gait (Jenkyn *et al.*, 2008); outlining high individual variability in this metric. It remains unknown how much patients with KOA should toe out to achieve dose-response effects. A biofeedback gait retraining which offers real-time EKAM and/or KAAI data may provide subject-specific information for gait modifications (Cheung *et al.*, 2018) which might be an appropriate way for tailoring FPA modification training. Future studies should address this in a clinical practice. It will be shown in the 'Future Work' section of this chapter that this is an avenue of research the MSKBRF at Cardiff University will be undertaking as a direct response to this PhD.

The current thesis adopted a self-selected magnitude when asking the participants to adopt their gait to a toe out gait style. This thesis has highlighted that a varus aligned cohort with mKOA pre-HTO change their FPA from 16° to 28° when adopting a toe out gait style (indicating a mean increase of 12° FPA). 12 months post-surgery, patients went from a ~17° to ~25° FPA (indicating a mean increase of 8° FPA). This would indicate, although not statistically compared within this thesis, that a patient pre-surgery was able to adopt a larger FPA than when asked to do so 12-months post-surgery. Future work could assess whether these changes in self-selected changes in FPA when adopting a toe out gait has a dose-response effect. This would indicate that for this study, participants tended to walk with a greater toe out angle compared to other studies but is consistent with Whelton et al.'s FPA of ~14° for the control group, ~19° for pre-HTO unaltered gait and ~31° for the pre-HTO toe out gait (Whelton *et al.*, 2017). This would indicate that individuals awaiting a HTO should not be compared simply to research that involves KOA patients and should in fact be a phenotype of patients themselves and research should be patient-cohort specific.

Contrast to the above findings that are widely accepted (Whelton *et al.*, 2017; Wang *et al.*, 2020), the work by Legrand et al. concluded that FPA's of 10°, 15°, and 20° significantly increased EKAM in both early and late stance whilst consequently increasing hip frontal moment in early, mid, and late stance and decreased the ankle sagittal moment in early stance and hip sagittal moment in late stance (Legrand et al., 2021). It is also of interest that within this Legrand study, 2 healthy participants were not able to successfully adopt either a toe out gait or a trunk lateral lean.

For both pre- and post-HTO visits, during the first half of stance, peak external dorsiflexion moment was significantly reduced when adopting a toe out gait style when compared to an unaltered gait style. Additionally, for both pre- and post-HTO, in the transverse plane, adopting a toe out gait resulted in a significantly reduced peak external internal rotation moment and significantly increased peak external rotation moment compared to an unaltered level gait. However, unlike no changes for the pre-HTO analysis, post-HTO toe out gait style resulted in an increased peak external ankle eversion moment.

For both pre- and post-HTO visits, during the second half of stance, peak external plantarflexion moment significantly reduced when adopting a toe out gait style. Additionally, in the frontal plane, for both pre- and post-HTO visits, peak external inversion moment was significantly increased when adopting a toe out gait when compared to an unaltered level gait. Additionally, adopting a toe out gait post-HTO resulted in a significant reduction in peak external eversion moment compared to an unaltered gait post-HTO. In the transverse plane, for both pre- and post-HTO visits, adopting a toe out gait resulted in a significantly increased peak external rotation moment compared to an unaltered level gait.

To the knowledge of Charlton et al. (2019), prior to their study no data existed examining biomechanical differences at the ankle joint during toe in gait and toe out gait walking in people with KOA. Given that most of the rotation during toe in and toe out walking originates distal to the knee joint, it is important to better understand how these modifications affect areas of the lower limb apart from the knee if such gait modification strategies are to be implemented clinically.

It is difficult to directly compare the findings of the ankle moments from the present thesis to that of Charlton et al. (2019) as Charlton et al. did not directly compare the different toe out angles to the control group. The control group was presented as a reference and not included in any statistical analyses.

Charlton et al. found that toe out gait increases rearfoot eversion angles but did not exhibit different ankle eversion loading relative to toe in walking, whereas toe in gait increased ankle/rearfoot inversion angles and moments relative to toe out walking. During toe out walking, the ankle moment would tend to cause eversion. The results in Charlton et al. constituted small differences that in some cases were within the range of expected error.

The Charlton et al. study demonstrated that toe in gait and toe out gait are performed with small differences in ankle/rearfoot joint kinematics and kinetics. The main findings are that toe out gait increased peak rearfoot eversion angles throughout stance. The implications of these differences are currently not known, particularly as the differences were small and the relationship between increased rearfoot eversion and discomfort, pain, or lower extremity injury is not yet clear. Charlton et al.'s results support to further assess FPA as a viable conservative treatment strategy for mKOA due to the relatively small differences in ankle/rearfoot biomechanics.

The study by Legrand et al. (2021) assessed a toe out gait strategy on the impact on lower limb joints for a cohort of healthy participants. Among the 23 participants, the data of 21 were included in this study. Two participants were not able to perform enough successful gait cycles. A FPA of 10°, 15° and 20° significantly decreased the ankle sagittal moment in early stance. A FPA of 10° significantly increased the ankle sagittal moment mid-stance. No significant alterations were shown in the frontal external moments when adopting a toe out gait style.

In the current thesis, for both pre-HTO and post-HTO visits, when asked to adapt their gait to a self-selected toe out gait style, during the first half of stance, peak external hip adduction moment significantly reduced when adopting a toe out gait style. Additionally, both visits showed that a toe out altered gait significantly increased peak external hip abduction moment compared to their respective unaltered level gait. In the transverse plane, both pre-HTO and post-HTO visits, when asked to adopt a toe out gait resulted in a significantly reduced peak external hip internal rotation moment when compared to a pre-HTO unaltered level gait.

This thesis has also shown that during the second half of stance, for both pre-HTO and post-HTO visits, peak external hip adduction moment significantly reduced when adopting a toe out gait style compared to an unaltered gait. For the pre-HTO visit, there was a significant reduction in peak external hip internal rotation moment to a pre-HTO unaltered level gait. For the post-HTO visit, adopting a toe out gait resulted in a significantly increased peak external hip abduction moment.

The study by Legrand et al. (2021) established that for a healthy cohort, A FPA of 10°, 15° and 20° significantly increased the hip sagittal moment in late stance. A FPA of 10°, 15° and 20° significantly increased the EKAM in early and late stance. It also significantly increased the hip frontal moment in early, mid, and late stance. The work by Legrand et al. would need to be taken with caution for any recommendations due to the increase in EKAM1 and EKAM2 as well as the sample population used was a healthy cohort and the limitations to compare to a cohort of individuals pre- and post-HTO.

To the authors knowledge, this thesis is the first time a study has reported on hip moments on individuals performing a toe out gait on individuals with mKOA and so the findings cannot be compared to any previous literature. Therefore, the work presented in this thesis affords the opportunity to understand the effect of a toe out gait and the consequences at the hip joint loading for the first time. This work should be considered when designing any gait retraining programme that would adopt a toe out gait. Notwithstanding the new and novel findings from this thesis, it is the first time an analysis has been undertaken on individuals pre- and post-HTO to quantify hip and ankle moments. The findings should however be considered as of secondary importance for the further work recommended to understand the effects of toe out gait retraining on individuals pre- and post-HTO in reducing medial compartment knee joint loading.

As per previous literature, this work has indicated that adopting a toe out gait may reduce EKAM2. However, when assessing the internal joint load adaptions due to adopting a toe out gait, internal joint loading simulation suggests that at SP, total knee and medial compartment knee contact forces were significantly decreased. Additionally, medial compartment mean pressure and maximal pressure were significantly decreased when adopting the toe out gait style. Adopting toe out gait did not alter lateral compartment joint loading and remained elevated compared to the control group. Additionally, total knee and lateral knee compartment medial-lateral point of direction changed significantly between the pre-HTO unaltered level gait and pre-HTO toe out gait style with the point of application occurring more laterally when adopting a toe out gait. At SP, total and medial contact area significantly decreased when adopting a toe out gait compared to an unaltered level gait. It is currently unknown what the clinical significance of a reduced second half of stance joint loading means in terms of preserving the medial compartment knee joint.

8.2.2.1.1 **Wide stance gait**

This thesis presents novel results from a study that, for the first time, explored the effects of wide stance gait retraining in individuals pre- and post-HTO surgery. The study revealed that, like adopting a toe out gait style, a wide stance gait style did not significantly alter EKAM1, but it did significantly decrease EKAM2 when compared to an unaltered gait pre-surgery. Post-surgery, adopting a wide stance gait decreased both EKAM1 and EKAM2. Adopting a wide stance gait following surgical correction complimented the effects of surgery by further reducing medial joint loading parameters and so the present data suggests that wide stance gait might be a suitable mechanism to prolong the medial knee joint unloading benefits of HTO surgery. The reason for the difference in EKAM1 pre- and post-HTO can be attributed to the lateralising of the COP and/or medialising the knee joint during the first half of stance gait and thus reducing the frontal plane knee moment arm.

Limited research has been undertaken assessing the feasibility of an increased step width as an altered gait intervention to reduce medial compartment knee joint loading. Two single-subject studies evaluated the effect of increased step width, achieved by increasing the frontal plane distance between feet during consecutive steps (Fregly, Reinbolt and Chmielewski, 2008; Reinbolt et al., 2008).

In addition to Fregly et al. (2008) and Reinbolt et al. (2008), Favre et al. (2016) highlighted the interactions of a general combination of gait modifications (increasing step width, toeing-in, and increasing trunk sway) associated with reductions in EKAM1. These interactions are particularly important because, as shown in this study, some gait variables are difficult to modify without inducing involuntary secondary changes in other gait variables.

Understanding that gait modifications are not isolated to a particular gait measure will aid in the design of gait retraining programmes and in the guidelines provided to the participants of these programs, as it demonstrates the importance of considering an overall scheme of altered walking mechanics. When asked to walk normally, participants walked with a step width of 0.036m and a gait speed of 1.3 m/s. Participants successfully followed the instructions to modify gait, as they significantly increased step width when they were instructed to do so. Retraining programmes should not instruct participants to modify a particular gait variable without considering secondary changes to other gait variables.

This is the first time COMAK has been used to predict internal tibiofemoral joint loading for individuals pre- and post-HTO whilst altering their gait to a wide stance. During the first half of stance, total knee contact force, mean pressure and maximum pressure increased when a wide stance gait was adopted compared to unaltered gait both pre- and post-HTO. This research is the first to establish that pre-HTO, medial compartment maximum pressure significantly increased when adopting a wide stance gait style compared to pre-HTO unaltered level gait. This was not the case post-HTO. During the second half of stance, medial compartment knee contact forces, mean pressure and maximum pressure were significantly decreased when adopting a wide stance gait compared to unaltered level gait pre-HTO. Additionally, lateral compartment contact force, mean pressure and maximal pressure were significantly increased when adopting the wide stance gait style. However, post-HTO only second half of stance medial compartment contact force significantly reduced when adopting a wide stance gait style.

The work in this thesis has therefore shown contradictory findings in terms of adopting a wide stance gait for pre- and post-HTO. For the pre-HTO analysis, adopting a wide stance gait compared to an unaltered level gait pre-HTO resulted in a significant increase in medial to total contact force ratio whilst for the post-HTO analysis there was a significant reduction in the medial to total contact force ratio at first peak. However, at second peak, both time points (pre-HTO and 12 months post-HTO) resulted in a significant reduction in medial to total contact force ratio and significantly increased lateral to total contact force ratios when compared to an unaltered level gait.

Richards et al. aimed to establish the relationship between EKAM and internal medial compartment knee joint forces when performing a wide stance gait (Richards, et al., 2018). The method of extracting peak contact forces differed to the method the current thesis used. In the Richards et al.'s paper the authors extracted peak values for the medial knee contact force and total knee contact force during the first (1-50%) and second (51-100%) half of the stance phase. Using these timings, Richards et al. (2018) identified peak values in the EKAM. Thus, for each gait trial, multiple values were extracted for EKAM and the external knee flexion moment, where each value corresponded to a peak in the medial or total contact force. Notwithstanding the methodological differences, individuals went from having 1.9 mKCF and 2.1 mKCF at first and second peaks respectively and were not significantly changed because of adopting a wide step gait style.

However, when assessing the FP medial to total force ratio, it was determined that a wide step gait style significantly reduced the ratio whilst showing no significant differences at SP. To their knowledge, Richards et al. paper was the first to report changes in KCF during gait modifications in KOA patients. Previous studies have reported effects of gait modifications on knee moments (Bowd et al., 2019; Simic et al., 2011) or on KCF post-TKA (Walter et al., 2010), or in healthy controls (Ogaya et al., 2015). Richards et al. found that walking with a modified wide stance gait did not reduce the KCF compared to normal walking. However, medial to total KCF ratio was significantly reduced.

The work in the current thesis has shown contradictory findings in terms of adopting a wide stance gait for pre- and post-HTO. For the pre-HTO analysis, adopting a wide stance gait compared to an unaltered level gait pre-HTO resulted in a significant increase in medial to total contact force ratio whilst for the post-HTO analysis there was a significant reduction in the medial to total contact force ratio at FP. However, at SP, both time points (pre-HTO and 12 months post-HTO) resulted in a significant reduction in medial to total contact force ratio and significantly increased lateral to total contact force ratios when compared to an unaltered level gait.

In the Richards et al. (2018) study, patients were asked to perform 3 different gait styles: one of them being with a wider stance gait. Visual feedback with target step width projected on screen, and position of patients' feet shown relative to the target. Within this study, forty patients with mKOA underwent 3D gait analysis on an instrumented treadmill, while receiving real-time feedback on the peak knee adduction moment. The target step width was set based on the reduction in EKAM in the direct EKAM feedback trial. Richards et al. (2018) concluded that significant changes in the peak ankle adduction moment were noted during the final two trials. The clinical relevance of this change is unclear, given the small changes in absolute values. Importantly ankle flexion moment was not reduced, which may be an important factor, to maintain the required power generation for initiation of the next step.

For both pre-HTO and post-HTO visits, when asked to adapt their gait to a self-selected wide stance gait style, during the first half of stance, peak external hip adduction moment significantly reduced and significantly increased peak external hip abduction moment when adopting a wide stance gait style. In addition to the above adaptations when adopting a wide stance gait, when adopting a wide stance gait post-HTO first half of stance, peak external hip extension moment significantly reduced and peak external hip rotation moment significantly increased.

During the second half of stance, for both pre-HTO and post-HTO visits, peak external hip adduction moment significantly reduced and peak external hip abduction moment significantly increased when adopting a wide stance gait style compared to an unaltered gait style. It is important for the clinical implications of these changes to be determined when defining a gait retraining programme.

In contrast, the study by Richards et al. (2018) concluded that adopting a wider stance during real-time feedback did not significantly alter hip joint moments. In comparison with baseline condition, Richards et al. (2018) found no significant increases in either the hip frontal or sagittal plane moments, suggesting no contra-indications for this type of training regarding risk of increased loading at the hip.

Like toe out gait, at SP, medial compartment knee contact forces, mean and maximum pressure were significantly decreased when adopting a wide stance gait compared to unaltered level gait. Additionally, lateral compartment contact force, mean pressure and maximal pressure were significantly increased when adopting the wide stance gait style. However, at FP, total and lateral compartment contact force, mean and maximum pressure increased because of adopting a wide stance gait compared to an unaltered level. Medial compartment maximum pressure significantly increased because of adopting a wide stance gait style compared to pre-HTO unaltered level gait. Clinically, it is paramount to establish whether this increased total and lateral compartment joint loading in the first half of stance is of concern and to what extent the second half of stance decrease in medial compartment joint loading is of clinical relevance.

Like performing a toe out gait retraining method, the work presented in this thesis affords the opportunity to understand the effect of a wide stance gait and the associated consequences at the hip joint and should be considered when designing any future gait retraining programme that adopts a wide stance gait. Before being implemented within a clinical setting, research should be undertaken to establish what the consequences are of adopting a wide stance gait during tibiofemoral joint loading in the first half of stance as an increased medial compartment maximum pressure may be advantageous.

8.2.2.1.2 **Medial thrust gait**

This thesis is the first time medial thrust gait retraining has been addressed in a cohort of individuals pre- and post-HTO. Unlike the relative ease of adapting to a toe out gait style or a wide stance gait style, a medial thrust gait was not successfully adopted by the full cohort both pre-surgery and at 12 months post-surgery. Pre-HTO, 20 out of 30 patients attained a significantly reduced peak knee adduction angle (the metric adopted to indicate successful medial trust), during the first half of stance but it remained significantly higher than that found for the control group. Post-HTO, 14 out of 28 patients achieved a significantly reduced peak knee adduction angle during first half of stance and was within the range found within the control cohort.

This new research has produced the following interesting and contradictory findings. Adopting a medial thrust gait pre-HTO significantly reduced EKAM1 and EKAM2 when compared to an unaltered level gait pre-surgery, which is arguably due to the reduction in gait speed that was associated with a medial thrust gait style. However, when predicting internal knee joint loading, at first peak, total, medial, and lateral compartment contact forces, mean pressures and maximum pressures significantly increased when adopting a medial thrust gait compared to an unaltered gait. These contradictory results emphasise the need to understand the relationship between joint moments and the predicted internal joint contact forces. Previous research suggests only moderate association at best exists between the measured EKAM and estimated tibiofemoral contact forces (Richards, Andersen, et al., 2018). The discrepancies are attributed to the resultant joint moments not considering muscle, ligament and soft tissue loading influences whilst the COMAK framework predicts muscle, ligament and soft tissue loading to predict tibiofemoral contact forces.

Gerbrands et al. (2014) assessed several altered gaits and their effects on EKAM parameters; one of which was undergoing a medial thrust gait. Thirty-seven healthy participants underwent 3D gait analysis. The discrete metric used to determine if a medial thrust gait was achieved was a reduction in maximal knee adduction angle extracted from the first 50% of the stance phase compared to normal gait. The same discrete metric used within this thesis. All participants were healthy and finished the protocol from start to finish with ease. Five minutes of practice seemed to be sufficient to instruct participants on all four strategies. Medial thrust was the most difficult strategy to instruct to participants as the gait style required postural coordination. In addition to these challenges, the participants in the present thesis may also have secondary undesired adaptations to their gait which may have resulted in the patients not being able to adopt the medial thrust gait and that future research should assess the feasibility of this gait style as a potential intervention.

Gerbrands et al. corrected for walking speed and concluded a significant change in EKAM when adopting a medial thrust gait style. In late stance, EKAM decreased significantly for medial thrust gait style. Medial thrust affected both overall peak and impulse and showed the greatest EKAM reduction. This suggests that aligning the knee centre in the frontal plane has the highest potential to reduce both peak and cumulative knee load during gait.

At the time Gerbrands et al. (2014) was written, their findings were in line with previous work regarding EKAM peak fall when adopting a medial thrust gait (2.4% (Schache et al., 2008) to 54% (Fregly et al., 2007)). In relation to the findings of the current thesis, medial thrust also resulted in a significant reduction in gait speed as well as significant reductions in both EKAM1 and EKAM2. However, due to the cohort having mKOA, it was deemed not appropriate to look to correct for gait speed when assessing EKAM changes due to the gait style as argued by Wilson (2012). It is argued by Wilson (2012) that when considering accounting for gait speed in statistical analysis for KOA patients, critical assumptions of the model are violated making the methods inappropriate and the results misleading. Studies that experimentally control speed can answer interesting questions, particularly when interpreted with the results of studies that do not control for speed. However, it is important that we are aware that conclusions made from speed-controlled studies often cannot be generalised to our understanding of the natural environment of the joint in which the disease developed.

In addition to this, the training the individuals received to adopt their gait to a new gait style was quite short in duration which may have led to an exaggeration of the changes needed to produce the style. If given a longer duration to learn and practice the medial thrust gait, patient participants may have been more successful in adopting the gait style which may have resulted in a ‘smoother walking style’ and less exaggerated style that was often seen.

When assessing the effectiveness of adopting a medial thrust gait 12 months post-HTO, at first peak, total and lateral compartment contact force, mean pressure and maximum pressure significantly increased because of adopting a medial thrust gait compared to an unaltered level post-HTO. In addition to this, medial compartment knee mean pressure also significantly increased. Whilst at second peak, adopting a medial thrust gait resulted in a significant decrease in medial compartment knee contact force and mean pressure compared to unaltered gait.

Fregly et al. (2009) investigated the effectiveness of a medial thrust gait pattern for reducing medial compartment contact force in the knee. This study reported findings for just one participant. This participant had a force-measuring knee replacement performed overground gait with simultaneous collection of internal knee contact force and external ground reaction force data (Fregly et al., 2009). The patient was tested 3.5 years after implantation for primary knee osteoarthritis. Therefore, it is difficult to make comparisons between Fregly et al. (2009) and the current thesis.

In the Fregly et al. (2009) study, most quantified changes relative to normal gait were statistically significant, and their magnitudes suggest that they may be clinically significant as well. Several possible explanations exist for the reduced medial contact force achieved by the medial thrust gait. Fregly et al. (2009) stated that 'the trend toward increased lateral contact force suggests that medial thrust gait shifted a portion of the medial contact force to the lateral compartment, like what one would expect from HTO surgery'.

Kinney et al. (2013) assessed medial thrust gait and the effect this had on internal knee joint loading with one participant. The subject was instructed not to increase knee flexion during stance. In this study, gait speed was not altered when walking with a medial thrust compared to the participants unaltered gait. Although the changes were not statistically significant, medial thrust reduced both medial and lateral contact forces by greater than 10%, which may be clinically significant. Medial thrust achieved average medial contact force reductions of 14%, at 25% of stance phase (Kinney et al., 2013). Further investigation of the medial thrust gait results revealed that during five of the ten stance phases analysed, medial contact force at 25% of stance phase was reduced between 22% and 25% relative to normal gait.

This result is likely because the subject was instructed to medialise his knees without increasing knee flexion significantly, as recommended by a previous study, making knee medialisation more difficult to achieve consistently during late stance. In the current thesis, adopting a medial thrust gait compared to a pre-HTO unaltered gait resulted in a significant increase in peak flexion external moment. Accordingly, the results of a reduced gait speed may be hidden within the increased flexion moment values and future research should aim to address this.

Long-term training and feedback have been shown to improve long-term performance of medial thrust gait and, therefore, may improve the effectiveness of medial thrust gait at consistently reducing in vivo medial contact forces. While the findings from Kinney et al. (2013) provides important insight into changes in medial and lateral contact force through gait modification, the results are based upon data from a single subject implanted with a cruciate-retaining TKR. Therefore, the extent to which these results can be generalised to other individuals with healthy or implanted knees is unknown.

This research addresses, for the first time, a medial thrust gait style in individuals pre- and post-HTO. The gait retraining that the individuals received to adopt their gait was short in duration which may have led to an exaggeration of the changes needed to produce a medial thrust gait. If given a longer time to learn and practice the medial thrust gait, participants may have been more successful in adopting the gait style which may have resulted in a 'smoother walking style' and less exaggerated style that was often seen. Therefore, this research has contributed knowledge to the research community with respect to ease of training and learning how to adopt an altered gait style and it would be embedded into future research. Nevertheless, this is the first study to show that medial thrust gait may be effective in reducing the EKAM pre- and post-HTO. As demonstrated in Chapters 6 and 7, pre-HTO external moments and kinematics (Figure 45, Figure 46) and post-HTO external moments and kinematics (Figure 57 and Figure 58) all three planes at the hip and ankle joints were altered due to adopting a medial thrust gait. Adopting a medial thrust gait appears to increase the external hip sagittal moment throughout stance, reflected also by an increased hip flexion angle throughout stance. The changes are also met with a drastically increased knee flexion moments and angles throughout stance. Adopting a medial thrust gait is also shown to alter external ankle dorsiflexion moments and joint angles. Significant alterations are also observed in the transverse plane moments and kinematics at the hip, knee, and ankle joints. The extent to which these joint alterations are (a) due to the medial thrust gait as opposed to the alterations in gait speed, and (b) detrimental to gait are not examined further in this thesis.

For both pre- and post-HTO medial thrust gait, during the first half of stance, peak external dorsiflexion moment significantly increased when adopting a medial thrust gait style, as well as a significant decrease in peak external plantarflexion moment. Again, for both pre- and post-HTO medial thrust gait, in the frontal plane there was a significant increase in peak external inversion moment when adopting a medial thrust gait compared to a pre-HTO unaltered level gait, as well as significantly reducing peak external eversion moment. In the transverse plane, there were significant changes when adopting a medial thrust pre-HTO, but these changes were not seen when adopting a medial thrust 12 months post-HTO. Adopting a medial thrust gait resulted in a significant increase in peak external rotation moment compared to a pre-HTO unaltered level gait.

During the second half of stance, the only significant difference between pre-HTO unaltered level gait and adopting a medial thrust gait was in the frontal plane peak external eversion moment which significantly decreased when adopting a medial thrust gait. This change was also seen when adopting a medial thrust gait compared to an unaltered gait at 12 months post-HTO. However, for the 12-month post-HTO visit there were also significant differences when adopting a medial thrust gait in a reduced peak external dorsiflexion moment and a significant reduction in peak external internal rotation moment.

Richards et al. (2018) assessed peak frontal and sagittal plane external moments. In the frontal plane, the effect sizes should be interpreted with caution because of the very high standard deviation. Sagittal plane moment indicated a null effect for medial thrust gait. Gerbrands et at. (2017) found that a medial thrust gait significantly reduced gait speed. Gerbrands et al. concluded that a medial thrust gait significantly reduced late stance ankle inversion moment (−3%). For both pre-HTO and post-HTO visits, adopting a medial thrust gait significantly increased peak flexion moment when compared to pre-HTO unaltered level gait, as well as significantly reducing peak extension moment.

For both pre-HTO and post-HTO visits, during the first half of stance, peak external hip adduction moment significantly reduced when adopting a medial thrust gait style and significantly increased peak external hip rotation moment compared to pre-HTO unaltered level gait. In addition to these changes, at the post-HTO visits, adopting a medial thrust gait also resulted in a significantly increased peak external hip internal rotation moment compared to post-HTO unaltered level gait.

During the second half of stance, for both pre-HTO and post-HTO visits, peak external hip adduction moment and a significant reduction in peak external hip abduction moment when adopting a medial thrust gait style. Additionally, for both pre-HTO and post-HTO visits, peak external hip internal rotation moment significantly decreased when adopting a medial thrust gait style, whilst peak external hip external moment significantly increased when adopting a medial thrust gait style.

In the study by Richards et al. (2018) assessed the effect of real-time biofeedback on peak knee adduction moment in patients with medial knee osteoarthritis whilst performing a medial thrust gait style. Visual feedback with target knee position projected on screen and actual position of knees shown relative to the target. Patients were instructed to bring their knees closer together during the stance phase, while trying to avoid an excessive increase in knee flexion. As previously the target distance was defined based on the EKAM visual feedback trial. If the patient was successful in reducing the EKAM during the direct feedback trial using a change in the knee frontal plane position (i.e., medial knee thrust), then the target was set to the distance used during the direct EKAM feedback trial. If this was not the case, the target distance was decreased by up to 5cm depending on the percentage reduction during the direct EKAM feedback trial. Note: targets were set separately without combining the modifications. Use of medial thrust did not result in significant changes in the hip adduction moment. Similarly, peak external hip flexion moment was not significantly changed through walking with a medial thrust. In comparison with baseline condition, Richards et al. (2018) found no significant increases in either the hip frontal or sagittal plane moments, suggesting no contraindications for this type of training regarding risk of increased loading at the hip. Gerbrands et at. (2017) found that external hip adduction moments were not significantly increased after correcting for walking speed.

For this thesis, the influence of change in muscle activation, an important determinant for knee joint loading, is neglected. Internal knee joint loading is increased by co-contracting muscles without affecting the value of the net external moment. This is because concurrent activation of agonist and antagonist muscles will cancel out their individual contribution to the joint moment but add to their contribution to the knee reaction force. While co-contraction can enhance stabilisation of the knee joint, it increases knee loading, which is not reflected in the EKAM. Consequently, muscle co-contraction is an important outcome parameter that should be considered in future studies which assess interventions that target mKOA progression such as gait retraining.

Long-term training and feedback have been shown to improve long-term performance of medial thrust gait and, therefore, may improve the effectiveness of medial thrust gait at consistently reducing internal knee joint loading. While the findings from Kinney et al. (2013) provided important insight into changes in medial and lateral contact force through gait retraining, the results are based upon data from a single subject implanted with a cruciate-retaining TKR. Therefore, the extent to which these results can be generalised to other individuals with mKOA who undergo HTO surgery is unknown.

Adopting a medial thrust gait pre-HTO or post-HTO may not be of clinical benefit. Adopting a medial thrust gait results in an individual walking slower which may not be a desirable consequence of the gait style. This also begs the question as to how easily medial thrust gait can be taught as it may take time to adjust to the gait style to maintain the same gait speed as that obtained in unaltered level gait. Irrespective of gait speed, pre-HTO only 19 out of 30 patients and post-HTO only 14 out of 30 patients were able to successfully adapt to a medial thrust gait style as determined by a reduction in maximum knee adduction angle in the first half of stance. However, it must be noted that patients only had a one-off visit to learn this gait alteration and so the time of learning how to walk with a medial thrust gait may require multiple visits to be effectively implemented.

Interestingly, the reductions seen in EKAM1 and EKAM2 when compared to an unaltered level gait pre-surgery were not reflected in the internal joint loading analysis. At FP, total, medial compartment, and lateral compartment contact force, mean pressure and maximum pressure significantly increased because of adopting a medial thrust gait compared to an unaltered level.

At MS, total and lateral compartment contact force, mean pressure and maximum pressure significantly increased when adopting a medial thrust gait compared to an unaltered level gait. In the medial compartment, maximum pressure increased because of adopting a medial thrust gait style compared to pre-HTO unaltered level gait. At SP, the only significant change that occurred due to adopting a medial thrust gait was that the medial compartment knee maximum pressure significantly reduced compared to unaltered level gait; no other changes occurred.

The findings have established that, for this cohort of patients, medial thrust gait pre-surgery may result in an increased load rather than an intended decrease and recommending this gait style for patients awaiting HTO should be done so with caution.

8.2.3 Acceptability of altered gait patterns from previous research

In chapter 3 of this thesis, it was shown that very little and limited work focusing on gait retraining focuses on the acceptability of the desired altered gait style. In this thesis, three altered gait styles were introduced on a cohort of individuals pre- and 12 months post-HTO. Anecdotally, it is the author of this thesis view that individuals' pre-surgery found walking with their normal gait difficult, let alone adapting their gait to either wide stance, toe out or medial thrust. With that said, not many individuals complained or made comments on the difficulty of the wide stance or toe out gait styles. Contrast to this, the acceptability of learning and adapting to a medial thrust gait was limited. Several participants did not accept this gait style as being a style that would be easily adopted outside of the motion laboratory. It is paramount that future research in this area focuses on gaining the participants feedback on the acceptability, ease and comfortless of adopting their gait, both in a single session scenario and over a longer period.

The findings have established that, for this cohort of patients, medial thrust gait pre- and post-surgery may result in an increased load and recommending this gait style for patients post-HTO should be done so with caution.

8.3 Limitations

8.3.1 Variability

Within the methodology of this thesis, the aim was to acquire six clean gait cycles for each limb, with clear single-limb force plate strikes from heel strike to toe-off. When processing the data, in some cases, one or more of the gait cycles were not suitable for processing. While several attempts may have been necessary to achieve clean force plate data during a data collection session, pathological patients would occasionally start to experience pain during the session and were unable to complete all six walks. This was evident for the pre-HTO cohort, as well as for patients post-HTO who were awaiting a contralateral HTO. In these instances, the averages were calculated on at least three gait cycles and therefore the results may be less representative of the participant's average gait cycle.

While every effort was made to provide a comfortable environment for the subject recruited into the study, gait analysis can feel like a very unnatural experience. This can be exacerbated by the addition of passive markers, and EMG electrodes. Subjects were asked to walk along a 10m walkway at a self-selected pace. Dummy markings on the floor of the walkway and starting points managed by the research support team were introduced so that the patient was not aware of any targeting. While it might not have been possible to distinguish force plates, the flooring within the centre of the walkway was visually distinct.

8.3.2 Hardware changes

As the data presented in this thesis spanned a decade, there were hardware changes within the motion analysis laboratory during the subject data collections included within this study. The resultant effect of adding additional force platforms into the motion analysis laboratory was not quantified; however, they were the same make and model and, anecdotally, there was no obvious change in the GRF data. The upgrade in cameras from ProReflex infrared cameras (Qualisys, Sweden), to the Oqus 3 cameras (Qualisys, Sweden), to the Oqus 700+ cameras (Qualisys, Sweden) was likely to have an impact. Average residual marker trajectory errors can be calculated during initial calibration, and these errors decreased from 1.2mm in general to lower than 0.8mm with the updated cameras. It is not known what effect this might have had on calculations of dynamic joint biomechanics; however, it is acknowledged that these changes are much smaller than the errors induced through soft tissue artefact, which have been reported as high as 30mm on the thigh and 15mm on the tibia (Peters *et al.*, 2010).

8.3.3 Inter-operator errors

Subjects used within this study were part of ongoing data collection over several years. Within this time, several researchers have been involved in the data collection and processing. Clear SOPs have continually been in place, and new researchers have undergone training and assessment before being signed off as competent to collect and process data in a repeatable and consistent way. Inter-operator variability is, however, possible in addition to the unavoidable intra-operator variability of motion analysis techniques. The study of (della Croce, Cappozzo and Kerrigan, 1999) identified intra and inter-operator errors when identifying anatomical landmarks in the range 6-21 mm and 13-25 mm, respectively.

8.3.4 Musculoskeletal modelling

A complex 12-DoF knee model was used in this thesis (Lenhart et al., 2015; Smith et al., 2018). This model includes ligaments, as well as articular cartilage that allows the computation of contact pressures. Furthermore, the secondary tibiofemoral kinematics (tibiofemoral translations and non-sagittal rotations) and patellofemoral kinematics are load dependent as they evolve as a function of muscle and ligament forces, and cartilage contact.

Some limitations need to be considered when interpreting the results of this study. In terms of the methodology, the model that was used in the current study comprises a generic knee model, with a uniformly distributed cartilage thickness. Consequently, differences in cartilage thickness on the medial compartment in this cohort is neglected when calculating contact pressure distribution. As such, medial compartment KOA, whereby cartilage thickness is reduced, and accordingly will increase contact forces may be under-represented within this study. Future work should aim to incorporate patient-specific geometry of the patient's knee.

Additionally, the optimisation algorithm used in the current study did not account for subject-specific muscle contractions. This would require the use of an EMG-driven modelling approach. Future work should be undertaken to establish whether individuals with varus deformity and medial compartment KOA have co-contraction of the lower-limb muscles to establish what effect this may have on estimating contact forces.

The results derived from the COMAK pipeline must be interpreted in view of some methodological limitations, as inherent to the model used. First, a single generic knee model was scaled to represent the anthropometry of the participants instead of considering the subject-specific articular geometries, including those of the tibia plateau. Our model does not account for mKOA induced changes in the articular geometry, such as thickness and mechanical properties of the cartilage, or changes in the muscle and ligament properties. Consequently, the reported differences in tibiofemoral contact forces and contact pressures only result from altered kinematic and kinetic behaviour. Bone deformities, ligament laxity or changes in cartilage induced by joint degeneration were not considered and they might produce an effect on contact pressures. Second, although the secondary tibiofemoral kinematics and patellofemoral kinematics were calculated as a function of muscle forces, ligament forces, and cartilage contact as well as only knee flexion being tracked in the gait simulation, the method may still present some sensitivity to soft tissue artifacts. Third, although the validation of the model has shown a good agreement between the calculated and experimental kinematics and contact forces in healthy subjects and patients following TKR (Lenhart et al., 2015), this validation cannot easily be extended to a varus deformed mKOA population.

This PhD attempted to incorporate a varus lower limb alignment. The ankle is translated to a new location to effectively introduce varus, and then this work corrected foot alignment, so it lands flat on the floor. However, whether this implements the true varus angle for the patient cannot be answered in this thesis and should be addressed in future work. Alterations in tibiofemoral varus angle has a direct impact on calculated joint moments, muscle force production and estimated joint contact forces and pressures. Therefore, it is of paramount importance to have this consideration in mind when interpreting the results.

The presence of increased co-contraction, bone deformities or changes in cartilage mechanical properties, and the potential presence of ligament laxity induced by joint degeneration were not evaluated. Therefore, this model might present specific limitations when used in patients with varus deformity and mKOA, especially those known to present increased co-contraction resulting in an underestimation of the joint loading.

The validation of this model compared estimated knee kinematics (closed kinetic chain movement) with in vivo knee kinematics collected during supine posture tasks (open kinetic chain movement) using dynamic magnetic resonance imaging (Lenhart et al., 2015). In this model, the ligaments are represented as nonlinear spring elements, one-dimensional discrete elements, rather than deformable 3D representations that account for spatial variations in strain. Instead, some wrapping surfaces were included to improve wrapping around the bony structures but no ligament–ligament interactions were incorporated. The thickness of the cartilage surface was assumed constant, which is a simplification since cartilage thickness varies, in particular when considering KOA. This simplification might result in differences in terms of contact pressures and contact areas. Further, the knee model does not include menisci, which are known to distribute pressure in the tibiofemoral joint. It has been shown that inclusion of the menisci provides a small improvement in kinematics and a significant difference in the distribution of tibiofemoral loading during activates (Guess et al., 2010; Kia & Guess, 2011). This may indeed suggest that the menisci should be included in investigations where cartilage loading is important. Therefore, the absence of menisci might increase the peak contact pressures in the knee joint surface.

Secondly, inclusion of subject-specific characteristics into the knee models for the individuals with varus malignment and mKOA was limited. Generic models that were scaled to represent only the anthropometry of the subjects were used. Subject specific articular geometries, muscle-tendon and ligaments properties were not considered in our approach since there was no data available for the cohort used. Therefore, the models do not account for KOA induced changes in the articular geometry, such as thickness and mechanical properties of the cartilage, or changes in the muscle and ligament properties. Consequently, the reported differences in knee contact force and contact pressures only result from altered kinematic and kinetic behaviour. Bone deformities, ligament laxity or changes in cartilage induced by joint degeneration were not taken in account and they might produce an effect on contact pressures (Smith et al., 2016).

Thirdly, limitations resulted from the static optimisation techniques used to calculate muscle forces. In optimisation methods, the same cost function is assumed for both healthy and mKOA subjects. The COMAK algorithm minimises the weighted sum of squared muscle activations and the net cartilage contact elastic energy. These cost functions are based on previous research (Challis, 1997) that showed that the minimisation of effort yields muscle activation patterns like those observed experimentally. However, it is unknown how much they represent the true muscle coordination strategy, especially, when the analysed motion deviates from the normal walking pattern. Thus, the calculated muscle forces did not necessarily capture the effect of co-contraction patterns (Hubley-Kozey, Deluzio and Dunbar, 2008) which may reduce the estimated knee loading. Furthermore, although computationally inexpensive, as an inverse dynamics problem, static optimisation neglect muscle activation and contraction dynamics. However, static and dynamic optimisation solutions have been proven to provide similar muscle forces during gait (Anderson and Pandy, 2001).

In COMAK, inverse kinematic measurement techniques are first used to compute the coordinates, speeds, and accelerations of the primary model DOF. Thereafter, numerical optimisation is performed to simultaneously solve for the secondary kinematics, muscle, ligament, and articular contact forces that generate the primary joint accelerations while minimising a cost function that resolves inherent muscle redundancy. Therefore, questions are raised as to how accurate does the tibiofemoral adduction angle that is simulated reflects the varus angle of the individual. If there are disparities within these measurements this will have a direct impact on the predicted joint moments and joint contact force and pressures. Accordingly, these assumptions may mean that the results found in this thesis are not a true reflection of the individual.

One last consideration to state for the use of COMAK is that the majority of patients included in this thesis have had previous arthroscopies, knee ligament surgery, and/or other lower limb operations. These previous operations may mean that the predictive manner of the COMAK model is not simulating for this.

8.3.5 None aged and BMI matched control group

The control cohort involved in this study were not aged matched. Although the mean age of the control cohort is lower than the patient cohort, this study shows that following HTO, biomechanical measures of knee loading are improved to values like the younger control group, adding weight to the proposed merits of HTO surgery.

Additionally, the control cohort in this thesis were not BMI matched to the pathological group; the HTO cohort had significantly larger BMI compared to both pre- and post-HTO. Body mass influences joint moment magnitudes. This PhD attempted to factor this potential confounding variable by presenting joint moments as %BW*h and contact forces as a multiple of body weight.

8.3.6 One assessment visit

Lack of multiple visits may be viewed as a limitation. It was the purpose of this thesis to establish the potential mechanical impact of adopting a particular gait style in reducing knee joint loading and the consequences of this on the hip and ankle joints. The thesis has highlighted key novel findings which form the base of future work to address patient-specific gait adaptations with real-time feedback to better understand gait retraining. The next step would then be to address the questions around the long-term feasibility of altering gait on this specific cohort pre- and post-HTO.

The study by Shull et al. (2013) assessed ten participants with mKOA and self-reported knee pain participated in weekly gait retraining sessions over 6 weeks. This study found that a 6-week gait retraining programme can reduce the EKAM and improve symptoms for individuals with medial compartment knee osteoarthritis and knee pain. Other research assessing gait retraining programmes over multiple visits include Hunt et al. (2018) which performed a 4-month programme to increase walking activity with (toe out) or without (progressive walking) concomitant toe-out gait modification. The study found that although both groups experienced improvements in self-reported pain and function, only the toe out group experienced biomechanical improvements. In addition to this, Shull et al.'s (2014) systematic review emphasises that wearable feedback can improve walking stability and reduce joint loading and that work should implement in natural environments such as home or work. Going forward, longitudinal assessments of a gait retraining programmes could be undertaken remotely within a natural environment.

8.3.7 Acknowledging the effect of altering the critical alpha value to 0.01 from 0.05

Critical values for a test of hypothesis depend upon a test statistic, which is specific to the type of test, and the significance level, α, which defines the sensitivity of the test. A value of α = 0.05 implies that the null hypothesis is rejected 5 % of the time when it is in fact true. The choice of α is somewhat arbitrary, although in practice values of 0.05, and 0.01 are common in the human biomechanics' literature. Critical values are essentially cut-off values that define regions where the test statistic is unlikely to lie; for example, a region where the critical value is exceeded with probability α if the null hypothesis is true. The null hypothesis is rejected if the test statistic lies within this region which is often referred to as the rejection region(s). This PhD acknowledges that if a critical alpha level of 0.01 was implemented instead of 0.05, the following conclusions of the thesis may have been presented. Note that the below is not an exhaustive list. The reader is guided to the individual result chapters of this work for further checks.

When comparing pre-HTO and 12-month post HTO, gait speed would not have changed due to surgery (p = 0.026) along with there being no significant differences with stride length and stride width. The key take-home messages when comparing EKAM and KAAI metrics would have remained the same, i.e., HTO surgery significantly reduces EKAM and KAAI metric values. Some ankle and hip rotation and external moment metrics would not have made significant differences. In terms of the COMAK analysis, less significant differences would have been present, especially in FP tibiofemoral joint loading parameters. These are:

- FP medial compartment knee: Contact force [BW] (p = 0.017);
- FP medial compartment knee: Mean pressure [MPa] (p = 0.046);
- Point of Application of the Contact Forces, FP, medial compartment knee (p = 0.016);
- Contact area: FP, medial compartment (p = 0.030);
- Contact area: FP, lateral compartment (p = 0.030).

In terms of adopting an altered gait either pre-HTO or post-surgery, several discrete metrics that were shown in the thesis to be statistically different in comparisons would have changed in adopting a critical alpha value of 0.01. These values were focused on hip and ankle external moments as well as several important internal joint loading metrics derived from the COMAK analysis. The reader of this thesis is directed to chapter 6 (pre-HTO altered gait styles) and chapter 7 (post-HTO altered gait styles) to determine these specific metrics.

8.4 Future work

8.4.1 Include an aged-matched control group

Future research must incorporate an aged-matched healthy cohort. This PhD had planned to recruit an aged-matched healthy control cohort at the end of 2019 and throughout 2020. The recruitment led to approximately 10 aged-matched controls being recruit before COVID-19. It was envisioned that an aged-matched comparison would add more clinically meaningful results.

8.4.2 Principal Component Analysis and the Cardiff Classifier

The PCA and Cardiff Classifier work presented in this thesis was an exploratory study. Future work should input the variables that have been shown to be discriminatory between healthy and OA shown in Biggs (2016). In addition to this, joint moments would be changed from the whole of the gait cycle to that of just the stance phase. Future PCA and Cardiff

Classifier work should also assess whether gait retraining changes the BOA pre- and post-HTO.

8.4.3 Inclusion of muscle co-contraction using electromyography driven simulations

In this project, muscle forces were calculated using a static optimisation approach. This approach determines the muscle forces that produce the joint moments calculated using inverse dynamics by minimising a cost function. Although static optimisation has shown satisfactory results in calculating muscle forces during gait, studies have shown that individuals with KOA exhibit different muscle activation patterns compared to healthy subjects (Lewek, Rudolph and Snyder-Mackler, 2004; Hubley-Kozey, Deluzio and Dunbar, 2008; Heiden, Lloyd and Ackland, 2009). High muscle co-contraction has been observed in KOA patients that might result in higher joint loading (Hubley-Kozey, Deluzio and Dunbar, 2008). Using static optimisation, these co-contraction patterns are not accounted for.

Since muscle forces are the main contributors to joint contact forces, muscle coordination strategy is expected to highly influence joint contact loading. Therefore, the inclusion of individual muscle force activity in the muscle force and consequent tibiofemoral contact force calculations in subjects suffering from KOA during gait must be the next step.

EMG-constrained static optimisation can improve muscle force estimation by matching the muscle activity patterns collected from EMG-sensors. To include muscle co-contraction as derived from surface EMG recordings, the cost function can be extended with an additional term penalising the difference between the simulated and measured activity patterns or some constraints can be added. These constraints are derived from EMG recordings collected from the knee extrinsic muscles to constrain the solution space of available solutions throughout the gait cycle. Since EMG signals were collected for the different studies, it would be possible to use EMG-constrained static optimisation to evaluate contact loading during gait in a follow-up study.

8.4.4 Inclusion of subject-specific characteristics into the knee models

Joint loading can be affected by joint geometry, mechanical properties of the cartilage and bone, muscle strength/weakness and ligament properties, which are likely to be altered in individuals with KOA (Andriacchi *et al.*, 2004). Therefore, subject-specific geometric characteristics in the musculoskeletal models used to calculate knee loading are required.

Indeed, by changing the generic to a subject-specific geometry, moment arms and forces of the muscles surrounding the knee are affected due to changes in the muscle-tendon paths and in the knee joint centre position. Consequently, the muscle-tendon paths need to be adjusted based on medical imaging (e.g., MRI and/or ultrasonography).

In addition, it is well-known that patients with KOA complain of muscle weakness, which might influence calculated muscle forces and, ultimately, effect the resulting knee contact forces. Therefore, a measurement of muscle strength examined by dynamometry might also be considered to reflect patient specific force production. Finally, it is known that forces in the ligaments can vary significantly between individuals due to subject specific gait characteristics and knee joint geometries. This is also relevant for individuals with KOA, who commonly present increased passive knee laxity (Lewek, Rudolph and Snyder-Mackler, 2004). Therefore, further investigation is recommended to better understand the importance of having a subject-specific model in estimating knee contact forces on the knee joint for individuals with KOA.

8.4.5 Multiple visit gait retraining with real-time feedback and clinician input

As outlined in the systematic review undertaken in this thesis, there is a severe lack of understanding of gait retraining on a patient cohort and the effects this may have on adjacent joints (Bowd et al., 2019). This thesis has added valuable insight into the potential benefits and detriments of introducing three altered gait styles (1) toe out gait (2) wide stance gait and (3) medial thrust gait. The work in this thesis assessed data that was collected at two time points for all patients; one just prior to undergoing an HTO, and second at approximately 12 months post-HTO. It is unknown how much the conclusions that are drawn from these single time points would reflect those from multiple visits over a longer period.

8.5 Thesis conclusion

The aim of this thesis was to explore the biomechanical effectiveness of gait retraining pre-HTO in patients who had mKOA and varus deformity and then 12 months post-HTO where lower limb realignment surgery has been undertaken. Prior to assessing the merits of adopting gait retraining pre- and post-HTO, the biomechanical merits of HTO were assessed.

HTO surgery resulted in biomechanical changes in all three planes at the hip, knee, and ankle joints. Post-HTO, medial knee loading was reduced by ~10% and ~16% when assessed using COMAK at both peaks in stance. HTO surgery reduced the classification belief in mKOA for 20 out of 22 patients, indicating biomechanical improvement occurs due to realignment surgery.

The literature review and systematic review identified a lack of studies that assessed the effectiveness of gait retraining on individuals with mKOA. Only one study was identified which involved patients with mKOA and lower-limb varus deformity.

It is apparent from the results in this thesis that gait retraining, at least in the form of the three gait styles assessed in this thesis, all have different effects on knee joint loading and subsequent effects on hip and ankle biomechanics. This study has demonstrated that two separate analyses assessing knee joint loading, external knee moments vs predictive internal knee joint loading, can result in different recommendations being suggested.

Toe out gait pre- and post-HTO reduced EKAM2 (~12% pre and -11% post) and second half of stance internal medial loading peak (~12% pre and ~7% post). Pre-HTO, adopting a toe out gait also increased medial internal joint loading in early stance by ~6%. Wide stance gait reduced medial compartment loading in late stance when adopted pre or post HTO (10-13% reduction in EKAM2 and ~7% reduction in medial internal tibiofemoral joint loading). Medial thrust gait reduced EKAM pre- and post-HTO. The reductions in EKAM were met with significant alterations at the hip and ankle joint moments and kinematics. Contrary to EKAM, medial thrust resulted in a reduced gait speed and conflicting findings with predictive internal joint loading.

This study is the first to investigate the influence of gait alteration on medial compartment loading pre-to-post HTO surgery. It reveals a set of novel clinically important findings and provides preliminary data supporting future development of patient specific gait retraining aimed at clinical translation.

CHAPTER 9: REFERENCES

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CHAPTER 10: RESEARCH CONTRIBUTIONS, AWARDS, FUNDING AND VISITS

Peer-reviewed research articles:

Bowd, J., Biggs, P., Holt, C., Whatling, G. (2019). Does Gait Retraining Have the Potential to Reduce Medial Compartmental Loading in Individuals with Knee Osteoarthritis Whilst Not Adversely Affecting the Other Lower Limb Joints? A Systematic Review. Archives of Rehabilitation Research and Clinical Translation 1(3-4).

Selected international conference abstracts:

Bowd, J., Biggs, P., Holt, C., Whatling, G. (2019). Wide Stance Gait Style Compliments High Tibial Osteotomy in Reducing Knee Joint Loading. OARSI World Congress on Osteoarthritis 27, S123-S124., article number: 146. (10.1016/j.joca.2019.02.181).

Bowd, J., Biggs, P., Holt, C., Whatling, G. (2019). Does Gait Retraining Have the Potential to Reduce Medial Compartmental Loading in Individuals with Knee Osteoarthritis Whilst Not Adversely Affecting the Other Lower Limb Joints? Systematic Review. Podium presentation. International Combined Orthopaedic Research Societies (ICORS).

Bowd, J., Rossom, S., de Vecchis, M., Williams, D., Wilson, C., Elson, D., Jonkers, I., Holt, C., & Whatling, G. (2020). Knee Joint Contact Forces During Gait for Patients Undergoing High Tibial Osteotomy. OARSI World Congress on Osteoarthritis, 250–251.

Bowd, J., Biggs, P., Whatling, G., Wilson, C., Elson, D., de Vecchis, M., & Holt, C. (2020). Podium presentation. Waveform Analysis Using Principal Component Analysis to Better Understand Biomechanical Factors Affecting Varus Deformity of the Knee. Orthopaedic Research Society (ORS).

Bowd, J., Williams D., De Vecchis, M., Wilson, C., Elson, D., Whatling, G., Holt, C. (2021). Does Varus Knee Deformity Only Effect Frontal Plane Biomechanics? Podium (virtual) presentation. 26th Congress of the European Society of Biomechanics, Milan, Italy.

National conference proceedings:

Bowd, J., Biggs, P., Cathy, H., & Gemma, W. (2018). Combined Effect of Wide Stance Gait and High Tibial Osteotomy on Knee Adduction Moment in Patients with Varus Knee Deformity. Podium presentation. Engineering The Knee: Innovation at the Interfaces for Improved Surgery and Rehabilitation.

Bowd, J., Biggs, P., Elson, D., Metcalfe, A., Wilson, C., Holt, C., & Whatling, G. (2019). High tibial osteotomy (HTO) and wide stance (WS) gait reduces knee joint loading in individuals with varus knee deformity. 5th Joint Conference of the Bone Research Society (BRS) and British Orthopaedic Research Society (BORS).

Bowd, J., Rossom, V., Wilson, C., Elson, D., Jonkers, I., Whatling, G., & Holt, C. (2020). Peak Knee Contact Forces Pre-To-Post High Tibial Osteotomy and the Relationship to Peak External Knee Adduction Moments. British Orthopaedic Research Society (BORS).

Funding awarded:

2019: 1 x £1,000 CITER Conference Travel Bursary. Awardee. To present at International Combined Orthopaedic Research Societies (ICORS), Montreal, Canada 2019.

2019/2021: 2 x £2,000 CUROP student summer placements. Co-supervisor. (1) 2019 - Muscle Activity, Muscle Force and Their Application to Patients with Knee Joint Pathology (2) 2021 - Does Gait Retraining Alter Muscle Activation Patterns And Co-Contraction Around The Knee In Individuals Awaiting High Tibial Osteotomy.

2019: 1 x £2,000 CITER student summer placements. Co-supervisor. Understanding High Tibial Osteotomy and Non-Surgical (Gait Retraining) Interventions in Patients with Varus Knee Deformity.

2019: 1 x £3,000 Early Career Researcher placement funding from OATech+ Network 2019. 6-week placement at KU Leuven, Human Movement Biomechanics Research Group.

2021: £20,000 EPSRC IAA. Co-investigator. Gait Retraining for Patients Undergoing High Tibial Osteotomy Surgery.

Workshops:

C-Motion, Visual 3D Workshop, CMUG Meeting, Cardiff, UK, April 2018.

The Oswestry ORLAU Gait Course, Oswestry, UK, June 2018.

CAMS-Knee OpenSim Workshop, Zurich, Switzerland, February 2020.

Delsys Europe User Group Training, Manchester, UK, January 2020.

Invited BBRCVA talks:

2018 – December - BBRCVA Research Seminar - Combined Effect of Wide Stance Gait and High Tibial Osteotomy on Knee Adduction Moment in Patients with Varus Knee Deformity.

2019 – December - BBRCVA Research Seminar- Knee Joint Contact Forces During Gait for Patients Undergoing High Tibial Osteotomy.

2020 – December - BBRCVA Research Seminar - Can Altering Gait Just Prior To High Tibial Osteotomy Reduce Knee Joint Loading? What We Know So Far.

2020 – June - BBRCVA Research Seminar - Using Musculoskeletal Simulation to Inform Surgical Outcomes.

2021 – June - Does Varus Knee Deformity Only Effect Frontal Plane Biomechanics?

CHAPTER 11: APPENDIX A: SYSTEMATIC REVIEW DATABASE SEARCH KEYWORDS

Syntax was adjusted appropriately for use in multiple databases. Keywords were identical for all searches

The following keywords were grouped and searched in all fields with conjunction "OR" in each group to ensure that all relevant articles were obtained. Group one consisted of keywords "walk*" OR "gait". Keywords "knee" OR "adduction moment" built up the second group. Group three consisted "osteoarthriti*" OR "arthriti*" OR "osteo arthriti*", OR "OA". Group four included "hip" OR "ankle".

In the second stage, the searched results of each group were combined with conjunction "AND" in all fields. CINAHL subject headings were "walking" for the first group, "knee" and "adduction" for the second group, "osteoarthritis" and "knee" for the third group, and, "ankle" and "hip" for the fourth group. All searches were initially carried out in any language in their titles, abstracts and full-length articles and later assessed for English language only versions.

CHAPTER 12: APPENDIX B: PATIENT INFORMATION SHEET AND CONSENT FORM

CARDIFF UNIVERSITY
PRIFYSGOL CAERDYDD

GIG CYMRU NHS WALES
Bwrdd Iechyd Prifysgol Caerdydd a'r Fro
Cardiff and Vale University Health Board

Biomechanics and Bioengineering Research Center Versus Arthritis

PATIENT INFORMATION SHEET

Assessment of joint function in patients with joint problems using three-dimensional motion analysis techniques

<u>We would like to invite you to take part in a research study</u>.

- Before you decide if you would like to take part it is important for you to understand why the research is being done and what it will involve.
- Please take some time to read the following information sheet carefully and discuss it with friends or relatives if needed.
- It is your decision whether or not to take part.
- Ask a member of the study team if you have any questions about the research.
- If you decide to take part in this research but later change your mind you are free to withdraw at any time. This will not affect any of your NHS care.

<u>Important Information about this Research</u>

- Taking part in this research will not change your NHS treatment in any way.
- This research is part of a series of studies being conducted by the Biomechanics & Bioengineering Research Centre Versus Arthritis (BBRCVersusArthritis) at Cardiff University.
- Participating in this research could involve visits to Cardiff University School of Engineering, or School of Healthcare Sciences, additional to any NHS care.
- During study visits you may be asked to complete some questionnaires.
- We would also like to collect information about your diagnosis and treatment from you and from your medical records
- We do not expect there to be any direct benefit for people who take part in this research
- The information we collect in the research will help improve our understanding of how people with joint problems move.

BIOMECHANICS & BIOENGINEERING RESEARCH CENTRE VERSUS ARTHRITIS

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What is the purpose of this research?
This research is part of a series of studies being carried out by the Centre Researchers, Orthopaedic Surgeons and Physiotherapists.

We are interested in knowing more about how people with joint (e.g. knee) and back problems move when performing normal activities such as walking, standing, bending etc. We aim to investigate how treatment (operation or physiotherapy) changes the way you move and how your movement compares with people without joint or back problems.

We are interested in learning about changes that happen within the affected joints. In order to do this we may ask if you are willing to take part in some of the related Centre studies.

We hope that the information we collect in this research can be used to develop new tools to help orthopaedic surgeons and other health professionals with the diagnosis and management of joint and back problems.

Why am I being asked to take part?
You have been asked to take part because you fall into one, or more, of the following categories:

- Are currently on a waiting list for orthopaedic, physiotherapy or rheumatology treatment
- Have received treatment for a joint or back problem
- Have previously taken part in Centre research.
- Have a joint problem we are interested in looking at with this technique

If you are on a waiting list for surgery, your surgeon has agreed that you may be suitable to take part in this research.

What does taking part involve?
This research is being carried out in a number of different settings. If you decide to take part in the research you will be asked to attend one of the following locations;

- **The Musculoskeletal Biomechanics Research Facility** (Cardiff University School of Engineering), or;
- **The Research Centre for Clinical Kinaesiology** (Cardiff University School of Healthcare Sciences).
- **A relevant clinical setting such as an NHS clinic**

The number of times you will be asked to attend will depend on your specific joint problem. Patients with back problems may only be asked to attend a single session. If you are waiting for an operation you may be asked to attend a session before your operation and further sessions during your post-operative recovery. You may be asked to attend a maximum of six sessions over a period of five years.

Before any study activities are performed you will be introduced to the research facility. A researcher will talk you through the specific requirements of the study. You will have an opportunity to ask questions about the research and the study setting. Each session will last between 30 minutes and three hours. The length of the visit will depend on the joint and treatment under investigation. After attendance at the session you will be reimbursed for reasonable travel expenses

If you are happy to take part in the study we will ask you to sign a consent form

After you have signed a consent form we will ask you about your joint problem and take some measurements (e.g. height and weight, limb circumference)

You may also be asked to complete some questionnaires and be asked to answer some questions on your joint problem and how it affects daily life.

To prepare for the movements you will be asked to change into **appropriate clothing**. For lower limb problems this is usually loose fitting shorts and t-shirt. Patients with back problems may be asked to wear a sports bra or swimming costume. If you do not have appropriate clothing this can be provided by the researcher. Your modesty and dignity will be respected throughout the visit.

A selection of **reflective markers** will be placed at specific points on your; feet, legs, lower back, spine and arms. These markers are held in place with sticky tape. The markers help the motion capture system track your movements – see pictures on the next page.

For the final part of the visit you will be asked to perform a selection of movements that will be appropriate to your specific joint or back problem.

Introduction to facility, Opportunity to ask questions.

↓

Consent form signed

↓

Information collected

↓

Questionnaires completed

↓

Change into appropriate clothing

↓

Sensors and markers applied to skin

↓

Performance of study activities

↓

End of visit

Throughout the session you will be given the opportunity to rest and take regular breaks. During the session you will not be expected to perform any activities that cause you pain and discomfort.

Throughout the visit you will be asked to perform repeated movements. The number of times each movement is repeated will vary depending on the joint under investigation. These movement tasks may include:

Back	**Hip, Knee and Ankle**	**Shoulder and Elbow**	**Wrist**
Bending	*Walking*	*Lifting light objects*	*Grip*
Stretching	*Up and down stairs*	*Range of motion*	*Range of motion*
Sit to stand	*Sit to stand*	*Reaching for objects*	
	Standing on one leg		
	Up and walk from a chair		

Some of these movements may be performed on a special treadmill. The treadmill is set at floor level and can rotate in multiple directions to replicate uneven ground. In some circumstances, these treadmills will be set within a virtual reality environment. When using the treadmill you will be asked to wear a safety harness to prevent falls.

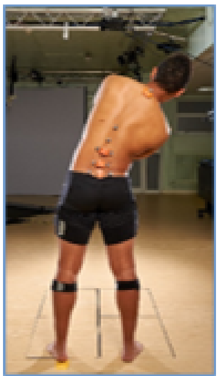
The treadmill is not used for patients awaiting joint replacement.

Depending on the joint being studied; muscle activity, function and joint strength may also be measured. Measuring muscle activity involves the placement of muscle sensors, called electrodes, on the surface of the skin. The location of the sensors will depend on the joint that is being studied. In some circumstances, it may be necessary to shave hair from the area where the senor is to be placed. In some cases, muscle function is measured with the use of a small electrical muscle stimulus during certain movements to activate the muscle and produce a change in movement. This may cause a strange feeling but will not be painful. To test joint strength, you may be asked push or pull against a resistance. This may involve your limb being strapped into a machine while performing movements.

During the session, your movement may also be recorded using standard audio-visual equipment (e.g. video camera). These recordings are used to verify the data collected by the motion capture system. If recordings are used in any presentations or publications, digital masking (removal of features) will be used to ensure that you cannot be identified from the video files.

You may also be asked to perform the following movements as fast as you can without pushing yourself to overexertion and within a short set time: standing and sitting from a chair, standing from a chair and walking, walking on level ground, ascending and descending stairs.

Before you decide if you would like to take part in the research a member of the study team will talk you through the exact requirements for your study visit(s).



Examples of Sensor and Marker placement for Low Back Pain (left picture) and lower limb (right picture)

What are the potential risks and benefits of taking part?

The reflective markers and sensors are placed with sticky tape or adhesive silicon rubber. The removal of these items may cause some mild discomfort, similar to removing a sticking plaster.

You may experience some temporary pain in the affected joint associated with the study activities. This risk will be limited by allowing breaks during the visit and limiting the activities performed to those which you are comfortable with

There is no intended clinical benefit for people taking part in the study. The information we collect from patients may help us to provide future patients who have joint disease or injury with improved treatment options.

Risk associated with the COVID-19 pandemic

Due to the COVID-19 pandemic, new safety measures have been put in place at our Cardiff University research facilities to avoid the spread of COVID-19. Research visits are usually arranged via a phone call. Due to the COVID-19 pandemic, we will call you twice before the research visit. The first call will be for checking if you are at high risk of developing complications from a COVID-19 infection and to screen whether you have experienced COVID-19 symptoms in the previous 14 days. If you do not have COVID-19 symptoms and are not at high risk of developing complications due to COVID-19 then a visit date will be arranged.

The second call will happen within 24 hours before the visit to check if you have developed COVID-19 symptoms or have met anyone who has had COVID-19 since the first phone call. If you or anyone you met have COVID-19 symptoms, the research visit will be cancelled and you will be advised to follow the Welsh Government guidelines on Test, Trace and Protect. If you do not have COVID-19 symptoms, the visit will go ahead as planned. The research team will also be screened for COVID-19 symptoms before your visit takes place and in case any of the researchers has possible COVID-19 symptoms, the visit will be cancelled. Research visits will only happen if it is safe to do so for you and the research team.

During the research visit, our team will be wearing full personal protective equipment (PPE) to keep you and ourselves safe. This equipment includes eye protection, face masks, disposable aprons, and gloves. Following the Government guidance on safe working, additional cleaning and disinfection procedures will be performed after every volunteer visit. The number of people within the facility will be kept to a minimum (researchers involved in the assessment and you).

You might be asked to wear a facemask and disinfect your hands at arrival to our research facility. Social distancing measures will be applied where possible. However, during the assessment preparation and for some parts of the assessment, social distancing may not be possible.

What will happen to my information?

After you have signed a consent form you will be assigned a unique number. from then on, this number will be used to identify you throughout the study.

All electronic data will be held securely on NHS or University computers. Access to this information will be restricted to members of the research team.

As well as the data collected at the study visits we may also collect some routine data from your medical records. This may include information about your operation, diagnosis and treatment, where it is relevant to your participation in this study.

Cardiff University is the sponsor for this study based in the UK. We will be using information from you and your medical records in order to undertake this study and will act as the data controller for this study. This means that we are responsible for looking after your information and using it properly. Cardiff University will keep identifiable information about you for up to 15 years after the study has finished.

Your rights to access, change or move information are limited, as we need to manage your information in specific ways in order for the research to be reliable and accurate. If you withdraw from the study, we will keep the information about what we have already obtained. To safeguard your rights, we will use the minimum personally-identifiable information possible. You can find out about how we use your information by contacting the project lead detailed on the next page.

You can find out more about how we use your information at: https://www.cardiff.ac.uk/public-information/policies-and-procedures/data-protection or by contacting the University's Data Protection Officer: inforequest@cardiff.ac.uk

The NHS will use your name, NHS number and contact details to contact you about the research study, and make sure that relevant information about the study is recorded for your care, and to oversee the quality of the study. Individuals from Cardiff University and regulatory organisations may look at your medical and research records to check the accuracy of the research study. The NHS will pass these details to the Biomechanics and Bioengineering Research Centre (Cardiff University) along with the information collected from you and/or your medical records. The only people in Cardiff University who will have access to information that identifies you will be people who are conducting the research, those who need to contact you about the study or audit the data collection process.

The NHS will keep identifiable information about you from this study for at least 10 years after the study has finished..

With your consent, anonymous data collected in the study may be shared with other institutions, including Universities and commercial organisations.

You will not be identified in any reports, presentations or publications relating to this research.

Other Useful information about this study.

Occasionally, during a research project, new information may become available. If this happens you will be contacted by a member of the research team to explain how this may affect you and your participation in the research.

We do not routinely send a letter to your GP to inform them that you are taking part in this research. However, we would still like to collect the details of your GP for the study. This will only be used to ensure that it is still appropriate to contact you for study follow-up visits.

This research has been reviewed approved by Wales Research Ethics Committee 3 (REC3) and is managed by Cardiff University.

If something goes wrong and you are harmed due to negligence, you may have grounds for legal action. If you wish to make a complaint about the way you were approached or the treatment you have received within the study please contact Cheryl Cleary: Centre Manager 029 2251 0265. If you feel your complaint is not adequately addressed, you may escalate your complaint by writing to: The School Manager, School of Bioscience, Cardiff University, Museum Avenue, Cardiff, CF10 3AX

As well as being asked to take part in this research you may also be asked if you are interested in taking part in some of the other Centre studies.

For each of these studies you will be provided with a further information sheet and have the opportunity to ask questions. For each additional study you will be asked to sign a consent form before and research activity is performed.

What happens next?

This information sheet covers research into a wide range of joint and back problems. The study requirements vary depending on the joint under investigation and the planned treatment.

If you still have questions after reading this information, please contact a member of the research team.

Contact Details:

Centre Manager
Cheryl Cleary
Biomechanics and Bioengineering Research Centre Versus Arthritis
Cardiff School of Biosciences
Cardiff University
Cardiff
CF10 3AX
Tel: 029 2087 5419 or 029 2087 4986
email: ArthritisCentre@Cardiff.ac.uk

NHS Site
Cardiff and Vale Orthopaedic Centre - Research Office
University Hospital Llandough
Tel: 029 2071 6370

Project Lead / Contact
Jake Bowd (High Tibial Osteotomy Project)
email: BowdJ@cardiff.ac.uk
Marina De Vecchis (Total Knee Replacement Project)
email: DeVecchisM@cardiff.ac.uk
Musculoskeletal Biomechanics Research Facility
Cardiff School of Engineering
Cardiff University

Thank you for taking time to read this information sheet

More information about the Biomechanics and Bioengineering Research Centre Versus Arthritis can be found by visiting:
http://www.cardiff.ac.uk/arthritis-biomechanics-bioengineering-centre

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PATIENT CONSENT FORM

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Assessment of joint function in patients with joint problems using three dimensional motion analysis techniques

Centre ID: ___________________ **Project Name:** __________________________

You DO NOT have to sign this document. Please DO NOT sign this document unless you fully understand it. If there is ANYTHING which you do not understand please do not hesitate to ask for a full explanation.

To confirm agreement with each of the statements below, please <u>initial</u> each box and delete where applicable:

1. I confirm that I have read and understand the information sheet dated 03 August 2020 (Version 12.2) for the above study. I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily..	☐
2. I understand that my participation in the study is voluntary and that I am free to withdraw at any time, without giving any reason, and without my medical care or legal rights being affected but any data collected up to the point of my withdrawal will be kept.	☐
3. I understand that relevant sections of my medical notes and other data routinely collected by the NHS related to my treatment may be looked at by individuals from Cardiff University, from regulatory authorities and from NHS Organisations where it is relevant to my taking part in the research. I give permission for these individuals to have access to my medical records.	☐
4. I agree for my movements to be recorded using audio-visual equipment. I understand that digital masking will be used to ensure my anonymity if the footage is used in any publication or presentation.	☐

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PATIENT CONSENT FORM

5. I understand and agree that the research team will securely store my identifiable details in order to contact me in future regarding this study (e.g. telephone/text/email). Identifiable details, including a copy of the consent form, will be available only to the research team, other than for purposes of monitoring and audit. ☐

6. I agree to take part in the above study. ☐

Optional

I agree that anonymous information collected during the study may be shared with external collaborators in the UK and abroad, including commercial companies. ☐

I agree to be contacted in the future to ask if I would be interested in taking part in future research into my joint/back problem. ☐

Name of Participant	Date (dd/mmm/yyyy)	Signature
Name of person obtaining consent	Date (dd/mmm/yyyy)	Signature

Original Investigator Site File / Trial Master File, 1 copy for the participant; 1 copy for the patient notes (where applicable), 1 copy researcher

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