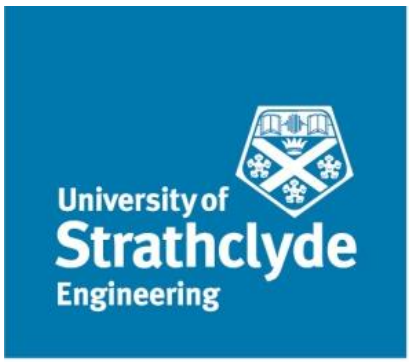
UNIVERSITY OF STRATHCLYDE, BIOMEDICAL ENGINEERING



Clinical Investigation of the Functional Outcomes of Fixed Bearing Versus Mobile Bearing Knees.

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PhD Biomedical Engineering, 2022.

This thesis is submitted in fulfilment of the requirements for the degree of PhD in Biomedical Engineering.

This thesis is the result of the author's original research. It has been composed by the author and has not been previously submitted for examination which has led to the award of a degree.

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Date:

Acknowledgements

Writing the acknowledgements is the loveliest conclusion to this project and as with any research project there are many people without whom this thesis would not have been completed. Firstly, a huge thank you to all participants who kindly gave up their time to take part in the study, hear my rubbish jokes in the lab, and bared with any technical difficulties, always with a smile. The photo of the artwork on the next page came from one of our participants and I always considered myself to be privileged to have a project involving people, and what a wonderful group I had.

Next, Dr. Philip Riches, who would probably say I have been in his life for far too long! Thank you so much for all your patience – and importantly for allowing me to be the recipient to your terrible jokes in our meetings. I always felt you believed in me when I did not believe in myself and I will always appreciate having you as a supervisor. Thank you also to my other University of Strathclyde supervisors: Dr Andrew Murphy and Dr. Craig Childs. Andrew, you brought me onto this project, and I couldn't have asked for a better person to introduce me to motion capture. Craig, you picked up where Andrew left off and so many times, I was convinced you were the CAREN whisperer!

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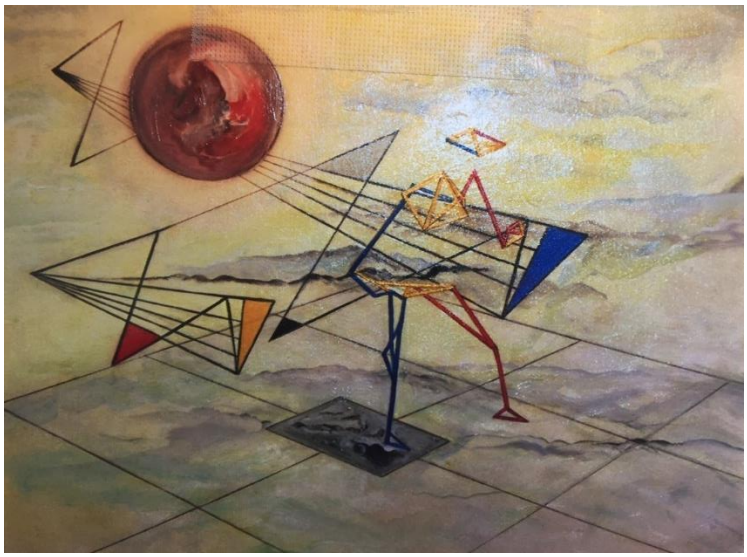

Publications

Komaris, D.-S., Govind, C., Murphy, A.J., Clarke, J., Ewen, A., Leonard, H., Riches, P., 2021. Implant design affects walking and stair navigation after total knee arthroplasty: a double-blinded randomised controlled trial. Journal of Orthopaedic Surgery and Research 16, 177.

Komaris, D.S., Govind, C., Clarke, J., Ewen, A., Jeldi, A., Murphy, A., Riches, P., 2020. Identifying car ingress movement strategies before and after total knee replacement. International Biomechanics 7, 9–18.

Komaris, D.-S., Govind, C., Murphy, A., Ewen, A., Riches, P., 2018. Identification of Movement Strategies During the Sit-to-Walk Movement in Patients With Knee Osteoarthritis. Journal of Applied Biomechanics 34, 96–103.

Conference Proceedings

Govind, C., Komaris, D., Childs, C., Ewen, A., Jeldi, A., Clarke, J. V., Picard, F., Riches, P. E. (2017). Short-term Lower Limb Mechanics During Stair Descent and Downhill Gait in Total Knee Arthroplasty Patients. International Society of Biomechanics Congress, Brisbane.

Govind, C., Komaris, D., Ewen, A., Clarke, J., Picard, F., Riches, P. E. (2016). Short-term Knee Flexion During Stair Ascent in Total Knee Arthroplasty Patients. Presented at British Orthopaedic Research Society. University of Glasgow.

Project Background

This project was a collaboration between the University of Strathclyde, and the Golden Jubilee National Hospital (GJNH), Clydebank and B.Braun, the project funders. This was a randomised controlled study to investigate the functional outcomes of three commercially available total knee replacement implants. Myself and another student from the University of Strathclyde were sub-contracted alongside the Orthopaedic Research team at the Golden Jubilee to achieve the project deliverables. Whilst recording TKA participants' motion during activities of daily living (ADLs) was stipulated in the contract, the study protocol, data collection and analysis was carried out independently from the funders.

The study protocol was larger than presented here, results from gait-based tasks were presented in this thesis however participants also carried out activities that were not gait-based. Between both researchers there was a divide in the tasks, and the methodology for recording each task was independently researched and presented in meetings to gauge agreeability. Once ethical approval was received patients were recruited by members of the orthopaedic research team at the GJNH, and controls were recruited by the PhD researchers. Participants were recorded in a single session for both researchers' tasks, and data capture was carried out together. Data processing and analysis was subsequently completed independently. All motion capture data for the gait-based tasks was processed by myself, and all processing pipelines and MATLAB scripts were created by myself with no overlap to the non-gait based activity analysis.

Where findings have been published this was firstly drafted by the relevant researcher and sent around for comment to the rest of the research team. For the paper titled "Implant design affects walking and stair navigation after total knee arthroplasty: a double-blinded randomised controlled trial" this was drafted by the other researcher using the data from this thesis.

Abstract

Total Knee Arthroplasty is a high-volume and high-cost procedure, with persisting limitations to patient satisfaction. Prosthesis designs aim to restore function whilst providing stability, without joint constraint. This double-blinded randomised controlled trial is the first of its kind where the functional performance of a low congruent fixed (CR DD), ultra-congruent fixed (UC), and ultra-congruent mobile (UCR) bearing Columbus Total Knee Systems were assessed. The pre- and postoperative function of twenty-four osteoarthritic patients was evaluated against nine control participants whilst carrying out activities of daily living. Spatiotemporal, kinematic, and kinetic gait parameters during walking, stair navigation and sloped walking were extracted using fully instrumented motion capture. Questionnaire responses were also recorded. Across all ADLs, postoperative patient function improved, although not to control levels. The average postoperative increase in range of sagittal knee motion across all tasks came to: 7.3±3.1^{o} (CRDD), 4.9±4.9^{o} (UC), 0.7±7.7^{o} (UCR), and peak knee flexion was mostly reduced at postoperative. Both fixed bearing implants presented larger post-surgery hip and ankle kinetics in magnitude, and improved distinction between knee adduction moment maxima, linked to improved loading to the mobile bearing group. Overall, the CRDD group showed more significant changes to preoperative and any significant inter-implant differences at post-surgery was also to this group. The UC and UCR groups showed less improvements during challenging activities, with the UCR group showing some limits to knee extension. The UCR group also self-reported more difficulty, pain, and tiredness than the fixed bearing groups. Kinematic cross talk error significantly impacted the interpretation of non-sagittal kinematics, and small and unequal sample sizes reduced statistical power. Despite the limitations it was concluded that both fixed bearing implants initially outperformed the mobile bearing joint and the CRDD group showed the most prominent improvements. Clinically relevant thresholds for all parameters, would further determine whether functional advantages exist between implant bearing types.

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List of Abbreviations and Notations

3D	Three Dimensions
AJC	Ankle Joint Centre
COM	Centre of Mass
CR DD	Cruciate retaining deep – dish (implant)
GRF	Ground Reaction Forces
HJC	Hip Joint Centre
HS	Heel Strike
KAD	Knee Alignment Device
KAM	Knee Adduction Moment
KFHS	Knee Flexion at Heel Strike
KFM	Knee Flexion Moment
KFTO	Knee Flexion at Toe off
KJC	Knee Joint Centre
MKFS	Minimum Knee Flexion angle during Stance
OA	Osteoarthritis
OKS	Oxford Knee Score
PKFS	Peak Knee Flexion angle during Stance
PiG	Plug-in Gait
ROM	Range of Motion
SD	Standard Deviation
SE	Standard Error
TO	Toe Off
TKA	Total Knee Arthroplasty
UC	Ultra-congruent (implant)
UCR	Ultra-congruent rotating (implant)

Anatomical Planes

Coronal/Frontal	Divides the body into its anterior and posterior portions
Sagittal/Median	Divides the body into left and right portions.
Transverse/Horizontal	Divides the body into superior and inferior parts.

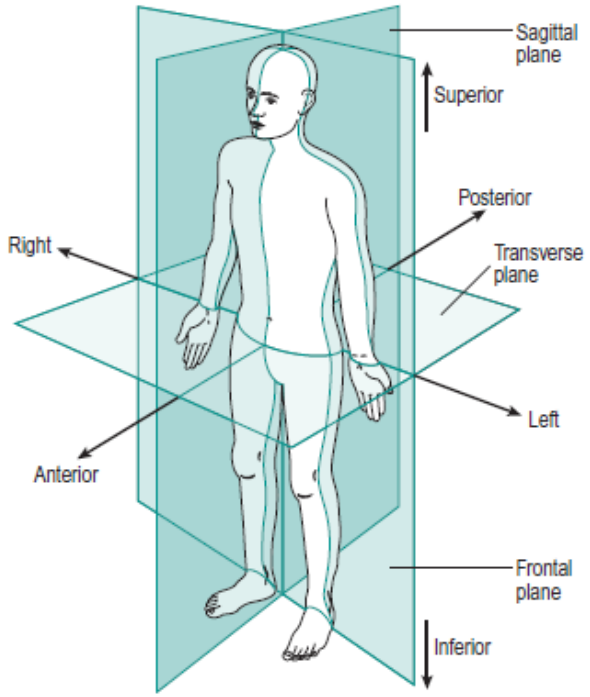

Figure 0-1: Human anatomical planes and six fundamental directions. (Whittle, 2006)

Nomenclature

Anthropometry	Measurement of size, weight, and proportions of the human body
Kinematics	Study of joint motion and angles
Kinetics	Study of forces and moments exerted on to rigid bodies
Spatial	Changing in relation to space
Temporal	Changing in relation to time

1. Introduction

Total knee arthroplasty (TKA) is a joint replacement surgery performed on patients with end-stage degenerative joint disease. In England, Wales, and Northern Ireland, Between April 2003 to December 2019, 1,145,000 primary TKAs were carried out (National Joint Registry, 2020). The major recorded indication for surgery was osteoarthritis (OA) reported by 97.4% of patients. As the population increases in age and size there will inevitably be a rise in the number of OA sufferers. For younger sufferers, the demands of a TKA are slightly different than, say, those over 65 years of age. The activity levels and lifestyle are different in younger sufferers who may be working or partaking in sports meaning that the joint is placed under a greater demand, requiring implants with improved functional performance.

Clinical knee scoring systems, such as the Oxford Knee Score (OKS), are questionnaires that report the success of an individual TKA. These are subjective and evidence suggests the OKS is insensitive to postoperative differences between different interventions (Jenny and Diesinger, 2012). This may increase the chance of a type II error where it could be concluded that no difference exists when one does exist. To accurately determine the functional outcome of a TKA, multidimensional gait analysis methods can be used. This potentially allows small postoperative biomechanical differences between intervention groups to be accurately quantified.

Within the natural knee joint, the bones articulate in a motion difficult to replicate in a prosthetic design. Components of an implant used for TKA include the femoral and tibial metal prostheses along with a polyethylene insert which provides a bearing surface between them. The bearing may be of a fixed or mobile design whereby in the fixed bearing (FB), the insert is locked to the tibial prosthesis and is well established with successful long term follow up in OA patients. In contrast, mobile bearing (MB) implants allow longitudinal rotation of the insert to better attempt replication of natural knee movements. Although there are theoretical benefits to the MB design, many of these are yet to be verified as many studies documented no

improvement in outcomes compared to FB implants (Jacobs et al., 2012; Mahoney et al., 2012; Post et al., 2010)

Currently there is little research using current movement analysis techniques to investigate bearings with high and low congruencies during a range of activities of daily living (ADLs). Any studies doing so primarily analyse joint motion during level walking which is not a demanding ADL. Therefore, the higher ranges of joint motion that may highlight differences between designs would not be achieved. This study used quantitative and functional testing to determine the performance of implant designs in different bearing and congruency configurations over a range of ADLs. The results from this study may be used to better inform implant manufacturers on areas of further research to improve existing designs.

2. Review of Literature

2.1. Anatomy

The knee joint is a weight bearing, diarthrodial joint which works harmoniously with the hip and ankle joints, allowing smooth motion throughout various activities. Three main bones articulate within this joint: the femur (the thigh bone), tibia (the shin) and the patella (the kneecap). Many other supporting structures are present, such as the ligaments and muscles, which are described further here. Reference here has been made to Drake et al., 2009; Moore et al., 2009 and Crouch and McClintic, 1976.

2.1.1. Skeletal Structures

The main bones in this joint, the femur and tibia are both long bones with similar internal structures. The proximal femur forms the ball and socket hip joint by articulating with the pelvis. Distally, the rounded femoral condyles articulate with the plateau-like distal tibia (Figure 2-1) and the patella slides against the intercondylar fossa between the femoral condyles where ligaments also pass through. The lower limb muscles stabilise the joint and insert into various sites on the bone surface via tendons. The tibia is medial in the lower limb and forms parts of the knee and ankle joints. The distal tibia widens to form the medial malleolus and the lateral malleolus is formed by the distal fibula.

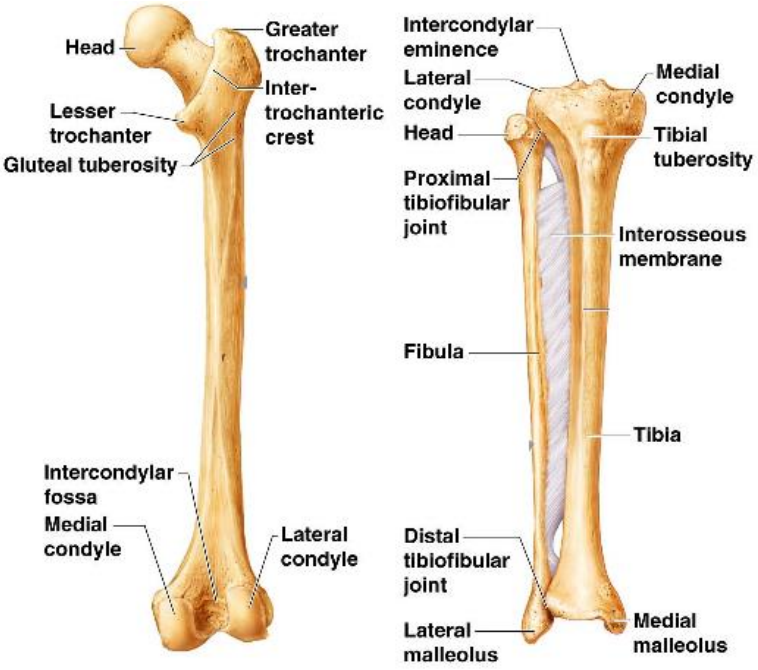

Figure 2-1: Bony landmarks on the femur (left) and tibia and fibula (right). Image from Marieb and Hoehn, 2009.

The patella is a triangular sesamoid (tendon-embedded) bone found within the quadriceps femoris muscle tendon. The patella's articulation with the femur allows the pull of the quadriceps femoris to be directed anteriorly over the knee to the tibia without causing tendon wear.

Other skeletal structures found in the lower limb are the pelvis and ankle. The acetabulum (a deep lateral socket) of the pelvis articulates with the femoral head to form the multi-axial and ball and socket hip joint as shown in Figure 2-2.

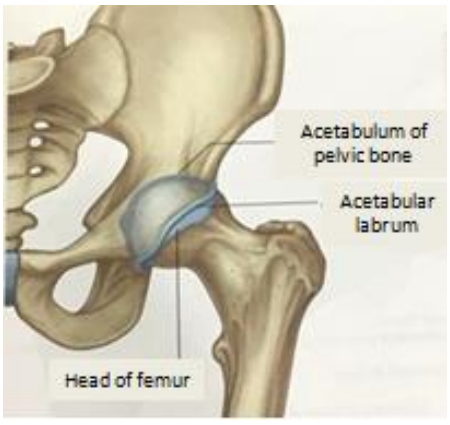

Figure 2-2: The hip joint (Drake et al., 2009).

The ankle joint includes the distal fibula and tibia and the talus of the foot (Figure 2-3). The distal fibula and tibia are joined by strong ligaments forming a bracket-like structure for the upper-expanded part of the talus body to fit into. Forces are transmitted between the tibia and the talus.

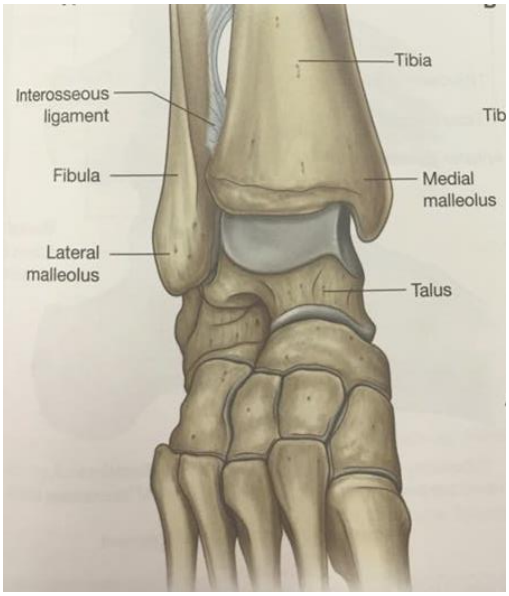

Figure 2-3: Anterior view of the ankle joint (Drake et al., 2009).

2.1.2. Structures of the Knee Joint

The knee is the largest synovial joint in the body and each complex articulation is reinforced by additional structures such as the muscles, ligaments, and menisci which keep the joint as stable as possible.

2.1.2.1. The Synovium

Hyaline articular cartilage (AC) covers the main interacting surfaces of the joint in order to protect it from wear. The surface of AC is extremely smooth, giving an extremely low friction surface as the bones move through their range of motion. A capsule containing synovial fluid unites the interacting bones as shown in Figure 2-4. Synovial fluid provides nutrition and primarily lubricates the joint. When compressive forces act on the knee, the synovium may act to absorb shock to minimise damage and cushion the joint.

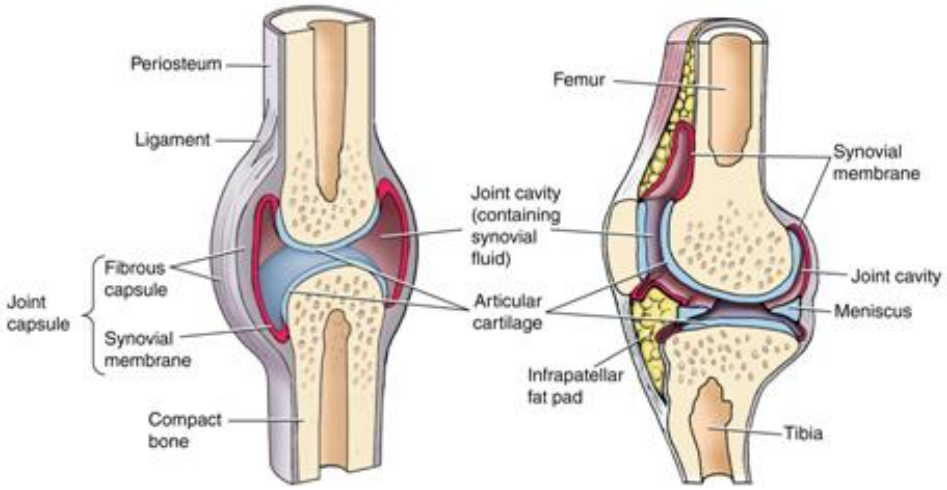

Figure 2-4: The knee joint in the synovial capsule (Moore and Daly, 2009).

2.1.2.2. The Ligaments

The ligaments stabilise the knee and prevent over-displacement of the femur with respect to the tibia. Any ligamentous laxity is known to contribute to osteoarthritis (van der Esch et al., 2005). The four main ligaments of the knee (as depicted in Figure 2-5) are the:

- Lateral collateral ligament (LCL),
- Medial collateral ligament (MCL),
- Anterior cruciate ligament (ACL),
- Posterior cruciate ligament (PCL).

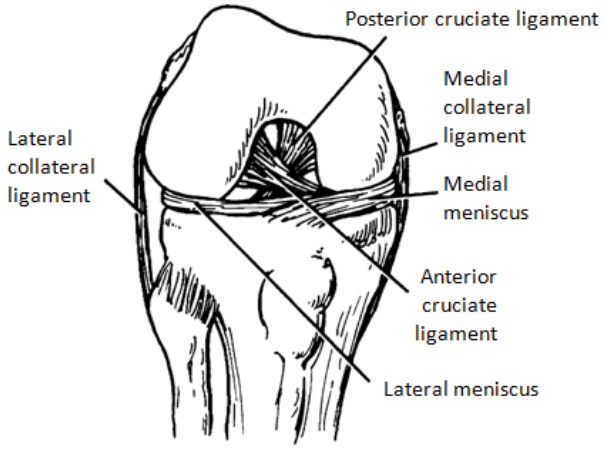

Figure 2-5: The menisci and main ligaments of the knee joint.

Each ligament arises from the femur and inserts into the tibia, apart from the LCL which inserts into the fibula. The collateral ligaments stabilise the “hinge-like” motion

of the joint and are at their tightest when the joint is in full extension to protect from varus or valgus directed forces. The cruciate ligaments govern the knee's natural roll and glide motion and protect against shearing and torsion forces. Each cruciate ligament is made up of different bundles of fibres which are taut at different angles of knee flexion. The PCL is the strongest ligament in the knee with a tensile strength nearly twice that of the ACL (Kennedy et al., 1976).

2.1.2.3. The Menisci

The menisci lie on the tibial plateau and are fibro-cartilaginous crescent-shaped pads which improve the congruency between the femur and tibia (Figure 2-6). They have impact absorbing characteristics and reduce joint pressure by increasing contact surface area between the bones. In order to distribute load more effectively during gait, the menisci are able to change their shape and move in an anteroposterior direction.

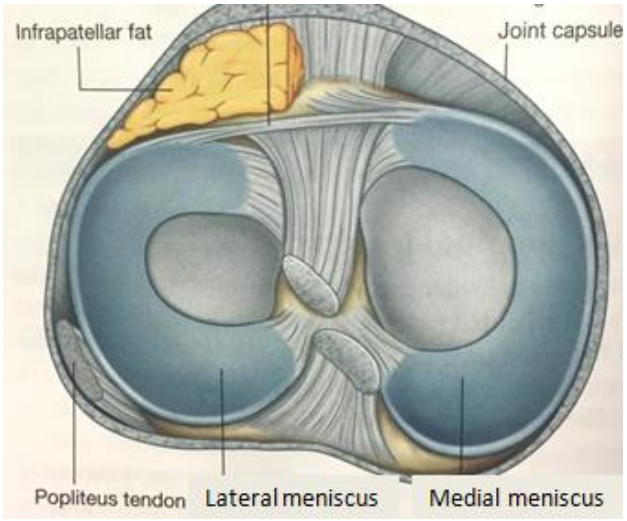

Figure 2-6: A transverse view of the knee showing the lateral and medial menisci (Drake et al., 2009)

The menisci pack the space between the femur and tibia converting the tibia into a shallow socket allowing for increased stability. As the femoral condyles are loaded and the joint becomes compressed, this forces the wedge-shaped menisci to move radially outwards. As the menisci have powerful attachments to the tibia, this causes resistance to the extruding force allowing the force to be transmitted from the femoral to the tibial articular surfaces which are not in contact.

2.1.2.4. Musculature

The musculature of the lower limb controls motion which is mostly carried out by the posterior flexor (hamstrings) and anterior extensor (quadriceps) groups (Figure 2-7). The four main extensor muscles insert via a common quadriceps femoris tendon to their location above the patella, which then connects to the tibia via the patella tendon. The main quadriceps muscles are:

- *Vastus medialis*
- *Vastus intermedius*
- *Vastus lateralis*
- *Rectus femoris*

The flexor hamstring muscles counteract the extensor muscles and are comprised of:

- Semitendinosus
- Semimembranosus
- Long head of *biceps femoris*
- Short head of *biceps femoris*.

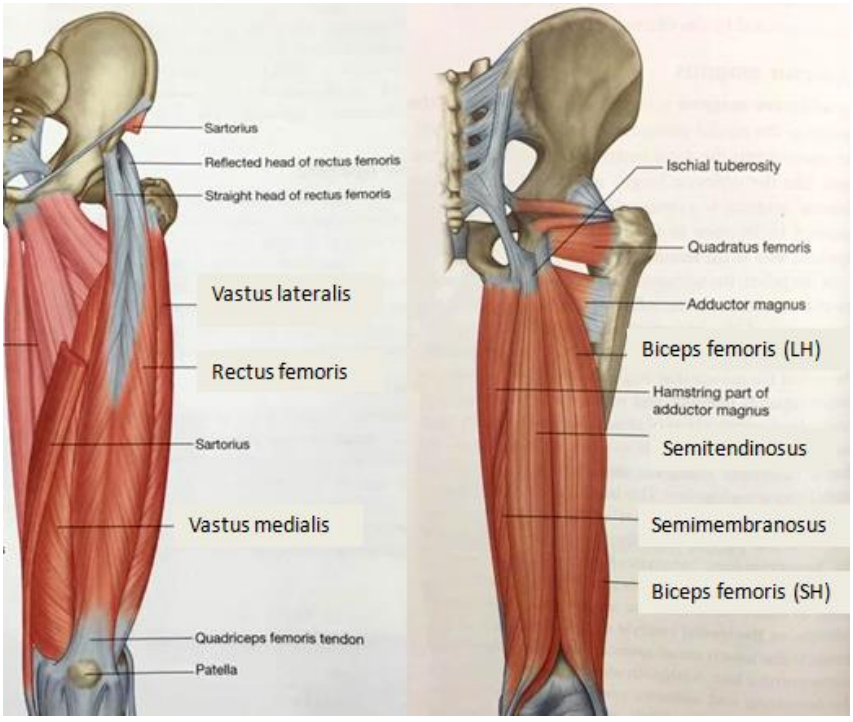

Figure 2-7: An anterior and posterior view of the thigh showing the main extensor (left) and flexor (right) muscles. LH/SH refers to the long and short head of bicep femoris respectively (Drake et al., 2009).

2.1.3. Alignment

Anatomically, the tibia is a near vertical bone and the femur runs obliquely from the pelvis to the tibia allowing for a larger range of motion. The orientation and position of each bone is crucial for efficient load transmission and various axes and the angles between them describe different alignment aspects as depicted in Figure 2-8.

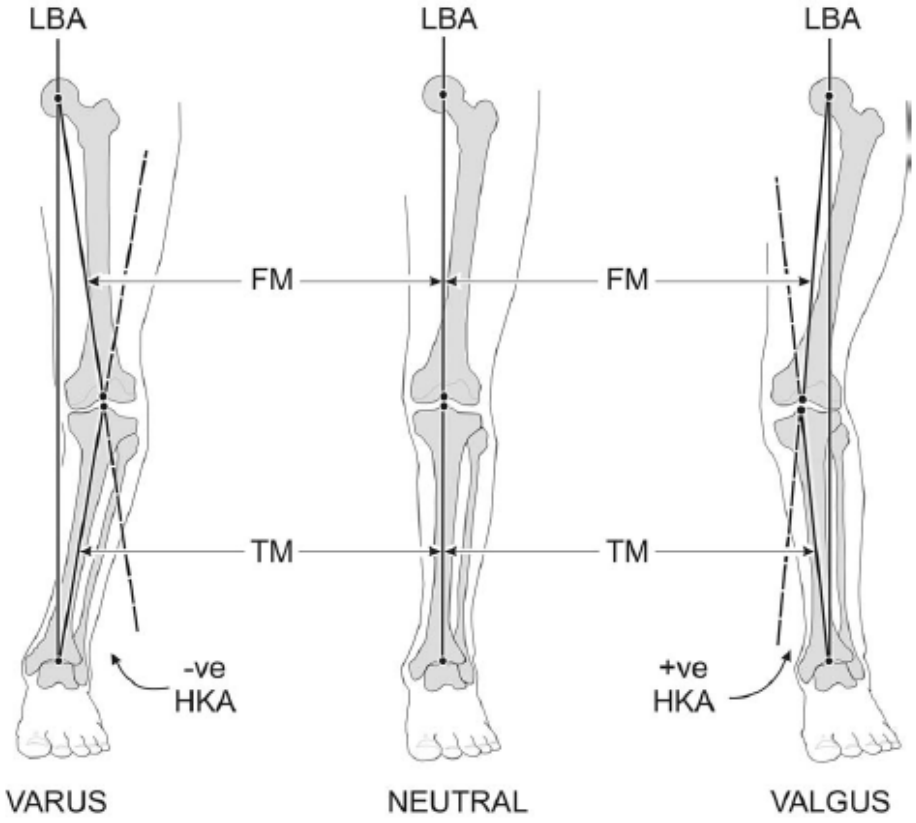

Figure 2-8: Lower limb alignments. LBA = Load Bearing Axis, FM = Femoral Mechanical Axis, TM = Tibial Mechanical Axis, HKA = Hip-Knee-Ankle angle (also the femorotibial angle). (Cooke et al., 2007).

Three key joint centres and the axes between them relate to lower limb alignment: the hip joint centre (HJC), the centre of the femoral head, the ankle joint centre (AJC), commonly the centre of the medial and lateral malleoli, and the femoral or tibial knee joint centres (KJC). The femoral KJC is usually a mid-condylar point on the femur between the cruciate ligaments (Takahashi et al., 2004) and the tibial KJC is the centre of the tibial plateau.

The load bearing axis (LBA) is defined as the line running from the HJC to the AJC (Cooke et al., 2007). In healthy people, the LBA correlates with the femoral and tibial

mechanical axes (FM and TM respectively) which are neutrally aligned with respect to each other. The FM axis runs from the HJC to the femoral KJC, and the TM axis centre runs from the tibial KJC to the AJC. Together, these axes make the mechanical axis of the knee. The angle between the FM and TM is the hip-knee-ankle (HKA) angle (or the mechanical femorotibial (MFT) angle) and in a neutrally aligned limb, the MFT angle will be 180^{o}, overlapping with the LBA.

A healthy MFT angle range during standing is quoted as 180±3^{o}, in healthy knees there is a natural varus alignment of 1.1-1.5^{o} (Takahashi et al., 2004) often measured from full limb, weight-bearing radiographs. If the MFT angle deviates far from the healthy range, then the limb can be described having a *genu varus* or *genu valgus* deformity. In a varus knee the LBA is shifted medially causing force to be transmitted through the inner compartment of the knee. Conversely, in a valgus knee the LBA is shifted laterally causing higher stress at the outer compartment of the knee. In osteoarthritic patients, deformities causing an altered LBA are common. Abnormally high forces produced from both conditions causes increased cartilage wear and disease progression in either the medial or lateral compartment which can spread throughout the joint.

2.2. Osteoarthritis

Osteoarthritis (OA) is a degenerative joint disease and approximately 8.5 million in the UK people have painful joints due to OA (NICE, n.d.). OA may affect any joint, and commonly affected areas are the knee, hip, ankle, foot, hand, and wrist. The ends of the bones in healthy joints are covered with articular cartilage (AC) which ordinarily allows near frictionless movement over each other. AC can also absorb energy from movement impact and in OA there is commonly degradation of the cartilage of articulating bone surfaces. When AC loss is localised to the knee then pain is experienced during load-bearing activities such as walking or standing. In severe cases, loss of joint function is possible which greatly affects the patient's quality of life.

This risk of developing OA is multifactorial and some of the associated risk factors are:

- Age: as cartilage breakdown becomes faster than formation.
- Sports or work trauma: sudden or repetitive damage of the supporting ligaments or meniscus may alter lower limb alignment resulting in localised OA damage.
- Genetics: It has been suggested that there is a 40-60% heritability factor (Williams and Spector, 2006) showing the importance of genes in OA development. And so, OA is likely to run in families.
- Obesity: carrying extra mass is a form of a static load which accelerates joint wear (Felson, 1996).

2.2.1. The Osteoarthritic Joint

In healthy knees, there is an equilibrium of cartilaginous matrix formation and breakdown on articular surfaces which respond to factors such as increased joint forces or load line shift (Michael et al., 2010). Mechanisms are in place which compensate for the effect of OA-causing influences, namely by modifying the metabolic activity of chondrocytes in the AC (Goldring, 2012). When damaging influences exceed the capability of the system to compensate then the first step in the development of OA occurs: irreversible degradation of the cartilaginous matrix. A cascade of disease progression is triggered including worsening of the joint congruency, which increases lower limb malalignment and worsens the condition further beginning a cycle of deterioration that directly affects knee motion and function (Figure 2-9).

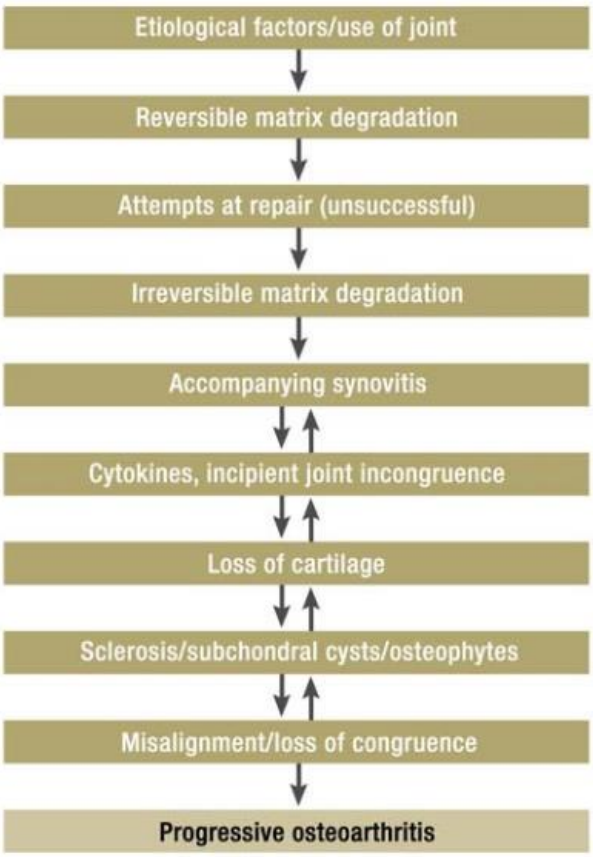

Figure 2-9: The pathogenesis of OA (Michael et al., 2010).

2.2.2. Diagnosis and Management

Often, prior to diagnosis a person with OA will already experience knee pain, stiffness or swelling by which time the majority of joint damage will have occurred. At this point the patient will visit their doctor for diagnosis and evaluation of their stage of OA using clinical questionnaires, and analyse the change in strength and ROM. Finally, weight-bearing radiographs are taken to assess the degree of narrowing in the joint space and to exclude any other causes of pain.

Current treatments are not able to cure OA or reverse AC damage. Their aim is to relieve pain, improve joint function and reduce the contributing symptoms that worsen OA. Such treatments include pharmaceutical interventions (to alleviate pain or swelling), keeping active and mobile to improve knee ROM, and weight loss to reduce joint loading (Van Manen et al., 2012). Once these non-operative treatments cease to be effective then surgery is the next viable option.

2.3. Knee Arthroplasty

For those with severe degenerative joint disease, knee arthroplasty is a surgical procedure carried out to alleviate pain and disability and restore knee motion and lower limb alignment. Here, the knee's articulating surfaces are replaced with an endoprosthesis which mimic natural movement. Arthroplasties, or replacement surgeries, may be total (TKA) where the whole joint is replaced or, partial (UKA) if the damage is localised to one or more compartments of the knee (Figure 2-10).

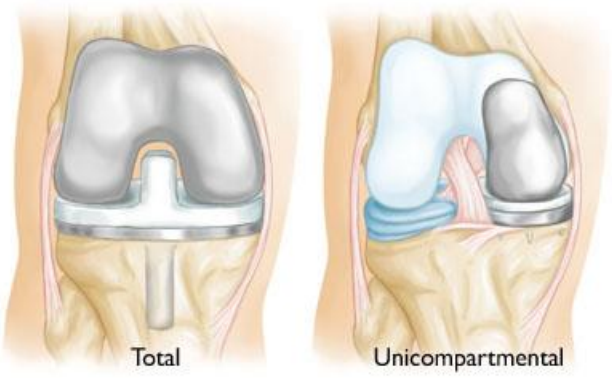

Figure 2-10: A depiction of a total (left) and unicompartmental (right) knee replacement ("Unicompartmental Knee Replacement-OrthoInfo - AAOS," 2010).

TKA may be carried out using conventional (or traditional) instrumentation or, more recent techniques utilise intraoperative computer/robot assistance. The procedure requires bone resection using four basic cuts: the distal, posterior, and anterior femur (Figure 2-11) and the proximal tibia. All cuts ought to achieve good alignment of the femoral, tibial, and patellar components in each plane.

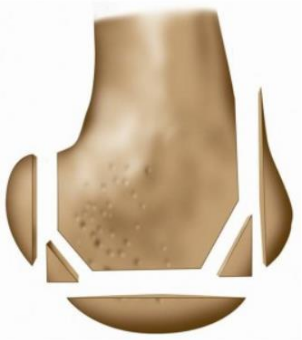

Figure 2-11: The three main femoral cuts (anterior, posterior, and distal) and chamfer cuts ("Custom Made Knees," n.d.).

The mechanical axis and the rotational and translational alignments of the femoral and tibial components and an equalised flexion/extension gap are important for good functional outcome (Sparmann et al., 2003). Following implant insertion, good tissue balance that allows stability without movement limitations or excessive compression of the implant is a further task for the surgeon. Balancing of the joint by the contraction or release of soft tissue and ligaments is also important in fixing preoperative deformities. The release of the medial soft tissue (including the deep and superficial MCL) will correct varus deformities. For the correction of valgus deformities, the LCL, iliotibial band, and/or the posterolateral capsule may be released.

2.4. Knee Prostheses

As well as accurate surgical technical skills, the success of a TKA is also dependent on implant design. There are at least 150 different TKA implant designs (Carr and Goswami, 2009). The first replacements used hinged prostheses which had a short lifespan (of around 2 years) due to mechanical failure and high rates of loosening (Shetty et al., 2003). To balance biocompatibility with strength requirements more appropriate materials have since been selected for implant design. The use of titanium alloys, cobalt chrome, and ultra-high molecular weight polyethylene (UHMWPE) improved implant design. However, there are still issues due to wear, loosening and other factors which limit the implant's performance.

The first surface replacing TKA implants emerged in the 1950's (Carr and Goswami, 2009) which sought to replace damaged AC on the distal femur and proximal tibia. The natural distal femur has a convex surface and the tibial plateau has a shallow concave surface, meaning these surfaces are not congruent and only specific, localised portions of the articulating surfaces transmit load at any point. Advancements in knee biomechanics knowledge led to designs which give femoral implants which are polycentric, that preserve intact ligaments, and are better sized - all of which greatly improves joint motion. Understanding that the femoral condyles roll and glide on the tibial plateau with multiple instant centres of rotation, rather

than rotate around a single axis is crucial for implant design (Frankel et al., 1971; Fu et al., 2016; Gerber and Matter, 1983; Hollman et al., 2002).

There are six degrees of freedom in a normal knee joint and prostheses are required to balance stability with the freedom of joint movement. A more constrained implant will be more stable, however the reduced freedom to move increases sheer force at bone/cement interfaces which may lead to loosening (Sathasivam and Walker, 1999). Also there is the issue of polyethylene wear debris which is greater in implants with a higher contact area (Brockett et al., 2018). Increased wear is linked to lower joint longevity as foreign particles trigger inflammatory responses within the joint causing implant failure due to osteolysis and implant loosening (Chakrabarty et al., 2015). Overall, it is vital the implant replicates the healthy joint as closely as possible and some factors to consider for implant design are:

- Posterior cruciate ligament (PCL): sacrificing or preserving.
- Fixation method.
- Bearing: fixed or mobile.

2.4.1. PCL Sacrificing vs. Retaining

Most TKA procedures require removal of the ACL however whether or not the PCL is preserved is dependent on level of PCL damage and the implant used. Implant designs based anatomical models are designed to be cruciate retaining (CR) and are theoretically considered better than PCL sacrificing implants. CR implants purportedly provides more natural knee kinematics and stability as anterior translation of the femur on the tibia is prevented (Kolisek et al., 2009). This allows more demanding tasks to be carried out with greater confidence (Song et al., 2013). Furthermore, revision surgeries are made simpler with PCL preserving types since there is less bone resection (Kolisek et al., 2009).

Functional designs which sacrifice the PCL attempt to replace the PCLs role with a femoral cam and tibial post (Figure 2-12). This mechanical interaction involves the post to extend centrally from the tibial insert into the cam placed between the

femoral condyles. This design gives non-anatomical joint surface geometries which intend to reduce point polyethylene stress and maximise contact surface area (Robinson, 2005). As a result of the increased contract surface area, this prosthesis type is slightly more constrained. However, this constraint is necessary to prevent disengagement of the cam and post and allow expected implant motion (T. Brown et al., 2014). A last advantage to this design is that the TKA procedure is less technically demanding, reducing the risk of postoperative complications (Kolisek et al., 2009).

The design of the cam and post controls sagittal and coronal (if a larger post is used) plane kinematics (Williams et al., 2010). At flexion angles of around 60-70^{o}, the tibial polyethylene post comes into contact with the cam of femoral component inducing posterior displacement of the femur giving satisfactory roll-back and knee flexion (Freeman and Railton, 1988). Conversely to cruciate retaining implants, revision surgeries are more difficult with PCL sacrificing implants due to bone loss at the femoral intercondylar notch to accommodate the cam-post design. There are also additional risks of wear at the cam-post interface, and of “cam over post” jump. The jump may occur in posterior stabilised implants with loose flexion gaps, or during periods of hyperextension the cam could rotate over the post and dislocate. The jump would then be treated by carrying out an anterior drawer manoeuvre or a revision surgery to remedy the loose flexion gap (Hagedorn and Levine, 2012).

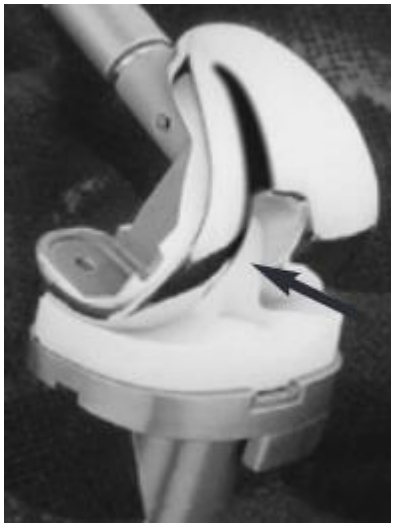

Figure 2-12: A typical posterior stabilised TKA implant showing a larger femoral post (arrow) to improve coronal plane stability (Williams et al., 2010).

Another implant type utilises variable congruency of the polyethylene insert allowing the PCL to be resected whilst maintaining knee kinematics and joint stability. This involves the lateral and medial sides of the insert to be highly congruent to the curvature of the femoral condyles (Laskin et al., 2000). When combined with proper soft tissue balance the PCL can be resected without requiring the cam and post mechanism and may give the benefits of cruciate retaining procedures where PCL preservation is unachievable (Stronach et al., 2019).

Individual studies comparing PCL retaining vs sacrificing implant types report conflicting results, however comparing many at once shows some similarities. A meta-analysis by Jacobs et al., in 2005 compared the results of eight randomised controlled trials and found flexion ROM was significantly higher in the PCL sacrificed implants with a post and cam mechanism by 8.1^{o}. Another meta-analysis of six studies carried out by Bercik et al., (2013) found a significant difference in peak and range of flexion in favour of PCL sacrificing implants. However, no significance was found when comparing complication rates. Similarly, Jiang et al., in 2016 assessed seven studies (four of which were also reported by Bercik et al.) and found significantly higher knee ROM in PCL sacrificing implant group compared to CR, mean difference: 7.7^{o}.

A retrospective study carried out by Stronach et al., in 2019 compared 161 Attune TKAs (Depuy Synthes, Warsaw, IN) carried out by a single surgeon. Here the PCL sacrificing implant was explicitly stated to not require a cam and post mechanism. In contrast to the above, no significant change in knee flexion ROM from preoperative between the PCL retained (n = 104) and sacrificed (n = 57) groups was seen. With no clear advantage of one implant type over another, further research is needed into surgical outcomes and patient populations.

2.4.2. Implant Fixation

The method to affix the prostheses to the distal femur and proximal tibia is crucial as implant loosening is the most common cause for TKA failure (Wright and Chitnavis, n.d.). Implant fixation may include the use of a bone cement on both the femoral and tibial components, be cementless, or utilise a mixture of the two where commonly

the tibial plate is cemented and the femoral component is not (Aprato et al., 2016). Bone cement fixation is achieved with polymethyl-methacrylate (PMMA) cement. Cement use allows secure and consistent fixation, necessary for the number of arthroplasties carried out. Cement also fills gaps at the bone-implant interface and transfers loads whilst resisting compressive forces (Jackson and Pagnano, 2012). However, concerns over the long-term tolerance of cement led to the manufacture of non-cemented fixation designs. Over time, cement may crack or wear away at the bone-cement interface causing osteolysis to occur which lends itself to further complications due to foreign particles in the operated joint (Akan et al., 2013).

Non-cemented implants have a surface topography which is coated or textured to be conductive to bone growth. Screws or pegs may also be used to stabilize the implant until bony ingrowth occurs (Song et al., 2013). Since these implants rely on new bone growth for stability, a longer healing time (up to three months) compared to cemented implants is required. This has considerable drawbacks regarding rehabilitation and returning to independent living. In addition, using un-cemented components requires healthy bone stock with high metabolic activity to promote biological fixation, potentially better suited for younger patients. Benefits of non-cemented implants include bone conservation, a shorter operative time, ease of revision and none of the disadvantages associated with cement fixation (Jackson and Pagnano, 2012). Early non-cemented implants demonstrated higher failure rates compared to cemented types due to aseptic loosening associated with micromotion (Matassi et al., 2014). Higher failure rates in non-cemented designs may also be attributed to slight cutting imperfections which cemented implants accommodate better. A bone-implant gap less than 1mm is required with cementless designs, so surgical technique error may be a factor here. In addition, the increased cost (up to three times) of the bioactive surfaces found in these implants (Aprato et al., 2016) may further explain why cemented implants are preferred as the cost of healthcare rises worldwide. Whilst cementless implants may be considerably more expensive than a cemented one, the time and costs saved in the operative theatre and cement

have been found to offset the cost of cementless components to give a differential of $150 (Kamath et al., 2011).

Since cemented implants have been in use longer than non-cemented implants, there are few studies directly comparing fixation types after long-term use. A study carried out by Rand et al., in 2003 analysed the survivorship of 11,000 TKA's performed between 1978 and 2000. The authors found cemented implants gave an estimated ten-year survivorship of 92%, significantly higher than 61% seen for cementless knees. However, this is now a dated study and with recent advancements in surgical techniques results may differ. Another study comparing the long term (11-16 year) survivorship probability of 100 cemented and cementless TKAs showed no statistical differences between the types (Prudhon and Verdier, 2017). However, cemented knee survivorship was 5.2% higher than cementless. All surgeries took place between 2003 and 2006, and both implants were identical except the inner portion of the cementless knee were coated with hydroxyapatite.

Materials research incorporated highly porous metals (such as tantalum) into implant design. Such metals could enhance bone growth allowing non-cemented implants to be used in younger patients. Bobyn et al., (2004) showed promising short-term results for a mono-block tibial implant (where the polyethylene insert is directly compression moulded to a tantalum base with fixation pegs). After two years, 101 knees with a mono-block tibia showed no evidence of loosening. Cement fixation of all implant components remains the gold standard and further research into low-cost, durable fixation could be carried out to improve current cement techniques.

2.4.3. Mobile vs. Fixed Bearing

The relationship between the polyethylene insert on the tibial tray further differentiates implants. The difference in bearing refers to whether the polyethylene insert is rigidly attached to the tibial insert (fixed bearing) or is partly unattached and free to translate and/or rotate around the tibial prosthesis (mobile bearing) as shown in Figure 2-13. Traditional fixed bearing (FB) implants have good success rates however, long-term survival studies generally include elderly participants with low

activity levels. In younger patients, FB implants have been associated with problems regarding wear of the polyethylene insert after long-term use (Song et al., 2013).

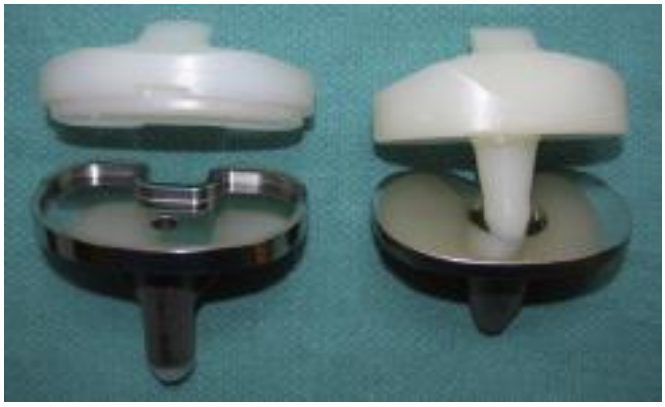

Figure 2-13: Examples of the conventional fixed- and mobile-bearing knee replacement implants (left and right respectively. Post et al., 2010).

FB knees may be further divided into low or high conformity designs, each of which have different mechanical characteristics (Figure 2-14). A high congruency knee indicates the polyethylene insert is better cupped to the femoral component and a low congruency knee indicates a flatter polyethylene surface. Where there is lower conformity between the two components, there is less constraint reducing the risk of early implant loosening. However, there is also less contact between the femoral component and polyethylene at various degrees of flexion and so, more local contact stress.

Conversely, a more conforming polyethylene insert gives a greater area for overall force distribution at the bone-implant interaction thus reducing local contact stress. This conformity constrains the movement of the implant preventing sliding and rotating and produces a higher torque (Kohn and Kusma, 2012) which predisposes the implant to loosen.

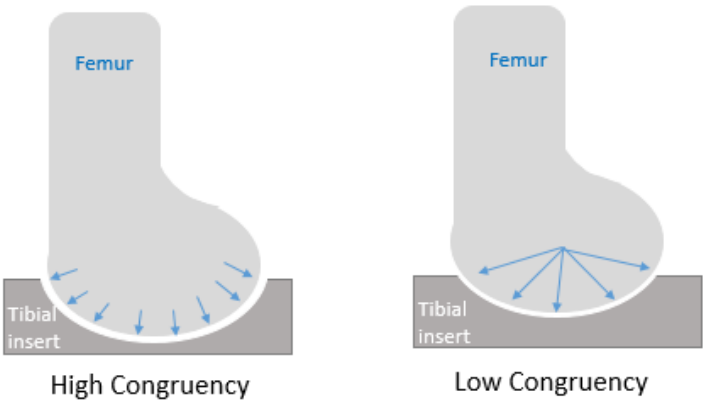

Figure 2-14: Schematic showing an example of high and low congruency polyethylene inserts. Blue arrows indicate contact stress. Stress is greater and concentrated in the low congruency implant, and more spread out in the high.

A natural knee would ordinarily move in a multidirectional manner, and any FB implant limits joint to move unidirectionally. The development of mobile-bearing (MB) implants reflects the efforts made by manufacturers to optimise wear whilst also addressing the complexities of function. The first MB implant was developed in 1977 by Buechel and Pappas (Robinson, 2005). Hypothetically, by maintaining high congruency between the articulating surfaces, local contact stress and wear is reduced. This high congruency would ordinarily result in a decreased range of motion, overcome by allowing the polyethylene insert to slightly translate and rotate. This is crucial considering natural knee flexion involves external rotation of the lower limb.

For both bearing types the operative procedure is the same however, MB implants are more dependent on surrounding soft tissue and ligaments to prevent dislocations, and may be subject to a bearing spin out of the rotating platform (Beverland et al., 2002). Spin out is associated with joint laxity allowing femoral and tibial disengagement and unconstrained axial rotation of the polyethylene, hence for FB implants spin out is less of an issue. The risk of spin out is reduced when there is correct ligament tension and balance particularly during flexion. During dislocation, either the lateral or medial condyle remains in contact with the insert causing the flexion gap to be tighter on one side. Coupled with a shear force, excessive insert rotation is driven by the femoral condyle on the tighter side. In terms of incidence

rates, Capella et al., 2016 reports between 0-9.3% and treatment involves either an open or closed reduction (physical manipulation) to correct the joint.

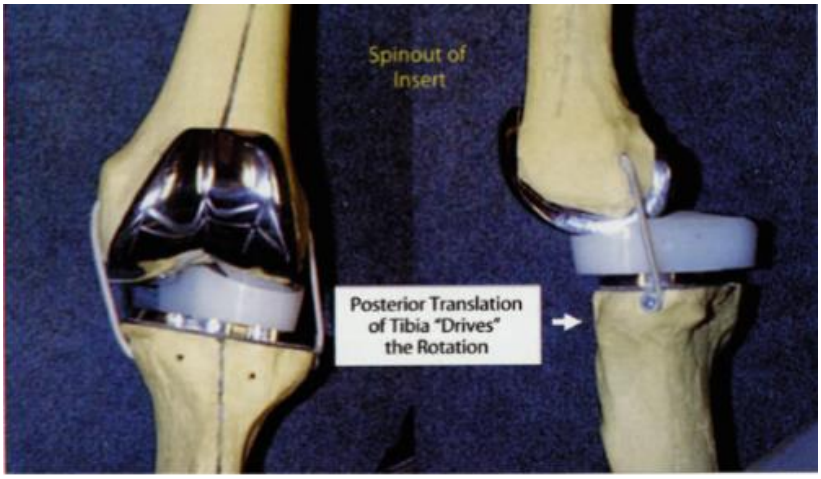

Figure 2-15: Saw bone model showing spin out of the rotating platform (Beverland et al., 2002).

Despite the theoretical advantages of MB designs, there is conflict in the literature regarding significant advantage over FB implants. Tibesku et al., in a randomised controlled study in 2011 looked at sixteen MB and seventeen FB Genesis II implants (Smith and Nephew, Germany) at two year follow up. The cruciate was retained for both groups and the MB group showed significantly better clinical Knee Society scores. However, other results (from gait analysis, or other functional measures such as knee flexion ROM) showed no significant advantages. At a longer, nine year follow-up Poirier et al., in 2015 assessed around thirty MB and FB knees each and similarly found no significant clinical differences between bearing type when assessing ROM or outcome measurements.

A meta-analysis carried out by Smith et al., in 2011 found that from thirteen randomised controlled studies, only one indicated the MB implant gave a significantly better outcome than FB knees at 2-3 years follow up (Kim et al., 2009). The FB implant used was the Medial Pivot fixed bearing prosthesis (Advance^{TM}, Wright Medical, Arlington, TN), this was compared to a press-fit condylar (PFC) Sigma rotating platform knee (DePuy, Warsaw, IN) and 92 patients had one of each implanted. At follow up the FB knee gave significantly lower flexion ROM than the MB: mean difference: 12^{o}. A literature review by Capella et al., (2016) looked at 27 recent papers

and concluded that MB implants showed no advantages in terms of function, radiological outcome, ROM, pain or patient satisfaction to FB knees. In addition to this, MB implants showed an additional risk of polyethylene spinout as well as a greater risk of revision at first year postoperative.

2.4.4. The Columbus Total Knee System

The Columbus® Knee System by B. Braun Aesculap (Tuttlingen, Germany) was launched in 2003 initially as a standard fixed platform which may be implanted using conventional manual techniques or using the computer navigated system the OrthoPilot (Hakki et al., 2013). It now provides a suite of implants including PCL sacrificing or retaining, and fixed or rotating bearing configurations (Figure 2-16). The implant options are interchangeable and provide various degrees of stabilisation or range of motion to allow appropriate treatment of most knee conditions.

Figure 2-16: An overview of the Columbus® TKA products. Image from B. Braun Columbus® Brochure.

Several studies investigated the Columbus knees, either between bearing types or other commercially available implants. A summary of findings is presented in Table 2-1:

Table 2-1: Summary of papers comparing Columbus knee implants. Acronyms in the Prosthesis Used column are as so: CR = cruciate retaining, CR DD, cruciate retaining, deep dish, RP=rotating platform, PS = posterior stabilised, UC = Ultra Congruent, FP=Floating Platform. n = number of knees. KSS = Knee Society Score (standard deviations have been omitted for readability), ROM = flexion range of motion. Significant difference refers to significance between FB and MB knees.

Authors	Prosthesis Used	Bearing	n	Follow up (y)	Pre functional KSS	Post functional KSS	Pre-clinical KSS	Post clinical KSS	Pre ROM (°)	Post ROM (°)	Significant difference
Lampe et al., 2011	Columbus CR	FB	52	1	52.0	88.0	30.0	85.0	111.0 ± 15	113 ± 13	N
	Columbus RP	MB	48		42.0	87.0	29.0	88.0	109.0 ± 12	115 ± 11	
Goebel and Schultz, 2012	Columbus Knee	FB	109	4	/	175.6*	/	/	/	/	Y
	NexGen Full Flex	FB	22		/	183.4*	/	/	/	/	
Jung et al., 2012	Columbus PS	FB	197	2	42.1	79.0	45.3	91.1	128.4 ± 16.2	131.8 ± 10.7	N
	Scorpio PS	FB	187		40.1	80.6	45.0	91.5	124.5 ± 19.9	130.2 ± 14.2	
Hakki et al., 2013	Columbus DD/UC	FB	79	5	44.4	81.8	45.5	89.9	94.7	110.4	n/a
Marques et al., 2014	Columbus CR	FB	45	4	53.7	85.0	29.5	85.1	110.6 ± 15.5	114.3 ± 9.3	N
	Columbus RP	MB	42		42.9	85.0	29.4	87	109.4 ± 12.7	117.7 ± 10.9	
Luzo et al., 2014	Columbus PS	FB	196	0.5	44.1	76.9	/	/	/	/	Y (between pre + postop
Kim et al., 2016	Columbus UC	FB	73	5	34.0	88.3	32.3	93.6	115.5 ± 10.3	125.6 ± 9.1	N
	E.motion UC	MB	73		34.9	86.9	31.7	92.8	114.1 ± 9.6	123.7 ± 9.7	
Lützner et al., 2017	Columbus UC	FB	63	1	42.5	65.3	36.6	86.1	102.9 ± 14.6	112.2 ± 11.8	N
	Columbus PS	FB	64		47.3	64.3	36.0	81.6	102.1 ± 13.1	115.1 ± 13.0	
Yoon and Yang, 2018	Columbus UC	FB	233	8	/	/	54.0	92.0	125.0 ± 13		
Yoon and Yang, 2018	Columbus UC	FB	105	~ 6	54.0	94.0	52.0	95.0	125.0 ± 13	127 ± 8	N
	E-motion FP	MB	95		56.0	95.0	53.0	95.0	129.0 ± 10	120 ± 5	
Fuchs et al., 2018	Columbus CRDD	FB	187	5	49.6	84.0	38.1	85.6	106.3 ± 20.2	114.0 ± 12.1	Y (between pre + postop)

- refers to total knee score.

More recently the National Joint Registry 15th Annual Report 2018 for England, Wales and Northern Ireland gives three and five year survival rates of 98.29% and 97.52% respectively of over 12,000 cemented Columbus implants. The studies above which compared both FB and MB B.Braun implants appear to favour the MB knee overall (Lampe et al., 2011, Marques et al., 2014 and Yoon and Yang, 2018). However, in all studies this was not a significant finding. A study by Jung et al., (2012) showed comparable knee scores and ROM between bearing types, and at two year follow up a disproportionately higher rate (5.1%) of complications were seen in the Columbus PS implant compared to the Scorpio PS implant, and indeed five other studies (range 0.2 – 3.8%).

Whilst Lützner et al., in 2017 found no significant difference in ROM or KSS scores for both Columbus FB implants at follow up, the patient-reported Oxford Knee Scores were significantly better in the UC group than the PS, also the UC TKA procedure was significantly shorter than the PS TKA by 7 minutes as no cam/post mechanism was required. Luzo et al., in 2014, assessed ~200 navigated TKAs with the same posterior stabilised Columbus implant and found improvements in KSS between operative states. As determining implant function was not the focus, details about longer term follow up and ROM for instance was not analysed.

A drawback to the above studies is that only functional/clinical scores are given. Questionnaires and measurements like this are often subjective, not as accurate as other outcome measurements, and may be insensitive to small differences (Jenny and Diesinger, 2012). Complication/revision and survival rates are also beneficial to analyse although not included in the table above as not every study reported this parameter. Studies have also investigated other properties of implants such as wear behaviour, implant kinematics, and radiological outcomes such as mechanical femorotibial angle.

Sciberras et al., in 2013 assessed the clinical results and survivorship at five years of 219 navigated Columbus knees (unspecified implant type). Twenty-one

complications were recorded, with five of these undergoing revision surgeries, giving a complication rate of 2.3% similar to the National Joint Registry 2011 report of 2.5%. Radiolucent lines may appear on radiographs as dark lines where x-rays have passed. In TKA the presence of these lines in the early postoperative period have been linked to poor cement fixation into cancellous bone (Guha et al., 2008), usually under the tibial component. Radiolucent line progression is associated with early joint failure as joint fluid or wear debris many enter the bone-component interface. In Sciberras' study fourteen patients initially presented with a radiolucent line and at one-year postoperative this number decreased to five which may relate to the five revision surgeries previously mentioned.

Grupp et al., in 2009 investigated the wear behaviour of two implants *in vitro*. The Columbus CR (FB) and RP (MB) knees were analysed, using a hydraulic wear simulator for five million cycles. No significant difference between the wear rates of both implants was found, however the normalised wear per unit area was significantly lower in the MB knee than the FB. This correlates to a similar study assessing wear in three FB configurations (flat, curved, and dished polyethylene) where it was found that as congruency and contact areas increased, peak contact stresses and wear decreased (Thomas M. Grupp et al., 2009).

This study will analyse knee implants of the following configurations: a low congruency fixed bearing (LCF), a high congruency fixed bearing (HCF), and lastly a high congruency mobile bearing (HCM). The LCF bearing implant is the Cruciate Retaining, Deep Dish implant (CR DD). This is a PCL retaining implant with a slightly more concave meniscal component than the CR-only implant as evident in the sagittal plane in Figure 2-17.

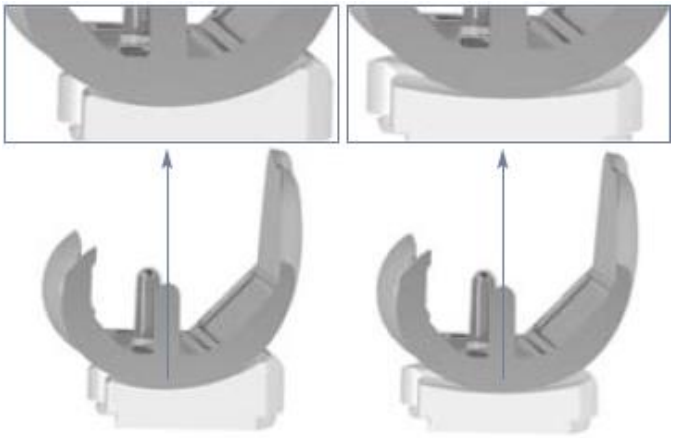

Figure 2-17: Sagittal view of the LCF CR DD implant (left) and the flatter CR design (right).

The HCF bearing implant is the PCL-sacrificing Columbus UC (Ultra Congruent) implant. Contrary to other PCL resecting implants the UC is easier on the femoral bone as no additional resection for a post-cam mechanism is necessary. Instead the polyethylene has a ventrally elevated edge of 12.5mm (Grupp and Schwiesau, 2009, Yoon and Yang, 2018). The anterior polyethylene lift increases the area of articulation, expanding the circumference to stabilise the femur during flexion (Figure 2-18).

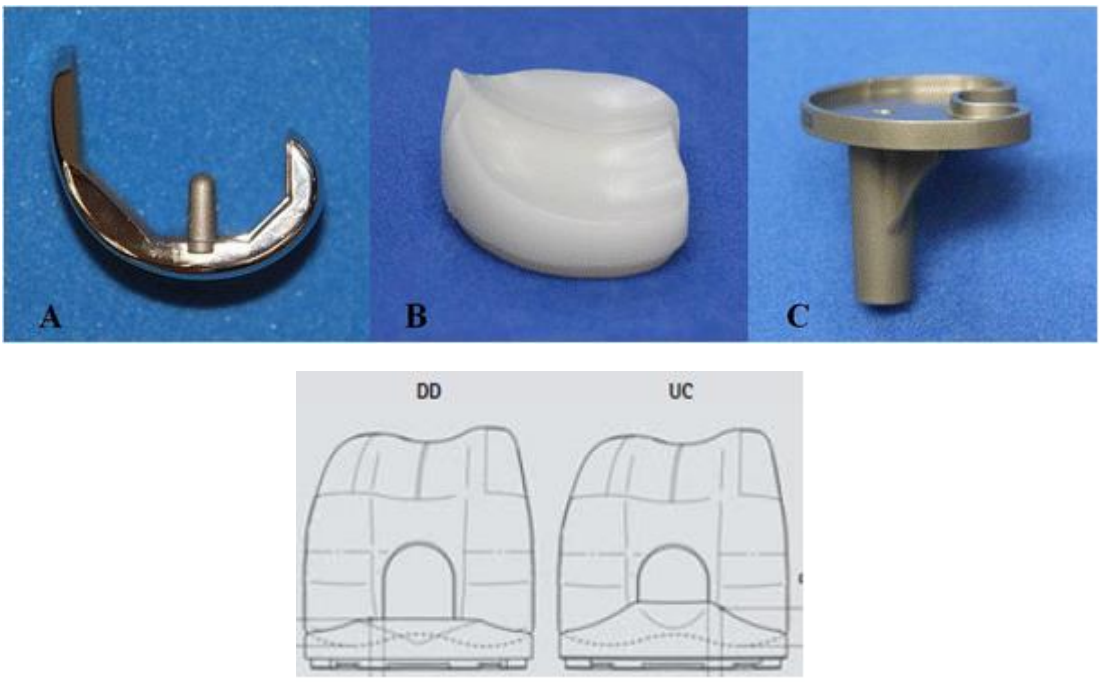

Figure 2-18: Top: The femoral (a), polyethylene (b) and tibial (c) components of the UC implant. Below: a comparison of the UC polyethylene shape to the DD insert (left).

The HCM bearing implant as seen in Figure 2-19 and Figure 2-20 is the Ultra Congruent Rotating implant (UCR). This is a PCL-sacrificing implant which gives axial rotational freedom of ±20°. Rotational movement is facilitated via a curved stop (Figure 2-20) which gives higher stability during rotation. The polyethylene component also has a ventrally elevated edge which gives high anteroposterior stability similar to the UC implant.

Figure 2-19: The Columbus® UCR implant and a depiction of the rotation of the polyethylene.

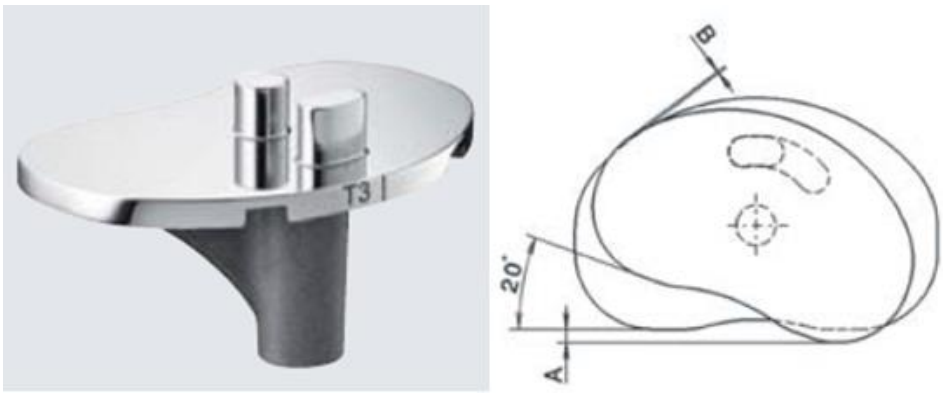

Figure 2-20: The UCR implant's tibial component with the rounded rotation stop (left) and tibial overhang at maximum rotation (right).

The existing literature shows some conflicting conclusions regarding Columbus implants, especially around the benefits of a MB implant. To the author's knowledge, these particular implants are yet to be investigated *in vivo*. Assessing implant function should confirm any theoretical advantages to the MB implant.

2.5. Outcome Methodologies and Technologies

There is considerable interest in determining the success of a TKA to patients, surgeons, and healthcare professionals. As this procedure is high cost and high

volume, understanding the efficacy, safety and effectiveness of an intervention is crucial, whether this be cost or patient satisfaction. Monitoring outcomes is also a useful way of comparing various implants, and surgical techniques and methods.

In TKA there are high rates of patient dissatisfaction, at least 20% of TKA patients are unhappy with their outcome (Baker et al., 2007) and/or experience long lasting pain which impedes on their rehabilitation (Wylde et al., 2018). These patients are less likely to display healthy knee biomechanics due to a reluctance to weight-bear on the operated limb, which may go on to impact the non-operative limb. It is not uncommon for some patients to never regain healthy joint function at postoperative which is clearly a cause for patient dissatisfaction (Benedetti et al., 2003; Jevsevar et al., 1993).

To assess TKA success, outcome measures may be patient-reported and/or performance-based and must be sensitive enough to detect changes between interventions. They should be recorded at least once pre- and post-intervention in order to make a valid within-subject comparison. Outcome measures should also be reproducible and interpretable (Fitzpatrick et al., 1998). Clinically, the common measures used are questionnaires and radiography however, no single tool is suitable to measure outcome alone and using a combination of tools is more reliable.

2.5.1. Knee Scores

There has been a shift in the determinants of TKA success, from radiographic variables and physical examinations to a more patient-centred assessment (Tilley and Thomas, 2010). An example of a patient reported outcome is the Oxford Knee Score (OKS). The OKS was developed especially for TKA patients through interviews with patients, focussing the questionnaire on their needs. The score is generated from the answers from a twelve item questionnaire (Dawson et al., 1998). Each question asks the patient to state how difficult it is to carry out an activity and has five possible answers ranging from “No pain/difficulty” to “impossible/severe pain”. Five questions assess pain and seven assess function and the best outcome score is 0 and 48 is the worst. Some care must be taken when comparing results as previously

scores of 12 and 60 was interpreted to be the best and worst outcomes respectively. The OKS is freely available and widely used due to its simplicity and brevity and because of this, gives high response rates (Dawson et al., 1998). A difference of 5 points between OKS results is thought to be clinically significant (Liow et al., 2003)

Another available score is the Knee Society Score (KSS). This is a widely used instrument to measure the success of a TKA procedure (Insall et al., 1989). It is split knee and functional scores, each providing an outcome between 0 and 100, with 100 giving the best result. This score is subdivided into pain (which contributes 50% of the score), ROM, and joint stability/alignment. As many studies report the KSS score as an outcome measurement, this allows for direct comparison between studies comparing different treatment plans or implants. This is based on the patient's self-report as well as the perception of the assessor.

The Western Ontario and McMaster Universities osteoarthritis index (WOMAC) is another outcome measurement tool, this requires a license to use which is free to obtain. It comprises of a twenty-four-item questionnaire divided into three subscales measuring: pain, stiffness, and physical function. Each question has five possible responses scored from zero to four, and similar to the OKS a better outcome is given by a lower score.

Patient-reported outcomes are commonly used as they can be fast, inexpensive, and easy to carry out although evidence suggests there is a lack of sensitivity to detect functional changes due to their subjective nature (Dowsey et al., 2013). Where an assessor's observation is required in the score, this may show high inter- and intra-observer variation, particularly for less experienced observers. Another consideration is that the timing of the application of the knee score can vary between studies making direct comparisons more complicated (Dowsey et al., 2013). Finally, if a questionnaire is long, this will give greater understanding of a patient's functional ability although completing them can be time consuming leading to low completion rates.

Although it is not feasible to determine the outcome of a TKA solely from a questionnaire, the information gathered is satisfactory for recording patient-reported measures and satisfaction. To fully characterise a patient's TKA outcome scores ought to be complimented with objective measurements to also quantify change in physical function and other information not captured by patient reported measurements.

2.5.2. Objective Measurements

Objective measurements typically give a unit amount which, within a standard deviation of error, should give the same impartial result when repeated. When applied to TKA, these measurements may include the range of motion (ROM) of a joint or alignment measured from a radiograph.

2.5.2.1. Joint Range of Motion

Measuring the change in ROM of the affected knee at pre- and postoperative is an insightful outcome. Motion is typically measured in the sagittal plane (flexion/extension) using a goniometer (Figure 2-21), and motion may be active or passive. Active motion is where the participant carries out the motion independently and passive motion involves an external force to find the maximum ROM. Again, this may be susceptible to inter and intra-observer variability if measured by less experienced observers.

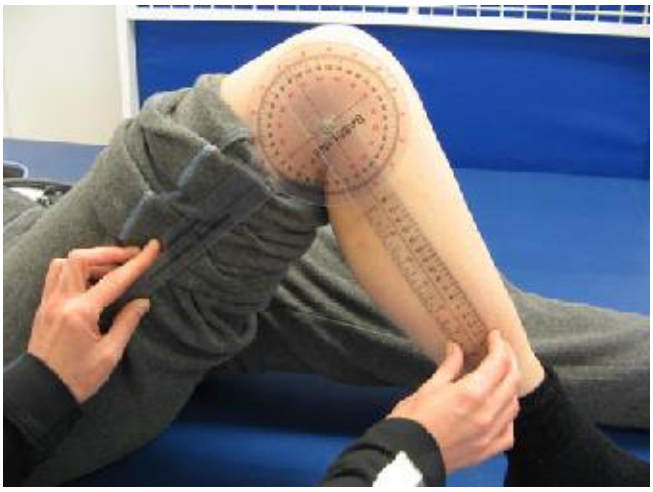

Figure 2-21: Recording knee flexion using a goniometer ("What Is Range of Motion (ROM)?," n.d).

Up to 150^{o} of knee flexion and 5^{o} of knee extension or more would be expected from a healthy person during activities of daily living such as walking or sitting down (Hyodo et al., 2017) . OA may limit motion at one or both extremes of the range and a preoperative TKA patient may experience joint stiffness and/or some degree of flexion contracture deformity giving a limited ROM. Reducing stiffness to give a healthy ROM is a goal of TKA although achieving a healthy range of motion may be ambitious for a TKA patient.

The obstacles to knee extension are often almost completely eliminated or reduced following surgery (Tew and Forster, 1987). For flexion, surgery may also remove the pathological obstacles although these may be replaced by mechanical obstacles in the prosthesis design or soft tissue tension (Tew et al., 1989). Studies researching flexion following a TKA often have somewhat contradictory results, however it can be concluded that knees which had a high preoperative flexion mostly had better postoperative flexion, although others may have lost flexion (Anouchi et al., 1996; Rowe et al., 2005; Sancheti et al., 2013). Whereas those with poor preoperative flexion generally improved at postoperative. Whilst preoperative ROM gives a good indication for postoperative ROM, by itself it is not an accurate predictor of TKA success (Garg et al., 2013) so measuring ROM at both instances is a vital part of assessing improvement following TKA.

2.5.2.2. Radiography

Similar to how lower limb malalignment increases OA progression in patients, in TKA postoperative lower limb malalignment also increases the risk of joint failure. Correct implant alignment is vital in all planes as well as being rotationally matched and small errors in one instance may affect the others. Mis-alignment can alter joint mechanics which may cause polyethylene wear, joint instability and implant loosening (Zihlmann et al., 2005). Long-leg, weight-bearing radiographs (Figure 2-22) are commonly taken to assess alignment. Here, two-dimensional images of the lower limb are taken allowing the MFT angle to be calculated. A disadvantage to radiography is these only show static situations and cannot be used dynamically. Radiographs are used to aid

preoperative planning of implant placement and joint alignment and postoperative radiographs measure alignment and confirm implant positioning (Eckstein et al., 2014).

Measuring alignment from radiographs may also be subject to intra- and inter-observer error depending on experience level (Bach et al., 2001; Bowman et al., 2016). As radiographs are two dimensional, the effects of lower limb rotation on true alignment may be negated causing measurements to be inaccurate. Although a seemingly low intra-observer error within ±2° has been found (Lonner et al., 1996) when implant accuracy ought to be within ±3° of healthy alignment then these observer variations could have a considerable impact on the outcome.

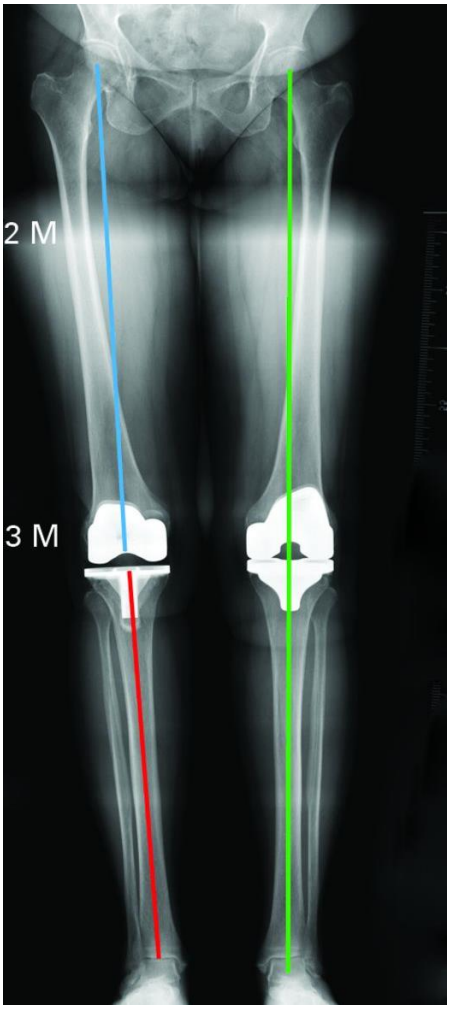

Figure 2-22: Long leg, weight-bearing radiograph of a bilateral TKA patient. Blue and red lines gives the femoral and tibial mechanical axes. Green line shows the mechanical alignment (Oussedik et al., 2015).

Computerised Tomography (CT) scans give three-dimensional information about a patient's anatomy, achieved by compiling sequentially taken two-dimensional images. These are taken with the patient lying supine which may not accurately reflect the weight-bearing characteristics of the lower limb. The reliability and repeatability between inter- and intra-observer is also highly varied (Konigsberg et al., 2013).

Ultimately TKA aims to place the implant so that postoperative joint kinematics is restored to that of a non-arthrosed joint (Dossett et al., 2012). If the implant is placed incorrectly, this will affect motion at the joint, increase the risk of failure, and so, recording lower limb alignment is a key factor in determining the success of a TKA.

2.5.3. Timed Functional Outcome Measures

Timed functional measures are patient-generated outcomes which are performance based and record the functional capacity of a patient. In OA groups, the ability to carry out activities of daily living (ADLs) may be compromised due to pain and disability. As the aim of TKA is to restore function and mobility, recording patients carrying out tests that mimic an ADL can give picture of recovery. When patients carry out these tests they are instructed to do so as they normally would in keeping with safety and comfort of the patient.

The 6 Minute Walk test measures the maximum distance a patient can walk on a level surface in this time. In TKA patients who have mobility issues this may be a maximal test however; any post-surgery improvement should be apparent. In addition, the use of an assistive device is permitted and rest periods may also be taken when required to minimise the burden on the patient (Mizner et al., 2011).

The Stair Climbing test is a more demanding performance measure which records the time taken for a patient to ascend and descend a flight of stairs. Bannister use is also permitted and the test-retest reliability has been found to be high (Rejeski et al., 1995).

The Timed Up and Go test measures the time taken for a patient to rise from a seated position, walk 3 metres, turn, and return to sitting in the chair. The patient is also allowed to use the arm rests for support if necessary. This is a widely used mobility test in older adults and also has high test-retest reliability (Steffen et al., 2002).

Although these tests are largely inexpensive and easy to carry out, measuring the time to carry out a task may not truly reflect the success of a TKA or allow comparison between different interventions. There may also be a discrepancy in patient-reported outcomes and objective outcomes where a patient is pleased with their surgery and is unaware that their functional ability has decreased.

2.5.4. Joint-based Functional Outcome Measures

Where the above outcome measurements are either patient reported or give limited functional data, they also may be subjective or have discrepancies between assessors. To provide objective joint-based biomechanical data one or a combination of the following systems may be utilised:

- Electrogoniometers
- Video Fluoroscopy
- Inertial Measurement Units
- Motion Analysis systems

2.5.4.1. Electrogoniometers

Electrogoniometers (EGs) are an electronic version of manual goniometers routinely used in a clinical setting to measure, for instance, knee flexion in patients. EGs are advantageous to manual goniometers in that kinematics in more than one plane can be measured. They are also unconstrained allowing data to be collected outside of a clinical laboratory setting - useful for recording joint ROM through activities not easily replicated in a laboratory. Furthermore, since the results from EG are immediately available they can be displayed to patients at the time of testing. Measurements using manual goniometry are typically recorded when the patient is lying supine or

seated, which are both non-weight bearing and may not reflect normal joint ROM (Rowe et al., 2001).

There is a potential difficulty with regarding EG attachment to the patient, so that they do not restrict or affect natural movement. The accuracy of EGs is highly dependent on correct placement: since some longitudinal rotation occurs during knee flexion/extension, this may cause readings to have some error since the system assumes motion to be in one plane. To measure multi-axially, multiple EG's must be mounted in different planes. EG devices typically attach via a cuff to a limb segment above and below the joint- as close to the joint axis as possible. Assumptions are made about the skin and underlying bone movement being identical, however soft tissue movement also contributes to a measurement error (Bronner et al., 2010).

Figure 2-23: The electrogoniometry system used by Myles et al., (2002).

EG systems have been validated against motion capture systems in studies investigating the ROM of various joints. Gurney and Kersting (2007) investigated elbow angles measured with an infra-red camera based Vicon (an example of an optical motion capture system further detailed in Chapter 2.5.4.4) and EG systems during cricket bowling and found significant differences between the recorded angles. Cross-talk between the EG axes is the likely cause of this discrepancy,

highlighting the importance of accurate EG placement. Rowe et al., (2001) compared knee flexion ROM during gait in five healthy participants. The data were collected from an EG and a Vicon system (Oxford Metrics, Oxford, UK) and the mean difference between systems was 1.5°±2.8°, showing high similarity and small errors that are acceptable within clinical practice.

Rowe et al., (2000) also used EG to analyse knee flexion ROM in 20 elderly healthy patients during eleven activities of daily living including level, stair, and slope walking, stepping into and out of a bath, and seat ascent/descent. It was deduced that a range of 138° of knee flexion is required to complete these tasks, with entering and exiting a bath requiring the most knee flexion. If using a standard bath is not important then a clinical range of 110° will be sufficient to complete the other activities. Although no comparison was made to a motion capture system, the merits of using EG in a clinical setting was presented and encourages targeting rehabilitation to allow knee function to be safely restored to these values.

Using EG, TKA patients' flexion ROM whilst carrying out the same eleven ADL's as the previous study was also recorded at pre- and postoperative (Myles et al., 2002). Compared to the healthy age matched control participants who reached a full ROM of ~138±7.3°, it was found preoperative TKA patients gave an active average knee ROM of 98.6±14.6°, which worsened to 96.1±13.7° at up to two years postoperative. The change between operative states was not significant. When exiting a bath (the most demanding challenge), the control group showed a ROM of 135.4±14.2°, whereas the patients at pre- and postoperative gave knee ROM's of 74.5±23° and 72.9±19.9° respectively. This was 20° less than the active ROM measured in these groups, and suggests patients are reluctant to use their full potential ROM available to them.

EG's have been shown to be a simple and repeatable method of recording knee function. Although EG is more expensive and time consuming than patient reported questionnaire-based outcome measures, it is considerably simpler and cheaper than conventional motion analysis whilst still reporting insightful data. There is scope for

EG to be used in clinical practice, the previous studies have encountered few technical difficulties, and EG's themselves are found to be comfortable to wear and acceptable to patients.

2.5.4.2. Video Fluoroscopy

Video Fluoroscopy (VF) combines traditional fluoroscopy and video technology to provide *in vivo* motion x-ray of a body part. This technology reveals more information compared to standard static radiographs and is commonly used to assess swallowing. VF has also been utilised in biomechanics to improve TKA by analysing pre- and postoperative knee kinematics during dynamic activities and comparing to a healthy knee. By directly tracking the motion of the bones and implant, errors which commonly occur during other forms of motion analysis are minimised in VF, namely skin motion artefact. In addition, motion of the tibia or femur with six degrees of freedom may be recorded, which are often too small to be detected with other motion analysis systems. As established, knee flexion is more complex than a hinge movement, and better condylar tracking is possible with VF giving the potential to improve prosthesis design (Komistek et al., 2003).

Knee kinematics during loaded and unloaded conditions may be assessed to a high accuracy with VF (Banks and Hodge, 1996; Komistek et al., 2003; Li et al., 2004; Lu et al., 2008). Banks and Hodge (1996) estimated three-dimensional knee position and orientation from sequential single plane, two-dimensional fluoroscopy images (Figure 2-24). Early accuracies of 1^{o} for knee rotations, and 0.5mm for sagittal knee translations were reported. Using a single plane image, motion along axes parallel to the image plane show good accuracy, motion perpendicular to the image plane less so. Using modern techniques improved accuracies to 0.1mm and 0.1^{o} (Li et al., 2004) where six degree of freedom (6DOF) knee kinematics were calculated using two orthogonal images during a weight bearing lunge (Figure 2-25). Limitations were evident as motion was recorded as a series of static images rather than dynamically. Also, participants were required to remain still for four seconds whilst images were taken, which may not suit all participants especially those with mobility issues.

Nevertheless, the total time to image the joint was less than thirty minutes, and radiation exposure is thirty times less than a standard CT scan.

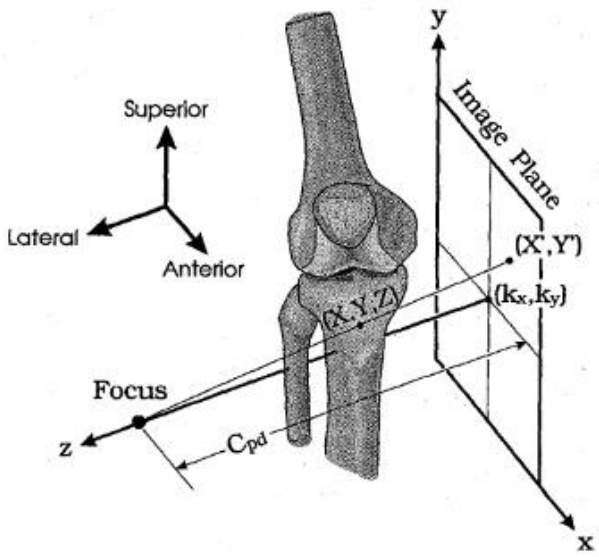

Figure 2-24: Single plane fluoroscopy and conventions as depicted by Banks and Hodge (1996).

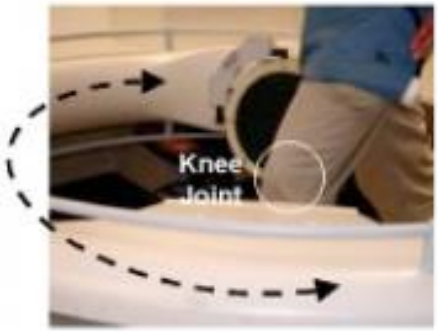

Figure 2-25: Depiction of how two-dimensional fluorscopy images were recorded during the weight bearing lunge (Li et al., 2004).

Lu et al., (2008) compared three-dimensional kinematics during loaded and unloaded conditions in eight healthy participants. Participants were seated and asked to actively extend their knee from a fully flexed to extended position and in the loaded scenario, a 5kg mass was attached to the ankle. The pose and contact points of the femur and tibia were assessed and it was found that at high degrees of flexion (>75^{o}), the loaded condition significantly affected the lateral condyle's contact position on the lateral tibia. This also reduced the asymmetry of surface kinematics between the condyles.

Knee implant designs have also been investigated using fluoroscopy (Bellemans et al., 2002; Dennis et al., 1998; Kessler et al., 2007; Stiehl et al., 1995; Uvehammer et al., 2000). Advancements in VF systems allow motion during other activities such as stair navigation and level walking to be recorded with the use of movable systems (List et al., 2017) as seen in Figure 2-26. These track the knee joint as the patient carries out slightly demanding activities.

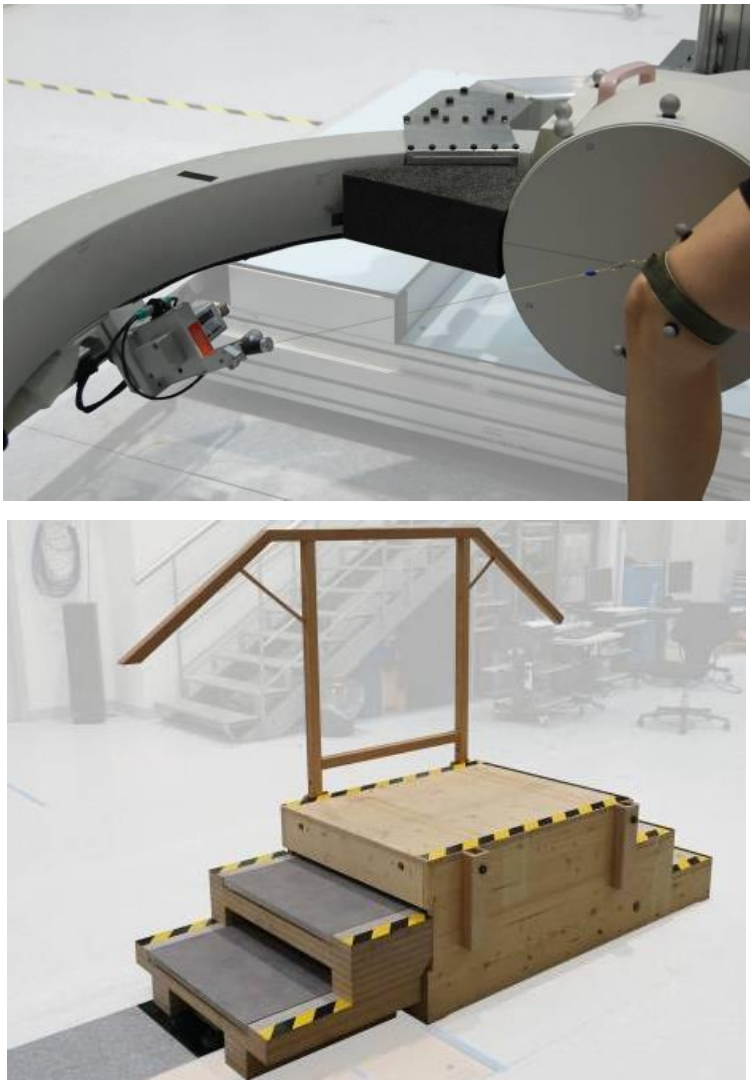

Figure 2-26: Moving fluoroscopy system and instrumented staircase and ramp (List et al., 2017).

Zihlmann et al., (2005) combined force plates with a movable VF system to produce kinetic and kinematic outputs which was compared to a Vicon system. Error in knee moments were shown to be reduced by sevenfold with this method compared to the Vicon system, giving a case to improve inverse dynamic calculations of joint loading.

To record a greater number of gait cycles, Li et al., in 2009 integrated a dual fluoroscopic imaging system and treadmill (Figure 2-27) and successfully produced three-dimensional anatomical knee models. With this, 6DOF kinematics can be obtained accurately.

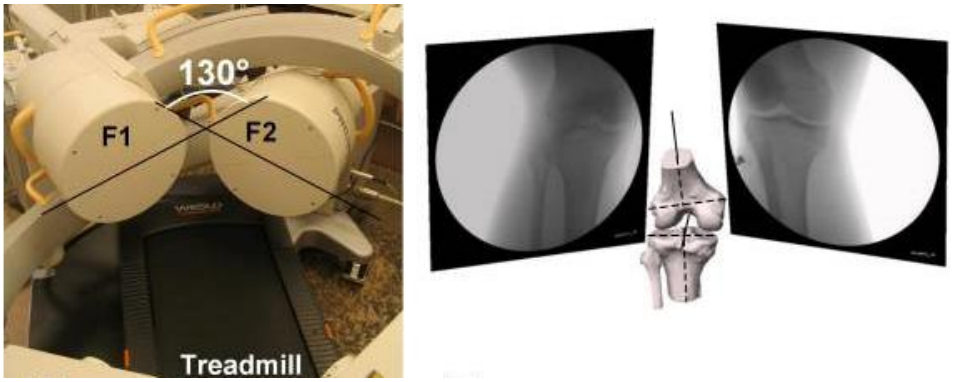

Figure 2-27: Set up of the dual fluoroscopy system and treadmill (left) and virtual knee model reconstructed from the dual fluoroscopy system (right). Image from Li et al., 2009.

A drawback to VF is the limited field of view of the system meaning knee motion is mostly limited to less challenging tasks for a short period of time. An exposure to radiation and use of sophisticated equipment is required which may not be suitable for routine follow up. In addition, only bones are able to be imaged: cartilage, ligament, and meniscus geometry is unable to be recorded which have significant roles in healthy motion. As VF is a direct measurement of bone motion, the accuracy of results is the highest from all motion analysis methods. The findings from VF may therefore be used to improve knee prosthesis designs, or to help increase the accuracy of non-radiographic motion capture methods.

2.5.4.3. Inertial Measurement Units

Another example of wearable devices to measure human motion are inertial measurement units (IMU's) such as gyroscopes, accelerometers, and magnetometers. IMU's have the benefits of being low cost, lightweight, and easy to use, making them ideal for a clinical or even a home setting. Gyroscopes utilise the principles of angular momentum and measure or maintain orientation. Accelerometers measure non-gravitational acceleration but cannot measure rotation and magnetometers are able to measure the direction of a magnetic field in space.

Triads of accelerometers and/or gyroscopes can be placed in an orthogonal pattern allowing motion to be recorded in three dimensions. Three magnetometers may also be included to improve dynamic orientation calculations. Algorithms are then used to estimate sensor orientation and, by comparison, joint angles (Seel et al., 2014). A disadvantage of these units is that they are prone to accumulate error causing a drift to the readings. Either regularly resetting the system or utilising filtering algorithms minimises this error (Tong and Granat, 1999). As movement data is prone to measurement noise, it is common to pass data through a filter to reduce noise (Schreven et al., 2015). Typically, low-pass filters removes high frequency signal components and accurately determining the cut off frequency is vital so as not to filter out actual information, or even introduce artefacts to the data (Sinclair et al., 2013). When assessing movement during more demanding tasks, or movement in those with mobility disorders, the differentiation between noisy and actual data is less clear so using the appropriate filtering methods is crucial.

When compared to a motion analysis system (Findlow et al., 2008; Leardini et al., 2014; Nüesch et al., 2017) IMU's were comparable when recording lower limb kinematics with mean differences less than 5^{o}. This gives IMU's potential to be utilised as part of a clinical or home-based rehabilitation program. This is convenient for the clinician as well as the patient. However, much like EGs, any malpositioning of the units on body segments may hinder system performance and cause human error. In addition, trailing wires and the accumulation of drift are factors that should be improved upon for routine use.

Rahman et al., (2015) investigated the use of IMU's to assess gait in TKA patients as a functional outcome measure. IMU's were found to be repeatable and accurate when measuring knee joint ROM and temporal parameters. However, participants were recorded walking in a straight line over a level surface and it was found that IMUs may not be as effective for more challenging activities. In addition, a full three-dimensional analysis of the whole of the lower limb was not possible (such as hip and ankle angles). However, they were successful in carrying out basic and inexpensive

gait analysis in a busy outpatient clinic which could be the first step to making gait analysis a routine outcome measure.

Although able to record less data than gait analysis systems, IMU's provide a low cost and time efficient method of assessing joint kinematics. Wireless IMU's also allow for unconstrained analysis outside of a laboratory setting. Although IMU's are subject to drift induced by the integration of accelerometer and gyroscope signals, magnetometers are able to correct the drift. A slight disadvantage to magnetometers however is their susceptibility to be disturbed by ferromagnetic materials, often present in hospitals or other clinical settings (van der Straaten et al., 2018). These settings are typically where data capture sessions would be likely to take place in order to assess larger patient cohorts so this should be considered when using this system.

The overall accuracy of IMU's is dependent on senor positioning on the body, complexity of the movement, and the applied biomechanical model. As models and calibration techniques constantly improve to reduce errors and improve usability, with time IMU's may readily become used for rehabilitation purposes.

2.5.4.4. Instrumented Motion Analysis

As discussed, analysing the performance of a TKA patient carrying out various activities of daily living gives a good indication of the success of the procedure. Performance-based tests such as the 6 Minute Walk test, Stair Climbing test and the Timed Up and Go test gives an insightful albeit basic indication of functional improvement. However, these tests are not sensitive enough to measure small changes in the affected joint which would allow quantitative comparison between different interventions.

Motion analysis aims to collect quantitative information about the biomechanics of a body during motion, using kinematic and kinetic data (Cappozzo et al., 2005a). Motion analysis may be carried out with a range of tools and hardware configurations including systems which may be inertial-based, electrogoniometry-based, video-

based (covered earlier), as well as optical and infra-red systems. Processing this data allows for the analysis of certain parameters during tasks, and comparing different parameters highlights any key differences from normal function.

Different measurements systems can be combined to give a comprehensive picture of motion. For instance, in infra-red systems a three-dimensional kinematic model is created by combining skin-affixed marker positions and anthropometric data allowing the participant's motion to be studied. Such markers are retroreflective and placed over palpated anatomical landmarks. The three-dimensional marker coordinates can be tracked and processed to give parameters such as joint flexion angles. Dynamometers (such as force plates) show the functional demands on the lower limb during weight bearing periods and gives kinetic data (external ground reaction forces) when combined with a kinematic model. Dynamic electromyography gives the period and relative intensity of muscle function. Each system serves as a diagnostic technique for one feature of gait. Combining systems provides more information than each individual system such as analysing the relationship between segments and forces with inverse dynamics calculations (Figure 2-28). Selecting which systems to use depends on the needs, costs, and staffing of the clinical or research study.

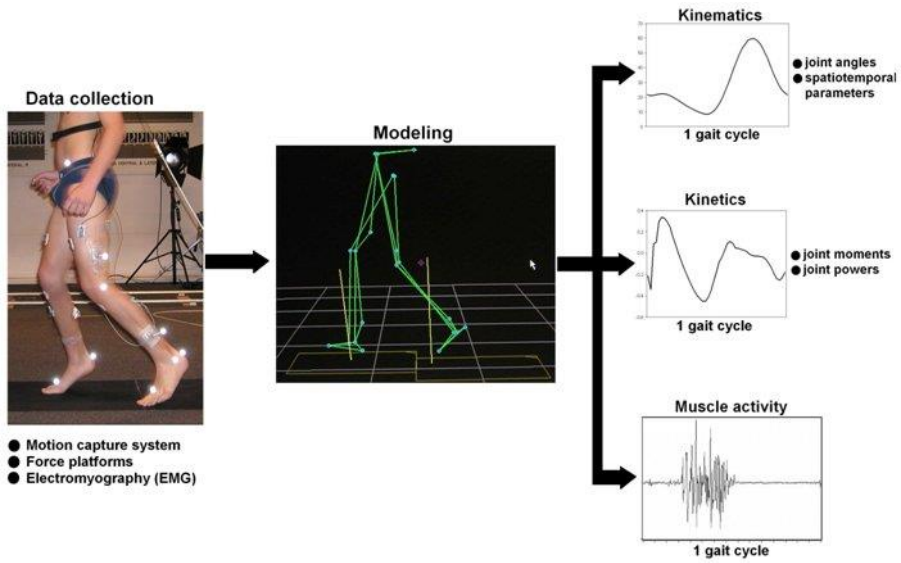

Figure 2-28: Instrumented motion analysis, from the participant (wearing retroreflective markers) and computer model to the output biomechanical data: image from Schweizer, n.d.

Advantages to three-dimensional motion analysis include: the assessment is non-invasive and does not involve exposure to radiation, whole body movement in all planes can be recorded, repetition of movements can be recorded, and large quantities of data can be gathered. Data can also be presented to patients to allow progress to be visualised which may be a motivating factor towards rehabilitation.

To obtain quantitative information about the body during motion, an anthropomorphic model is created. From this model, quantitative descriptions of locomotor functions can be calculated or estimated including joint kinematics, load transmission across or between body segments, and muscular work. The model consists of a kinematic chain of links, where each link represents a body segment. In reality these segments are made of bony and soft tissues. While the bony part is considered non-deformable, soft tissues may or may not be considered as such. Most of the literature chooses to consider the whole segment as a rigid body, making analysis straightforward. However by ignoring the deformability the data is affected by inaccuracies (Alexander and Andriacchi, 2001; Lucchetti et al., 1998).

Until soft tissue deformity is accounted for accurately in human movement modelling, motion analysis currently assumes the body is composed of rigid body segments. To ensure that the skeletal system is reconstructed realistically, and that kinematic calculations are correct, morphological data of each segment is required. This is given by representing the segment as a series of points relative to an orthogonal set of axes known as its local coordinate system (LCS). The LCS may be also represented with respect to an arbitrary, global coordinate system (GCS). When given both coordinate systems, calculation of the position vectors of the segment and transformation of vectors or coordinates from the LCS to the GCS, or vice versa is possible.

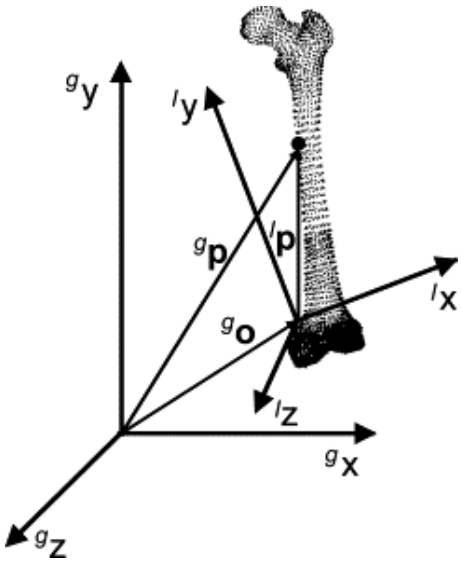

Figure 2-29: A position vector of a point (p) shown in both the global and local coordinate systems, labelled ^{g}p and ^{l}p respectively. (Cappozzo et al., 2005b).

In motion capture, when assessing multiple segments (such as the lower limb), each position vector and LCS is collated relative to the GCS. The GCS is given from the coordinates of set markers and is defined during calibration of the system. In human movement analysis any example of a local orthogonal coordinate system follows a right-hand rule, for instance where the X-axis points forward (in the direction of progression), the Y-axis points vertically upwards, and the Z-axis pointing right by convention. As different systems measure various biomechanical properties (such as dynamometers and marker positions), all coordinate systems should be represented by the same system to interpret results. This requires determining the position vector and orientation matrix of all secondary coordinate systems relative to the GCS. The relationship between the GCS and subsequent LCS is:

$$P_g = [R]_g P_l + O_g$$

Where P_g is the calculated position vector of point P in the GCS, $[R]_g$ is the orientation matrix defining the orientation of the LCS relative to the GCS, P_l is the position vector of the point to be transformed in the LCS, and O_g is the position vector of the origin of the LCS relative to the GCS (Cappozzo et al., 2005b). Recording segmental movement during motion analysis requires the creation of LCS's. This requires the identification of position vectors of a segment, which ought to correlate with

anatomical landmarks to make the process repeatable. For this reason, superficial, bony landmarks identified by palpation are what markers are affixed to for motion analysis. Internal landmarks are often estimated from the location of superficial landmarks and predictive models (Davis III et al., 1991).

Retrieving kinematic data about segments under relative motion requires information from at least two joined segments, one proximal and one distal. This describes relative motion in terms of segment orientation and position of one to the other with respect to a global coordinate system. When combined with dynamometers kinetic data is given such as intersegmental joint moments by using inverse dynamic principles described in Appendix 1.1.4 (Cappozzo et al., 2005b).

2.5.4.4.1. Biomechanical Models

Many different biomechanical models are available to carry out three-dimensional optical motion analysis. Variations in the models can arise due to different marker positions and configurations, calculations for joint centre and axes (predictions based on regression equations or functional approaches) and even data processing may have slight effects on the resulting outcome measures. The most validated model in current clinical use is the Plug-in-Gait model (PiG), a variant of the conventional gait model (CGM). A systematic review of the repeatability of models carried out by McGinley et al., in 2009 saw fifteen out of twenty three papers using variant of the CGM (such as the Gage/Helen Hayes/Davis/Newington/Kadaba/Gilette models).

The origins of the CGM along with the calibration to give estimated joint centres and use of inverse dynamics to estimate joint moments may be traced to the early 90's (Davis III et al., 1991; Õunpuu et al., 1996). The current PiG axis conventions and calculations to determine joint centres are given in Appendix 1: Plug in Gait Biomechanical Model and it should be noted that the origins of widely used scaling equations were not determined from a diverse population, and so estimates of particularly deep internal joints (such as the hip) position performed badly when compared to functional methods in healthy populations (Sangeux et al., 2014, 2011).

PiG is also considered to have over-simplistic foot and upper body modelling, and inadequate compensation for soft tissue over pelvic landmarks (Baker et al., 2018), however the latter issue is not unique to PiG alone.

In the CGM and PiG the single point markers used are susceptible to significant errors on biomechanical data. The overall effects of marker misplacement by 5mm onto static joint kinematics are described in Table 2-2

Table 2-2: Summary of effects on static outputs seen when as a result of marker misplacement. Changes of less than 0.1° are left blank (Baker et al., 2018). X Y Z refers to flexion/adduction/internal rotations respectively. Positive values refer to flexion, adduction, and internal rotation.

Marker moved	Pelvis (°)			Hip (°)			Knee (°)			Ankle (°)		Foot (°)
	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z
RASI up	-0.9	1.4	0.1	-1.2	1.8	-0.5	-0.2	-0.4				
RASI laterally	0.4		0.2	-0.1		-0.3	-0.2	-0.1				
RKAD int				-0.5	-0.1	2.8	-0.9	1	-0.1	-0.1	-2.7	
RKNE up							-0.2	-0.1	0.4	-0.2	-0.4	0
RKNE ant				1.3	0.1	-1.8	2.2	-0.9		0.8	1.9	0.1
RANK up										0.1	0.1	0.1
RANK ant							-0.9		-0.1	-1	-1.1	0.1
RTOE out										0.1	-4.6	4.7

The static errors seen above considerably magnify for dynamic angles during gait, and changes up to 25° have been reported (Szczerbik and Kalinowska, 2011) with the largest change in ROM in the frontal and transverse planes (Kadaba et al., 1990). Another study similarly found large non-sagittal errors associated with knee marker misplacement: up to 7.59° and 5.17° per 10mm of displacement in the transverse and frontal knee planes respectively (Osis et al., 2016). This is due to a shift in the segment coordinate systems causing motion artefacts where rotations in one plane are accounted for in another, known as kinematic cross talk. The clinical effects of a posterior knee marker misplacement may present knee hyperextension, internal hip rotation, and external ankle rotation and vice versa with anterior knee marker placement. Antero-posterior displacement of the thigh (or shank) markers are large contributors to cross talk error (Baudet et al., 2014; Schwartz et al., 2004). The effects of thigh marker offset between ±15° is shown below in Figure 2-30.

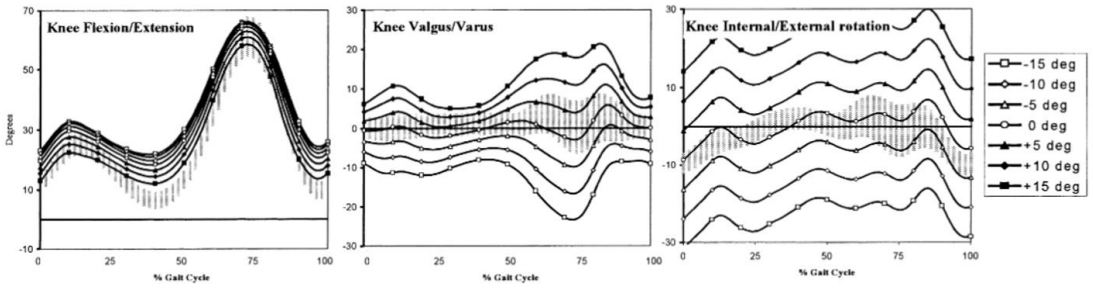

Figure 2-30: The changes in three-dimensional knee kinematics with change in thigh marker position (Baker et al., 1999).

As expected, the change in knee kinematics is opposite when the thigh rotation offset is positive to negative. Anterior thigh misplacements bring the knee flexion axis rotated internally which translate knee flexion traces upwards, and frontal and transverse rotation shift downwards towards abduction and external rotation. Hip rotation angles also have an internal offset. The spread of data is largest during stance in the sagittal plane, during swing in the frontal plane and transverse figures have equal spread throughout. Effects of the tibial/shank marker misplacement are less elucidated in the literature, although smaller changes in knee kinematics are seen, and generally affect longitudinal rotation (Wen et al., 2018).

To minimise the errors associated with thigh and shank marker misplacement with PiG, a Knee Alignment Device (KAD) may be utilised during static calibration. The KAD gives the plane of the KJC so accurate thigh and shank marker is less vital, further description is given in Appendix 1.1.1. No studies have directly compared gait data recorded with and without the KAD, however one study found no noticeable differences in sagittal kinematics and sagittal or frontal kinetics when mixed-ability assessors placed the KAD on participants (Stout et al., 1996). This is despite the change in offset angles compared to the most experienced assessor ranging between 0.2 – 17.4^{o}. The authors concluded transverse plane kinematics are considerably affected by KAD misplacement, particularly reflected in altered hip rotation plots which may be checked then remediated during a participant assessment if necessary.

Where PiG and the CGM use a linked hierarchy of joints, a common alternative is a six degree of freedom model, Calibrated Anatomical Landmark Technique (CAST) (Figure 2-31) where each segment is tracked independently (Cappozzo et al., 2005b).

Figure 2-31: Example of CAST model where rigid marker clusters are worn on each of the thigh, shank, and feet segments (out of view is the pelvis segment here worn posteriorly).

Here, clusters are placed onto each relevant segment, and their relative position to anatomical landmarks captured during calibration. As dynamic trials are carried out, the pose and orientation of each cluster (representing the segment it's attached to) is recorded and based on its known distance from anatomical landmarks kinematic data is given. This is different to single marker models where segment embedded axis systems are determined, as it is assumed that the cluster markers never move independently from the segment, the three or more cluster markers calculate reference axes within each cluster. Theoretical advantages to CAST are that clusters can be placed so as to reduce soft tissue artefacts and reduce inter-subject variability (Stief, 2018), and by using medial markers during static calibration the knee joint axis is better defined. However, this can be incorporated to non-CAST models easily.

Direct comparisons between PiG and CAST model outputs gives similar findings where, for out of sagittal motions, the CAST model shows a lower variance than PiG. Variation in frontal knee ROM of 14° and 22° for PiG and CAST respectively has been

reported (Duffell et al., 2014) as well as high sagittal correlation (coefficient of multiple correlation (CMC) >0.95). The authors noted that use of a KAD to reduce PiG variability could have been employed. A study comparing five marker sets (Ferrari et al., 2008) presented good correlations in sagittal data, and worse correlations for non-sagittal rotations. The authors presented correlations between PiG and CAST, and correlation coefficients for most kinematic and kinetic parameters were close to 1, transverse hip and knee CMCs were particularly low (reaching less than 0.01), frontal knee angles showing CMCs of -0.55 and -0.303 for the right and left limbs respectively.

Whilst not assessing gait, a study comparing CAST and PiG during a double leg drop jump similarly found large differences between the models. Most prominently in the frontal (R^2 = -0.28 (right knee), -0.02 (left)) and transverse (R^2 = 0.56 (right), 0.58 (left)) planes. Again, good sagittal agreement was seen (R^2 >0.95 for both limbs). A similar study assessing the agreeability of six different models during landing showed non-sagittal knee kinematics showed the largest differences across marker sets (Kerkhoff et al., 2020), however no numerical or statistical information was provided, a visual inspection of the graphs shows the peak frontal knee angles were considerably varied from ~12º in PiG and ~4º in the 6DOF cluster model.

A final study assessing repeatability found that apart from pelvic tilt, all segment and joint angles showed high intra- and inter-session repeatability for both CAST And marker (Helen Hayes) models (Collins et al., 2009) where CMC >0.8. Whilst CAST overcomes some of the limitations from marker systems some problems associated with skin artefact and landmark definition persist – particularly the optimal location for thigh clusters.

Inverse kinematic models have also been reported (Charlton et al., 2004; Reinbolt et al., 2007), here an optimisation technique (optimised lower-limb gait analysis- OLGA) is used to fit marker positions to a predefined linked segment rigid body. Much like CAST no single model has been widely used or validated, and as this a modelling method, artefacts in the data may arise which may be difficult to source and correct.

Charlton et al., found reduced kinematic cross talk effects however, only sagittal data were recorded and the reduction in cross talk effects was based on an observed difference of 6^{o} greater knee flexion ROM in the optimised model compared to CGM. Another study assessed whether OLGA could reduce errors brought about following 10mm marker misplacements (Groen et al., 2012). Comparing the root mean square of the differences between unadjusted, then adjusted markers showed that OLGA significantly reduced the RMS for the hip, knee, and ankle in most planes. OLGA reduced errors due to thigh and knee marker shift, but less so for the ankle maintaining adequate marker position is fundamental for accurate 3D motion analysis.

Ultimately, disadvantages to biomechanical models (such as repeatability, joint centre and axis estimation, and the effects of STA) will apply to each model to some degree. Often assessor skill and experience will determine data quality. Improving calibration processes and standardising post processing techniques will allow for more direct comparison between models. Ultimately any model that assume a superficial object reflects motion of underlying bone will still show an absolute error (Barré et al., 2013). Determining the parameters to be analysed with the participant group in mind can affect model choice. It is likely a combination of techniques which aim to be participant specific that should be used to develop and improve biomechanical models further.

2.6. The Gait Cycle and Analysis

Examining a patient's motion through an activity reflects any restoration or worsening of function and one of the most common activities to measure is walking. Clinically, gait analysis gives the degree to which a patient's gait has been affected by an existing disorder, meaning it is used as an evaluation tool rather than a diagnostic tool (Davis III et al., 1991). The following describes key events during a gait cycle for various activities and gait features that can be used for further analysis.

2.6.1. The Basic Gait Cycle

The gait cycle starts at the point of initial contact with the ground and ends with the second ipsilateral foot contact. During a cycle there is an ordered sequence of events which allows the gait cycle to be divided into phases. To start, the cycle can be divided to when a limb is in contact with the floor (stance phase), and when it is not (swing phase) as depicted in Figure 2-32.

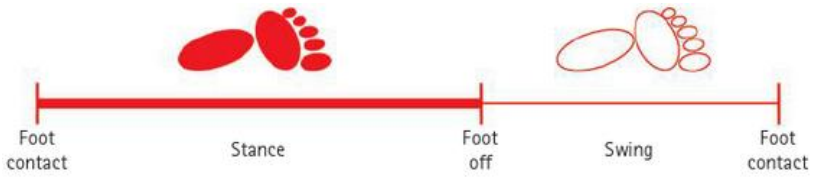

Figure 2-32: A single stride divided into its stance and swing phases (Baker, 2013).

Stance phase takes up approximately the first 62% of the gait cycle with swing phase taking up the remaining 38% (Gamble et al., 1994). Stance phase is where the limb provides support for the contralateral limb undergoing swing. In this period the stance limb balances the weight of the body whilst allowing forward progression. During swing the limb undergoes higher knee flexion to raise and propel that limb forwards.

In walking, periods of double and single support are seen (Figure 2-33). During double support, the leading leg has made foot contact and the trailing leg is about to commence swing. In a single gait cycle the body is in double support for around 25% of the cycle (Whittle, 2006).

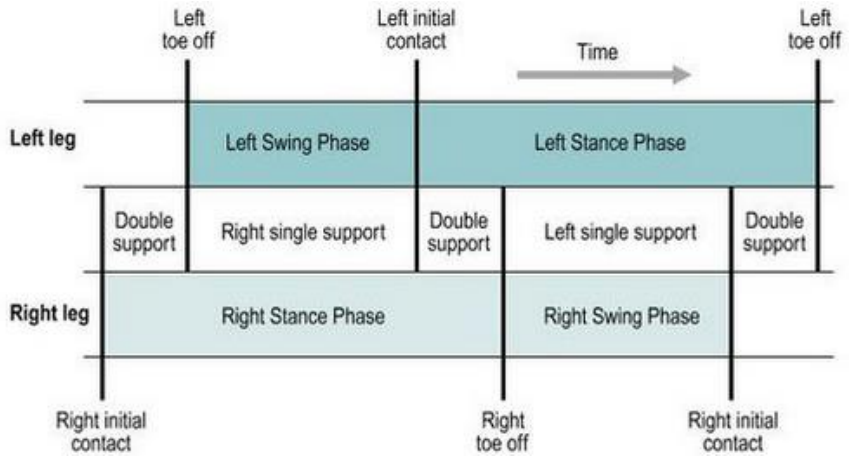

Figure 2-33: The timing of single and double support during one gait cycle of both limbs (Levine et al., 2012).

Further subdivision of the stance and swing phases is possible (Figure 2-34). Stance is commonly divided into four periods: Loading response (following initial contact), mid-stance, terminal/late stance, and pre-swing. The subdivisions of the swing phase are initial swing, mid-swing, and terminal swing. Swing phase has two distinct periods: acceleration to mid-swing and deceleration from mid-swing to terminal swing until the following initial contact.

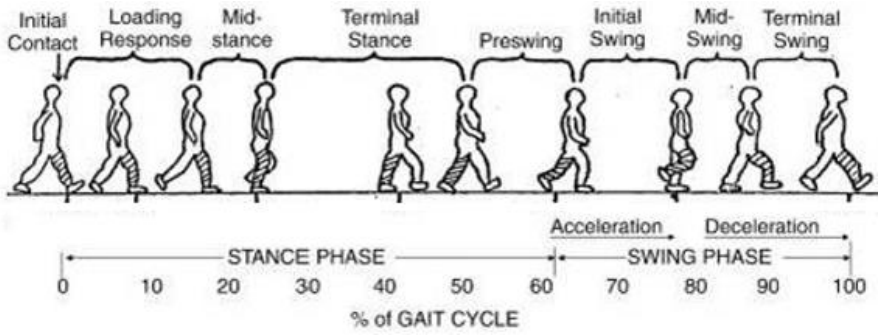

Figure 2-34: The subdivisions of a gait cycle of a single side (De Koster, n.d.).

The events and what happens during each period are as follows (reference here is made to Perry and Burnfield, 2010; and Whittle, 2006):

2.6.1.1. Stance phase

- Initial Contact (0-2%) - the beginning of the gait cycle, frequently known as foot or heel strike and is the immediate onset of weight acceptance onto that limb.

- Loading Response (2-12%) – directly following initial contact, is a period of double support. The loading foot is lowered to the ground by ankle plantarflexion and the contralateral limb is lifted off the ground. The loading limb is placed under various demands: to absorb shock, maintain stability and preserve progression.
- Mid-stance (12-31%) – the contralateral limb begins swing phase and the body is supported by the single ipsilateral loading limb. The body is preparing to propel the body forward and ends when the body weight is aligned over the loading limb.
- Terminal stance (31-50%) - this begins when the heel of the loading limb lifts from the ground. Terminal stance ends when the contralateral limb completes its swing phase and contacts the ground.
- Pre-swing (50-62%) - double support occurs again as the contralateral limb makes initial contact. Pre-swing ends when the primary foot leaves the ground, allowing the load to be rapidly transferred to the secondary limb to allow the ipsilateral limb to begin swing phase.

2.6.1.2. Swing phase

- Initial swing (62-75%) – the period between when the ipsilateral foot is lifted from the ground (i.e., when toe off occurs in healthy gait) to when that limb passes the stance limb. Maximum knee flexion occurs during this phase to allow adequate toe clearance. The contralateral limb is in mid-stance.
- Mid-swing (75-87%) – from the moment the feet are adjacent at the end of initial swing, until the point where the tibia of the swinging limb is vertical and in front of the contralateral limb is mid-swing.
- Terminal Swing (87-100%) – the ipsilateral limb follows on from vertical tibia and completes the swing phase by contacting the ground, starting the next gait cycle.

During the gait cycle, various parameters may be analysed to give an idea of a participant's walking ability. These include spatiotemporal parameters (STP) related

to space and time, such as walking speed, stride length and cadence (Robinson and Smidt, 1981). A description of some of STPs are given below:

- Walking Speed – calculated by dividing the distance covered (in metres/kilometres) by a unit of time taken (often seconds/minutes). A walking speed of around 1.4m/s is expected in healthy adults (Bohannon and Williams Andrews, 2011).
- Cadence –the number of steps taken in a time frame (often per minute). In healthy adults, a cadence of ~120steps/min is expected (Perry and Burnfield, 2010).
- Stride length – The distance (often in metres) between two consecutive heel strikes of the ipsilateral side. To adjust to the surrounding environment stride length is constantly altering, along with walking speed and cadence. A formula relating the three is as follows (Levine et al., 2012):

$$Stride\ length(m) = \frac{Walking\ Speed\ \left(\frac{m}{s}\right) x\ 2\ x\ 60}{Cadence(steps/\min)}$$

  Multiplying by 60 converts minutes to seconds and multiplying by 2 converts steps to strides.
- Step length – the distance in metres between two consecutive contralateral heel strikes, i.e., between a left heel strike and the following right heel strike. Figure 2-35 depicts the difference between step and stride length.
- Stride/step time: the time taken to carry out one stride or a step, usually measured in seconds.

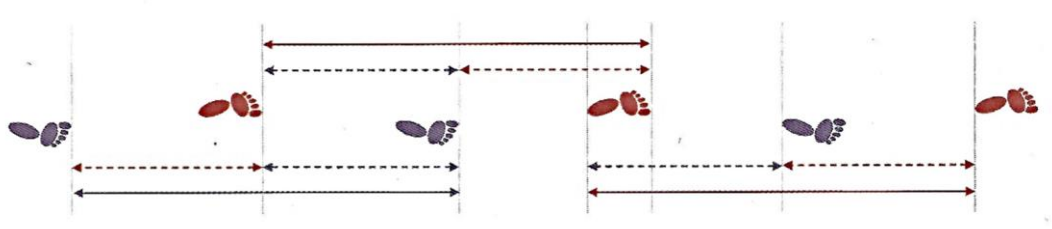

Figure 2-35: A schematic showing stride length (solid arrows) and step length (dashed arrows) (Baker, 2013).

By comparing these parameters at pre- and postoperative, recovery progress may be tracked. Whilst a person may show a near-healthy walking speed for instance, this may be achieved with asymmetrical and ineffective movements of both limbs. Studying joint movements in higher detail may be necessary to obtain a clearer picture of function.

2.6.2. Level Walking

Level walking is the basic locomotor function, aiming to maintain upright stability, propel the body forwards, minimise the impact of the ground reaction force, whilst conserving energy. Raising the complexity of the task (for instance navigating stairs or an incline) increases the requirements from the locomotor system to carry out the task safely.

As well as spatiotemporal parameters, the participant's joint kinematics and kinetics can also be analysed with motion analysis. Kinematic parameters give data about the angles of any given joint during motion, and kinetic parameters use force plate data to calculate joint moments and powers. Where available, kinematic and kinetic data may be given in the sagittal, coronal, and transverse planes. Examples of sagittal joint kinematics and kinetics during a gait cycle during level walking is given in Figure 2-36 and Figure 2-37 respectively. Reference is made to Levine et al., 2012.

Lower limb angles are defined as so: the angle between the femur and tibia is the knee angle and ankle angles are taken between the tibia and an arbitrary line in the foot. Hip angles defined either as the angle between the vertical and the femur, or the pelvis and the femur. During gait each joint undergoes characteristic trends. Overall, the hip flexes and extends once per gait cycle, maximum hip flexion (of ~30^{o}) occurs around mid-swing and maximum hip extension occurs during terminal stance. The knee shows a trend of two flexion and extension peaks. The first flexion peak occurs at the start of mid-stance (around 18% of the gait cycle), then goes towards full extension at the beginning of late stance, and flexes again to a peak (~50^{o} or more) during initial swing. The ankle displays a smaller ROM compared to the hip and knee. Following initial contact, the ankle plantarflexes to bring the foot flat to the

ground during loading response. As the tibia progresses over the foot during mid-stance the ankle undergoes dorsiflexion. Proceeding contralateral initial contact, the ankle reaches a peak plantarflexion angle of around 20^{o} until toe off. During swing the ankle returns to dorsiflexion, then remains near neutral until the next initial contact.

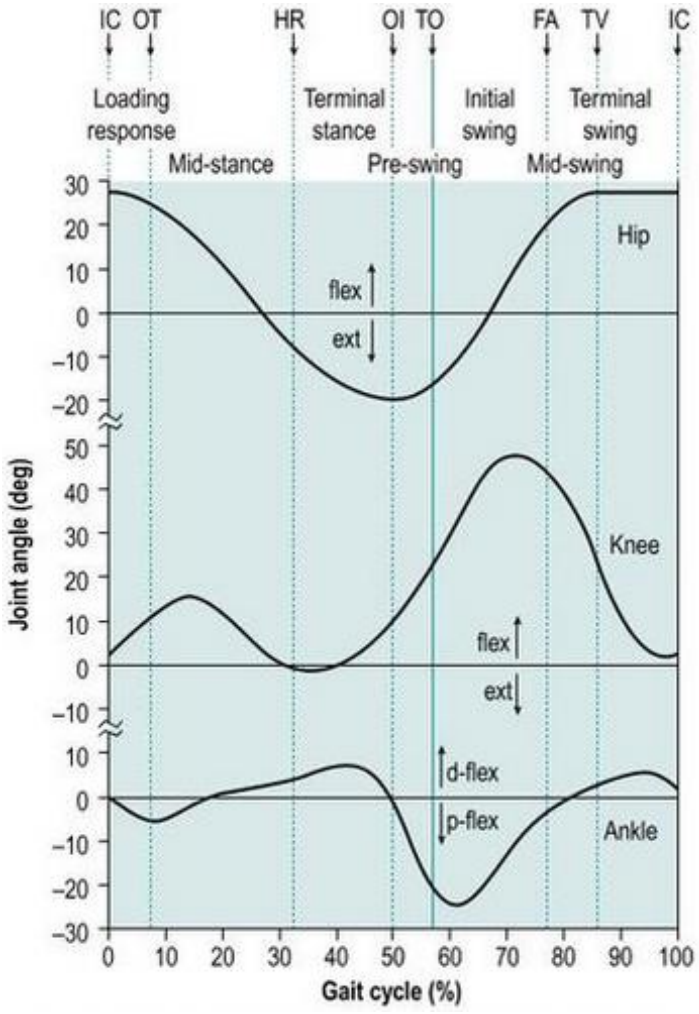

Figure 2-36: Sagittal plane joint angles during a single gait cycle. Flexion and dorsiflexion are positive. Key: IC = initial contact, OT = opposite toe off, HR = heel rise, OI = opposite initial contact, TO = toe off, FA = feet adjacent, TV = tibia vertical. (Levine et al., 2012)

Motion in the coronal and transverse planes act to moderate stability and force transmission to keep gait as efficient as possible. The lower limb is modelled as an interconnected multi-segmental system and changes in one location impacts the joints above and below. During stance, the hip adducts relative to the pelvis, and abducts relative to the pelvis during swing. At the same moment, on the contralateral

side the pelvis will be undergoing a complimentary motion to minimise movements to the torso and above for balance and equilibrium (Dicharry, 2010).

Joint kinetics describe how external ground reaction forces (GRFs), inertia and gravity interact with internal structures acting to stabilise the joint (such as muscles and ligaments). The weight and anthropometrics of the participant allows for estimates of body segment mass and moments of inertia to be made. Along with the relative positions and accelerations of body segments from kinematic measurements the joint kinetics can be calculated, and often are normalised to body weight and/or leg length. Joint moments may be presented as internal moments: referring to the moment applied by the proximal segment to the distal segment, or external moments: which is the perpendicular distance of the GRF to the joint centre (Kowalk et al., 1996). An external moment is matched with an equal and opposite internal moment and care should be taken when interpreting results between authors using different conventions. Joint moments are calculated using inverse dynamics from the centre of pressure (COP) and ground reaction forces (GRFs) both measured using force plates, and joint kinematics. Joint powers are the velocity of a joint moment, demonstrating the work done by the muscles.

Sagittal moments act to bring flexion or extension as seen in Figure 2-37, this depends on whether the GRF is behind or in front of the joint centre (Kirtley et al., 1985). During gait, the hip shows an internal extensor moment (or external flexion moment) from initial contact to halfway of mid-stance, after which they become internal flexor moments until mid-swing. A peak flexion moment occurs at opposite initial contact. From feet adjacent to the second initial contact the moments become internal extensor moments.

The knee produces a biphasic sagittal moment during gait. An internal flexor moment is seen at initial contact as the hamstrings contract to prevent hyperextension following swing. At opposite toe off, the force vector is behind the joint giving an internal extensor moment generated by the quadriceps. Knee extensor moment peaks during mid-stance and declines to a flexor moment during terminal stance as

the knee angle changes from flexion to extension during single limb support. As the knee flexes during terminal stance the force vector is again behind the joint and the moments change from internal flexor to extensor until feet adjacent. To prevent knee hyperextension from the inertia of the shank in swing phase, an internal flexor moment is evident from mid-swing to the end of the gait cycle.

During loading response, the ankle produces a small internal dorsiflexor moment allowing the foot to be lowered to the ground. An increasing plantarflexor moment proceeds, reaching a maximum at terminal stance to oppose the high external dorsiflexor moment as the force vector is at the forefoot. During pre-swing, as the ground reaction force declines, the plantarflexor moment decreases rapidly and falls to zero at toe off, remaining until initial contact.

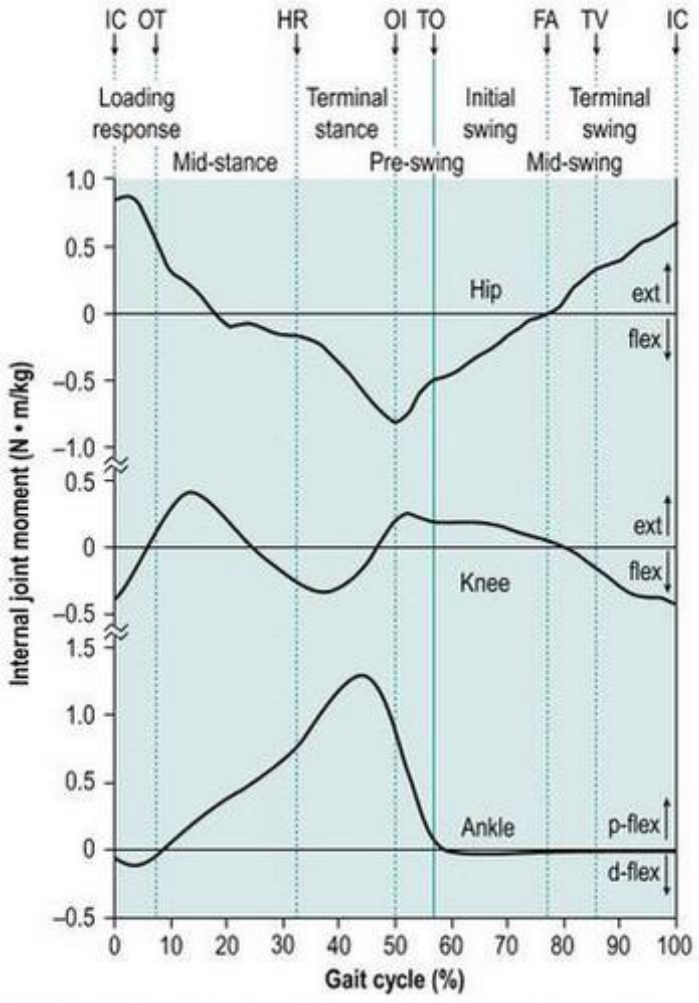

Figure 2-37: Internal sagittal plane joint moments during a single gait cycle, extensor and plantarflexor moments are positive. Key: IC = initial contact, OT = opposite toe off, HR = heel rise, OI = opposite initial contact, TO = toe off, FA = feet adjacent, TV = tibia vertical.

In knee OA many gait parameters are affected, the amount by according to the severity of the disease. Symptoms such as knee pain, joint stiffness, and decreased ROM cause significant adaptations to gait since the individual would tend to avoid loading their affected side with their whole-body weight to avoid triggering further pain. Compared to healthy matched controls, OA patients commonly show reductions in walking speed, cadence, stride length and knee ROM, as well as increases in double support and knee flexion at heel strike (Heiden et al., 2009). Regarding kinetics, OA patients are reported to have smaller external knee flexion moments during early and terminal stance (Kaufman et al., 2001; Smith et al., 2004), and increased external knee adduction moments (Astephen and Deluzio, 2005). Typically the expected biphasic sagittal moment pattern is diminished in OA which

may be either due to quadriceps avoidance or overuse (McClelland et al., 2007). Quadriceps overuse is where a flexion moment is primarily present throughout stance, and a primarily extension moment during stance is linked to quadriceps avoidance (Figure 2-38).

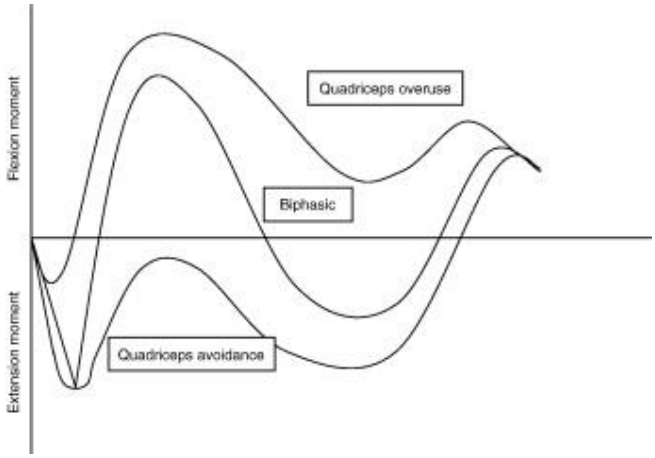

Figure 2-38: Graph showing the variations of sagittal plane knee moment trace: normal (biphasic), quadriceps avoidance, and quadriceps overuse.

Another important parameter is the external knee adduction moment (KAM), a combination of the GRF and perpendicular distance from the KJC in the frontal plane (Figure 2-39). During gait the KAM is thought to correlate with load distribution during stance between the medial and lateral compartments of the proximal tibia. During level walking, forces are not evenly distributed across the knee, around 70% of the load passes through the medial compartment which may contribute to the higher incidence of OA found in this region (Foroughi et al., 2009). In healthy people, the KAM during level walking typically tends to adduct the knee to a varus position (Kutzner et al., 2013). The KAM has a characteristic double peak or biphasic pattern, where the first peak is greater than the second (Figure 2-40) due to increased loading after swing and in OA there is less distinction between peaks (Rutherford et al., 2008). Higher external KAMs relate to increased medial loading relative to the lateral side. A relationship between varus deformities from radiographs and peak KAM has been suggested and in dynamic movement OA patients tend to present larger KAMs than healthy people (Teichtahl et al., 2003).

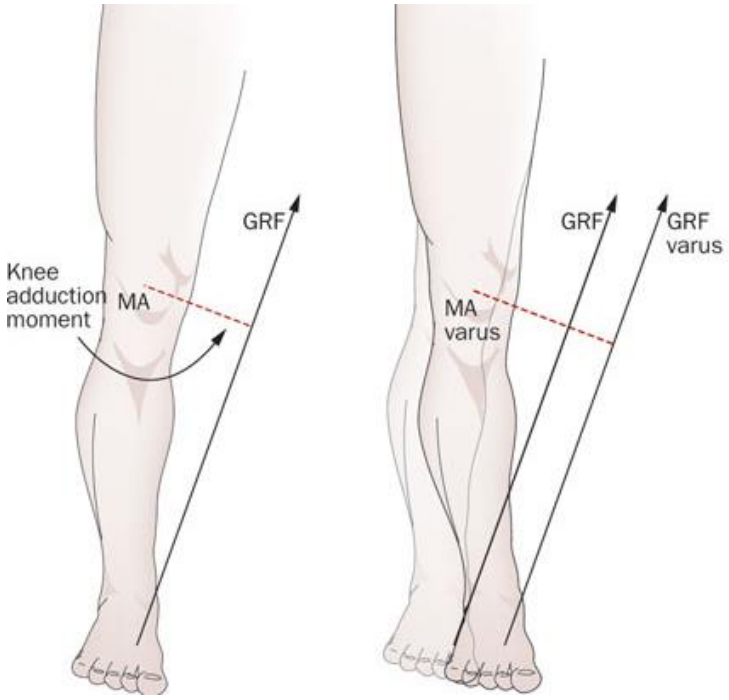

Figure 2-39: Frontal view of a neutrally aligned (left) and varus (right) lower limb and the direction of motion of the external KAM. MA = Moment Arm indicted by a red dashed line. (Reeves and Bowling, 2011).

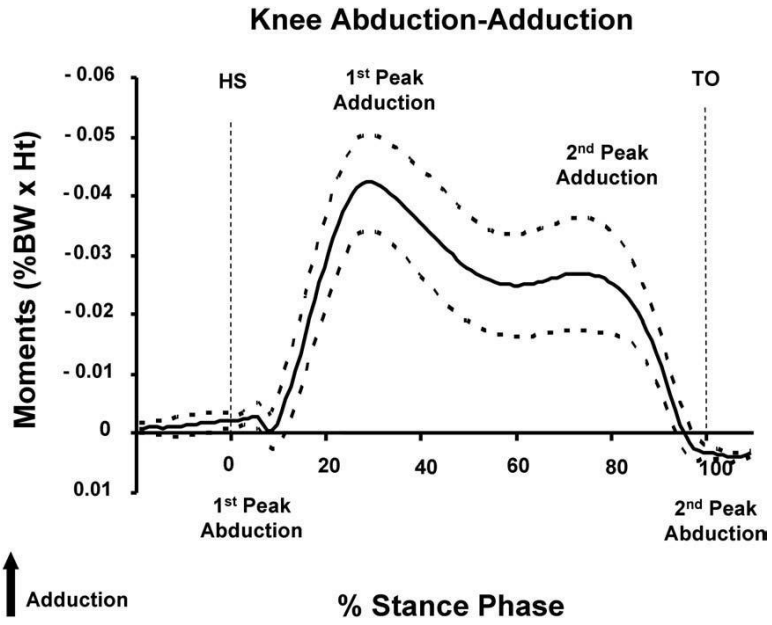

Figure 2-40: External knee adduction moment (KAM) during stance, dashed line indicate ±95% confidence interval (Foroughi et al., 2009).

A considerable factor that affects gait parameters is walking speed. Studies may opt to control for walking speed or allow participants to walk at their natural pace. Walking speed can be set to a fixed pace on a treadmill, using a metronome or simply instructing participants to walk faster/slower than normal for overground walking.

Generally faster walking speeds than normal is complemented by increased cadence and stride length, faster stride times and smaller durations of stance phase and double support. Larger kinematic and kinetic amplitudes are also noted, with the peaks occurring earlier in the gait cycle. Notable affected parameters are sagittal hip angles and moments, knee flexion angle, knee flexion and adduction moments, absorptive knee power, and ankle dorsiflexion angle, moment and concentric ankle powers in healthy participants (Fukuchi et al., 2019; Stoquart et al., 2008). The converse is true for slower walking speeds compared to natural walking. Due to the changes to biomechanics mentioned, using a fixed walking speed across a participant group (particularly, where the group has limited mobility) is not advantageous as this will inherently be too fast or slow for some participants to walk naturally. In able-bodied groups with similar heights and weight, or repeatability studies, fixing the walking speed is advantageous as controlling this variable means any variations in data are due to other interventions.

2.6.3. Stair Navigation

Ascending and descending stairs is a common activity of daily life. This task is more demanding than level walking as it requires a larger joint range of motion (up to 85^{o} of knee flexion) and higher joint moments (Laubenthal et al., 1972). A large amount of energy has to be generated during ascent to raise the body upwards, and absorbed during descent due to the effects of gravity (Riener et al., 2002). Stair descent is considered more demanding than ascent due to high external knee flexion moments which generates due to high knee joint loading (Andriacchi et al., 1980; Draganich et al., 1999).

The gait cycle differs slightly compared to level walking, again there are two main phases: a stance (64%) and swing (36%) phase. Stance phase can be further divided into three sub-phases: Weight Acceptance, Single Limb Support (SLS, or Pull-up for ascent and controlled lowering for descent), and Forward Continuance (or Swing Limb Advancement, SLA). Swing phase can be divided into Foot Clearance and Foot

Placement (Figure 2-41). Unlike level walking, it is usually it is the forefoot that is first to contact the step rather than the heel.

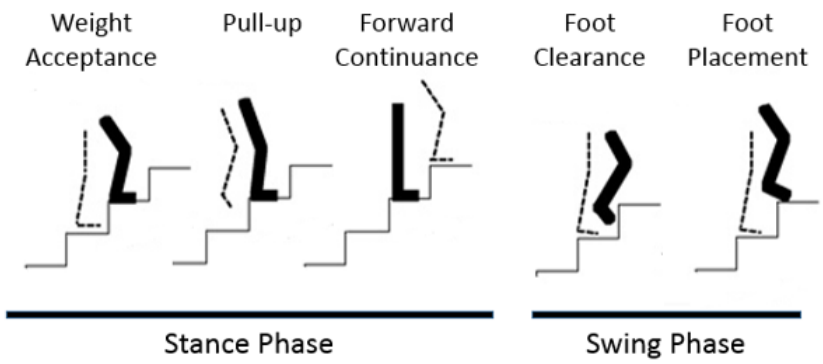

Figure 2-41: The two main phases and five sub-phases of a stair ascent referring to the limb in bold.

Stair ascent gait patterns differs from level gait. Initial contact and weight acceptance begins with the hip and knee at high degrees of flexion, the hip may be between 50-60^{o} of flexion and the knee between 50-70^{o} (Perry and Burnfield, 2010). Both joints promptly undergo extension until swing where flexion is at a minimum (10-15^{o} for the hip and 5- 15^{o} for the knee) to propel the body upwards. During swing, all lower limb joints flex allowing sufficient foot clearance to ascend to the next step, and the hip and knee then extend to prepare for foot contact on the next step. Regarding kinetics, immediately following foot contact there is a small internal knee flexor moment which transitions to a higher extensor moment up till mid-stance to displace the body upwards. This can peak from 0.5Nm/kg to 1.5Nm/kg (McFadyen and Winter, 1988; Protopapadaki et al., 2007; Riener et al., 2002). During the last half of SLS the moments become flexor moments.

The gait pattern for stair descent is more similar to level walking, with a much higher peak knee flexion angle during swing (Figure 2-42). At weight acceptance, energy due to gravity is absorbed as the body is lowered while the ankle dorsiflexes (from 20^{o} plantarflexion to 10^{o} dorsiflexion) and the knee flexes by 10^{o} to lower the upper body. The hip remains stable (~20^{o} flexion) during this time. During SLS, the stance limb controls the lowering of the body while the contralateral side descends the step. The knee flexes and ankle dorsiflexes further and peaks in internal knee extensor moment are given at the beginning and end of SLS. In the second half of SLS the hip goes from

a plateau of movement to flexing as the limb descends. In SLA, knee flexion rapidly advances to a peak of ~90°. Whilst the limb is in swing, hip and knee flexion abruptly changes to extension to reach the descending step (Perry and Burnfield, 2010).

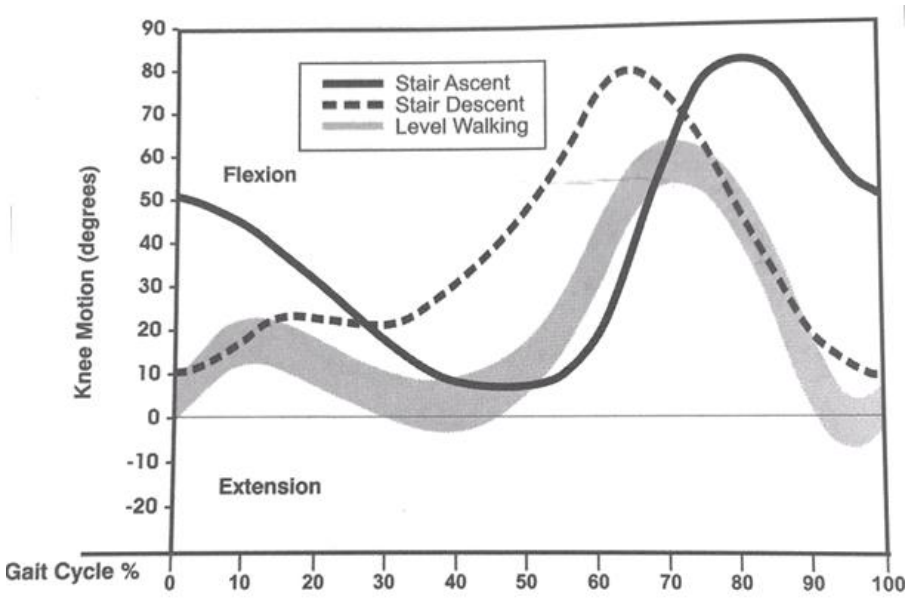

Figure 2-42: Knee flexion during level walking, stair ascent and stair descent. (Perry, 2010).

Compared to level walking, stair walking was found to generate up to three times higher internal knee extension moments (Riener et al., 2002) and around 1.1 times higher adduction moments (Yu et al., 1997) in healthy people. Stair ascent is directly related to quadriceps femoris function, characterised as the external knee flexion/internal knee extension moment (Asay et al., 2009). Reduced muscle strength is common in knee OA and it was noted knee extensor moments were smaller in OA patients compared to controls when carrying out stair navigation (Kaufman et al., 2001).

It is not uncommon for OA patients to use compensatory strategies to complete stair negotiation. Where a healthy person could ascend and descend stairs with a step-over-step fashion, severe OA patients may use a step-by-step strategy where there is a leading and trailing limb in order to reduce pain. In addition, to reduce loading and/or increase stability, OA patients would likely use either one or two banisters or a walking aid when navigating stairs, increasing safety whilst reducing knee moments.

2.6.4. Sloped Walking

Walking on an incline/decline is similar to both level walking and stair navigation as the person walks in a cyclical fashion and the body's centre of mass is raised or lowered with each stride. Work is done either against gravity during slope ascent or is absorbed during slope descent. In addition, the upper body interacts with the lower body to achieve efficient gait. The gait patterns observed for sloped walking are similar to overground level walking with also an increase in magnitude of flexion angles. The amount peak angles increase by relates to the gradient of the slope (McIntosh et al., 2006). Studies have reported that with downhill walking, stride length and walking speed may decrease (Kawamura et al., 1991), while the opposite is true for walking uphill (Leroux et al., 2002; Sun et al., 1996).

Similar to stair ascent, all kinematic parameters and postural changes reflect the need to raise the limb high enough for adequate toe clearance and heel strike to raise the body uphill. Compared to level walking, when ascending an incline, the knee (as well as the hip and ankle) is more flexed at initial contact. Whereas in level walking, and even more so for downhill walking, at heel strike the knee may be near fully extended (Kuster et al., 1995). To raise the foot up an incline, higher hip flexion is required and maximum hip extension occurs around contralateral heel strike (McIntosh et al., 2006). Pelvic tilt also increases as uphill incline increases, which is thought to be related to trunk motion: to maintain forward momentum and to position the body's COM further anteriorly (Leroux et al., 2002). The ankle remains dorsiflexed during stance and the knee shows higher flexion during mid-stance to lift the body up (Lay et al., 2006).

Decline walking, like stair descent, requires more control to lower the body safely, primarily achieved by the knee joint. Compared to level walking, peak knee flexion during stance is higher when downhill walking and often the pre-swing knee extension period is less distinguished or even absent in slope descent. Maximum hip flexion is also reduced, a compensation for the lowered point of contact compared to level walking (Kuster et al., 1995). The posture of the trunk and pelvis shows a

backwards tilt, likely a mechanism to counteract the mechanical effect of the gravity and the downward slope tending to accelerate the body forward and down. (Leroux et al., 2002). Stride length often decreases to act as a brake to the forward momentum of the body.

Few papers report joint kinetics whilst incline walking. Whilst downhill walking, the external knee moments are flexor and decrease through stance (Kuster et al., 1995; Redfern and DiPasquale, 1997). When ascending a slope, Lay et al., (2006) found knee and ankle moment patterns were similar to level walking levels, but the hip extensor moment is significantly greater and the transition to a flexor moment is delayed. During stance, the knee moment is extensor when walking uphill, and when walking downhill higher knee extension moments are required which also influence ankle and hip kinetics, such as generating higher hip extension and ankle dorsiflexor moments (Kuster et al., 1995; Redfern and DiPasquale, 1997).

2.6.5. Total Knee Arthroplasty Gait

Comparing preoperative gait may predict postoperative gait in TKA. When level walking, OA patients have slower walking speeds compared to healthy peers, TKA can increase walking speeds compared to preoperative states but not to the extent of control participants (Andriacchi et al., 1982; Levinger et al., 2012). This may be related to the success of the procedure as walking faster requires greater sagittal motion. Compared to controls, TKA patients show less knee excursion during stance, decreased peak flexion during swing, and altered sagittal plane moments (McClelland et al., 2007; Milner, 2009). This suggests knee mechanics are still abnormal at post-surgery and patients may not recover to the pre-pathological state.

Another outcome of TKA is to restore lower limb alignment to within a healthy range. The knee adduction moment is commonly thought to indicate load distribution to the medial compartment of the knee. Postoperative improvements in lower limb alignment would be indicated by a reduction in KAM from preoperative. Orishimo et al., in 2012 found reductions in peak KAM in TKA patients at six months postoperative, which increased at one-year follow up to near preoperative levels.

This may be due to accelerated polyethylene wear in the medial compartment (Collier et al., 2007). Alnahdi et al., in 2011 looked at frontal plane mechanics of TKA and non-operated limbs in patients compared to age matched control group. The study found throughout stance, the operated and control limbs were more abducted compared to non-operated limbs. A more distinguished biphasic pattern (adduction peaks with a minimum between them) was observed in the operated and control limbs whilst the non-operated limb frontal showed less peak distinction. This may lead to OA progression in the non-operated side.

Instability in the operated knee may also affect gait, and has been reported to be the third most common mode of TKA failure requiring a revision procedure (Chang et al., 2014). Instability is attributed to component loosening, incorrect implant size, implant malpositioning, bone loss or ligament laxity, prosthesis wear or breakage, or periprosthetic fracture. Patient related risk factors include preoperative deformity which required aggressive ligament release or other pathologies such a rheumatoid arthritis or osteoporosis (Rodriguez-Merchan, 2011). Vince et al., (2006) suggests instability may be differentiated by the following types depending on the direction of force: 1) varus/valgus, 2) recurvatum, 3) anteroposterior (in flexion) and 4) global types. Finding the cause of instability determines the treatment. Usually, a revision procedure ensuring appropriate alignment and gap balancing (mediolateral/flexion extension) will be necessary, or even the use of a more constrained implant.

Carrying out motion analysis of TKA patients whilst carrying out demanding activities is important to detect potential; compensatory strategies which would not be as obvious when assessing level walking only. Stair navigation is a common activity analysed in TKA groups and is said to produce the highest knee compression loads and force up to 3.5x body weight (Komnik et al., 2015). Walsh et al., in 1998 also found TKA patients were around 50% slower than healthy controls when stair climbing. Implant types may affect performance in this activity as Andriacchi et al., in 1982 found patients with cruciate retaining implants displayed normal ROM whilst patients with more constrained implants showed a reduced ROM.

3. Thesis Aim

There are limited studies analysing TKA patients walking on an incline, and none comparing implant bearing. Since inclines are common in everyday life and demand unique neuromuscular postural control it would be beneficial to determine the extent TKA restores function in this patient group. Studies in healthy participants show high kinetic demand on the lower limb muscles during this activity compared to level walking. This is relevant for TKA patients where muscle weakness is prevalent which would modify the expected gait pattern.

In order to detect potential compensatory mechanisms in TKA patients, motion of the lower limb ought to be analysed in its entirety as opposed to the knee joint alone. As the lower limb can be viewed as a chain of segments which affect one another, any defects in the operated joint or side may be compensated for by the non-operated joints/limb. For instance it has been shown hip kinetics are frequently increased in TKA patients, and abductor and ankle forces have an effect on the magnitude of knee adduction moments (McClelland et al., 2007). Analysing motion in three dimensions during a range of activities is key to assess implant function and TKA success.

3. Thesis Aim

The literature review described the pathology of knee OA and the use of TKA to reduce pain and disability associated with the disease. Many factors contribute to a successful TKA procedure, such as the surgical procedure itself or implant design. Part of gauging TKA success requires assessing function between pre- and postoperative states. Currently, few studies give detailed knee function, presenting knee scores and questionnaires over quantitative methods which suit busier clinics but are subjective and may not have the sensitivity to detect small changes.

Where available, instrumented gait analysis is currently regarded as the gold star system compared to other quantitative methods such as electrogoniometry or inertial measurement units. This technology gives joint kinematic and kinetic data in three dimensions and has been selected for use in this project investigating the functional performance of three knee prosthesis configurations from the Columbus® Knee System range (B.Braun Aesculap AG, Tuttlingen, Germany). All implants are currently used in TKA surgery and configurations include high congruency knees with fixed (HCF) and mobile (HCM) bearings and a fixed bearing low congruency knee (LCF). The LCF implant was the cruciate-retaining, deep-dish (CR DD) implant, the HCF was the ultra-congruent (UC) implant, and the HCM, the ultra-congruent rotating implant (UCR).

The biomechanical model selected for use in this study was the Plug in Gait model (Vicon), a widely validated and established model. As activities such as stair navigation and incline walking are more challenging than level walking, recording participants carrying out all of these activities should reduce possible ceiling effects. From achieving the aims and objectives an improved understanding of the functional properties of each bearing design should be realised and possibly provide surgeons and manufacturers with information regarding which implant is better suited for certain patients. Other studies comparing implant bearing have reported conflicting results particularly around the theorised benefits to a mobile-bearing knee, and to date, no study has compared the kinematics and kinetics of these Columbus implants.

3.1. Aims and Objectives

This project aimed to use three-dimensional motion capture technology to evaluate the working hypothesis that the mobile bearing B.Braun Columbus® knee had functional advantages to two fixed bearing implants in low and high congruency across a range of activities of daily living. A healthy age matched control group was also included for comparative purposes. The ADLs were level walking, stair navigation and sloped walking. As part of a second doctoral student's thesis with the same population group additional tasks such as sit to stand, sit to walk, and car ingress and egress were also carried out.

To determine to what extent the mobile bearing implant gave better performance to the fixed bearing knees, the mean differences of the biomechanical parameters outlined below between implant groups and to controls was calculated. The degree of improvement between pre- and postoperative for each implant group was also used to determine implant performance. Statistical analysis was then used to determine significance of findings.

The following parameters were analysed to achieve this objective:

- Spatiotemporal parameters: cadence, walking speed, stride time, opposite foot off and foot contact, period of double support.
- Joint kinematics: peak values (knee flexion, adduction, and rotation, and ankle dorsiflexion), range of motion (for pelvic tilt, hip and knee flexion, knee adduction and rotation, and ankle dorsiflexion). Also exported was knee flexion at heel strike and toe off, and maximum and minimum knee flexion during stance.
- Joint kinetics: maximum and minimum hip, knee, and ankle sagittal moments, knee adduction and rotation moments, and peak concentric and eccentric hip, knee, and ankle powers.
- Participant reported questionnaire answers for each activity.

4. Methodology

4.1. Motion Analysis

All data capture was carried out in either a conventional Vicon gait laboratory (Oxford Metrics, UK) or a Motek CAREN system (Motek Medical, Amsterdam, NL) at the University of Strathclyde. In both laboratories optical based infra-red Vicon cameras were used to record the motion of retroreflective markers attached to participants carrying out the activities of daily living (ADL's). The ADL's recorded in each laboratory were as follows:

1) Overground level walking (Vicon and Motek)
2) Stair navigation (Vicon)
3) Sloped walking (Motek)

4.1.1. Equipment

The Vicon laboratory was comprised of twelve Vicon T-series cameras which records motion at 100Hz. Four Kistler (Wintethur, CH) piezoelectric-based force platforms are embedded into the floor that collect kinetic data at 1000Hz. A 10-metre walkway across the capture volume allows activities to be recorded (Figure 4-1). To calibrate the laboratory and record data, Vicon Nexus software version 2.5 was used.

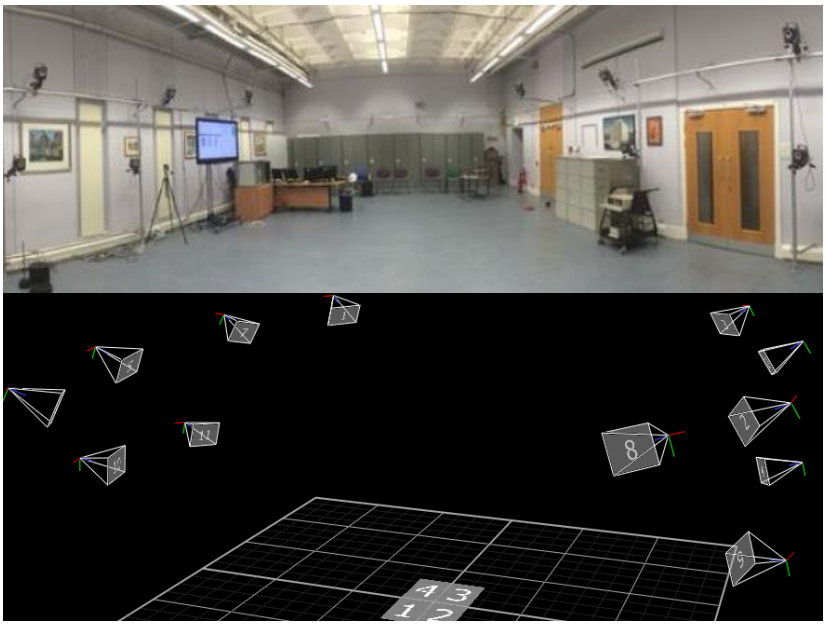

Figure 4-1: The conventional Vicon biomechanics laboratory.

The Motek CAREN system comprised of several components to generate a virtual immersive environment for the user. A six degree of freedom motion platform supported by electric actuators allows translation as well as pitch, roll and yaw rotations along any orthogonal axis. Embedded within the platform is a Motekforce link two metre dual-belt instrumented treadmill. The treadmill is capable of moving at a fixed speed (as determined by the operator), or a self-selected speed generated by participants' anterior-posterior position on the belt. By tracking markers on the participant, a feedback system alters the speed of the belt with the aim of bringing the participant to the centre of the belt. If the participant walks anteriorly the belt speed increased to bring them to the middle and vice versa. Two force plates under the treadmills are capable of collecting kinetic data at 0^{o} incline (Figure 4-2). A 180^{o} screen along with four image generators and projectors creates the virtual environment and twelve Bonita camera Vicon system recorded the participants motion (Figure 4-3).

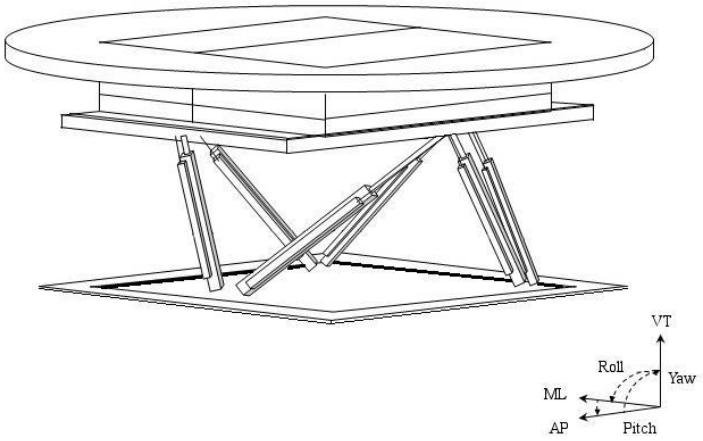

Figure 4-2: The platform, treadmill, and force plates of the Motek CAREN system.

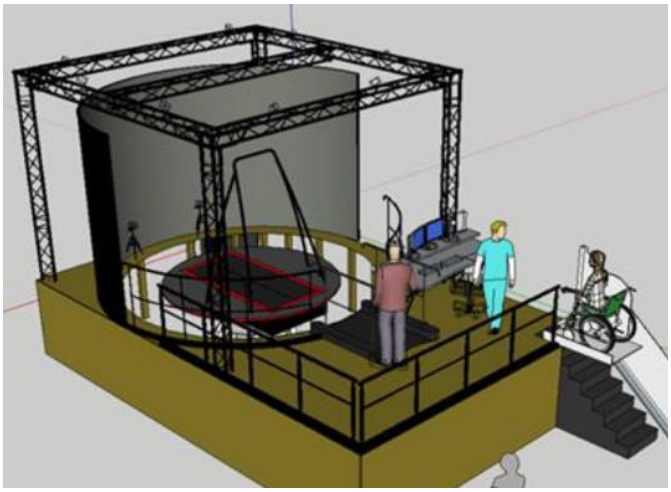

Figure 4-3: A schematic of the Motek CAREN system.

In order for both laboratories to capture motion accurately, each system must be calibrated prior to data capture. A dynamic calibration involves waving a calibration wand with markers of a known, set distance in the capture volume in view of all the cameras. This allows for each camera's position to be related to the capture volume using a direct linear transform which calculates the mapping wand marker positions in each camera's two-dimensional image into three-dimensional coordinates. Any marker within the capture volume may be tracked with high accuracy as long as a minimum of two cameras have the point in their field of view.

To set the origin of the laboratory, a static calibration is carried out using the same calibration wand placed in the origin of the laboratory. Computer software again calculates the relationship between the fixed three-dimensional marker positions of the wand markers, and the two-dimensional positions of the wand markers as seen in each camera's field of view. The cameras are then calibrated with respect to the global laboratory system.

To ensure no large system errors were present at each visit a calibration weight of 200N was placed on each force plate in the Vicon laboratory and vertical force (Fz) readings were recorded. From this, a percentage difference was calculated by (Fz-200N)/200N for each force plate and an average of the differences was calculated. After dynamic calibration of the cameras, the largest image error (in millimetres) of the Vicon and Motek lab was also noted at each session. This was given in Camera

Calibration Feedback in Nexus (under Image Error), a value for how close that single camera's two-dimensional view of a marker compares to the three-dimensional position . A summary of average findings is given in Table 4-1. Errors were low and similar across participant groups with a low standard deviation so hardware errors would be an unlikely source of significant error here. A one-way ANOVA found no significant differences between participant recording sessions.

Table 4-1: Average and standard deviation of force plate and camera errors of all groups in both motion capture laboratories.

Error	Control	Preoperative	Postoperative
Force plate (%)	1.4±0.7	2.1±0.9	1.9±0.9
Vicon cameras (mm)	0.189±0.017	0.215±0.082	0.168±0.027
Motek cameras (mm)	0.248±0.106	0.249±0.109	0.123±0.061

4.2. Ethical Approval, Patient Recruitment and Experimental Design

This study was carried out in partnership with the Golden Jubilee National Hospital, Clydebank. Ethical approval of the study protocol was granted by the departmental ethics committee at the Department of Biomedical Engineering, University of Strathclyde for control participants and the NHS Ethics committee, West of Scotland REC 5 for the patient groups (Appendix 5: Departmental Ethics Protocol and Appendix 2: NHS Research Protocol respectively). The patient participants for this thesis were part of a larger study group of patients who had TKA surgery at the Golden Jubilee National Hospital between August 2015 and June 2017 as part of a randomised controlled study (ClinicalTrails.gov identifier: NCT02422251).

Following ethical approval, patient and control participants who met the relevant inclusion and exclusion criteria (Table 4-2) were approached and provided information about the study (Appendix 3: Patient Information Sheet and Appendix 6: Departmental Participant Information Sheet). Control participants were recruited from various community groups (chess/bridge clubs, mature movers exercise

classes), and suitable patients were identified by a member of the direct care team, also a member of the research team:

Table 4-2: Each inclusion and exclusion criteria for potential participants to the study.

Control Group	
Inclusion Criteria	Exclusion Criteria
No pre-existing condition or injury likely to influence performance of test activities. Over 35 years of age. Willing to take part.	Previous lower limb joint replacement procedure. Unable to give written consent.
Patient Group	
Inclusion Criteria	Exclusion Criteria
Primary unilateral total knee arthroplasty. Suitable to have any of the three study implants. Over 35 years of age. Willing to take part. Able to return for follow up sessions. From one of the following NHS Scotland Health Boards: Ayrshire & Arran, Forth Valley, Greater Glasgow & Clyde, Highland, Lanarkshire, or Lothian.	Previous hip or knee replacement procedure if carried out in the previous six months. Unable to give written consent. Unable to attend the movement analysis sessions. Journey time from home to the university in excess of two hours. Previous ankle surgery. Any past neurologic history e.g., stroke, Charcot-Marie-Tooth disease.

Once participants agreed to take part, they gave informed consent and were recruited into the study (Appendix 4: Patient Consent Form and Appendix 7: Control Participant Consent Form). The patient was then randomly and blindly assigned either a LCF (CR DD), HCF (UC) or HCM (UCR) implant by a member of the orthopaedic research team at the Golden Jubilee National Hospital and data capture sessions was taken at preoperative and one-year postoperative. The control group visited the department once. The patient was also blinded to the implant received and implant analysis was also kept blinded until write-up. In each session participants were recorded carrying out various ADLs (level walking, stair navigation and incline

walking) using instrumented motion capture techniques described previously. The study was carried out in keeping with the Good Clinical Practice guidelines and with the ethical standards stipulated in the Declaration of Helsinki.

4.3. Surgical Procedure

All patients underwent routine practice for their TKA procedure. All procedures were completed as standard as per the expertise of three experienced orthopaedic consultant surgeons at the Golden Jubilee National Hospital using the three B.Braun Columbus knee types.

4.4. Study Protocol

Upon arrival to the laboratory, all participants changed into appropriate clothing and footwear for their functional assessment. Anthropometric measurements were recorded in accordance with the full body Plug-in Gait protocol described in Appendix 1: Plug in Gait Biomechanical Model. Thirty-five markers were affixed to various anatomical locations on the body. All participants underwent static calibration firstly using a knee alignment device (KAD) to obtain thigh and shank marker rotation offset angles with respect to the knee axis, then a second static calibration was taken where the KAD was replaced with lateral knee markers (Figure 4-4). When combined with anthropometric data the position of relevant joint centres could be calculated, used to produce kinematic and kinetic data. This model was used to be able to record upper body kinematics during the tasks recorded for the second doctorate candidate's thesis.

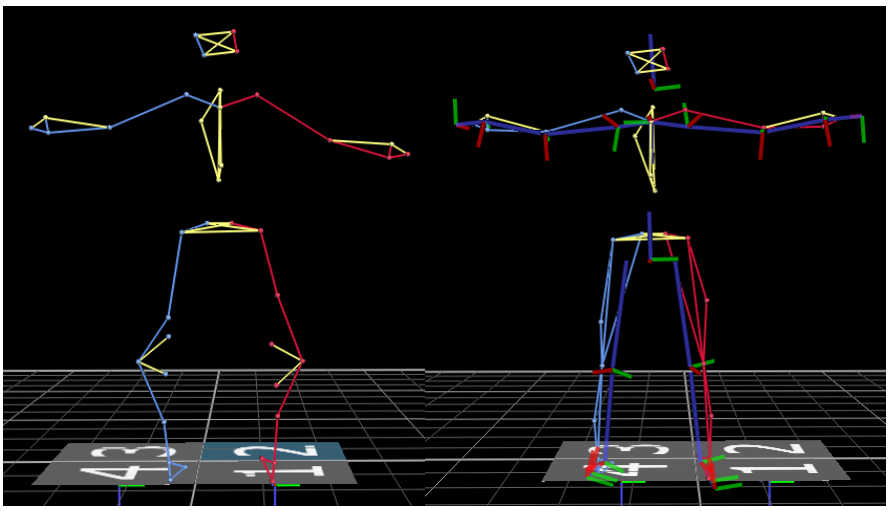

Figure 4-4: Two static calibration still frames, with the KAD (left) and without (right).

Once calibrated, the participant then completed the ADLs where possible. With each activity, notes were recorded in a case report form (CRF) given in Appendix 8: Case Report Form. After each activity questionnaires about the task were also asked and the responses were also recorded in the CRF.

1. Overground level walk

Participants were instructed to walk from one end of the Vicon laboratory to the other at a comfortable walking speed. The participant starting position was adjusted to increase the likelihood of the participant striking the force plates in the centre of the walkway cleanly. Clean strikes were where the whole of a foot strikes a force plate, and the contralateral foot strikes either a separate force plate or the floor next to the ipsilateral side. Participants were not told about the force plates to avoid potentially altered gait due to the effects of targeting. A minimum of six trials were recorded if the participants were able to walk that distance, enough to account for inherent variability without fatiguing the participant. Extra trials were recorded if there was insufficient kinetic data.

2. Stair Navigation

For this activity a custom-built instrumented staircase was used (Figure 4-5). The second step drilled into two force plates to collect ground reaction forces at that step. The height and tread of each step was 185mm and 280mm respectively.

Figure 4-5: The stairs used in the stair negotiation task. The second step was instrumented.

Each participant began the trial a couple of steps away from the first step to allow a natural motion whilst ascending the stairs. Participants were free to use any preferred strategy to complete the task, use of the bannisters was not restricted, and participants were free to ascend with a step-over-step, or step-by-step strategy. No instruction was given as to which foot ought to strike the step first. A minimum of five trials was recorded or less if the participant could not continue.

As the instrumented step had a height of 36.5cm above force plates 1 and 2, the centre of pressure on these force plates would be below the physical placement of the loading foot. Software was adjusted to virtually raise the vertical position of these force plates (recording surface) up to the height of the step and setting the distance between the sensor origin and force plate surface to 36.5cm ensured kinetic data were recorded correctly.

3. Incline walking

This task was carried out in the Motek CAREN system (Figure 4-6) and whether uphill or downhill walking was carried out first was decided randomly by coin toss. Participants were attached to safety rigs around the treadmill with a harness and a

colleague stood alongside the participant, also harnessed onto the rig, for assistance. The participant began with the platform at a level inclination (0^o) and the treadmill set in a slow fixed pace (around 0.3 - 0.7ms^{-1}). The platform pitch and treadmill states were controlled by Motek Medical software, DFlow. When the participant felt comfortable on the treadmill, a feature called self-paced mode was activated. Here the system related the location of the participant's pelvis markers and their anterior-posterior position on the treadmill. If the belt speed was slower than preferred, the participant may walk themselves forward, so feedback algorithms sped up the belt to bring the participant to the centre of the belt. Conversely, if the belt speed was too fast and the participant tended to walk to the posterior end of the belt, the speed of the belt slowed to allow the participant to walk to the centre of the belt. As the platform incline was altered a natural change in walking speed is expected which this feature accounted for.

Figure 4-6: Downhill walking on the CAREN system.

Once in self-paced mode the participant was encouraged to walk at a comfortable walking speed. Once a steady walking speed was reached a ten second recording was made in DFlow and a parallel recording in Vicon Nexus was simultaneously started. Clear instruction was made that with the participants' permission the platform inclination will change whilst they continued walking on the self-paced treadmill. Ten second DFlow recordings were made at$\pm 0^o$, 5^o, and 7.5^o inclinations or to the steepest inclination the participant felt comfortable walking at. As biomechanical data is retrieved from Vicon Nexus, the difference in time stamps between ten second DFlow

recordings were used to separate appropriate sections of interest in the Vicon Nexus trials. As the force plates were not integrated into Vicon Nexus at the time of recording, no kinetic data were available for this activity. In addition, when the platform was at a slope, the force plates' centre of pressure would undergo translational and rotational inertial movements. The effects of the platform motion gave incorrect kinetic data as these inertial movements were not compensated for.

4.5. Data Processing

Each trial was cropped in Vicon Nexus to include the relevant region of interest of data to be analysed. For level walking, the first and last frames were selected where all the lower limb markers were visible. For the stair tasks, the region of interest was such that a full gait cycle of each limb was visible (from heel strike to ipsilateral heel strike). In the sloped walking tasks, the regions of interest were the ten second period of walking at each inclination. Markers were then auto labelled using software pipelines and each recording was manually checked for any gaps or mislabels in the data. Trajectories were manually corrected in the event of mislabelling, and gaps in the data were filled using mathematic algorithms within the software.

For small gaps of less than ~ five frames, a Woltring quantic spline fill was used. This takes the last and next known marker position and interpolates the marker position within the gap. Rigid Body Fill (for pelvic markers) or Pattern Fill (for all other markers) were used for larger gaps which work by selecting a "Source" marker(s) that are present during the gap and filling the gap based on the relationship between the missing and present markers. To reduce noise and smooth trajectories, marker data were filtered using a 4th order Butterworth filter with a cut off frequency of 6Hz. Analogue GRF data were filtered using a low pass 4th order Butterworth filter with a cut off frequency of 300Hz. Both of these filtering steps were carried out before running the model and care was taken to only filter data once so as not to lose any real data by over-filtering.

Gait events (heel strike/foot contact and toe off) were detected either from force plate data (where available) or custom MATLAB (Mathworks Inc, US) scripts utilising the velocity of feet markers. For level and sloped walking, heel strike and toe off was determined by methods described elsewhere (Zeni et al., 2008). Here the change in antero-posterior velocity from positive to negative or vice versa was utilised. For stair ascent, the maximum vertical toe marker acceleration was set as the foot contact, and toe off was given by the local maximum in vertical displacement of the toe and pelvis. For descent, the minimum vertical velocity of the whole body centre of mass gave foot contact, and toe off was where the contralateral limb underwent maximum knee flexion (Foster et al., 2014). All gait events were also confirmed by manual visual inspection of each trial. MATLAB scripts for all tasks and data processing are included in the appendix (Appendix 9: Custom MATLAB Scripts).

Trial data were processed using MATLAB scripts to normalise kinematic and kinetic data to 100% of a gait cycle between heel strike to consecutive heel strike using the interpft function in MATLAB (Appendix 9.1.2). This carries out one dimensional interpolation using the Fast Fourier Transform method ("MATLAB interpft - MathWorks," n.d.). All data from trials of an activity type were stacked into a multidimensional matrix and at each percentage, data were averaged to give an average data set per person. Where data were to be specifically from stance or swing phase, each participants' toe off value was used to give individual stance phase regions. Appropriate discrete values were exported per participant and activity as described in Table 4-3 and collated with other participants receiving the same knee implant. The difference in mean gait parameter values between implant groups at either operative state or to controls was then analysed.

Table 4-3: Sections of the gait cycle where the peak/minimum parameter was taken.

Parameter	How exported
Peak Kinematics	Overall peak angle
Kinematic Range of Motion	Peak subtract minimum value
Flexion at Heel Strike	Value at 1% of gait cycle
Flexion at Toe Off	Value at individual participants' toe off
Peak Knee Flexion during Stance	Peak between 1-40% of gait cycle
Minimum Knee Flexion during Stance	Minima between 15-70% of gait cycle
Peak Hip Flexion Moment	Maxima during stance
Peak Hip Extension Moment	Minima during stance
Early Stance Knee Flexion Moment	Peak between 1-30% of gait cycle
Peak Knee Extension Moment	Minima between 20-50% of gait cycle
Late Stance Knee Flexion Moment	Peak between 40-80% of gait cycle
Early Stance Knee Adduction Moment	Peak between 1-30% of gait cycle
Mid-Stance Knee Adduction Moment	Minima between 20-50% of gait cycle
Late Stance Knee Adduction Moment	Peak between 40-80% of gait cycle
Knee Abduction moment	Minima during stance.
Ankle Dorsiflexion Moment	Maxima during stance
Ankle Plantarflexion Moment	Minima during stance
Peak Eccentric Hip Power	Maxima during stance
Peak Concentric Hip Power	Minima during stance
Peak Eccentric Knee Power	Maxima during stance
Peak Concentric Knee Power	Minima during stance
Peak Eccentric Ankle Power	Maxima during stance
Peak Concentric Ankle Power	Minima during stance

4.6. Statistical Analysis

After the data were grouped by activity, operative state, and implant type, descriptive statistics (mean, standard deviation) was calculated using the statistical software IBM SPSS version 21 (Chicago, Illinois, USA).

Due to the small samples in each implant group at each operative state, although non-parametric tests could be used, due to the low power of the sample sizes, higher power parametric tests were carried out (Bland, 2015). It is also well established that ANOVA tests are robust against violations for the normality of distribution (Glass et al., 1972; Khan and Rayner, 2003; Lix et al., 1996; Schmider et al., 2010). By opting for parametric tests throughout, normality and homoscedasticity of the distributions

were not assessed. When comparing preoperative or postoperative patient data to controls, for each dependent measure a one-way analysis of variance (ANOVA) test with a Bonferonni correction was carried out.

To assess differences between implant groups at each operative state, two-way ANOVA tests were similarly carried out for each dependent measure. If the interaction effects of implant * operative state were significant, these were explored further using a simple effects test of the estimated marginal means. As the operative state is only one level (between pre- and postoperative), no p value correction was necessary. As statistical significance is related to whether the confidence interval crosses 0, the mean differences and standard error of the mean difference are also reported. The standard error multiplied by 1.96 gives the 95% confidence interval which provides an idea of precision in the test.

When comparing between any two groups of data, if both the upper and lower bounds of the 95% confidence interval for difference of the means of these groups lies wholly outside of 0 (as in, the upper and lower limits are either both positive or both negative) this is considered statistically significant. However, it is important to consider clinically significant findings which may be insignificant. Based on work by Mahoney et al., (2012), a mean kinematic difference between groups of at least 5^{o} may be considered clinically significant and as such will also be analysed. Although differences of 5^{o} in non-sagittal planes are likely overconservative, in the absence of accepted minimal clinically important differences for these planes, using 5^{o} is indicative of at least a modest effect between groups.

Categorical data such as participant gender, affected side or participant reported questionnaire answers were analysed using multiple Fischer's Exact Test with a Bonferonni correction. This has benefits over the Chi squared test as it is better suited for smaller sample sizes (Kim, 2017).

For reference a table of PiG kinematic joint rotations and the convention that a positive value describes is given in Table 4-4:

Table 4-4: Lower limb kinematic variables and their convention in the Plug-in Gait model.

Joint Angle	Positive Value indicates:
Pelvic Tilt	That PSIS is above ASIS
Pelvic Rotation	Internal rotation- that side is in front of opposite side
Pelvic Obliquity	That side is higher above the opposite side
Hip Flexion/Extension	Hip flexion, knee is in front of body
Hip Ab/Adduction	Adducted (inward) leg
Hip Rotation	Internally rotated thigh (towards midline)
Knee Flexion/Extension	Flexed knee
Knee Ab/Adduction	Varus/adducted (outward) knee
Knee Rotation	Internally rotated knee (towards midline)
Ankle Dorsi/Plantarflexion	Dorsiflexion
Foot Inversion/eversion	Inverted foot
Foot Progression	Internal rotation (towards midline)

Furthermore, a table of kinetic outputs and their descriptions is provided below (Table 4-5) and moments are of the distal joint onto the proximal joint:

Table 4-5: The lower limb Plug-in Gait kinetic parameters (positive descriptor in brackets).

Kinetic Parameter		Description
External Moments	Ankle Moment	The moment of the foot on the tibia X - Dorsi/plantar flexion (dorsiflexion) Y - Ab/adduction (adduction) Z - Longitudinal rotation (internal)
	Knee Moment	The moment of the tibia on the femur X - Flexion/extension (external flexion) Y - Ab/adduction (adduction) Z - Longitudinal rotation (internal)
	Hip Moment	The moment between the femur on the pelvis X - Flexion/extension (external flexion) Y - Ab/Adduction (adduction) Z - Longitudinal rotation (internal)
Power	Ankle Power	The scalar energy transferred from the foot to the tibia (power generated)
	Knee Power	The scalar energy transferred from the tibia to the femur (power generated)
	Hip Power	The scalar energy transferred from the femur to the pelvis (power generated)

In all proceeding results tables, data will be presented as an average value ± standard deviation for easier interpretation. Any statistically significant values are indicated in bold and a grey shaded background. For gait graphs, any shaded region is twice the standard deviation as 95% of the data spread is captured as opposed to 66.7% for one standard deviation. Vertical lines within a gait graph corresponds to the percentage during the gait cycle toe off occurred. Symbols in superscript in tables indicate the following:

Table 4-6: Symbols used to show significant difference between pre- and postoperative, to control group and between implant groups.

†,††,†††	Significant difference to preoperative group of same implant type at p<0.05, <0.01 and <0.001 respectively.
1,2,3 11,22,33 111,222,333	Significant difference to the numbered implant group where CRDD=1, UC=2, UCR=3. Where an implant group is given in single, duplicate or triplicate indicates significant difference at p<0.05, <0.01 and <0.001 respectively.
*,**,***	Significant difference between that implant group to the Control group at p<0.05, <0.01 and <0.001 respectively.

5. Case Study Results

This chapter aims to provide confidence to the computational data analysis methods of this study. A case study and summaries of key biomechanical data of a single patient will be initially presented whereby full lower body kinematic and kinetic profiles are given for each ADL. Following this, the data respective to each implant group will be shown along with control data as a comparative reference. Whilst statistical analysis between operative states is presented, the differences reflect this individual only and is not representative of all TKA patients. Therefore, this chapter concentrates on demonstrating the quality of the recorded data, and the proceeding discussion focuses on how closely these data match literature values. Intra-subject repeatability is indicated by the standard deviation within each data finding.

This case study refers to a 64-year old female patient (BMI: 25.5kg/m^2 (preoperative), 25.4kg/m^2 (postoperative)) recruited into the study in January 2016 with the Participant Information Sheet shown in Appendix 3: Patient Information Sheet, and was assigned the study ID of #19. Total knee replacement of the left knee took place in February 2016, and she received a low congruency fixed bearing implant (CRDD). Patient #19 reported that her right knee remained asymptomatic throughout. The preoperative OKS was 35 which improved to 28 at 12 months post-surgery. Data is of the operated limb at pre- (red) and one-year postoperative (blue) time points. Significant differences between operative states are indicated with: † where $p<0.05$, †† where $p<0.01$ and ††† where $p<0.001$, found using an independent sample T-Test.

5.1. Overground Level Walking

The first activity carried out was overground level walking. As this activity comprised of multiple recordings, multiple spatiotemporal parameter (STPs) data were collated, and the average and standard deviations is shown in Table 5-1:

Table 5-1: Patient #19's mean and standard deviation STPs for the level walking activity at pre- and postoperative.

Parameter	Pre-Op	Post-Op	Pre - Post Percentage Change (%)
Cadence (steps/min)	106.5±2.3	**113.3±2.1††**	+6.2
Walking Speed (m/s)	1.09± 0.03	**1.22±0.05†††**	+11.7
Stride Time (s)	1.13±0.02	**1.06±0.02†††**	-5.78
Step Time (s)	0.58±0.01	**0.53±0.01†††**	-7.2
Opposite Foot Off (%)	13.6±1.5	**11.2±0.9††**	-17.6
Opposite Foot Contact (%)	48.9±0.6	**49.6±0.4†**	+1.4
Double Support (s)	0.29±0.03	**0.24±0.02††**	-18.9
Stride Length (m)	1.23±0.03	**1.29±0.02††**	+5.2
Step Length (m)	0.64±0.01	**0.67±0.01†††**	+4.6

Key: † = significance to preoperative where p<0.05

†† = significance to preoperative where p<0.01

††† = significance to preoperative where p<0.001

All STP data were significantly changed from preoperative. Cadence, walking speed, and step and stride length (p<0.001) increased postoperatively giving that Patient #19 walked faster, with a longer stride, at a higher frequency of steps. Significant decreases in double support (p=0.004), and stride and step time (p<0.001) corroborates this as less time is spent in stance for both limbs and a gait cycle is completed faster. The point of opposite foot off and opposite foot contact also changed at postoperative (p=0.009 and 0.021 respectively). As opposite foot off occurred earlier than preoperative, and opposite foot contact occurred later, this implies body weight was supported on the operated limb for a greater portion of the gait cycle (single support had increased). Step and stride length also significantly increased from preoperative, as these parameters are not height-normalised (VICON, n.d.), these parameters were included here but will not be part of the group analysis.

Patient #19's average full three-dimensional lower body kinematics during this activity is shown in Figure 5-1. N = 12 (preoperative), 15 (postoperative):

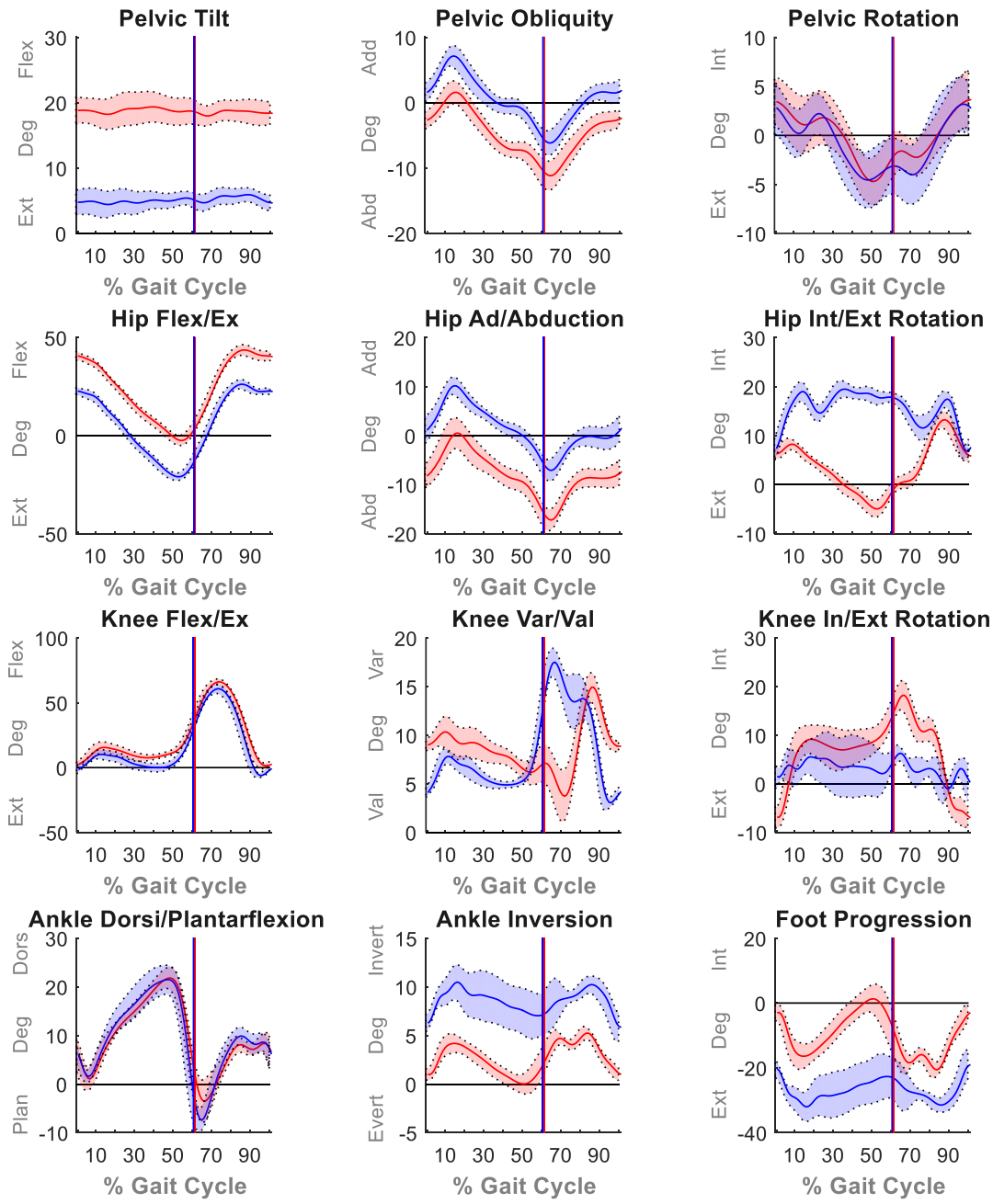

Figure 5-1: Patient #19's operated limb kinematics during overground level walking. Red traces = preoperative, blue = postoperative, vertical lines indicate toe off. Shaded bands = ±2 standard deviations from the mean.

Sagittal plane motions show the largest range in motion and gave smaller standard deviations giving confidence to gait event detection and data processing techniques. At postoperative, patient 19 showed reduced pelvic tilt and hip flexion and simultaneously more adduction in these joints. More knee flexion ROM during stance was seen at postoperative, and post-surgery peak flexion was slightly less than at

preoperative. In the frontal plane, the knee gave high adduction angles during swing phase, a common error associated with the Plug-in Gait (PiG) model. Rotational angles showed greater variability and suggested the knee and ankle was more internally rotated (toward the midline) at preoperative which became less internally rotated at postoperative.

A table of average maximum joint angles and range of motion (ROM) for each joint in each plane is given for Patient #19 at pre-and postoperative visits (Table 5-2).

Table 5-2: Average peak and range of joint angles of Patient #19 in all planes during overgroud level walking.

Joint Angle (°)	Pre-Op Peak Angle	Post-Op Peak Angle	Pre-Op ROM	Post-Op ROM
Pelvic Tilt	19.8±0.9	**6.2±0.6†††**	2.3±0.6	2.3±0.6
Pelvic Obliquity	1.6±0.9	**7.2±0.8†††**	12.8±0.9	13.4±1
Pelvic Rotation	4.5±0.8	3.8±1.2	9.2±1.5	8.7±1.6
Hip Flexion	43.8±1.3	**26.3±1.2†††**	46.3±1.9	47.3±1.6
Hip Ab/Adduction	0.6±1.5	**10.2±0.9†††**	17.8±1.5	17.4±1.4
Hip Rotation	13.3±0.8	**19.7±0.8†††**	18.3±1.3	**13.2±1.2†††**
Knee Flexion	66.2±1.1	**60.9±1.7†††**	65.1±1.3	**67.2±1.5†††**
Knee Ab/Adduction	15±0.8	**17.6±0.7†††**	11.4±1.8	**14.6±0.9†††**
Knee Rotation	18.5±1.3	**7.1±1.2†††**	26.4±1.4	**8.5±1.9†††**
Ankle Flexion	21.9±1.1	21.6±1.5	25.5±1.6	**29.2±1.5†††**
Foot Inversion/eversion	5.3±0.3	**10.8±0.6†††**	5.3±0.5	5.2±0.9
Foot Progression	1.4±2.1	**-18.3±2.2†††**	22.2±2	**14.4±2.5†††**

Key: ††† = significance to preoperative where p<0.001

Independent samples T-tests found that only maximum pelvic rotation (p=0.084) and ankle dorsiflexion (p=0.666) was similar between operative states. For all significantly different maximum and ROM parameters, all p values were <0.001. The largest change between pre- and postoperative was for peak hip flexion and peak ankle rotation (mean difference ± standard error (SE): 17.5±0.5^{o} and 19.8±0.8^{o} respectively)

and knee rotational ROM (17.8±0.6°). Peak knee flexion decreased by 5.3±0.6° at postoperative. Fewer joint ROM parameters were significantly changed at postoperative compared to peak values. Despite a reduced peak knee flexion, overall knee flexion ROM increased postoperatively by 2.1±0.6° indicating more knee extension was evident.

Table 5-3 shows further sagittal parameters for all lower limb joints at ipsilateral heel strike(HS), toe off (TO), and stance phase for knee flexion:

Table 5-3: Lower limb angles at heel strike and toe off. Maximum and minimum knee flexion during stance is also shown.

Joint Angle (°)	Pre-Op Peak Angle	Post-Op Peak Angle
Pelvic Tilt at Heel Strike	18.8±1	**4.8±0.9†††**
Hip Flexion at Heel Strike	40.7±1.1	**22.8±0.9†††**
Knee Flexion at Heel Strike	2.5±1.1	**-1.6±1.7†††**
Ankle Dorsiflexion at Heel Strike	5.9±1.7	6.2±0.9
Pelvic Tilt at Toe Off	18.6±1	**5.1±0.6†††**
Hip Flexion at Toe Off	3.3±1.6	**-13.2±1.5†††**
Knee Flexion at Toe Off	35.5±2.7	**30.7±2.7†††**
Ankle Dorsiflexion at Toe Off	2.8±2.3	**-2.7±2.3†††**
Max Stance Knee Flexion	15.8±2	**10.3±1.7†††**
Min Stance Knee Flexion	7.5±1.2	**-0.1±1.6†††**

Key: ††† = significance to preoperative where p<0.001

Only ankle dorsiflexion at heel strike was not significantly changed post-surgery (p=0.685). All differences for other parameters were p<0.001. Hip flexion at HS and TO shows the greatest postoperative decrease (± SE) of 17.9±0.4° and 16.4±0.6° respectively. At postoperative, generally all joints were more extended at HS and TO. During stance, the knee showed less peak (5.5±0.7°) and minimum flexion (7.6±0.5°) at postoperative confirming a more extended limb.

Average three-dimensional lower body external moments are given below in Figure 5-3. The sample sizes at preoperative was n = 4, and n = 5 at postoperative.

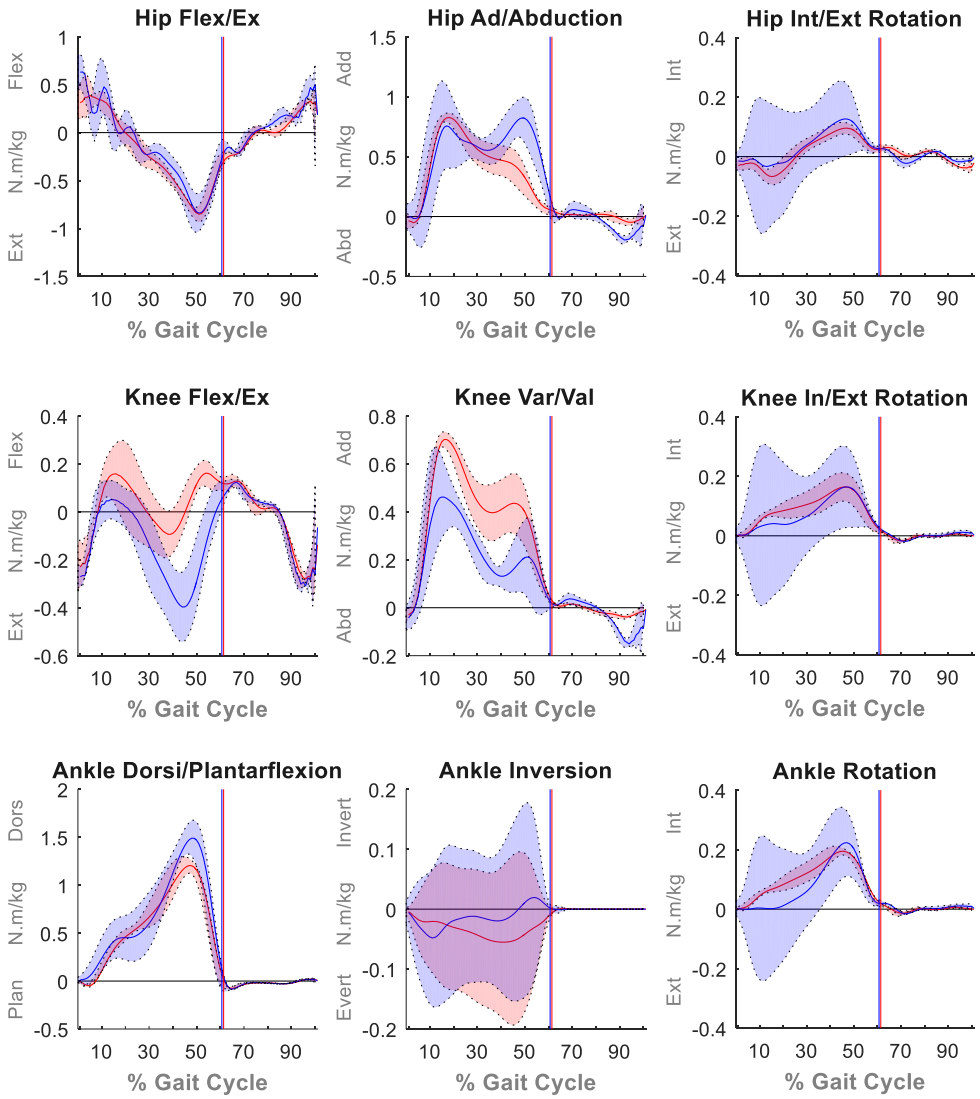

Figure 5-2: Lower limb joint moments of Patient #19 during overground level walking. Red traces = preoperative, blue = postoperative, vertical lines indicate toe off. Shaded bands = ±2 standard deviations from the mean.

Compared to preoperative, postoperative variability was greater as evident by the considerably larger shaded regions. The hip flexion moment showed a typical sinusoidal-type trace, beginning with a peak flexion moment at heel strike which became a peak extensor moment prior to toe off. The postoperative knee flexion moment showed a lower early stance flexion moment and greater midstance peak extension moment than at preoperative. The characteristic biphasic peaks of the

knee adduction moment (KAM) was evident, and greater KAM values and slightly less distinction between the peaks was seen at preoperative, typical of those with knee OA. Of all the joints, ankle dorsiflexion moments gave the largest values which increased further at postoperative.

The maximum and minimum joint moments of Patient 19 is given below in Table 5-4:

Table 5-4: Maximum and minimum joint moment data.

Joint Moment (N.m/kg)	Mean Pre-Op Peak Moment	Mean Post-Op Peak Moment	Mean Pre-Op Min. Moment	Mean Post-Op Min. Moment
Hip Flexion	0.41±0.08	**0.66±0.06††**	-0.85±0.04	-0.84±0.10
Hip Adduction	0.83±0.02	0.86±0.11	-0.05±0.02	0.05±0.09
Hip Rotation	0.1±0.01	0.13±0.06	-0.07±0.01	-0.07±0.07
Knee Flexion (Early/Midstance)	0.17± 0.06	**0.06±0.03†**	-0.23±0.05	**-0.40±0.07††**
Knee Flexion (Late Stance)	0.16±0.03	0.12±0.01	N/A	N/A
Knee Adduction (Early Stance)	0.7±0.02	**0.48±0.08††**	-0.03±0.01	0.04±0.03
Knee Adduction 2 (Midstance Trough)	0.38±0.06	**0.12±0.02††**	N/A	N/A
Knee Adduction 3 (Late Stance)	0.44±0.06	**0.22±0.08††**	N/A	N/A
Knee Rotation	0.17±0.02	0.17±0.07	0.0±0.0	-0.04±0.07
Ankle Dorsiflexion	1.21±0.04	**1.49±0.09†††**	-0.05±0.01	**0.01±0.02†††**
Ankle Inversion	0 ± 0.01	0.03 ± 0.06	-0.06±0.07	-0.06±0.04
Ankle Rotation	0.2 ± 0.01	0.23 ± 0.06	0.0±0.0	-0.04±0.07

Key: † = significance to preoperative where p<0.05

†† = significance to preoperative where p<0.01

††† = significance to preoperative where p<0.001

Significant differences were mostly in the sagittal plane. From preoperative, peak hip flexion moment significantly increased (p=0.004, 0.25±0.046 N.m/kg) and maximum and minimum early stance knee flexion moment (KFM) decreased (p=0.021, 0.11±0.03 N.m/kg and p=0.004, 0.17±0.04 N.m/kg respectively). Maximum and

minimum ankle dorsiflexion moment also significantly ($p<0.001$) increased at postoperative. Each preoperative knee adduction moment (KAM) was significantly greater than postoperative ($p=0.003$ for the KAM1 and KAM3 peaks, and $p=0.001$ for KAM2). Overall, the postoperative KAM peaks were around 0.24 N.m/kg lower than preoperative.

Further analysis of the KFM and KAM is depicted below in Figure 5-3 where T-tests carried out at each point of the gait cycle were carried out to find significant differences between pre-and postoperative.

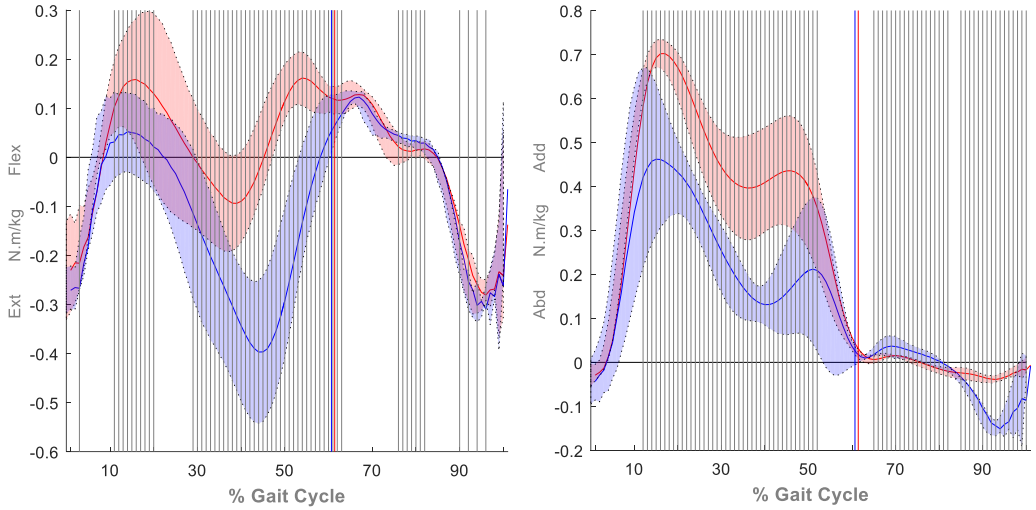

Figure 5-3: Pre-and postoperative knee flexion (left) and adduction (right) moments. Red traces = preoperative, blue = postoperative, vertical lines indicate toe off. Shaded bands = ±2 standard deviations from the mean.

Significance between pre-and postoperative at the relevant point in the gait cycle are indicated by grey lines. In both graphs, significance was seen in regions where peaks and mean differences were greatest: the peak early and late stance KFM and midstance peak knee extension moment, and between 11% to 55% of the KAM. In all significant findings the postoperative moment was lower (more negative) than preoperative.

Hip, knee, and ankle joint power throughout the gait cycle is given in Figure 5-4. Here positive values indicate concentric power, or power generated during an activity. Negative values correspond to eccentric power, or power absorbed during an activity.

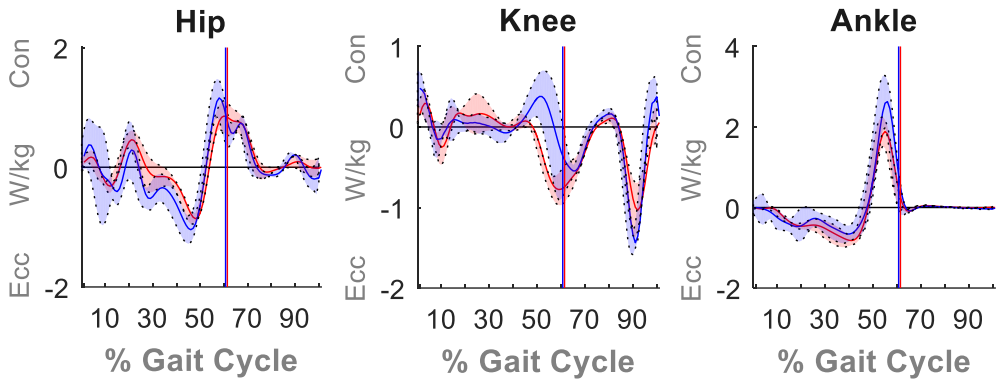

Figure 5-4: Lower limb joint powers at pre-and postoperative. Red traces = preoperative, blue = postoperative, vertical lines indicate toe off. Shaded bands = ±2 standard deviations from the mean.

Joint power is the scalar product of the joint moment and its angular velocity. Similar to the ankle moments, ankle power exhibited the greatest peak concentric powers representing propulsive plantarflexion which occurs during pre-swing. At post-surgery, the knee generated more power at pre-swing compared to preoperative (perhaps indicating improved quadriceps strength) where pre-swing power was mostly eccentric (absorptive).

Peak concentric and eccentric hip, knee, and ankle power during stance is shown in Table 5-5:

Table 5-5: Maximum concentric and eccentric powers during level walking.

Joint Power (W/kg)	Pre-Op Peak Concentric Power	Post-Op Peak Concentric Power	Pre-Op Peak Eccentric Power	Post-Op Peak Eccentric Power
Hip	0.86±0.12	**1.17±0.16†**	0.86±0.04	**1.05±0.12†**
Knee	0.31±0.08	**0.54±0.08††**	0.78±0.09	**0.37±0.18††**
Ankle	1.90±0.16	**2.69±0.29††**	0.83±0.08	**0.69±0.08†**

Key: † = significance to preoperative where p<0.05

†† = significance to preoperative where p<0.01

††† = significance to preoperative where p<0.001

All joints showed significantly higher generative powers at postoperative by p=0.01, 0.004 and 0.002 for the hip, knee, and ankle respectively. Maximum concentric power similarly showed significant changes for all joints at post-surgery, however

where hip showed an increase in power absorption (p=0.02), the knee and ankle showed less absorption (p=0.004 and 0.036 respectively).Summary

For level walking, Patient #19's spatiotemporal data presented strong improvements at postoperative. Peak joint kinematics tended to decrease whilst ROM showed no obvious trends. Sagittal angles at heel strike and toe off mostly significantly decreased (except for ankle dorsiflexion at HS which showed an insignificant increase) from pre-surgery meaning joints were less flexed at these events. This may indicate joint function had improved as straighter limbs at HS and TO allow for a more natural gait pattern. Postoperative patient kinetics exhibited greater inter-trial variability compared to preoperative. Postoperative knee flexion and adduction moment was significantly lowered from preoperative, and all joints showed more generative power also indicating healthier muscle function.

5.2. Stair Navigation

Following level walking, Patient #19 then carried out a stair navigation task, she carried out this activity with a step-over-step strategy and used both bannisters when ascending and descending the staircase. Data for stair ascent and stair descent is presented in Chapters 5.2.1 and 5.2.2 respectively.

5.2.1. Stair Ascent

The spatiotemporal parameters (STP) of Patient #19 during stair ascent is shown in Table 5-6:

Table 5-6: Patient #19's STPs and percentage change during stair ascent.

Parameter	Pre-Op	Post-Op	Percentage Change (%)
Cadence (steps/min)	76.4±5.2	85.5±10	+11.9
Walking Speed (m/s)	0.5±0.1	0.6±0.1	+20.0
Stride Time (s)	1.6±0.1	1.4±0.2	-12.5
Step Time (s)	0.8±0.1	0.8±0.1	0.0
Opposite Foot Off (%)	13.7±0.8	9.0±5.4	-34.3
Opposite Foot Contact (%)	48.1±4.0	46.7±2.2	-2.9
Double Support (s)	0.4±0.0	**0.3±0.2†**	-25.0

Key: † = significance to preoperative where $p<0.05$

As this activity requires forward and vertical translation of the body, the stride and step length parameters are omitted for both of the stair activities. Walking speed and cadence were increased post-surgery, not by a significant amount ($p=0.279$ and $p=0.12$ respectively). All other parameters decreased or showed no change from preoperative and mostly showed no significance except for the period of double support which was significantly reduced at postoperative ($p=0.033$). Spending less time in double support corresponded to increases in cadence and walking speed and a decrease in stride time. The point of opposite foot off, and time spent in double support showed the largest percentage decrease which is expected.

Three-dimensional joint kinematics during stair ascent are shown in Figure 5-5. N = 5 at both operative states:

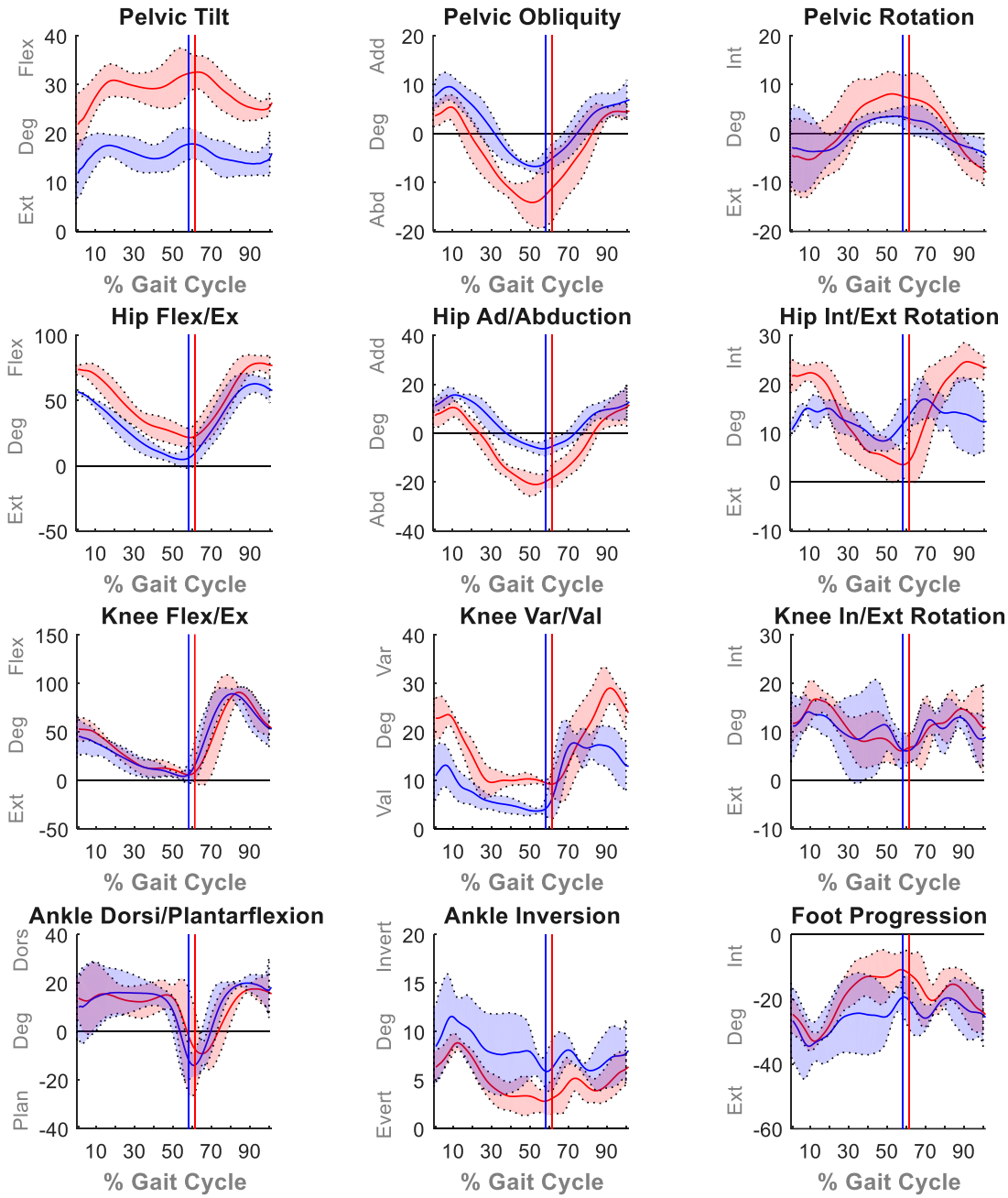

Figure 5-5: Lower limb three-dimensional kinematics during stair ascent. Red traces = preoperative, blue = postoperative, vertical lines indicate toe off. Shaded bands = ±2 standard deviations from the mean.

Traces here are similar to level walking, postoperative pelvic and hip sagittal data showed less flexion than at preoperative, whilst showing greater adduction in the frontal plane. Decreased postoperative pelvic tilt and hip flexion indicates a straighter lower limb alignment. Increased out of sagittal plane movement is expected during this activity whereby the hip adductor brings the limb towards the midline after foot strike and hip abduction then occurs after midstance to start to bring the limb away

from the midline at pre-swing. Transverse plane rotational angles followed similar patterns between operative states for all joints although postoperative traces consistently presented smaller maxima and minima. Knee flexion traces did not show a stance phase peak and trough as seen in level walking, and instead began at high degrees of flexion reaching near full extension before toe off. Peak flexion occurred during swing as the limb travelled upwards and forwards to contact the following step. Knee adduction angles again showed evidence of angle cross talk. The ankle remained in a state of dorsiflexion through the majority of the gait cycle until toe off where the ankle showed peak plantarflexion. At the same point the hip and knee showed a near neutral alignment.

Below in Table 5-7 the peak three-dimensional joint angles and ROM during this activity is given:

Table 5-7: Peak kinematics and ROM of all lower limb joints during stair ascent.

Joint Angle (°)	Pre-Op Peak Angle	Post-Op Peak Angle	Pre-Op ROM	Post-Op ROM
Pelvic Tilt	33.4±1.3	**18.2±1.5†††**	11.7±4	7±2.5
Pelvic Obliquity	6±0.7	**9.7±1.4††**	20.5±2.6	**16.5±1.7†**
Pelvic Rotation	8.9±2.1	**4.3±0.6††**	17.5±2.9	**10.1±1.9††**
Hip Flexion	79.2±3.1	**63±3.6†††**	58.1±6.1	58.5±1.8
Hip Ab/Adduction	13.6±0.8	**16.6±0.9†††**	35.2±2.8	**23.1±0.9†††**
Hip Rotation	24.8±1.9	**18.2±1.4†††**	21.7±2.4	**10±2†††**
Knee Flexion	93.3±3.9	90.5±2.7	89.2±4.1	86.5±3.7
Knee Ab/Adduction	29.6±1.7	**19±1.2†††**	21.4±2	**15.4±1.3†††**
Knee Rotation	17±1.8	15.5±0.6	11.9±2.5	10.9±2.9
Ankle Flexion	20.3±1.8	22.1±1.1	33.3±1.7	**37±1.5††**
Foot Inversion/eversion	8.9±0.5	**11.9±1.9†**	6.3±0.5	6.5±2.1
Foot Progression	-9.6±3	**-17.7±2.4††**	23.4±2.1	**17.5±5.3†**

Key: † = significance to preoperative where p<0.05

†† = significance to preoperative where p<0.01

††† = significance to preoperative where p<0.001

On the whole, peak and ROM values tended to decrease for each joint at postoperative and more significant differences were noted for peak kinematics than ROM. No significant changes in peak or range of knee flexion were observed (p=0.225 and 0.318 respectively) perhaps due to higher preoperative variance. Although peak ankle dorsiflexion was significantly similar at both operative states, dorsiflexion ROM was significantly greater indicating more plantarflexion was evident (p=0.006) as seen in Figure 5-5. The parameters with the largest change ± SE was the pelvis and hip peak flexion angle: 15.1±0.8^{o} and 16.2±2.1^{o} respectively.

Table 5-8 gives sagittal joint angles at heel strike and toe off, as there are no defined knee flexion/extension peaks during stance this was omitted for this activity.

Table 5-8: Sagittal joint angles at heel strike and toe off during stair ascent.

Joint Angle (°)	Pre-Op Peak Angle	Post-Op Peak Angle
Pelvic Tilt at Heel Strike	21.8±3.1	**11.8±3.1††**
Hip Flexion at Heel Strike	73.9±2.2	**56.8±1.4†††**
Knee Flexion at Heel Strike	53.3±6.5	45.4±11
Ankle Dorsiflexion at Heel Strike	13.8±4.5	10.2±6.9
Pelvic Tilt at Toe Off	33.2±1.2	**17.9±1.6†††**
Hip Flexion at Toe Off	21.9±3.7	**5.9±3†††**
Knee Flexion at Toe Off	5.3±1.7	5.9±2.8
Ankle Dorsiflexion at Toe Off	-10.7±2.6	-12.1±3.5

Key: †† = significance to preoperative where p<0.01
††† = significance to preoperative where p<0.001

All angles decreased from preoperative (indicating a more extended, straighter limb at heel strike and toe off) and hip flexion at heel strike showed the largest decrease ± SE of 17.1±1.1^{o}. All significant figures were where p<0.001 aside from pelvic tilt at heel strike where p=0.001. Knee flexion and ankle dorsiflexion at both gait events were statistically similar at both operative states.

Figure 5-6 shows the three-dimensional lower limb moments of Patient #19 during stair ascent, the sample sizes were n=2 at both operative states:

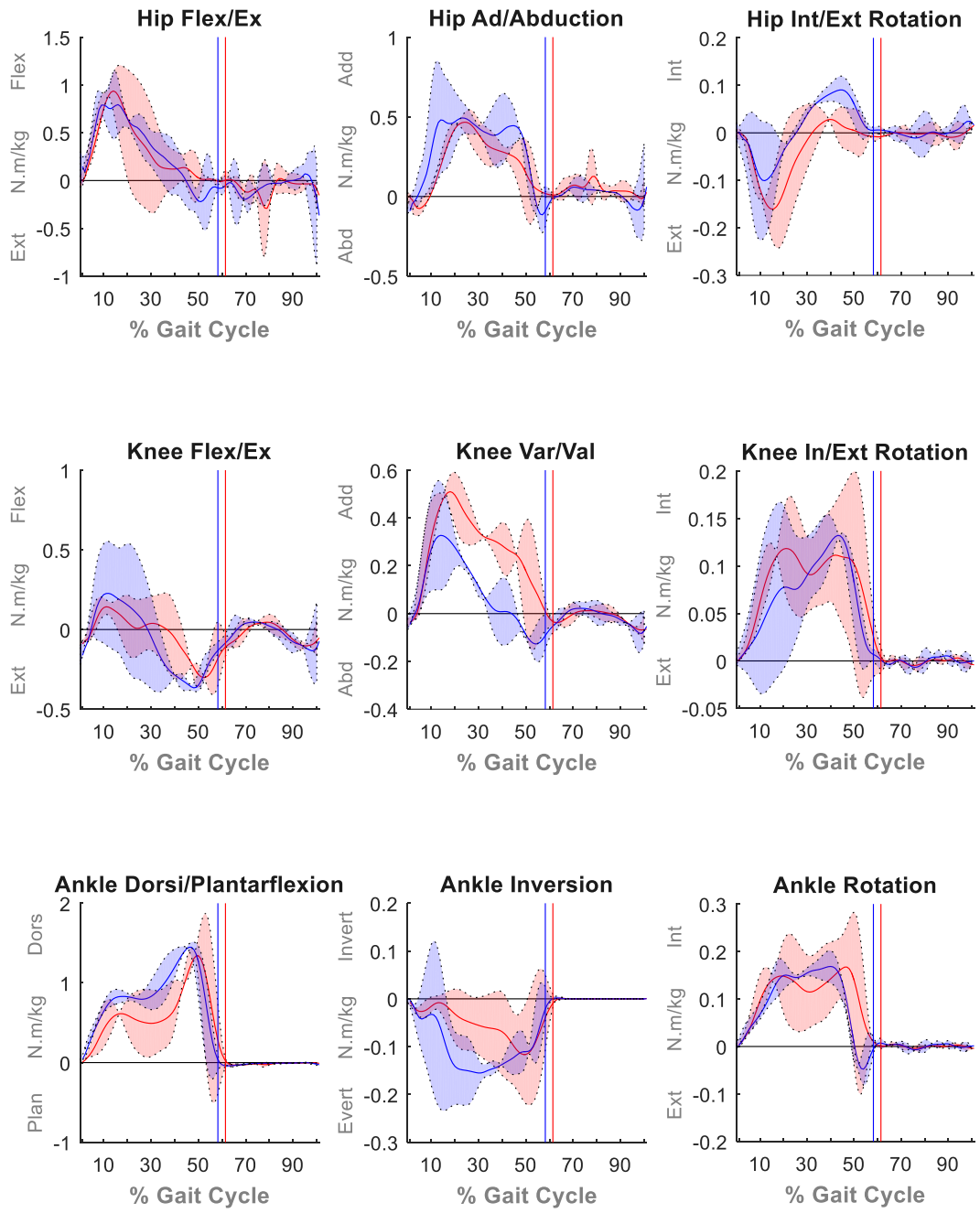

Figure 5-6: Lower limb moments during stair ascent. Red traces = preoperative, blue = postoperative, vertical lines indicate toe off. Shaded bands = ±2 standard deviations from the mean.

The spread of the shaded standard deviation regions shows similar performance across trials and good repeatability of data processing techniques. Hip moments are comparable between operative states, showing an expected hip flexion moment

peak after heel strike since the hip is flexed at the start of the gait cycle. The hip adduction moment facilitates out of sagittal plane motion for the lower limb and was larger at postoperative, corresponding to greater hip adduction angles seen in Figure 5-5. Knee moments showed greater changes at postoperative of all joints, namely the frontal moments which were considerably reduced at post-surgery. The postoperative knee flexion moment showed greater maximum and minimum values, however standard deviations were also greater, indicating higher variability. Lastly, postoperative dorsiflexion moments were consistently higher throughout the gait cycle compared to preoperative.

Significance at each point of the gait cycle between operative states for the knee flexion and knee adduction moments are presented in Figure 5-7.

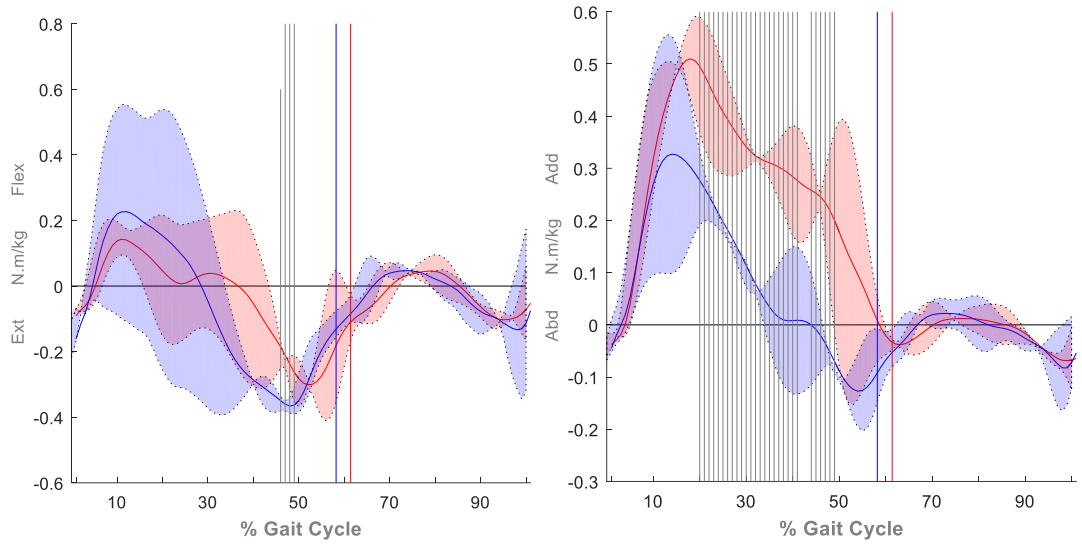

Figure 5-7: Significant differences (p<0.05) between pre-and postoperative for the knee flexion (left) and adduction moments (right) during ascent. Points of significance are depicted by a vertical grey line.

Vertical grey lines indicate points where p<0.05, evidently the knee flexion moment showed more similarities in moment traces between operative states compared to the knee adduction moment. However, significance for the knee flexion moment could be underreported due to the large shaded standard deviation regions. Significance between pre- and postoperative adduction moments were primarily during mid-stance.

Peak and minimum moment values are shown below in Table 5-9. As no minimum knee adduction moment between maxima was seen for this activity this parameter was not included:

Table 5-9: Maximum and minimum lower limb joint moments during stair ascent.

Joint Moment (N.m/kg)	Pre-Op Peak Moment	Post-Op Peak Moment	Pre-Op Min. Moment	Post-Op Min. Moment
Hip Flexion	0.95±0.06	0.83±0.13	-0.05±0.02	-0.23±0.09
Hip Adduction	0.47±0.02	0.53±0.11	-0.08±0.02	-0.12±0.04
Hip Rotation	0.03±0.0	0.09±0.01	-0.17±0.03	-0.1±0.06
Knee Flexion	0.14±0.03	0.23±0.26	-0.28±0.04	-0.37±0.01
Knee Adduction (Early Stance)	0.52±0.02	0.33±0.11	-0.04±0.01	-0.13±0.03
Knee Adduction 2 (Midstance Trough)	0.17±0.08	-0.09±0.0	N/A	N/A
Knee Adduction 3 (Late Stance)	0.28±0.05	0.05±0.02	N/A	N/A
Knee Rotation	0.13±0.02	0.13±0.0	0.0±0.0	0.0±0.0
Ankle Dorsiflexion	1.38±0.03	1.45±0.01	-0.02±0.03	-0.03±0.02
Ankle Inversion	0.0±0.01	0.01±0.02	-0.12±0.05	-0.16±0.02
Ankle Rotation	0.17±0.03	0.17±0.02	0.0±0.0	-0.05±0.01

Overall, no significant differences were identified for any parameters. The greatest change ± SE in peak moments were a decrease of around 0.23±0.06 N.m/kg for the late stance knee adduction moment. The greatest change ± SE in minimum moments was seen in the hip flexion moment by -0.18±0.07 N.m/kg at postoperative. This reduction corresponds to the significant decrease in peak hip flexion angle seen in Table 5-7. Minimum knee flexion and adduction moments decreased by 0.09±0.02 N.m/kg which were similarly related to a decline in peak flexion and adduction angles, although only the change in knee adduction angle was significant.

Hip, knee, and ankle power traces during gait and peak concentric and eccentric powers are displayed in Figure 5-8 and Table 5-10 respectively:

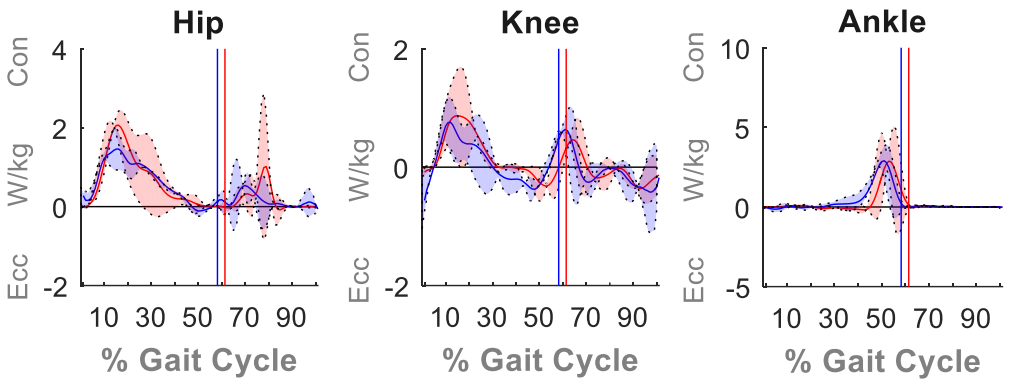

Figure 5-8: Lower limb powers during stair ascent. Red traces = preoperative, blue = postoperative, vertical lines indicate toe off. Shaded bands = ±2 standard deviations from the mean.

As this activity requires power to be generated in order to raise the contralateral limb to the next step, the joint powers were mostly in the concentric phase. Peak hip and knee power values occurred at around the same point of the gait cycle (~15%) where weight acceptance upon the single limb and contralateral swing was taking place. Another peak in joint power was seen in the knee and ankle around toe off, here the limb pushes off the step to be able to clear the next step when completing ascent.

Table 5-10: Maximum concentric and eccentric powers during stair ascent.

Joint Power (W/kg)	Pre-Op Peak Concentric Power	Post-Op Peak Concentric Power	Pre-Op Peak Eccentric Power	Post-Op Peak Eccentric Power
Hip	2.10±0.07	1.46±0.26	0.03±0.01	0.11±0.06
Knee	0.90±0.35	0.77±0.19	0.36±0.06	0.59±0.2
Ankle	3.29±0.01	2.94±0.29	0.20±0.08	0.15±0.07

Similar to the moment data no significance was seen for any power parameters. Overall, peak concentric powers decreased (less generative) for each joint and eccentric power became more absorptive at postoperative. The largest changes ± SE in power were the maximum concentric hip and ankle power which decreased at postoperative by 0.64±0.2W/kg (p=0.16) and 0.35±0.2 W/kg (p=0.39) respectively.

5.2.2. Stair Descent

Patient #19's spatiotemporal parameters during stair descent is shown in Table 5-11:

Table 5-11: Patient #19 STP data during stair descent.

Parameter	Pre-Op	Post-Op	Percentage Change (%)
Cadence (steps/min)	69.7±4.5	**117.3±17.8††**	+67.9
Walking Speed (m/s)	0.4±0.0	**0.7±0.1†††**	+75.0
Stride Time (s)	1.7±0.1	**1.0±0.2†††**	-41.2
Step Time (s)	0.9±0.1	**0.5±0.1††**	-44.4
Opposite Foot Off (%)	10.5±0.7	12.2±1.5	+16.2
Opposite Foot Contact (%)	44.6±2.2	**52.4±1.7††**	+17.5
Double Support (s)	0.4±0.0	**0.2±0.1††**	-50.0

Key: †† = significance to preoperative where p<0.01
††† = significance to preoperative where p<0.001

At postoperative, walking speed (p<0.001), cadence (p=0.003), and opposite foot contact (p= 0.003) all had significantly increased for this activity. In addition, the period of time spent in double support significantly decreased (p=0.006), also reflected by significantly shorter stride and step times (p<0.001). The largest percentage change from preoperative was for walking speed which nearly doubled.

Figure 5-9 gives the three-dimensional kinematics for Patient #19 during stair descent, n=5 at both operative states:

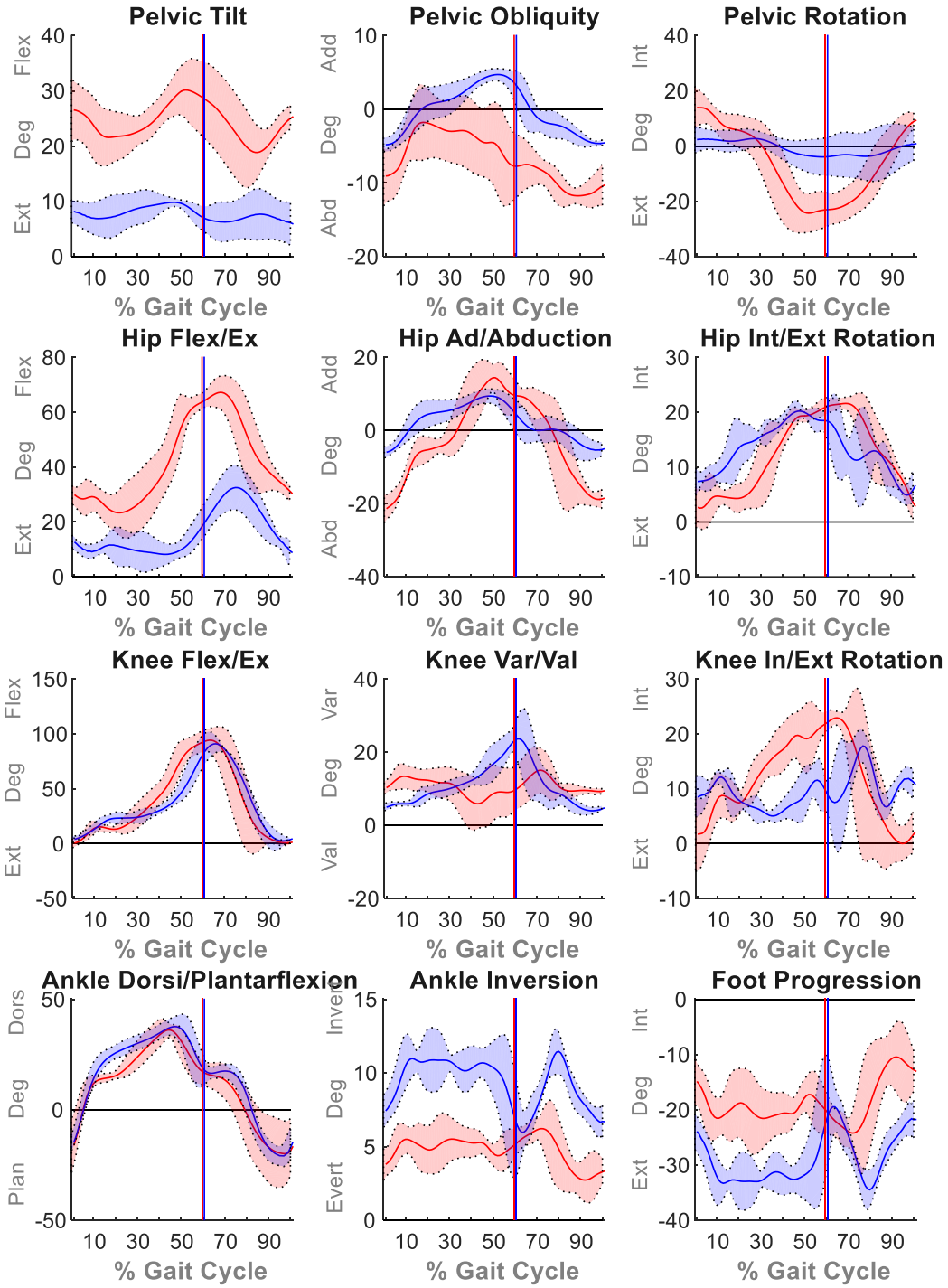

Figure 5-9: Lower limb kinematics during stair descent. Red traces = preoperative, blue = postoperative, vertical lines indicate toe off. Shaded bands = ±2 standard deviations from the mean.

Similar to stair ascent, postoperative pelvic tilt and hip flexion decreased from preoperative, implying Patient #19 had a straighter lower limb alignment when

navigating the stairs. Many parameters showed an opposite trace to stair ascent (Figure 5-5), such as pelvic obliquity, all hip angles and ankle dorsiflexion. Peak knee flexion during stair descent was highest out of all activities and interestingly the effect of kinematic cross talk was less prominent here compared to stair ascent. At initial contact, the ankle was plantarflexed then went into a period of dorsiflexion throughout stance and mid-swing, ankle motion in the sagittal field was comparable between operative states.

The peak and range of three-dimensional joint kinematics during stair descent is presented in Table 5-12:

Table 5-12: Peak and range of Patient #19's joint kinematics in all dimensions during stair descent.

Joint Angle (°)	Pre-Op Peak Angle	Post-Op Peak Angle	Pre-Op ROM	Post-Op ROM
Pelvic Tilt	30.8±2.1	**10.2±0.6†††**	12.9±1.9	**4.9±1.6†††**
Pelvic Obliquity	-1.3±2.3	**4.7±0.4††**	10.9±2.6	9.6±0.3
Pelvic Rotation	14.3±3.6	**3.8±1.8††**	39.1±3.9	**8.4±3.2†††**
Hip Flexion	67.9±2.5	**32.7±4†††**	45±3.8	**26±4.5†††**
Hip Ab/Adduction	15.2±2.1	**9.5±1††**	36.6±3.2	**15.5±0.6†††**
Hip Rotation	21.7±0.5	**20.5±0.7†**	19.9±1.1	**15.7±0.5†††**
Knee Flexion	95.5±4.5	91.5±5	96±4.6	90.1±7.5
Knee Ab/Adduction	15.9±2.8	**25±1.4†††**	11±1.9	**21±1.5†††**
Knee Rotation	23.2±0.9	**17.9±1.4†††**	24.7±1.7	**13.5±1.4†††**
Ankle Flexion	36.9±1.6	38±2.1	57.8±5.4	59±1.8
Foot Inversion/eversion	6.6±0.2	**11.9±0.7†††**	4.1±0.5	**6.5±1.5†**
Foot Progression	-9.6±2.2	**-17.7±3.3††**	15.9±2.2	17.6±4.2

Key: † = significance to preoperative where p<0.05

†† = significance to preoperative where p<0.01

††† = significance to preoperative where p<0.001

All peak angles were significantly changed post-surgery apart from peak knee flexion and ankle dorsiflexion (p=0.21 and 0.38 respectively). The largest difference ± SE from

preoperative was for peak hip flexion (-35.1±2.1°). All significant ROM changes were where p<0.001 (except for the foot progression ROM where p=0.018). Knee flexion ROM had not significantly changed postoperatively (p=0.178); however it had decreased by (mean difference ± SE) 5.9±3.9° which is a clinically relevant finding.

Patient #19's sagittal joint angles at heel strike and toe off is displayed in Table 5-13:

Table 5-13: Sagittal angles at heel strike and toe off during stair descent.

Joint Angle (°)	Pre-Op Peak Angle	Post-Op Peak Angle
Pelvic Tilt at Heel Strike	26.5±2.9	**8.2±1†††**
Hip Flexion at Heel Strike	29.9±1.8	**12.7±1.2†††**
Knee Flexion at Heel Strike	0.5±2.2	**4.1±1.3†**
Ankle Dorsiflexion at Heel Strike	-16.2±6.1	-14.8±1.8
Pelvic Tilt at Toe Off	29.3±2.6	**6.9±0.9†††**
Hip Flexion at Toe Off	62.6±1.8	**20.0±3.0†††**
Knee Flexion at Toe Off	90.5±3.7	**84±4.1†**
Ankle Dorsiflexion at Toe Off	18.8±3.1	17.2±2.9

Key: † = significance to preoperative where p<0.05

††† = significance to preoperative where p<0.001

All parameters showed less flexion (more extension) at postoperative. This was a significant finding for all joints except for ankle dorsiflexion at heel strike (p=0.64) and at toe off (p=0.42). Pelvic and hip flexion at both gait events gave p < 0.001 and mean differences between operative states were around 17° aside for hip flexion at toe off which showed a much larger difference ± SE of 42.6±1.5°. Knee flexion at heel strike (p=0.018, mean difference ± SE: -3.5±1.1°) and toe off (p=0.032, mean difference ± SE: -6.5±2.5°) was also significant, although the change in knee flexion at heel strike was not a clinically relevant result.

Full three-dimensional lower limb moment traces are shown in Figure 5-10 for the operated limb at pre-(n = 4) and postoperative (n = 3).

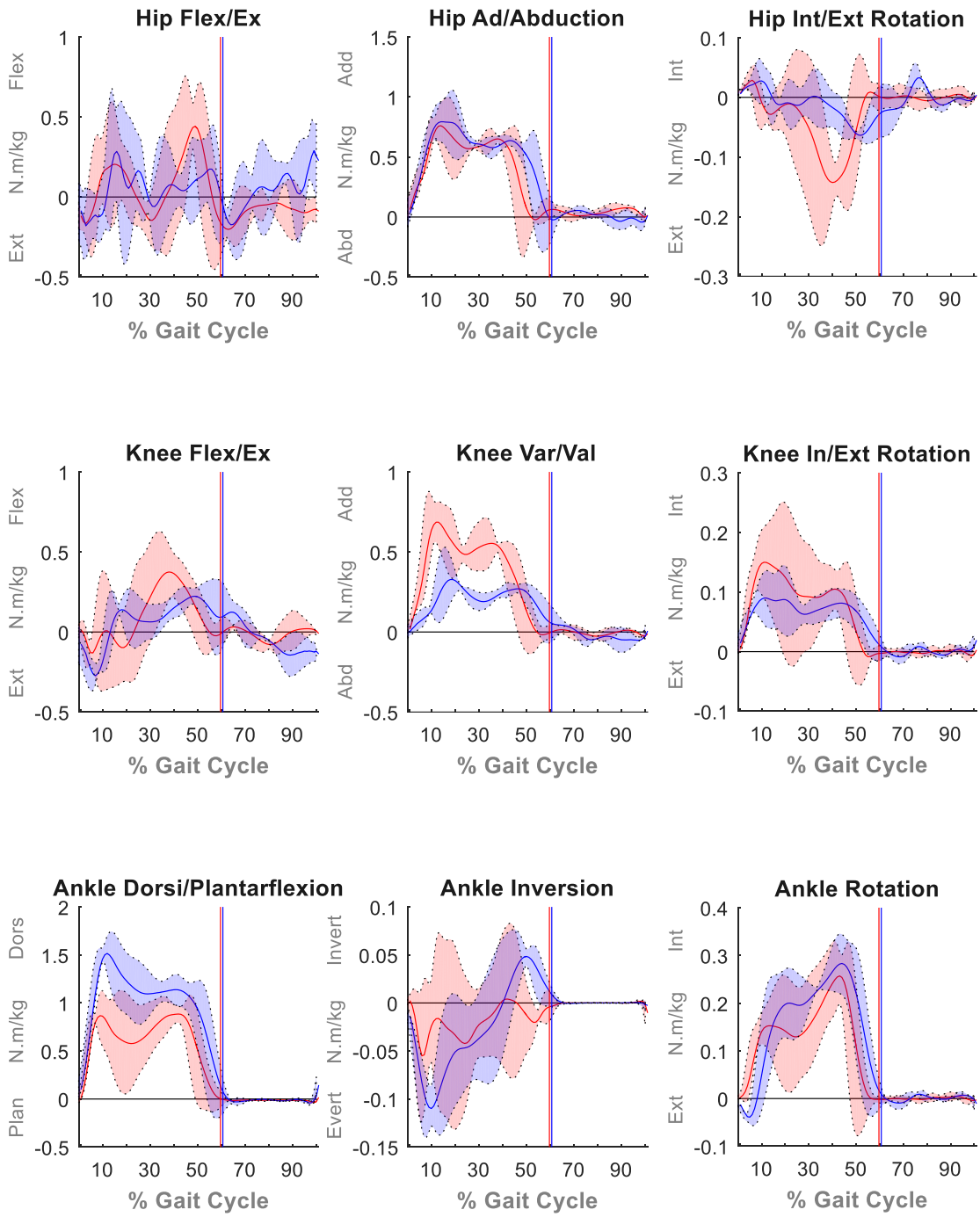

Figure 5-10: Lower limb moments during stair descent. Red traces = preoperative, blue = postoperative, vertical lines indicate toe off. Shaded bands = ±2 standard deviations from the mean.

Larger standard deviation regions were seen here compared to level walking and stair ascent (slightly more so at preoperative) which reflects variation in carrying out this activity. The postoperative knee flexion moment showed less variability and distinction in early stance peaks are clearer. The postoperative knee adduction moment showed more of a bisphasic pattern (similar to that seen whilst level

walking) than during stair ascent. This moment decreased from preoperative, and remained in adduction for a longer period before becoming an abduction moment. Sagittal ankle moments related to the dorsiflexion angle where there was an initial sharp increase into dorsiflexion which became gradual as the dorsiflexion moment reduced. In late stance the ankle remained dorsiflexed whilst tending towards plantarflexion at toe off hence the sharp decrease in dorsiflexion moment.

Table 5-14 displays peak and minimum three-dimensional moments of Patient #19.

Table 5-14: Maximum and minimum moments during stair descent.

Joint Moment (N.m/kg)	Pre-Op Peak Moment	Post-Op Peak Moment	Pre-Op Min. Moment	Post-Op Min. Moment
Hip Flexion	0.53±0.07	0.41±0.09	-0.23±0.06	-0.25±0.04
Hip Adduction	0.79±0.06	0.82±0.1	-0.06±0.04	-0.03±0.11
Hip Rotation	0.03±0.01	0.03±0.01	-0.16±0.02	**-0.06±0.0††**
Knee Flexion 1	0.22±0.14	0.17±0.04	-0.09±0.11	0.05±0.05
Knee Flexion 2	0.38±0.06	**0.17±0.04†**	N/A	N/A
Knee Adduction (Early Stance)	0.73±0.02	**0.35±0.1††**	-0.04±0.02	0.0±0.02
Knee Adduction 2 (Midstance Trough)	0.13±0.12	0.19±0.03	N/A	N/A
Knee Adduction 3 (Late Stance)	0.50±0.03	**0.28±0.01†††**	N/A	N/A
Knee Rotation	0.16±0.04	**0.10±0.02†**	-0.02±0.01	0.01±0.01
Ankle Dorsiflexion	0.94±0.06	**1.54±0.05†††**	-0.02±0.02	0.07±0.09
Ankle Inversion	0.02±0.02	0.05±0.01	-0.06±0.03	**-0.11±0†**
Ankle Rotation	0.27±0.02	0.28±0.03	-0.01±0	**-0.04±0.01††**

Key: † = significance to preoperative where p<0.05

†† = significance to preoperative where p<0.01

††† = significance to preoperative where p<0.001

Regarding peak moments, the late stance peak knee flexion moment was significantly (p=0.011) reduced post-surgery. Early and late stance KAM also exhibited significant decreases (p=0.007 and <0.001 respectively). Peak ankle dorsiflexion moment showed the greatest postoperative decline of all parameters with a mean decrease ±

SE of 0.6±0.04 Nm/kg ($p<0.001$). There were fewer significant changes between operative states in minimum joint moment data, firstly the ankle inversion moment showed a near insignificant finding ($p=0.049$). The minimum hip and ankle rotation moment p values were 0.002 and 0.004 respectively, however the mean difference in ankle rotation is negligible implying changes in ankle rotation moment were seen at more decimal places.

For the KFM and KAM traces, t-tests were carried out at each point of the gait cycle and Figure 5-11 shows points where significant differences between operative states where seen ($p<0.05$).

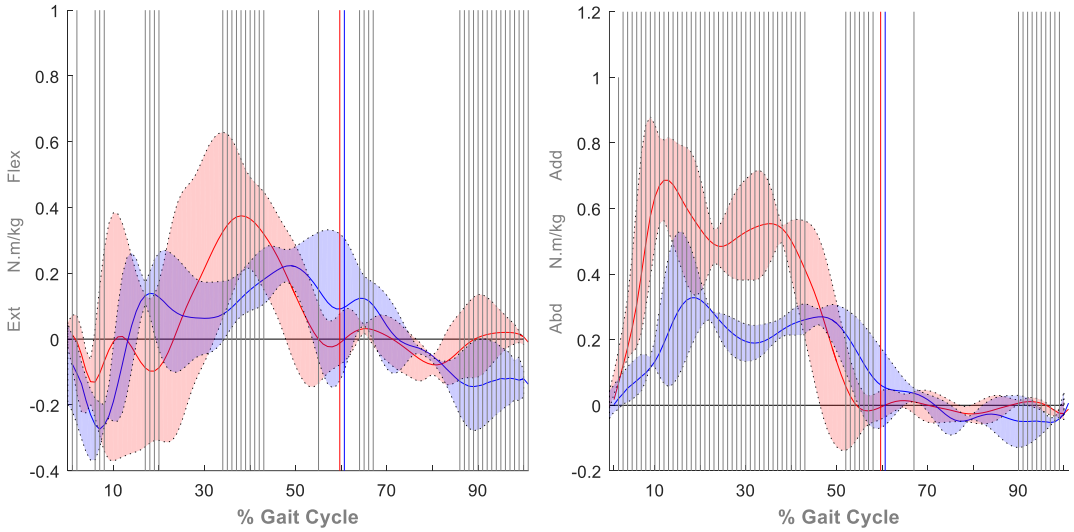

Figure 5-11: Significant differences ($p<0.05$) between pre-and postoperative for the knee flexion (left) and adduction moments (right). Points of significant difference are depicted by a vertical grey line.

Pre-and postoperative KFM traces showed fewer points of significant difference than the KAM trace. The preoperative KFM showed higher variability than postoperative which may explain why fewer significant differences was seen. Almost the whole of the postoperative KAM trace during stance was significantly lower than preoperative which may indicate the success of the procedure as a lower KAM is theorised to relate to improved joint loading.

Hip, knee, and ankle power traces is presented in Figure 5-12 and a table of peak concentric and eccentric powers are shown in Table 5-15.

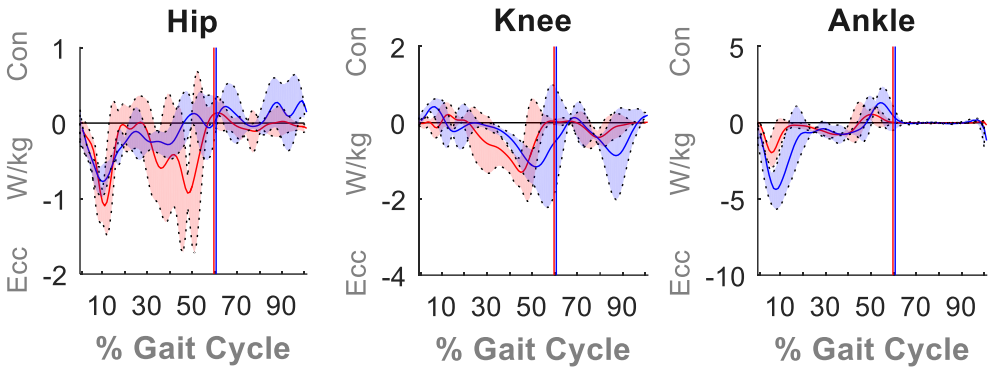

Figure 5-12: Joint powers during stair descent. Red traces = preoperative, blue = postoperative, vertical lines indicate toe off. Shaded bands = ±2 standard deviations from the mean.

As expected with a load absorbing activity, the majority of the power traces were in the eccentric phase. At preoperative, the ankle showed less power absorption which may have been offset by higher proximal joint power. This contrasts postoperative powers where greater eccentric ankle power and lower eccentric hip power was seen. Knee power traces stayed comparable pre-and post-surgery.

Table 5-15: Maximum concentric and eccentric powers during stair descent.

Joint Power (W/kg)	Pre-Op Peak Concentric Power	Post-Op Peak Concentric Power	Pre-Op Peak Eccentric Power	Post-Op Peak Eccentric Power
Hip	0.16±0.1	0.20±0.11	1.23±0.11	**0.80±0.10††**
Knee	0.28±0.11	0.43±0.09	1.43±0.13	1.27±0.36
Ankle	0.75±0.06	**1.51±0.13††**	2.01±0.41	**4.44±0.61††**

Key: †† = significance to preoperative where p<0.01

Peak concentric ankle power occurred around toe off and significantly increased from preoperative (p=0.004, mean difference ± SE: 0.76±0.08 W/kg). Maximum concentric hip and knee powers were not significantly changed post-surgery (p=0.66 and 0.11 respectively). Maximum eccentric hip power significantly decreased (more positive, or less absorptive) at postoperative (p=0.004). Conversely, peak eccentric ankle power was more absorptive (p=0.007, mean difference ± SE: 2.43±0.4 W/kg) compared to preoperative. Lastly, eccentric knee power was statistically similar between operative states (p=0.53).

5.2.3. Summary

Of both stair tasks, stair descent was found to be the more challenging activity. Patient #19's postoperative stair descent STP showed more significant changes (6 parameters) at postoperative compared to stair ascent (1 parameter). Compared to level walking, stair navigation uses a higher ROM, especially in the sagittal plane. Kinematic profiles between stair ascent and descent were generally opposite to each other, as expected based on the complimentary nature of these activities. Patient #19 utilised more pelvic tilt and hip flexion during stair ascent whilst greater knee flexion and ankle dorsiflexion was seen during descent. Whilst the peak and range of knee flexion had not significantly changed between operative states for both activities, peak knee flexion and ROM was reduced at postoperative. This was by ~5^{o} for flexion ROM during descent which was clinically relevant.

Sagittal knee data at heel strike and toe off showed no significant changes for stair ascent, however these parameters were significantly changed for the stair descent. Compared to preoperative more knee flexion at heel strike and slightly less flexion at toe off was reported. This finding may show a slight worsening of function as a flexed knee at foot contact could indicate less stability compared to preoperative. However. the more extended knee at toe off could imply a gain in function.

During stair ascent, no significant differences were observed for kinetic parameters, potentially due to the low sample size of 2 at both operative states, or perhaps there was no change in function here. No instruction was given to participants as to how to complete the stair task so as not to influence gait. Therefore, it was down to chance whether Patient #19 would strike the instrumented step with their operated limb. As significant differences were found in the kinematic data it likely that changes to kinetic data would be expected. More significant changes were seen in the stair descent kinetics, and both activities generally saw decreases in peak knee adduction and ankle dorsiflexion moments at post-surgery. Power trends reflect the generative/absorptive nature of stair ascent and descent respectively. During ascent

the ankle showed less postoperative generating power, and significantly more absorptive power during stair descent which a more flexed knee would allow for.

5.3. Slope Walking

Here, Patient #19 walked at an incline of +7.5° and -7.5° on the Motek CAREN system. Only kinematic data is presented as the force plates were not able to be integrated with the Vicon system. Inertial effects caused incorrect force readings. The treadmill speed was set to self-paced mode where feedback mechanisms sped up or slowed the belt to allow for a more natural walking speed.

5.3.1. Incline Walking

Table 5-16 shows the average spatiotemporal parameters taken from a ten second period when walking uphill at +7.5°. As this activity was carried out on a treadmill, stride and step length have been omitted from analysis.

Table 5-16: STP data for Patient #19 whilst walking uphill at 7.5°.

Parameter	Pre-Op	Post-Op	Percentage Change (%)
Cadence (steps/min)	80.3±0.4	75.7±1.7	-5.7
Belt Speed (m/s)	0.4	0.7	+75.0
Stride Time (s)	1.5±0.0	1.6±0.0	+6.7
Step Time (s)	0.8±0.1	0.8±0.0	0
Opposite Foot Off (%)	25.1±2.3	18.3±2.3	-27.1
Opposite Foot Contact (%)	49.8±5.9	49.2±1.6	-1.2
Double Support (s)	0.7±0.0	0.6±0.0	-14.3

It was not possible to carry out statistical analysis on the STP data for this activity as a single output file of averaged data were generated for each ten second walk. Notable findings were that walking speed near doubled at postoperative whilst cadence decreased implying longer strides took place, also reflected by the increased

stride time. Faster walking speeds also corresponds to the shorter time spent in double support observed and how opposite foot off occurs earlier in the gait cycle at postoperative.

The full three-dimensional lower limb kinematics of Patient #19 during this task is presented in Figure 5-13. N = 5 (preoperative) and 6 (postoperative):

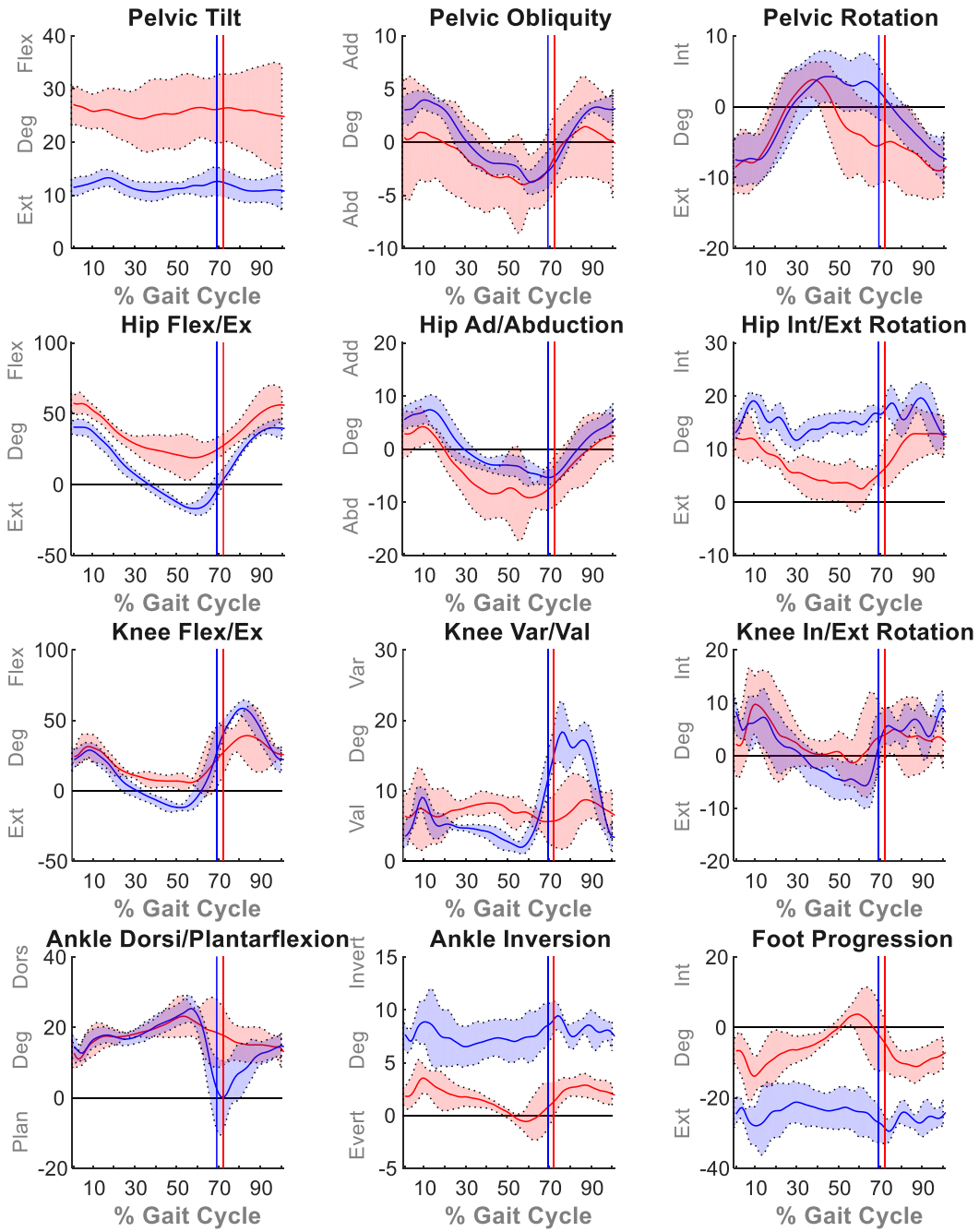

Figure 5-13: Lower limb kinematics during uphill walking. Red traces = preoperative, blue = postoperative, vertical lines indicate toe off. Shaded bands = ±2 standard deviations from the mean.

Traces here are similar to stair ascent kinematics. Preoperative data shows higher variability than postoperative particularly during swing for the pelvis, knee, and ankle. Patient #19 also showed less pelvic tilt and hip flexion, and greater peak and ROM of sagittal knee and ankle motion at preoperative. Postoperative knee adduction revealed larger cross talk effects than preoperative. Rotational knee angles ranged between internal and external rotations at postoperative whereas at preoperative the knee was mostly internally rotated throughout the cycle.

Maximum kinematics and ROM for all lower limb joints is given in Table 5-17:

Table 5-17: Peak kinematics and ROM of all lower limb joints during an uphill walk.

Joint Angle (°)	Pre-Op Peak Angle	Post-Op Peak Angle	Pre-Op ROM	Post-Op ROM
Pelvic Tilt	29.2±1.5	**13.6±0.9†††**	7.7±2.6	**4.1±1.4†**
Pelvic Obliquity	2.9±1.2	4.1±0.5	7.6±1.1	8±0.9
Pelvic Rotation	4.1±1.2	4.4±1.8	13.9±2.4	13.2±1.9
Hip Flexion	61.2±1.5	**42.9±2†††**	43.1±6.3	**60.1±2.1††**
Hip Ab/Adduction	5±0.8	**7.7±1.3††**	15.3±3.2	13.3±1.7
Hip Rotation	14.2±1.9	**20.4±1.2†††**	12.1±1.8	**9±1.5†**
Knee Flexion	43.5±5	**58.9±2.7†††**	38.8±4.8	**71.2±3.3†††**
Knee Ab/Adduction	9.5±1.2	**19±1.7†††**	5.3±1.3	**17.1±2†††**
Knee Rotation	10.2±3	10.4±1.4	12.4±3.3	**16.6±1.2†**
Ankle Flexion	24.1±1.9	25.4±1.9	13.8±1.9	**25.7±3.9†††**
Foot Inversion/eversion	3.8±0.6	**10.2±0.8†††**	4.5±0.5	4.1±1.1
Foot Progression	4.7±2	**-19.9±2.2†††**	19.3±2	**11.5±3†††**

Key: † = significance to preoperative where p<0.05

†† = significance to preoperative where p<0.01

††† = significance to preoperative where p<0.001

For all peak angles, significance was where p<0.001 aside from maximum hip adduction (p=0.002). Peak knee flexion increased by 15.3±2.5º (mean difference ± SE) at postoperative. All joint's sagittal ROM were significantly different at postoperative, the greatest mean difference was seen for the knee of 32.4±2.5º, near double preoperative ROM. Hip flexion (p=0.003, mean difference ± SE: 17.0±2.9º) and ankle dorsiflexion ROM (p<0.001, mean difference ± SE: 11.9±1.8º) also showed clinically relevant changes. Pelvic tilt ROM was the only sagittal parameter to reduce at postoperative, although this was a significant finding (p=0.03) the change from preoperative was 3.6±1.2º which is not clinically relevant.

Sagittal kinematics at heel strike, toe off, and during stance is displayed in Table 5-18:

Table 5-18: Extra sagittal lower limb kinematics during an uphill walk.

Joint Angle (°)	Pre-Op Peak Angle	Post-Op Peak Angle
Pelvic Tilt at Heel Strike	27.1±2	**11.6±0.8†††**
Hip Flexion at Heel Strike	57.5±3.5	**40.7±2.6†††**
Knee Flexion at Heel Strike	24.4±2.3	22.1±4.2
Ankle Dorsiflexion at Heel Strike	12.7±1.9	14.6±1.6
Pelvic Tilt at Toe Off	26.7±2.6	**12.6±1.3†††**
Hip Flexion at Toe Off	27.8±7.1	**-3.9±4.3†††**
Knee Flexion at Toe Off	29.5±5.3	26.4±7.5
Ankle Dorsiflexion at Toe Off	16.3±2.8	**2.8±4.6†††**
Maximum Stance Knee Flexion	31.9±4.4	29.5±2.4
Minimum Stance Knee Flexion	4.8±1.4	**-12.3±1.1†††**

Key: ††† = significance to preoperative where p<0.001

Joints were primarily more extended at postoperative aside from ankle flexion at heel strike. All significance was where p < 0.001. The largest mean difference and standard error from preoperative was seen for hip flexion at toe off (31.7±2.4º). All other

significant changes showed mean differences around 15° between operative states. Knee stance ROM at preoperative was approximately 27.1°, which increased at one year to 41.8° as the operated limb demonstrated increased hyperextension.

5.3.2. Decline Walking

Following the incline walk, Patient #19 then walked downhill at -7.5°. Table 5-19 shows the STP data from this activity:

Table 5-19: STP data for Patient #19 during downhill walking.

Parameter	Pre-Op	Post-Op	Percentage Change (%)
Cadence (steps/min)	116.5±0.0	117.7±1.6	+0.9
Belt Speed (m/s)	1.02	1.3	+27.5
Stride Time (s)	1.0±0.0	1.02±0.0	+2.0
Step Time (s)	0.5±0.0	0.5±0.0	0.0
Opposite Foot Off (%)	13.6±0.0	13.7±0.2	+0.7
Opposite Foot Contact (%)	50.0±3.4	50.5±1.4	+1.0
Double Support (s)	0.3±0.01	0.3±0.0	0.0

Much alike to incline walking, no statistics were able to be computed. The greatest percentage change (an increase of 27.5%) was seen for walking speed. As the other STP's showed postoperative changes no greater than 2%, the faster walking speed may have been facilitated by an increase in stride and step length. As this activity was carried out on a treadmill and the belt and limbs were moving to keep the body roughly centre of the belt there was no way to accurately verify this.

Below, the full lower limb kinematics during decline walking is given in Figure 5-14. N=7 (preoperative), 9 (postoperative):

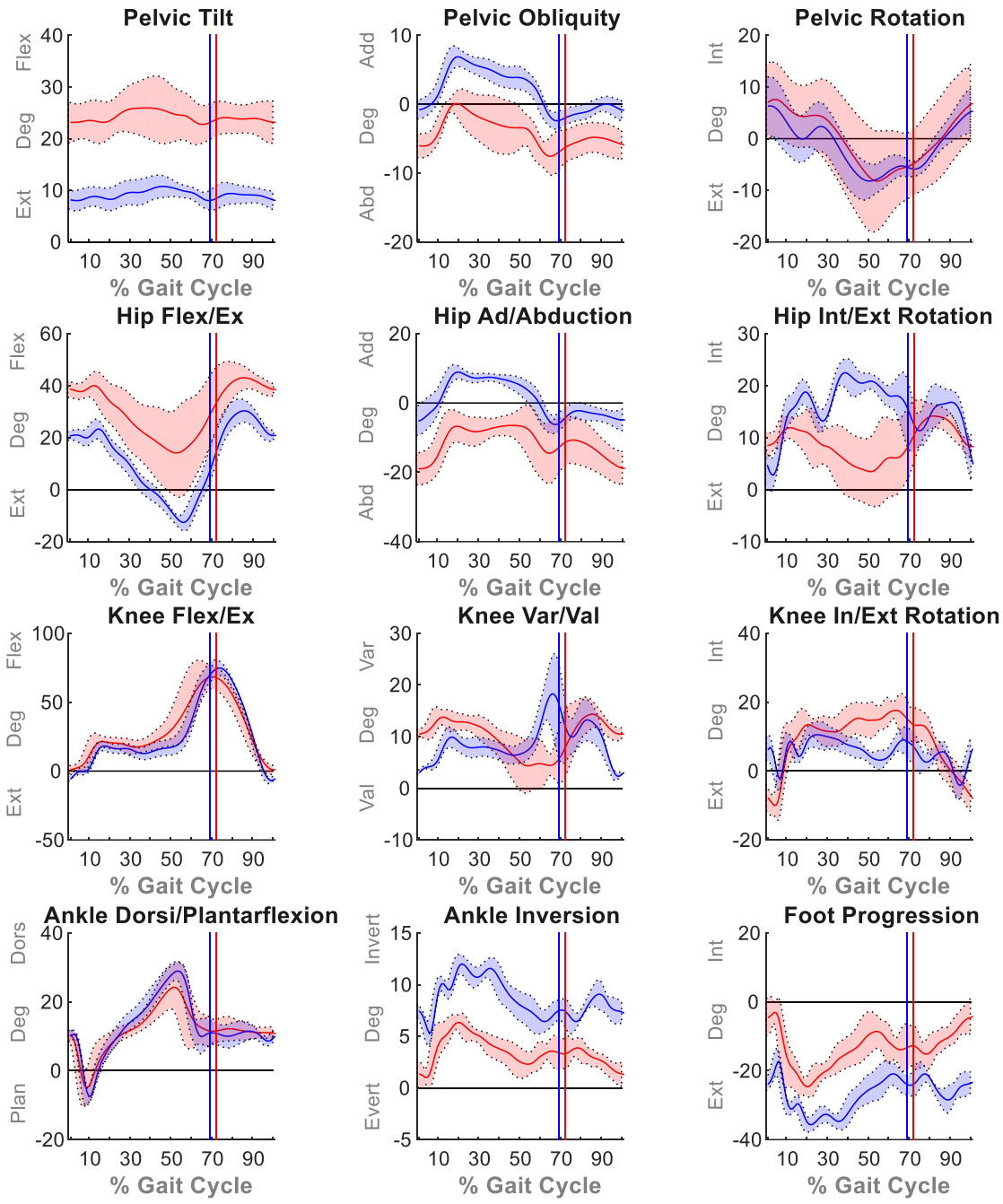

Figure 5-14: Lower limb kinematics during downhill walking. Red traces = preoperative, blue = postoperative, vertical lines indicate toe off. Shaded bands = ±2 standard deviations from the mean.

Overall, traces were similar to level walking and stair descent kinematics. Pelvic tilt and hip flexion graphs were also similar to uphill incline walking, although there was a greater difference between pre-and postoperative traces here. Ankle dorsiflexion here was uniquely different compared to previous activities, the foot was dorsiflexed at foot contact, then went into plantarflexion (keeping with the declined surface),

then returned to dorsiflexion during stance as the treadmill progressed the foot beneath the shank to a maximum dorsiflexion before swing. During swing the foot remained in some degree of dorsiflexion until the proceeding heel strike. Considering the similarities in motion to complete stair descent and decline walking, the discrepancy in hip flexion traces in both activities is substantial. Figure 5-9 shows a hip flexion peak around toe off, whereas Figure 5-14 shows a hip flexion minimum (or peak extension) at toe off similar to level walking. This finding may be due to the fact this activity was carried out on a treadmill where the stance limb was travelling backward meaning the COM remained in roughly the same location. Conversely, during stair descent the COM is progressively lowered, and it was necessary for the limb to be flexed at toe off to allow adequate foot clearance to reach the consecutive step.

Peak kinematics and ROM of all lower limb joints in three-dimensions is displayed in Table 5-20:

Table 5-20: Peak and ROM of all lower limb kinematics during downhill walking.

Joint Angle (°)	Pre-Op Peak Angle	Post-Op Peak Angle	Pre-Op ROM	Post-Op ROM
Pelvic Tilt	26.6±2.3	**10.8±1.1†††**	5.2±2	**3.4±1†**
Pelvic Obliquity	0.3±0.9	**7±0.7†††**	8.3±1	**9.4±0.9†**
Pelvic Rotation	9±3.2	6.9±2.6	17.4±4.6	15.3±2.9
Hip Flexion	43.7±1.9	**30.4±2.2†††**	30.8±7.7	**43.1±2.1††**
Hip Ab/Adduction	-4.9±1.9	**9±0.9†††**	15.6±2.1	16±1.6
Hip Rotation	14.7±0.9	**22.8±1.3†††**	11.9±3.1	**20.1±2.1†††**
Knee Flexion	69.4±3.8	**75±2.7††**	69.4±3.8	**82.1±2.1†††**
Knee Ab/Adduction	14.8±1.3	**19±2.6††**	12.2±0.9	**16.7±2.6††**
Knee Rotation	19.6±1.2	**11.3±1.8†††**	30.3±1.5	**16.3±2.3†††**
Ankle Flexion	25.2±3.1	**29.1±1.3††**	31.2±4.4	**37±2††**
Foot Inversion/eversion	6.6±0.4	**12.1±0.4†††**	5.9±0.7	**6.9±0.7††**
Foot Progression	-1.6±2	**-17.1±1.8†††**	23.8±3	**18.7±1.9††**

Key: † = significance to preoperative where p<0.05

†† = significance to preoperative where p<0.01

††† = significance to preoperative where p<0.001

At postoperative each peak angle except pelvic rotation was significantly different changed. Most p values were < 0.001 aside from peak knee flexion (p=0.004), knee adduction (p=0.001) and ankle dorsiflexion (p=0.008). Peak pelvic tilt showed the largest decrease ± SE between operative states of 15.8±0.9°. ROM significance was generally to a lesser degree than peak angles, and as well as pelvic rotation, the hip adduction ROM was also statistically similar at pre- and postoperative. Where the difference between eight maximum angles gave p<0.001, for joint ROM only five parameters gave the same p value. Both peak knee flexion and ROM significantly increased postoperatively by clinically relevant mean differences and standard errors of 5.6±1.6° and 12.7±1.5° respectively. Despite the similarity of motions, the same knee flexion findings were not reported for stair descent.

Further sagittal kinematics at heel strike, toe off and during stance is shown in Table 5-21:

Table 5-21: Additional sagittal kinematic data during decline walking.

Joint Angle (°)	Pre-Op Peak Angle	Post-Op Peak Angle
Pelvic Tilt at Heel Strike	23.2±2	**8.2±0.9†††**
Hip Flexion at Heel Strike	38.7±1.6	**20.7±0.8†††**
Knee Flexion at Heel Strike	0.9±1.5	**-5.9±0.9†††**
Ankle Dorsiflexion at Heel Strike	10.6±0.8	10±0.8
Pelvic Tilt at Toe Off	23.2±2.2	**9.1±0.6†††**
Hip Flexion at Toe Off	19.5±8.1	**-4.1±2.6†††**
Knee Flexion at Toe Off	57.5±10.7	49.6±5.9
Ankle Dorsiflexion at Toe Off	14±3.7	11.4±2
Max Stance Knee Flexion	22.3±2.1	**18.3±1.6†††**
Min Stance Knee Flexion	17.2±2.6	**12.6±2.2††**

Key: †† = significance to preoperative where p<0.01

††† = significance to preoperative where p<0.001

Significance was mostly where p<0.001 except for minimum knee flexion during stance (p=0.002). Hip flexion at toe off showed the greatest mean difference and

standard error from preoperative (23.4±3.0^{o}). Stance knee flexion ROM from preoperative (5.1^{o}) was similar to one-year post-surgery (5.7^{o}). Again, most sagittal parameters at heel strike and toe off were lower at postoperative indicating the limb was straighter (more extended) at these time points. Ankle dorsiflexion at heel strike however remained equivalent between pre- and postoperative.

5.3.3. Summary

Slope walking is a useful activity to analyse as the joints move in a similar manner to level walking, but as the walking takes place on an incline or decline the COM is being raised or lowered much like in stair navigation. Despite that kinetic data were not available, STP and kinematic parameters are useful to analyse. For slope walking the walking speed increased postoperatively, since cadence either remained the same or decreased it is possible stride and step length also increased to compensate but recording this on a treadmill would not be accurate.

Slope walking kinematics showed large changes from preoperative in particularly sagittal data, with decline walking showing more significance than incline walking. For both activities, the degree of significance was higher when comparing peak kinematic values than ROM (i.e., number of parameters giving a p value <0.001). For sagittal angles at heel strike and toe off, all parameters (aside from ankle dorsiflexion at heel strike when uphill walking and knee flexion at toe off for downhill walking) showed a reduction from preoperative (less flexed joint). The increased ankle dorsiflexion for incline walking corresponds to the nature of the activity, and the extension of the other joints indicates a confidence in stability.

5.4. Patient Reported Outcome Measurements

A couple of self-reported outcome measures were employed for patients, firstly the Oxford Knee Score (OKS) which ranged between 0 and 48 and a lower score indicates an improved outcome. Patient #19's preoperative OKS was 35 which decreased by seven points to 28 at one-year post-surgery showing improvements for this outcome measurement. After each activity, Patient #19's responses to how difficult and painful

she found each activity were also recorded. From five answers increasing in severity Patient 19's responses for each activity are shown in Figure 5-15.

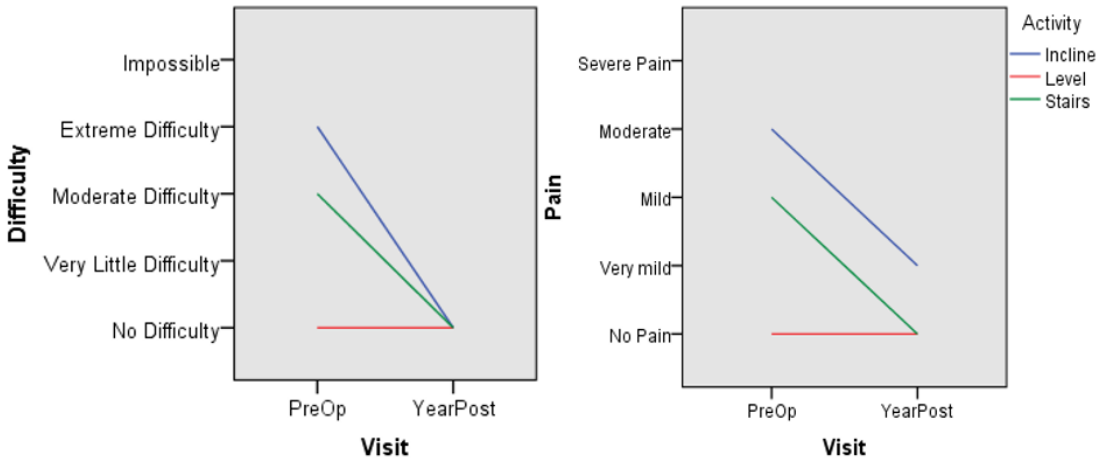

Figure 5-15: Patient #19 reported answers for how much difficulty and pain was experienced during each ADL.

Although not further divided into stair ascent/descent and uphill/downhill walking it is clear that the slope walking activity was the most difficult and painful activity. Level walking remained neither difficult nor painful at both operative states. At post-surgery no difficulty was experienced for any activities, and very mild pain during inclined walking was reported. Questions asked but not depicted was how tiring Patient #19 found each activity. Although across both operative states some difficulty and pain was experienced, no tiredness was reported by Patient #19 .

Further questions were also asked about the stair and incline tasks specifically, Patient #19 answered that the dimensions of the laboratory staircase resembled her own at home, and pre-surgery she would normally ascend a flight of stairs using two bannisters and descend using a single bannister. At postoperative this answer changed to using a single bannister for both ascent and descent. Bannister reliance indicates low confidence in joint performance so by reducing from two bannisters to one at postoperative indicates an improvement in joint function. For the incline task, Patient #19 stated that the inclinations walked on the CAREN system were similar to the inclinations experienced in everyday life, giving confidence that the study design is appropriate for this patient group.

5.5. Discussion

For Patient#19, some common findings were observed across all activities before and after the TKA. Regarding STPs, walking speed consistently increased post-surgery, and the period of double support simultaneously decreased. There was usually a complementary increase in cadence (Urwin et al., 2014), except for the sloped walking activities where cadence decreased. In this case, increases in speed and decreases in double support may have been facilitated by increases in stride and step length. However, it was not possible to analyse these parameters as this activity was treadmill-based.

Level and slope walking showed lower peak and range of sagittal knee angles compared to the stair tasks as in agreement with other studies (Jevsevar et al., 1993; Laubenthal et al., 1972). Intra-operative changes between peak and range of knee flexion angles were less predictable. For the stair tasks, both peak and range of knee flexion decreased at postoperative (although not a significant nor clinically relevant finding). During slope walking the opposite was seen where both parameters considerably increased at postoperative by a significant amount. The disparity in findings between the slope and stair activities may be due to preoperative unfamiliarity with the CAREN system particularly walking on a self-paced treadmill for a first time. Due to the less-able nature of TKA patients a full familiarisation period was not feasible as this could fatigue the participant prior to data capture so care was taken to analyse periods of most comfortable walking. The largest peak knee flexion and ROM was seen during stair descent and this activity is most likely to show differences between implant groups. During level walking, Patient #19's peak knee flexion significantly decreased whereas ROM significantly increased from preoperative indicating more extension was realised, typical with TKA (Tew and Forster, 1987).

The four pelvic markers orientate the pelvis in space and any vertical offset in marker position could cause the pelvis and hip kinematics to be misreported. For each activity's kinematic gait traces, a noticeable shift was evident between pre- and

postoperative sagittal pelvis and hip traces, and to a lesser extent, frontal pelvis and hip angles. Raising the ASIS markers raises the midpoint between markers such that an overly posteriorly tilted pelvis (relative to the laboratory global axis system) is reported along with an increase in hip extension (Baker, 2013). This change in hip angle is a result of the estimated position of the hip joint centre also becoming altered. Pelvic obliquity for that side will then be over reported and little effect on pelvic rotation is seen. Conversely, if the PSIS markers were too high then the pelvis will be tilted more anteriorly, and the hip will be seen to be more flexed than actual. Practically, around 3^{o} of tilt is seen for every 10mm of PSIS displacement. If a single pelvic marker is erroneously placed higher than the rest, then the effects are halved as midpoints are raised by half the misplacement of that marker. Lateral marker misplacement has little effect on gait traces since the midpoint between markers determines the anatomical reference system. In practice a small change of 1^{o} is seen for every 10mm of lateral marker misplacement. Since peak values may be less reliable for these joints, but ROM will still accurate, the group analysis will include only pelvic and hip ROM. As the KAD determines the knee and ankle coordinate systems, peak and ROM data will be presented for these joints in the group analysis. Whilst the knee kinematics also rely on accurate hip joint position inferred from the ASIS markers, the errors in peak knee kinematics values are considerably smaller than the effects on the hip and pelvis.

Kinematic cross talk (unfeasibly greater adduction angles at high knee flexion) was evident at both operative states for Patient #19. This stems from a misplacement in markers or, in this instance, the knee alignment device causing the orthogonal knee motions to not align to the correct planes. Whilst sagittal motion is largely less affected, frontal and transverse motion is misrepresented. During level walking, peak knee adduction was around 15^{o} and 18^{o} at pre- and postoperative respectively. Knee adduction ROM during this activity also rose from 11.4^{o} to 14.6 at postoperative. Generally, the peak and range of knee adduction increased at postoperative for all activities other than stair ascent where both parameters were reduced at postoperative. Conversely, knee rotational peak and ROM angles decreased post-

surgery for all activities except for uphill walking where more internal rotation was seen. Peak rotational angles and ROM during level walking was 18.5^{o} and 26.4^{o} respectively at preoperative which decreased considerably to 7.1^{o} and 8.5^{o} post-surgery.

Compared to studies investigating non-sagittal knee kinematics during level walking, the findings from Patient #19 are much higher, for instance Komnik et al., (2016) found peak knee adduction and rotation angles of ~2.4^{o} and 0.49^{o} respectively. Furthermore, the frontal and transverse plane ROM was 1.9^{o} and 11^{o} meaning the knee did not go into abduction, but did externally rotate, which was also seen in Patient #19. Although Patient #19's peak adduction and rotation angles was much higher than in Komnik's study it should be noted that they extracted values during stance only, whereas values from the whole of the gait cycle were extracted here. The gait traces in Figure 5-1 clearly shows that knee adduction during stance was still over 5^{o} and, in some instances reached over 10^{o} (at preoperative) which indicate some degree of error.

Studies investigating rotational knee angles in OA and healthy participants found OA patients had less knee rotational ROM than healthy controls (Bytyqi et al., 2014). Healthy control limbs tended to remain slightly internally rotated during stance which peaked during pre-swing, proceeded by external rotation during swing, reaching a peak at mid-swing. Patient #19 showed a similar trend, although preoperative ROM was considerably greater (26.4^{o}) compared to the OA patients in Bytyqi's study (7.6°). Patient #19's postoperative ROM (8.5^{o}) was acceptable especially compared to the control group in Bytyqi's study who showed a ROM of 9.3^{o}. No TKA procedures were carried out in Bytyqi's study so postoperative comparison is not possible.

Drawing conclusions from Patient #19's kinetic data were more difficult to do, specifically for stair ascent where the number of trials where operated limb kinetics was available was low (n = 2). And such, statistical analysis of this activity's kinetic findings is underpowered. Nonetheless, statistical analysis was carried out and some significance was found. On the whole, sagittal joint and knee adduction moments

showed more significant changes than the other joints and planes and greater peak moments were reported from level walking and stair descent than stair ascent. All knee flexion and adduction moments during level walking declined from preoperative (tending towards negative), in line with findings from Mandeville et al., (2008). Knee moments during stair descent also generally showed postoperative reductions aside from the peak knee extension moment which increased from preoperative (became more of an extension moment).

There is known to be a link between faster walking speeds and increased kinetics (Fukuchi et al., 2019). Although walking speed increased for all activities, knee adduction moment traces decreased from preoperative. This finding is corroborated by previous studies (Hatfield et al., 2011; Hilding et al., 1996; Orishimo et al., 2012) where it was concluded faster walking speeds with reduced peak moments indicate improved loading. For all relevant activities, Patient #19's peak dorsiflexion moment and peak and range of dorsiflexion angle increased from preoperative. As low dorsiflexion moments are associated with a stiffer (more co-contraction) gait (Hatfield et al., 2011; Lamontagne et al., 2000) it could be inferred that the higher dorsiflexion moment is a result of improved gait. Stair ascent and descent ankle dorsiflexion moment traces (Figure 5-6 and Figure 5-10) showed good similarity to those reported by Lin et al., (2004). Differences between traces may be attributed to the fact that Lin's study used healthy participants.

Level walking power trends generally showed more concentric postoperative powers (more generating and less absorbing), except for the hip which also showed more eccentric (absorptive) powers. Stair navigation showed typical power traces, where generating powers were greater during ascent and absorptive powers were greater in descent. Between operative states, powers during stair ascent were generally less concentric, except for the ankle which became slightly more generative. The reverse was found during descent where mostly peak powers became more generative and less absorptive between operative states. Again, this excludes the ankle which was significantly more absorptive. The power traces during stair navigation also

correspond well to those presented by Lin et al., (2004). Raising the body to ascend a step was achieved by concentric hip and knee power after heel strike and at pre-swing whilst stair descent required eccentric lower limb powers to allow the limb to support body weight and progress forwards and downwards. These powers are greater compared to level walking and are matched by a larger range in kinematic profile.

The level walking power traces in Figure 5-4 shows a similar trend to those presented by Levinger et al., (2013). Although no comparison to controls was made in this case study, the lower post-surgery generative hip power (particularly during early stance) and higher generative ankle power trends match Levinger's findings. A greater postoperative concentric ankle power and dorsiflexion moment was observed here, so it is likely that angular ankle velocity increased or remained similar between operative states. However joint velocities were not analysed in this study so this cannot be confirmed. Knee power between this case study and Levinger's findings were less similar, Patient #19 showed more concentric and less eccentric knee power from preoperative, contrasting the increased eccentric power seen in Levinger's study. Patient #19's peak eccentric knee power at pre-swing decreased postoperatively (tending towards zero) and minimum knee flexion during stance showed more extension. These two findings may reflect mechanical changes as a result of surgery where the quadriceps is better able to stabilise the joint during single leg stance.

5.6. Conclusions

This case study reported pre- and postoperative biomechanical data and questionnaire responses during various ADLs of a single patient recruited into this study. The parameters analysed included spatiotemporal parameters, three-dimensional joint kinematics (such as peak angles and ROM) and joint kinetics (moments and powers). The low standard deviation and error found here provides confidence in the data recording and processing techniques employed. Many results and trends were similar to literature figures providing further confidence to data

processing. Based on previous findings the following discrete parameters will be analysed for all patients:

- Spatiotemporal parameters: Cadence, walking speed, stride time, opposite foot off and foot contact, period of double support. (no stride/step lengths as these are not height or leg length normalised per participant by the data processing software, and it was assumed increased kinematics would relate to larger step/stride lengths)
- Joint kinematics: peak values (knee flexion, adduction, and rotation, and ankle dorsiflexion), range of motion (for pelvic tilt, hip and knee flexion, knee adduction and rotation, and ankle dorsiflexion). Also exported was knee flexion at heel strike and toe off, and maximum and minimum knee flexion during stance.
- Joint kinetics: maximum and minimum hip, knee, and ankle sagittal moments, knee adduction moments, and peak concentric and eccentric hip, knee, and ankle powers.
- Participant reported questionnaire answers for each activity.

Appropriate statistical analysis will be used to compare between implant groups' average findings (at the same operative state), each implant group to the control group, and intra-implant pre– and postoperative findings of the same implant group.

6. Group Results

Building on the previous case study, the same parameters were exported for all recruited patients carrying out the same ADL's. These were collated into groups depending on the implant received and averaged to give an average value per knee implant type. As the focus here is on the functional performance of each implant no comparisons to the unoperated limb were made as the patient's unoperated limb ranged from healthy to symptomatic could not be standardised.

6.1. Recruitment and Demographics

A total of 81 patients have been recruited into the study. Of these, 67 patients came for their preoperative testing session and fourteen patients did not attend their preoperative session. Seven out of the 67 later withdrew postoperatively for health reasons. 29 patients returned for one-year operative testing at Strathclyde. As three patients did not come for preoperative testing they were excluded from the analysis. In addition, two patients did not receive a B.Braun Columbus implant and were also excluded from the study giving a total of 9 control participants and 24 patient participants were analysed (Figure 6-1).

6. Group Results

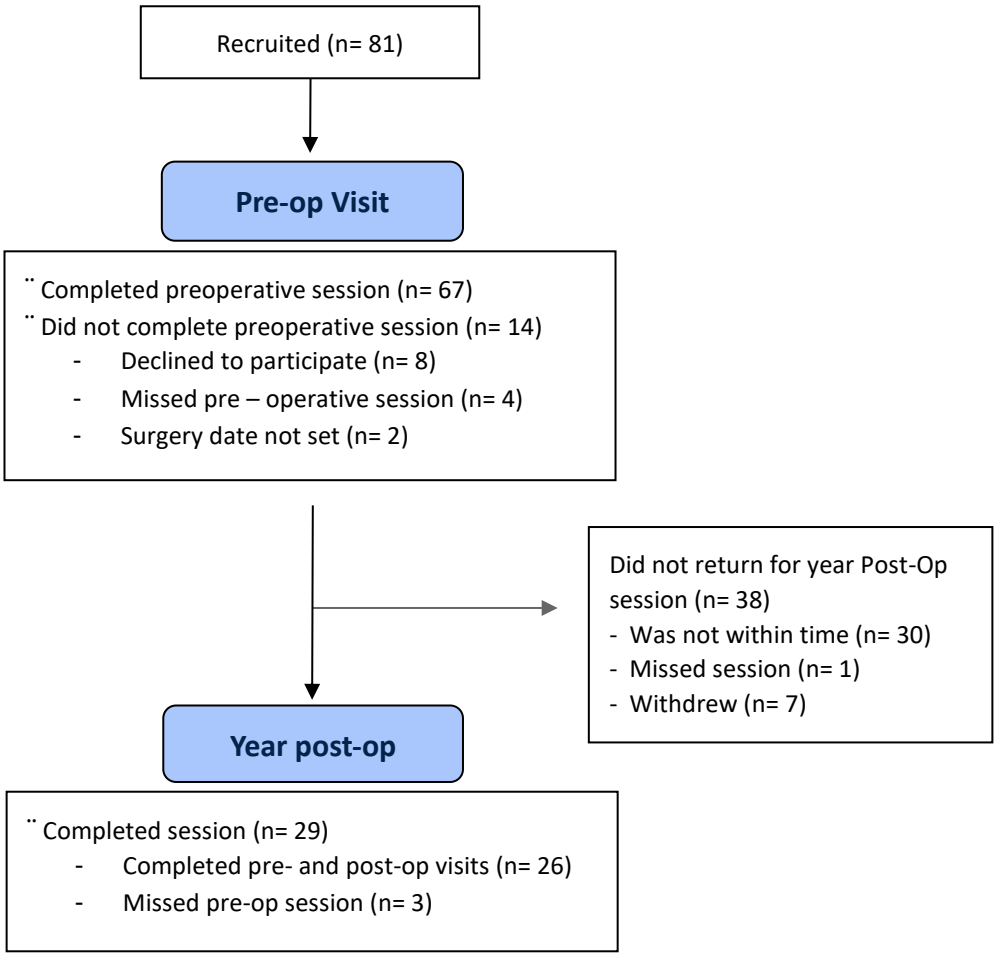

Figure 6-1: Flow chart depicting patient recruitment into the study.

The participant anthropometrics are given below in Table 6-1. The OKS recorded for each patient group records the pain felt during various activities, a lower score meaning less pain. Implant groups are as so: low congruency fixed bearing: cruciate retaining, deep dish implant (CRDD), high congruency fixed bearing: ultra-congruent implant (UC) and the high congruency mobile bearing: ultra-congruent rotating knee (UCR).

Table 6-1: Control and patient participant demographics.

Demographic	Control	CRDD	UC	UCR
n	9	12	5	7
Male/Female	**3/6[1]**	11/1	**1/4[1]**	**3/4[1]**
Age, mean±std (years)	70±6.4	66.6±4.0	68.4±8.8	70.1± 6.2
Affected side (L/R)	n/a	5/7	4/1	5/2
Pre BMI, mean±std (kg/m^2)	24.0±3	**30.0±2.5****	**32.0±5.9****	**30.3±3.3****
Year BMI, mean±std (kg/m^2)	n/a	**30.1±2.5****	**31.6±6.1****	**29.6±3.4***
Pre Oxford Knee Score (60)	n/a	35±7	42±4	36±5
Year Oxford Knee Score (60)	n/a	**24±8††**	**19±4†**	**22±8††**

Key: *** / ** / ***** = significance where p<0.05, < 0.01 and < 0.001 respectively between the selected patient group to controls.

† / †† / ††† = significance where p<0.05, < 0.01 and < 0.001 respectively for the selected implant group between preoperative and postoperative.

[1,2,3] = significance where p<0.05 between the selected group to CRDD (1), UC (2) or UCR (3) implant at the same operative state.

Nominal categorical data (gender and affected side) was analysed between groups using multiple Fischer's Exact test with a Bonferroni correction. Overall, the cruciate retaining, deep dish (CRDD) group showed a significantly higher gender imbalance compared to all other groups (p<0.05). Between implant groups, there were no significant differences in the operated side. Numeric age, BMI, and Oxford Knee Score (OKS) data were analysed between control and patient groups using a one-way ANOVA test with a Bonferroni correction. All patients showed a significantly higher BMI to controls (p<0.01) by 6kg/m^2. No significance in age and OKS was noted between groups. Between operative states, paired T-tests showed no change in BMI, however the OKS was significantly reduced (p=0.004, 0.012 and 0.002 for the CRDD, UC and UCR implants respectively). Despite showing a large postoperative decline in OKS (around double of the other implants) and smaller standard deviations, the degree of significance given by the UC group was less than the CRDD and UCR groups. This may be attributed to the smaller sample size in this group.

6.1.1. Experimental Considerations

Outlined below is additional information to be considered regarding the biomechanical data recorded during this study.

6.1.1.1. Stair Navigation Strategies

As patients ascended and descended the staircase with no instruction, a few variables influenced the analysis - namely the strategy utilised to carry out the activity. Firstly, whether participants used the bannister would possibly affect the kinetic data as some load is transferred through the upper body. In most cases this was assumed to be negligible as upon visual inspection most patients were resting their hand on the bannister as opposed to gripping and pulling themselves along. However, this is a factor that is unable to be accounted for. Another factor would be whether the patient used a step-by-step strategy (SBS) or a step-over-step (SOS) strategy. This would affect joint kinematics as a lower ROM would be seen, and joint kinetic data would be affected since both feet striking the same step would give error in the ground reaction force readings in SBS walkers. Lastly as only one step was instrumented, whether the patient struck that step with their operated limb was up to chance affecting data available for that limb reducing power for this parameter. The characteristics of participant groups are summarised below:

Table 6-2: Strategy and sample sizes of operated limb kinetic data for the stair navigation tasks.

Participant Group	# Participants who used a SBS strategy	Operated limb kinetics available during stair ascent	Operated limb kinetics available during stair descent
Control	0	n/a	n/a
Pre-op	5	12	12
Year-post	1	17	8

Four patients improved at postoperative to walking with a SOS strategy from SBS at preoperative. Only one patient (Patient #6) walked with a SBS strategy on ascent and descent and pre and postoperative. The gait characteristics of Patient #6 during this activity were not analysed as gait events (heel strike and toe off) identification was unreliable, and normalising Patient #6's data to find average values would be prone

to error. At preoperative, Patient #6 also ascended and descended the stairs using both bannisters, and at postoperative only a single bannister was sporadically used during stair descent indicating some improvement in joint function here.

From 24 patients, just half of the potential data were available for kinetic analysis during this activity at preoperative which slightly increased at postoperative. As this leaves this analysis to be considerably underpowered the kinetic results from this activity will not be included in the main result chapter and will be presented in Appendix 10: Stair Navigation Kinetic Data.

6.1.1.2. Incline Walking

For this activity, participants were asked to walk at various inclinations on the Motek CAREN system. Some were unable to carry out this activity due to either having limited mobility, or a laboratory hardware error. A table to summarise the number of participants who did or did not complete each incline and the reason why is given below.

Table 6-3: Summary of available datasets for the incline walk task at each incline.

Patient Group	# People completed Incline/ # People that had mobility difficulties/ # Incomplete due to hardware error				
	-7.5°	-5°	0°	+5°	+7.5°
Control	9/0/0	9/0/0	9/0/0	9/0/0	8/0/1
Pre-op	16/7/2	19/4/2	24/0/1	19/4/2	17/4/4
Year-post	14/5/6	16/3/6	20/0/5	19/0/6	17/2/6

The year postoperative group were most affected by hardware issues as up to six patients' data were not recorded. The available data from other patients were analysed although the power behind this analysis will be lower than other activities.

6.1.1.3. Sample Sizes

Continuing from the prior considerations a summary of all sample sizes for each ADL is given below in Table 6-4. The values below refer to both the available spatiotemporal parameter data and kinematic data used for the group analysis.

Table 6-4: Sample sizes of all participants' STP and kinematic data for each ADL.

Activity	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	12	12	5	5	7	7	9
Stair Ascent	10	12	4	5	5	6	9
Stair Descent	9	11	2	4	3	5	9
0º	12	7	5	4	6	7	9
+5º	10	7	5	4	5	7	9
+7.5º	9	7	4	4	4	5	7
-5º	11	7	4	3	3	5	9
-7.5º	10	7	4	3	2	4	9

STP and kinematic data were analysed for all participants for the level walking task. Sample sizes decrease from this as participants were either excluded from analysis for either, walking with a step-by-step strategy for the stair tasks or being unable to complete an activity due to joint pain or hardware issue (incline/decline walking on the CAREN system).

The average number of individual trials recorded per person was twelve for overground level walking (range: 11-14 in controls, and 7-24 in patients) and five each for stair ascent and descent. In the CAREN system, similar strides were seen when walking at 0 º and downhill (-5º and -7.5º) of ten strides in controls (range 7-12) and seven or eight strides in patients (range: 4-11). Patients walking uphill (+5 º and +7.5º) gave six strides on average (range: 4-10), and controls gave eight strides (range: 7-11).

For kinetic data, only the data from the patients' operated limb, and from the randomly selected control limb was included for analysis. Table 6-5 shows the sample sizes of all participants during level walking and stair navigation.

Table 6-5: Kinetic data sample sizes of each participant group for the level walking and stair tasks.

Activity	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	12	12	5	5	7	7	9
Stair Ascent	8	9	2	4	3	3	8
Stair Descent	8	3	1	3	3	2	7

As mentioned, the kinetic data from overground level walking will be presented in the main body of this thesis with stair kinetics in the appendix. The number of trials where kinetic data were available per participant averaged four trials in controls (range: 3-6), and three in the patient groups. A maximum of seven trials per patient was seen at both operative states, and a minimum of one (preoperative) and none at postoperative were also noticed on three occasions.

6.2. Comparison Overview

From Chapter 6.3 onwards STP and kinematic data will be pooled per parameter for the level walk, stair ascent and descent, and incline walking tasks at five different inclinations. The kinetic data is of the overground level walking task only, divided by peak moments and powers. The three tables below show a high-level overview of every significant finding in each patient group: compared to the healthy control group (Table 6-6), compared to the preoperative state of the same implant (Table 6-7) and between that implant and other implant of the same operative state (Table 6-8).

Below the number of activities where a significant difference per parameter was found is depicted, if a pre or postoperative implant group consistently showed a significantly different value to (for instance) controls for each activity, this is indicated with a darker red colour. Where fewer activities showed a significant difference to controls this is indicated with a gradually paler colour according to the key at the foot of the table, and green cells indicate where no activities showed a significant difference to the comparative group. The general direction of the change from the referenced group to the comparative group is indicated by a plus (+) or minus (-) symbol. For instance, referring to the row depicting significant cadence differences

in Table 6-6, each of the implant groups aside from the post-UC group showed a significantly lower cadence to controls. The pre-CRDD and UCR groups also showed a lower cadence in more activities (5 or 6) than all other relevant patient groups (3 or 4).

Table 6-6: Summary of all significant differences between the relevant patient group to controls only. Cell colour indicates the number of activities the patient group showed a significant difference to controls as indicated by the key.

Parameter	CRDD		UC		UCR	
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op
Spatiotemporal Parameters						
Walking Speed	-	-	-	-	-	-
Cadence	-	-	-		-	-
Stride Time	+	+	+		+	+
Opposite Foot Off	+	+	+	+	+	+
Opposite Foot Contact					+	
Double Support	+	+	+	+	+	+
Kinematics						
Peak Knee Flexion	-	-	-	-	-	-
Peak Knee Ab/adduction (whole cycle)		+				-
Peak Knee Ab/adduction (stance only)		+				
Peak Knee Rotation		-				-
Peak Ankle Flexion						
ROM: Pelvic Tilt					+	+
ROM: Hip Flexion	-		-		-	-
ROM: Knee Flexion	-	-	-	-	-	-
ROM: Knee Ab/adduction (whole cycle)		+				
ROM: Knee Ab/adduction (stance only)						
ROM: Knee Rotation		-	-	-	-	-
ROM: Ankle Flexion			-	-		
Knee Flexion at Heel Strike		-		-		-
Knee Flexion at Toe Off		-		-		
Stance Knee Flexion Peak	-	-	-	-		-
Stance Knee Flexion Minimum	+					
Kinetics (level walking only)						
Hip Flexion Moment	-		-		-	-
Knee Flexion Moment		-		-		-
Knee Ab/adduction Moment	+				+	
Ankle Flexion Moment	-		-		-	-
Hip Power	-	-	-	-	-	-
Knee Power	-	-	-	-	-	-
Ankle Power			-		-	

Key:	1 - 2	3 - 4	5 - 6	7 - 8	0

6. Group Results

Compared to controls, all patients showed a significantly lower walking speed, later point of opposite toe off, and longer period spent in double support. The point of opposite foot contact was largely close to controls except for the pre-UCR group. Except for opposite foot contact, each spatiotemporal parameter showed a significant difference to controls for at least one ADL between controls and the CRDD and UCR implant groups at both operative states. Of the two, the UCR group showed significant differences to controls during more activities than the CRDD group (darker red cells). The post-UC group showed STPs closer to controls than the other implants.

Peak patient angles showed more similarities to controls than ROM, particularly for the knee rotation and ankle dorsiflexion parameters. Both the peak and range of knee flexion angle were consistently significantly smaller in patients than controls. Postoperative peak knee flexion was significantly less than controls for more activities than preoperative (cell colour is darker postoperative). The same was seen in knee flexion ROM for the UCR group, whereas the other implants showed a significantly lower ROM to controls in less activities post-surgery. Significant additional sagittal knee data (flexion at heel strike and toe off, and maximum and minimum stance knee flexion), were lower in patients than controls aside from the preoperative minimum knee flexion during stance for the CRDD group. For these four parameters, the pre-UCR group were statistically similar to controls which changed at postoperative, possibly implying a worsening of joint function. Where patients presented significantly higher kinematics to controls for an ADL were the adduction angles in the CRDD group and pelvic ROM for the UCR group.

Kinetic data were only available for level walking and here maximum and minimum findings are merged together as trends were similar. Aside for the knee adduction moment, all significant differences to control were generally a decrease. Postoperative peak knee flexion and extension moments were significantly lower than controls for all implant groups. Significant patient concentric (generative) and eccentric (absorptive) power values also generally showed consistently lower (tending towards neutral) powers compared to controls.

Table 6-7 shows the number of activities where a significant intra-implant change was seen for a parameter between operative state for each implant group:

Table 6-7: Number of significant intra-implant changes per parameter per implant. Cell colour indicates the number of activities the patient group showed a significant difference to preoperative as indicated by the key.

Parameter	CRDD	UC	UCR
Spatiotemporal Parameters			
Walking Speed			
Cadence			
Stride Time			
Opposite Foot Off	-		
Opposite Foot Contact			-
Double Support		-	-
Kinematics			
Peak Knee Flexion			-
Peak Knee Ab/adduction (whole)	+		
Peak Knee Ab/adduction (Stance)	+		
Peak Knee Rotation	-		
Peak Ankle Flexion			
ROM: Pelvic Tilt			
ROM: Hip Flexion	+	+	+
ROM: Knee Flexion	+		-
ROM: Knee Ab/adduction (whole cycle)	+		
ROM: Knee Ab/adduction (stance only)	+		
ROM: Knee Rotation			
ROM: Ankle Flexion			
Knee Flexion at Heel Strike	+	-	
Knee Flexion at Toe Off			
Stance Knee Flexion Peak			
Stance Knee Flexion Minimum	-	-	
Kinetics (level walking only)			
Hip Flexion Moment	+		
Knee Flexion Moment	-	-	
Knee Ab/adduction Moment	-		
Ankle Flexion Moment	+		
Hip Power	+		+
Knee Power	+	+	
Ankle Power		+	

Key:	1 - 2	3 - 4	5 - 6	7 - 8	0

The number of parameters where no significant changes from preoperative were seen for a single activity were highest in the UC and UCR groups (respectively 22 and 23 green cells each). Comparatively the CRDD implant group showed considerably fewer parameters where no significance from preoperative was seen, fourteen parameters altogether. In this instance, either a positive or negative change from preoperative may be seen as either an improvement in joint function or worsening.

From the number of spatiotemporal parameters where a significant difference to controls was seen (Table 6-6), three of these were significantly higher than controls for a number of activities: stride time, the point of opposite foot off and total period of double support. Table 6-7 showed significant intra-implant reductions from preoperative for these same parameters. The point of opposite foot off and contact for the respective CRDD and UCR implants postoperatively lowered (meaning opposite foot off or contact occurred earlier in the gait cycle), and period of double support also significantly decreased for the UC and UCR implants for up to two activities. These findings imply an improvement in joint performance. No other spatiotemporal parameters showed significant intra-implant changes.

The CRDD group gave more significant kinematic changes from preoperative (nine parameters) as opposed to three changed parameters each for the UC and UCR groups. The CRDD group also showed significant changes in more activities compared to the other groups (depicted by the darker blue cells). The only parameter where all groups showed the same trend was for hip flexion ROM, where post-surgery increases was seen. For knee flexion ROM, compared to preoperative the CRDD group showed a significant post-surgery increase in ROM, the UC group presented no significant change, and the UCR group showed a significant decrease. Furthermore, the UCR group was also the only implant to show a significant decline in peak knee flexion angle, although this was only for two activities at most. Whilst no implant group presented a significant intra-implant change in peak knee flexion during stance, the CRDD and UC implant groups showed a significantly lower postoperative

minimum knee flexion during stance for up to four activities indicating more knee extension was realised.

Regarding the kinetics during level walking, the UCR group showed only one significant change from preoperative, which was an overall increase in hip power. Regarding moments, the UC group showed a significant decrease in sagittal knee moments only, and the CRDD group showed increases in sagittal hip and ankle moments. Decreases in knee flexion and adduction moments were also seen in the CRDD group. Postoperative power values compared to controls were generally significantly lower, and generally higher compared to preoperative.

Table 6-8 presents the significance of the referenced implant group to another group of the same operative state (inter-implant changes), for ease of comparison significant decreases to another implant was highlighted. As no peak angles, additional sagittal kinematics, nor power values showed any significant inter-implant differences, these parameters were excluded below.

Table 6-8: Number of significant inter-implant changes per parameter between implants of the same operative state (indicated in the bracket). Significantly lower differences are indicated by a yellow background.

Parameter	CRDD		UC		UCR	
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op
STP						
Walking Speed						
Cadence						
Stride Time						
Opposite Foot Off			- (CRDD)			
Opposite Foot Contact	- (UCR)					
Double Support					- (UC)	
Kinematics						
ROM: Pelvic Tilt		- (UC, UCR)				
ROM: Hip Flexion						
ROM: Knee Flexion				- (CRDD)		- (CRDD)
ROM: Knee Ab/adduction						
ROM: Knee Rotation						
ROM: Ankle Flexion			- (CRDD)	- (CRDD)		
Kinetics (level walking only)						
Hip Flexion Moment						
Knee Flexion Moment						
Knee Ab/adduction Moment						
Ankle Flexion Moment						- (CRDD)
Key:	A significant difference was seen for 1 – 2 activities				No significance was seen	

Significant inter-implant differences were less frequently seen compared to intra-implant differences or to controls. Pre-surgery function was generally similar between implant groups, with most significant differences occurring in the spatiotemporal parameters where the CRDD group showed a significantly later point of opposite foot off to the UC group, and earlier point of opposite foot contact to the UCR group. The UC group also showed a significantly longer period of double support than the UCR group. The only different pre-surgery joint kinematics were seen between the UC and CRDD groups where the former showed significantly higher ankle dorsiflexion ROM.

At one-year post-surgery, the CRDD group exhibited lower pelvic tilt ROM and greater knee flexion ROM than the other implants, and greater dorsiflexion ROM than the UC group. No changes were seen between the post-UC and UCR groups. Lastly, kinetic

parameters during level walking generally revealed no significant inter-implant differences apart from a significantly lower ankle dorsiflexion moment seen in the UCR group to the CRDD group at postoperative.

Whilst the above tables highlight the number of activities where a significant change to a particular group was observed for a certain parameter, it should be noted that the degree of significance is not depicted, and more detail will be given in the following sections. For Table 6-6 and Table 6-7 it is clear that paler, non-green cells show significant difference for fewer activities than a darker cell which is a further area to investigate. Additionally, insignificant differences that show similar or perhaps larger mean differences than those of significant values are not highlighted but are still valuable in gauging functional performance. Lastly, the considerably variation in sample sizes between participant groups will inherently reveal more significant differences in the CRDD group than the other two which have underpowered samples.

6.3. Spatiotemporal Parameters

This section gives various spatiotemporal parameters (cadence, walking speed, stride time, point of opposite foot off and contact, and period of double support) of all participants during each of the activities of daily living (ADL). Data is divided per parameter and all significant values (either to control group, preoperative, or another implant) are highlighted in bold with a shaded background.

6.3.1. Walking Speed

Below in Table 6-9, the walking speed of all participants for each activity is shown:

Table 6-9: Walking speed of all participants for each ADL.

Activity (m/s)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	**1.1±0.2*** **	**1.2±0.2***	**1.0±0.2****	1.2±0.2	**0.9±0.3*** **	**1.1±0.2****	1.5±0.1
Stair Ascent	**0.5±0.1***	0.6±0.3	**0.4±0.1****	0.5±0.1	**0.4±0.2****	0.5±0.2	0.6±0.0
Stair Descent	**0.5±0.1*** **	**0.5±0.1***	**0.4±0**.0**	0.5±0.1	**0.5±0.1***	0.5±0.1	0.7±0.1
0º	**0.9±0.2***	1.0±0.3	**0.6±0.3****	0.8±0.2	**0.5±0.4*** **	**0.8±0.3***	1.3±0.2
+5º	0.8±0.3	0.9±0.2	0.4±0.2	0.7±0.1	**0.6±0.3***	0.7±0.2	1.0±0.3
+7.5º	0.7±0.3	0.8±0.3	**0.4±0.2***	0.7±0.2	0.5±0.2	0.6±0.1	0.9±0.3
-5º	**1.0±0.3***	1.1±0.3	**0.7±0.4****	**0.9±0.3***	**0.8±0.1***	**0.8±0.3***	1.4±0.3
-7.5º	**0.9±0.3****	1.2±0.2	**0.7±0.4****	**0.9±0.4***	**0.7±0.1***	**0.9±0.3***	1.5±0.3

Key: *** / ** / ***** = significance where p<0.05, < 0.01 and < 0.001 respectively between the selected patient group to controls.

Participants generally walked faster when walking level or downhill, and slower during the stair tasks. Walking speed for all patients were consistently lower compared to controls, apart from the post-CRDD group who walked with the same average walking speed as controls during stair ascent (although more patient variation was seen). Furthermore, postoperative walking speed for each activity increased from preoperative for all implant groups and activities, except for the CRDD and UCR groups which showed no change in walking speed during stair descent (both implants), and -5º decline walking (UCR only). Treadmill level and downhill walking showed the largest disparity in average walking speed between the patient and control groups of up to 0.8m/s.

Any significance was to controls, and no significant changes were observed between implants (inter-implant), or between preoperative and postoperative for the same implant (intra-implant). Compared to controls, both the postoperative CRDD and UC groups showed significantly lower walking speeds for two activities each. The UCR group showed the same finding for four activities. However, the significant walking

speed and standard deviations seen for the post-CRDD group (during level walking and stair descent) were the same as the UC group which showed no significance to controls. Significant post-UC group walking speeds (seen for the decline walking activities) were slower than the CRDD group for the same activities. And so, it may be deduced that the CRDD group showed the better improvement in walking speed overall.

There is scope in comparing overground and self-paced treadmill level walking speeds from the Vicon and Motek CAREN systems. When the CAREN system is not set at an incline, participants should theoretically walk at similar speeds in both laboratories. Any large discrepancies between the laboratories may reflect unfamiliarity with the self-paced treadmill mode which may limit biomechanical interpretation of sloped walking. A summary of average speeds, and difference in mean findings across laboratories with standard error is presented in Table 6-10:

Table 6-10: Comparison of overground walking speed to self-paced level treadmill walking.

Walking Speed (m/s)	CRDD		UC		UCR		Control	Total
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op		
Vicon	**1.1±0.2^{++}**	1.2±0.2	**1±0.2^{++}**	**1.2±0.2^{++}**	**0.9±0.2^{++}**	**1±0.2^{+}**	**1.5±0.1^{+}**	**1.2±0.3^{+++}**
Motek CAREN	0.9±0.2	1.1±0.3	0.6±0.3	0.8±0.2	0.5±0.4	0.8±0.3	1.3±0.2	0.9±0.4
Difference ± SE	0.2±0.09	0.08±0.08	0.3±0.04	0.3±0.04	0.4±0.06	0.3±0.07	0.2±0.09	0.3±0.03

Key: **+/ ++/ +++** = significance where p<0.05 or < 0.01 respectively between the faster walking speed compared to the other laboratory for the same patient group.

Controls were found to walk on average 0.2 m/s slower when walking on the self-paced treadmill than overground walking. Paired T-tests found this to be a significant decrease (p=0.042). All patient groups also walked significantly slower when walking at self-paced mode compared to overground, apart from the post-surgery CRDD group (p=0.32). Aside from the aforementioned patient group, mean differences and standard error between laboratory walking speeds for all other patient groups were not comparable. Collating all walking speeds per laboratory (irrespective of

participant group), a paired T-test found a strongly significant difference between overground and self-paced treadmill walking speeds ($p<0.001$). This suggests participants were not walking at a natural pace in the Motek CAREN system. Further investigation into other biomechanical parameters will be carried out to assess the degree of disparity between data from both walking modalities.

6.3.2. Cadence

Below the cadence of all participants during all ADLs is given in Table 6-11:

Table 6-11: Cadence of all participants for each ADL.

Activity (steps/min)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	**104±10****	**105±10*****	**104±8***	111±6	**102±15****	**107±10****	123.7±9
Stair Ascent	77±12	87±17	**62±15****	75±20	**62±18*****	71±19	93.2±6
Stair Descent	**81±13*****	97±25	**72±7****	88±19	**79±17****	77±13	106±6
0º	**102±14***	105±17	103±5	111±3	90±24	**97±18***	118±11
+5º	99±16	**97±18***	101±8	108±3	89±32	**90±13****	116±8
+7.5º	**93±16***	90.9±22.3	94±8	107±3	**76±22****	81±22	116±10
-5º	**110±10***	108±16	108±4	115±1	**102±10***	**100±14***	123±10
-7.5º	114±14	**110±14***	111±2	118±5	101±25	115±9	130±12

Key: *** / ** / ***** = significance where $p<0.05$, < 0.01 and < 0.001 respectively between the selected patient group to controls.

Participants presented the greatest cadence during downhill walking and lowest cadence during stair ascent. Like walking speed, all significant differences were to controls only and generally cadence increased from post-surgery. Some exceptions to this were for the post-UCR group, where cadence slightly reduced during stair descent (by 1.2 steps/min) and -5º decline walking (by 2 steps/min). Likewise, the post-CRDD group's cadence for all sloped walking tasks decreased from pre-surgery, the largest difference for this group was during -7.5º walking by 5 steps/min, the only significantly different finding to controls for this activity ($p=0.017$, mean difference ± SE: 19.8±5.8 steps/min).

6. Group Results

The greatest mean difference between patients and controls was during +7.5° incline walking where the control group's cadence was 26 steps/min higher than the patients on average. Only the UC group showed an intra-implant increase in cadence consistently for all tasks, and no significant postoperative differences to controls, indicating cadence was at healthy limits. Despite the large difference in walking speed between level walking in the Vicon and Motek CAREN systems, cadence was particularly comparable between activities for the CRDD and UC implants. The UCR group cadences differed by ~10 steps/min between laboratories at both operative states, and control group cadence differed by ~6 steps/min between laboratories.

6.3.3. Stride Time

Table 6-12 displays stride time, the time between consecutive ipsilateral heel strikes, for all groups during each of the ADLs.

Table 6-12: Stride time of all participants for each ADL.

Activity (s)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	**1.2±0.1****	**1.2±0.1****	1.2±0.1	1.1±0.1	**1.2±0.2****	**1.1±0.1***	1.0±0.1
Stair Ascent	1.6±0.3	1.4±0.4	**2.1±0.6***	1.7±0.6	**2.1±0.7****	1.7±0.9	1.3±0.1
Stair Descent	**1.5±0.3****	1.3±0.5	**1.7±0.2***	1.3±0.5	**1.6±0.4***	1.3±0.4	1.1±0.1
0°	1.2±0.2	1.2±0.2	1.2±0.1	1.1±0.0	**1.4±0.5***	1.3±0.3	1.0±0.1
+5°	1.2±0.2	1.3±0.3	1.2±0.1	1.1±0.0	1.6±0.7	**1.4±0.2***	1.0±0.1
+7.5°	1.3±0.3	1.4±0.4	1.3±0.1	1.1±0.0	**1.7±0.5****	1.6±0.4	1.0±0.1
-5°	**1.1±0.1***	1.1±0.2	1.1±0.0	1.0±0.0	**1.2±0.1***	**1.2±0.2***	1.0±0.1
-7.5°	1.1±0.1	**1.1±0.2***	1.1±0.0	1.0±0.0	**1.2±0.3***	1.0±0.1	0.9±0.1

Key: *** / ** / ***** = significance where p<0.05, < 0.01 and < 0.001 respectively between the selected patient group to controls.

For all participants, stair ascent showed the longest stride time and -7.5° walking the shortest. Control stride time was shorter than all patient groups, more significantly so for preoperative groups. As with walking speed and cadence, all significant changes were to controls only. Intra-implant changes revealed a reduction in stride time for all activities by the UC group, and the UCR group mostly did the same, except

for the -5° downhill walking task where average stride time remained unchanged. The CRDD group showed no change in stride time for half the activities, a decrease in stride time for the stair activities, and an increase in stride time for the uphill walking tasks. The post-UC group was the only group to show showed no significance to controls and showed the shortest stride times of all implants. However, preoperative stride times for this group also did not show many significantly different stride times to controls so the preoperative function may have been higher compared to the other implants. The post-UCR group showed some of the largest stride times, as well as the largest total decrease from preoperative.

6.3.4. Opposite Foot Off and Contact

Table 6-13 and Table 6-14 gives the percentage during the gait cycle where the opposite unoperated foot off and foot contact respectively occurred.

Table 6-13: Percentage during the gait cycle opposite foot off occurred of all participants for each ADL.

Activity (%)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	**14.8±3.2***	**12.4±1.4†**	**16.0±3****	**13.6±1.5***	**15.4±2.4****	**14.4±2.0*****	10.8±1.4
Stair Ascent	16.6±4.4	15.0±3.5	18.5±5.4	14.5±2.6	15.0±4.5	15.2±3.7	12.6±1.3
Stair Descent	**15.1±2.7*,2**	**14.8±2.7***	10.9±0.0	11.0±3.3	12.6±2.2	12.8±2.5	10.7±1.3
0°	18.6±2.6	16.1±3.8	19.8±3.8	17.8±3.7	**22.4±9.5***	**19.4±3.2***	14.4±1.7
+5°	19.2±4.2	17.3±3.5	**21.8±2.7***	17.6±2.6	**21.9±5.1***	**19.4±2.1***	14.8±2.5
+7.5°	20.3±6.2	17.2±3.7	22.3±3.5	17.1±1.5	22.2±3.7	21.5±3.9	16.3±1.2
-5°	**17.7±3***	16.3±4.2	**20.1±4.5****	16.2±2.5	17.8±2.2	**19.6±4.2***	13.6±1.2
-7.5°	**17.3±3.2***	15.0±2.0	**17.8±4.8***	15.7±3.9	**20.4±3.6***	**16.3±2.1***	12.0.1±2

Table 6-14: Percentage during the gait cycle opposite (unoperated side) foot contact occurred of all participants for each activity for each ADL.

Activity (%)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	49.4±0.8	49.9±0.8	49.8±0.8	49.3±0.6	50.4±1	49.6±1.7	50.1±0.7
Stair Ascent	47.7±2.3	49.5±5.0	49.9±1.7	48.4±2.1	**52.6±5.5*,1**	**45.2±9.3†**	48.0±1.2
Stair Descent	49.1±2.5	51.7±2.9	47.0±8.9	48.2±2.6	49.0±0.6	46.2±7.7	50.7±2.2
0°	49.7±1.6	50.0±0.9	49.6±1.3	49.2±1.3	49.5±2.9	50.2±1.4	49.9±0.9
+5°	50.9±2.4	50.5±1.7	48.8±1.2	49.4±0.7	48.7±1.2	50.2±1.0	49.2±1.3
+7.5°	50.7±2.7	49.9±2.1	49.6±0.7	49.1±1.7	49.3±1.3	50.3±1.5	49.9±1.1
-5°	50.2±1.4	51.3±1.9	49.1±1.5	49.7±0.7	48.8±1.6	50.4±3.2	49.9±0.9
-7.5°	49.8±1.4	50.1±1.0	49.3±2.1	48.5±0.5	49.6±1.5	49.4±1.0	49.7±1.2

Key: *** / ** / ***** = significance where p<0.05, < 0.01 and < 0.001 respectively between the selected patient group to controls.

† = significance where p<0.05 for the selected implant group between preoperative and postoperative.

1,2,3 = significance where p<0.05 between the selected group to CRDD (1), UC (2) or UCR (3) implant at the same operative state.

During the gait cycle, whilst the operated limb is in the stance phase, the contralateral limb undergoes swing. This is initiated with opposite foot off then concludes with opposite foot contact. Opposite foot contact had low variability across all participant groups and only two significant findings were seen which were during stair ascent. Firstly, the pre-UCR group showed a later point of opposite foot contact to the control (p=0.041) and pre-CRDD groups (p=0.023). In addition, the post-UCR group showed a decrease in opposite foot contact from preoperative (p=0.02, mean difference ± SE: 7.3±3.0%). The higher variability and degree of significant findings in Table 6-13 suggests opposite foot off is the more dependent parameter which alters the duration of swing phase for the contralateral limb.

Controls had an earlier point of opposite foot off compared to all patients and post-surgery changes from preoperative were mostly an earlier opposite foot off point. An earlier point of opposite foot off, with a relatively unchanged point of opposite foot contact means a longer swing phase was carried out by the contralateral limb. This implies confidence in the operated limb to remain in single stance support. Generally,

level treadmill or uphill walking gave later opposite foot off points, overground level walking and stair descent showed earlier opposite foot off points. The majority of significant differences for this parameter were to controls aside from the pre-CRDD group which showed significant lower values during stair descent to the pre-UC group (p=0.041), and to controls (p=0.018).

At postoperative, the UC group's only significant finding was to controls where a later point of opposite foot off during overground level walking was seen (p=0.023, mean difference ± SE: 2.8±0.8%). For the same activity, the post-UCR group showed a larger significant difference to controls (p<0.001, 3.5±0.8%). The post-UCR group also showed significant differences to controls. Whilst the post-CRDD group showed a comparable number of significant findings to controls as the UC group, the UC group showed greater decreases from preoperative and may have improved function more than the other implants.

6.3.5. Double Support

The overall period (in seconds) of double support, where both limbs are in contact with the ground, is presented in Table 6-15:

Table 6-15: Period of double support of all participants for each activity for each ADL.

Activity (s)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	**0.3±0.1****	**0.3±0.1***	**0.4±0.1****	**0.3±0***	**0.4±0.1****	**0.3±0.1****	0.2±0.0
Stair Ascent	0.5±0.1	0.4±0.1	**0.8±0.5*,3**	0.5±0.3	0.5±0.1	0.4±0.1	0.3±0.1
Stair Descent	**0.4±0.1***	**0.4±0.1***	0.6±0.0	**0.3±0.1†**	0.4±0.1	0.4±0.0	0.2±0.0
0º	0.4±0.1	0.4±0.2	0.5±0.1	0.4±0.1	**0.7±0.4***	**0.5±0.1***	0.3±0.0
+5º	0.5±0.1	0.5±0.2	0.5±0.1	0.4±0.1	**0.8±0.6***	**0.5±0.1****	0.3±0.1
+7.5º	0.5±0.3	0.5±0.2	0.6±0.0	0.4±0.0	**0.8±0.4***	0.7±0.2	0.3±0.0
-5º	**0.4±0.1***	0.4±0.1	**0.4±0.1****	0.3±0.1	**0.4±0.0***	**0.5±0.1***	0.3±0.0
-7.5º	0.3±0.1	0.3±0.1	**0.4±0.1***	0.3±0.1	**0.5±0.1***	**0.3±0.0†**	0.2±0.1

Key: ***/ **** = significance where p<0.05 and < 0.01 respectively between the selected patient group to controls

† = significance where p<0.05 for the selected implant group between preoperative and postoperative

1,2,3 = significance where p<0.05 between the selected group to CRDD (1), UC (2) or UCR (3) implant at the same operative state.

6. Group Results

Controls showed a similar double support periods for all activities of either 0.2 or 0.3s throughout. Patients' double support period during stair ascent and uphill walking were longer compared to other tasks of up to 0.8s, particularly in preoperative groups. Controls consistently showed shorter double support periods compared to patients at both operative states. Intra-implant changes by the UC group showed consistent postoperative decreases in double support for all activities, the CRDD group mostly showed no change aside from a reduction in double support during stair ascent. Similar to the UC group, the post-UCR group mostly showed decreases in double support, except during stair descent where no change was seen, and -5^{o} walking where an increase of 0.1s was seen.

There were two instances where a significant decrease from preoperative occurred, the UC group during stair descent (p=0.048, mean difference: 0.28s, SE 0.14s) and UCR group during -7.5^{o} walking (p=0.047, mean difference ± SE 0.2±0.08s). One significantly different inter-implant finding was seen for this parameter, where the pre-UC group showed a longer period of double support to the pre-UCR group during stair ascent was (p= 0.046, mean difference ± SE 0.4±0.1s).

The UCR group at both operative states showed a similar number of activities where the period of double support was significantly greater than controls, suggesting low postoperative improvement. Since the UC group showed a decline in the number of activities that showed significance to controls (from four activities at pre-surgery to one at post-surgery) and a significant intraoperative improvement was seen for stair descent it may be deduced the UC group showed better postoperative joint function out of the implants. However, it is noted that the small sample sizes considerably limit interpretation of these results.

6.3.6. Summary

The spatiotemporal parameters of TKA patients at pre- and postoperative and a control group carrying out various ADLs was described above. Significance between all groups were reported and the majority of significant differences were between the patient and control groups. Pre-surgery data showed larger differences in STP

values to controls than at post-surgery, indicating some improvement for all groups. Compared to controls, patients consistently walked at a slower speed, decreased cadence, and greater period of double support. Although the magnitude of these differences lessened after surgery, control levels were not yet reached.

Significant intra-implant differences showed reductions in STP for at least a single activity each per implant (Table 6-7). The point of opposite foot contact during level walking was significantly earlier for the CRDD group, and the period of double support was significantly shorter for the UC and UCR implants during stair descent and -7.5^{o} walking respectively. The UCR group also showed a significantly earlier point of opposite foot contact than preoperative seen during ascent. Significant inter-implant differences were less frequently seen and were only seen at preoperative implying preoperative function was similar.

When using the number of significant differences to gauge which implant improved function the most, for some parameters this appear inconclusive. For instance, both the post-CRDD and post-UC implants showed significantly slower walking speeds for two activities to controls. The significant CRDD group walking speeds equalled the UC group speeds for the same activities where no significance to controls was seen. Conversely for the activities where significantly different walking speeds was seen between the UC and control groups, the respective speeds by the CRDD group were faster than the UC speeds and so it may be deduced that the patients receiving the CRDD implant performed better for that STP. For the other STPs, the post-UC group more often showed no changes to controls (implying data were comparable). The post-UCR group consistently showed greater instances of significant differences to controls than the other implants, and may be considered to improve function the least, or perhaps worsened function.

6.4. Joint Kinematics

This section focusses on joint kinematic parameters for all activities of daily living. Peak angles and joint range of motion in all dimensions for the pelvis, hip, knee and

ankle, and additional sagittal flexion angles at heel strike, toe off and knee flexion peaks during stance were extracted and analysed. However, not all results are presented in order to focus on significant and relevant findings. The sample sizes of each participant group were presented in Table 6-4. Whilst significant differences are described, differences larger than 5° are also explored as this is considered to be clinically relevant.

Compound plots of control data is displayed below (Figures 6-2 to 6-7) to roughly indicate key data points presented later. Red arrows indicate range of motion (indicated in all figures), and red stars indicate peak values (indicated for all knee kinematics, and ankle flexion only).

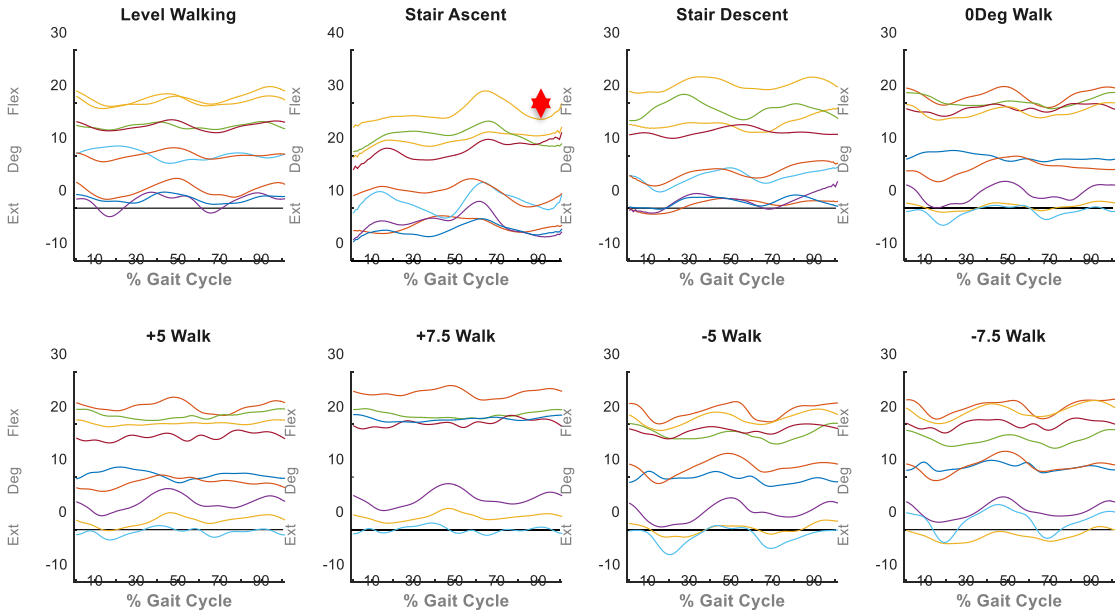

Figure 6-2: Pelvic flexion across all ADLs.

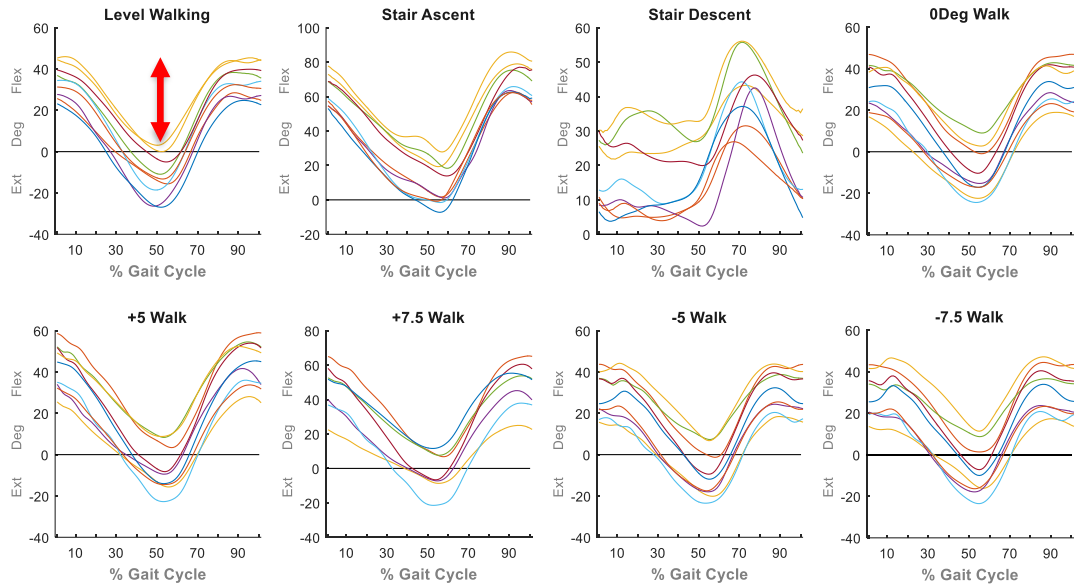

Figure 6-3: Hip flexion across all ADLs.

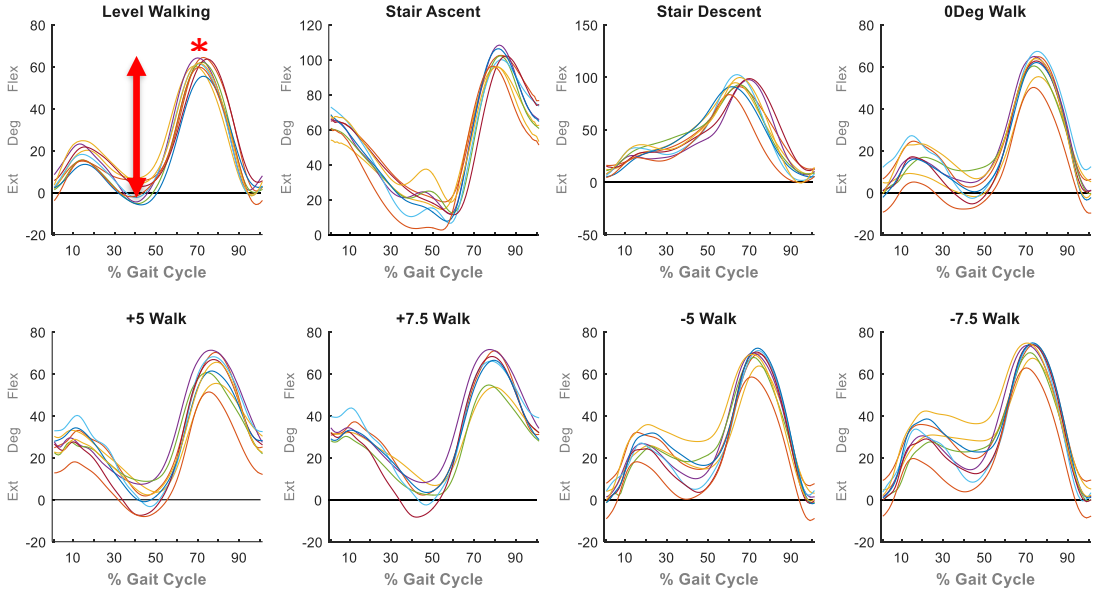

Figure 6-4: Knee flexion across all ADLs.

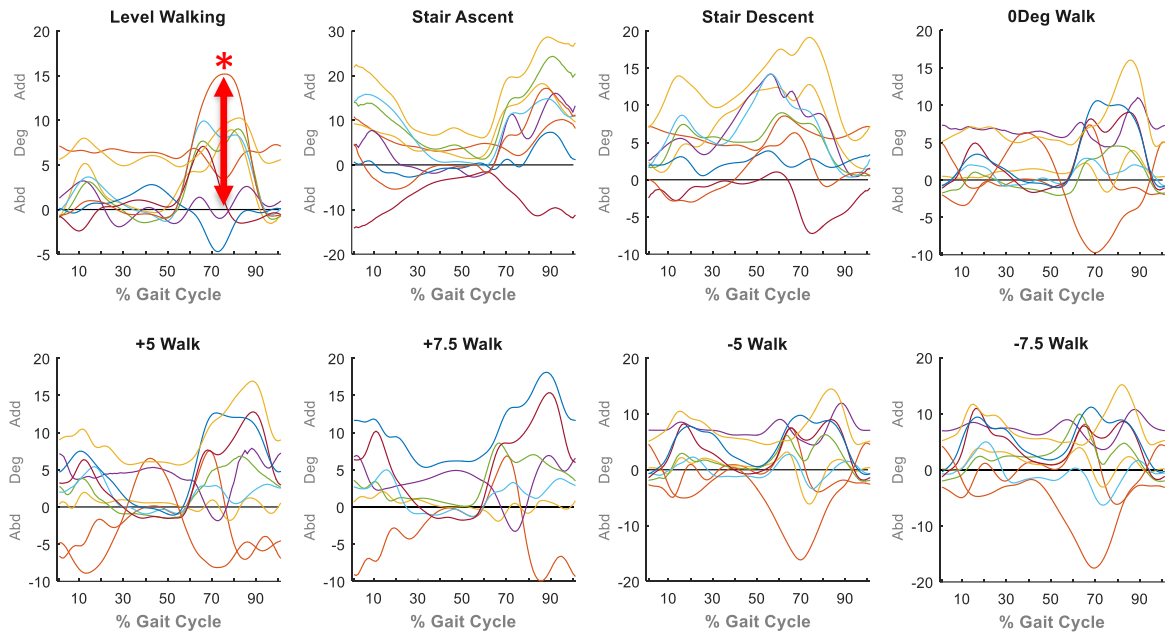

Figure 6-5: Knee adduction across all ADLs.

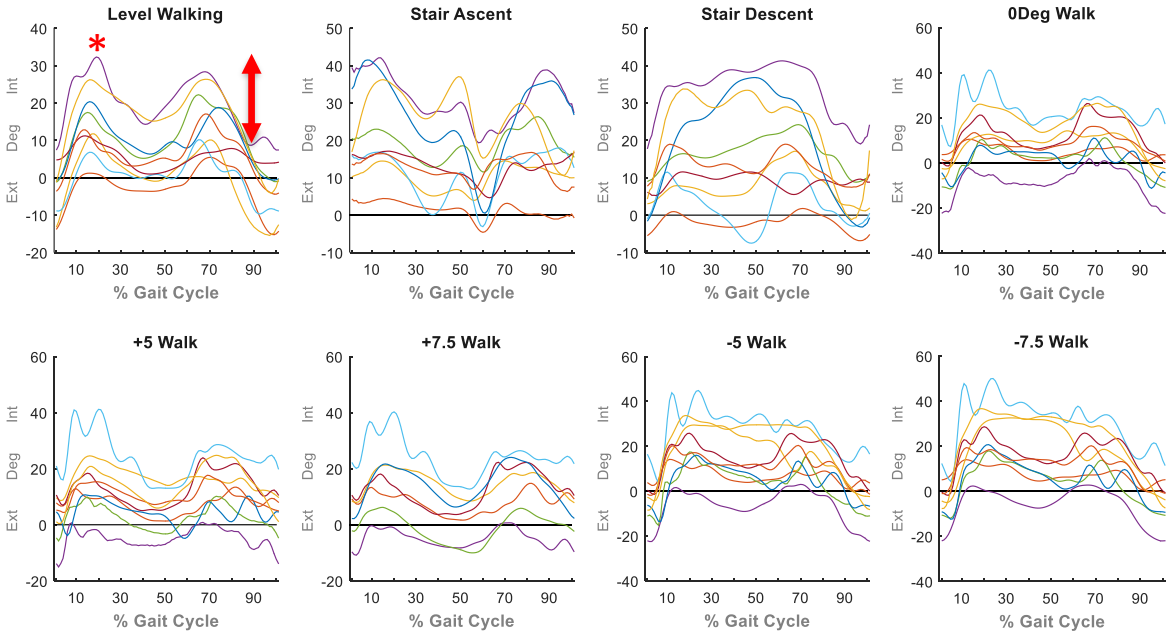

Figure 6-6: Knee rotation across all ADLs.

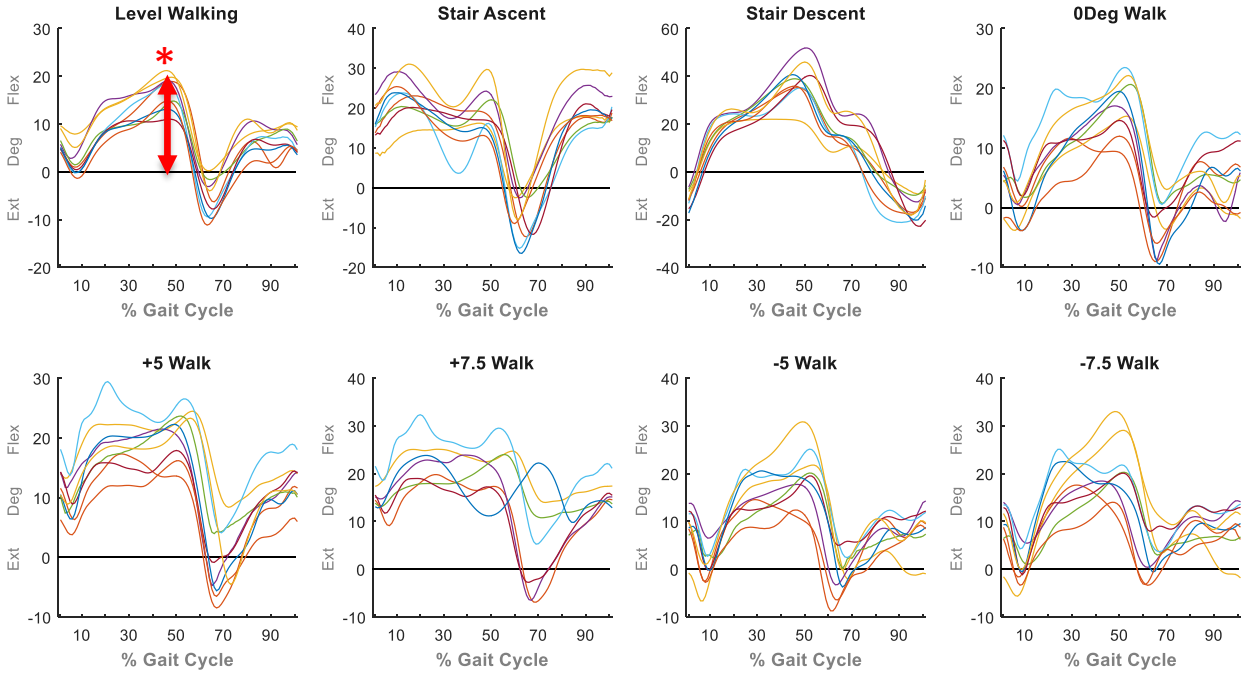

Figure 6-7: Ankle dorsiflexion across all ADLs.

6.4.1. Peak Kinematics

As deduced during the case study, peak hip and pelvic angles are susceptible to a large shift in values where a small vertical misplacement of marker position has occurred. Therefore, peak knee and ankle flexion values were presented, along with peak knee adduction and longitudinal rotation angles. As ROM is less affected by marker misplacement, this was displayed for all joints in Chapter 6.4.2.

6.4.1.1. Peak Knee Flexion

Peak knee flexion during all ADLs for participant groups is given in Table 6-16:

Table 6-16: Peak knee flexion of all participants for each ADL.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	55.7±6.2	**54.3±3.4****	53.1±9.6	**48.5±5.7*****	52.5±7.4	**53.7±6.0***	61.2±3.0
Stair Ascent	**85.2±8.6*****	**84.5±5*****	**88.5±3.7***	**82.2±4.3*****	**89.1±3.5***	**76.0±13.6***,††**	100.6±4.5
Stair Descent	**83.6±7.1****	**81.7±7.3****	87.8±5.2	**80.6±3.2***	82.9±6.7	**75.9±12.8*****	95.7±6.9
0°	55.2±7.2	54.9±5.4	50.1±10.3	**46.3±4.4*****	**46.5±12.2***	**51.0±6.0****	61.5±5.8
+5°	**51.8±7.4***	**51.5±4.9****	**47.2±12.4***	**45.7±3.8*****	**47.1±8.3***	**48.9±7.5*****	63.3±6.4
+7.5°	**50.8±8.6***	**51.6±3.5****	**47.5±10.9***	**47.0±3.5*****	53.2±3.9	**48.1±6.1*****	65.0±7.2
-5°	60.1±7.9	60.1±5.7	**49.8±13.5****	**51.0±8.5****	54.4±5.9	**52.3±6.2*****	68.2±4.5
-7.5°	62.1±8.9	63.7±6.9	**53.0±14.5***	**53.4±9****	58.4±5.3	**57.9±5.8****	72.1±5.0

Key: ***/ ** / ***** = significance where p<0.05, < 0.01 and < 0.001 respectively between the selected patient group to controls.

†† = significance where p<0.01 for the selected implant group between preoperative and postoperative.

Peak knee flexion was greatest during stair ascent– reaching 100° in controls, and ~80° in patients. There are discrepancies as to which activities showed the lowest peak knee flexion value, for controls this was during both level walking activities (~61°), and uphill walking in patients (~ 47°). In all instances, controls showed greater peak flexion to patients (by a clinically relevant amount), more so to postoperative groups than preoperative. The post-UC and post-UCR groups gave significantly lower flexion to controls for all tasks and for five activities respectively this was where p<0.001.

For all activities, the average difference in peak flexion between the post-CRDD, UC and UCR groups to controls was around 10.6°, 16.6° and 15.4° respectively. The largest mean difference ± SE to controls was seen in the post-UCR group during stair navigation (ascent: 24.6±3.7°, descent: 19.8±4.5°, p<0.001 for both). The post-UCR stair ascent flexion was also significantly lower than preoperative (mean difference ± SE: 13.1±4.5°, p=0.007). Aside from this, there were no other significant intra-implant differences. Whilst intra-implant changes generally decreased from pre-surgery, for the CRDD group no changes were clinically relevant (smaller than 5°). The other

implants showed clinically relevant decreases in peak flexion for both stair tasks (both implants) and +7.5^{o} walking (UCR group only).

Inter-implant comparisons showed that the post-CRDD group exhibited the highest peak knee flexion than the other groups. The post-UC group showed smaller peak flexion than the UCR group for six activities (by a clinically relevant amount during treadmill and overground level walking). The two activities where the post-UCR group showed the lowest peak knee flexion were also by clinically significant amount to the other implants. The post-CRDD group showed less significance to controls, and higher peak flexion values than the other implants. The UC and UCR groups performed similarly, however the UCR group showed slightly larger reductions in peak flexion to preoperative, controls and the other implants.

6.4.1.2. Peak Knee Adduction

The knee adduction angle (KAA) is a frontal plane parameter and is commonly misreported when using the PiG biomechanical model. A phenomenon known as kinematic cross talk causes sagittal plane motion to be interpreted as frontal plane motion causing flexion to be slightly underreported and knee adduction to be considerably over reported. Commonly this occurs during swing phase where flexion is greatest. Since stance flexion is lower and cross talk should affect adduction less, only peak KAA during stance is presented (Table 6-17):

Table 6-17: Peak knee adduction during stance of all participants during each ADL.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	9.4±4.9	7.9±3.8	6.1±8.5	5.3±2.8	9.1±3.3	6.2±4.8	5.3±2.7
Stair Ascent	15.5±8.1	**24.1±8.5*,†**	11.5±6.6	16.9±7.4	20.5±9.6	17.6±16	10.2±7.2
Stair Descent	16.9±6.7	**28.6±11.1*,†**	6.4±1.3	22.3±8.5	22.6±14.5	21.2±17.7	11.6±5.5
0°	9.6±4.1	11.4±7.9	7.2±8.3	5.1±3.9	9.8±8.0	5.4±3.8	5.2±3.4
+5°	9.6±3.1	10.4±6.9	6.2±8.4	6.3±3.5	9.6±7.5	4.6±3.7	6.3±2.7
+7.5°	8.8±4.5	11.1±7.9	2.2±2.8	7.4±2.9	6.9±3.9	6.2±5.8	6.7±3.4
-5°	10.9±6.3	14.4±8.7	8.5±9.3	4.9±7.4	8.3±3.1	8.0±5.0	5.9±4.5
-7.5°	11.2±4.4	15.7±9.0	8.5±9.6	6.2±5.4	8.3±3.8	6.2±3.1	6.6±5.5

Key: ***** = significance where $p<0.05$ between selected patient group and control group.

† = significance where $p<0.05$ for the selected implant group between preoperative and postoperative

The largest stance KAA was seen during stair navigation, and the smallest during level or uphill walking. There were few significant trends between participant groups overall. Control stance peak KAA was mostly lower than patient groups, except for up to three activities for the UC (both operative states), and post-UCR groups. Compared to preoperative, the UCR group showed non-clinically relevant reductions in peak KAA for all activities, the UC group showed a postoperative increase for half of the activities (three of which were clinically relevant), and the CRDD presented an increase in peak KAA (which was clinically and statistically relevant for the stair tasks) for all tasks except overground level walking.

The CRDD group showed the only significant findings from all groups, observed during the stair tasks. From preoperative, the mean change ± SE during stair ascent was 8.6±4.2°, ($p=0.047$) which is near insignificant, and 11.7±5.0° ($p=0.027$) during descent. The same peak KAAs were also significantly greater than controls, where the mean difference ± SE was 13.9±4.4° for stair ascent ($p=0.021$), and 17.0±4.9° for descent ($p=0.011$).

Taking a mean difference of 5° to be clinically relevant, whilst all significant differences displayed a mean difference greater than this, other differences existed that were greater than 5° and did not flag as statistically significant. The largest

clinically relevant differences were mostly seen during the stair tasks, namely the post-UC and UCR (stair descent) and pre-UCR (stair ascent) groups showed a mean difference to controls of around 10°. Between operative states the UC group also showed a large clinically relevant increase in KAA of 15.8° (p=0.11) during stair descent.

6.4.1.3. Peak Knee Rotation

All participants' maximum knee rotational angles during the whole gait cycle are presented in Table 6-18. Positive angles refer to an internally rotated joint (towards the midline) by convention:

Table 6-18: Peak knee rotation of all participants for each ADL.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	14.1±6.4	**8.3±4.0**,††	9.6±2.5	11.2±3.3	12.0±6.4	**8.5±4.9***	17.3±8.0
Stair Ascent	16.6±7.5	12.5±7.0	12.9±3.7	13.3±3.4	15.6±5.5	11.7±6.0	22.2±11.5
Stair Descent	16.6±6.8	11.4±6.1	14.3±4.6	13.4±3.7	14.9±6.0	11.7±6.7	21.5±11
0°	14.0±6.0	**8.6±5.0†**	7.4±4.9	11.5±4.4	8.9±6.1	7.9±5.6	18.5±11.2
+5°	12.8±6.1	7.8±4.1	7.3±5.2	11.3±2.3	8.8±4.5	7.6±6.6	17.7±11.2
+7.5°	12.2±6.6	7.2±4.4	8.2±5.3	11.5±2.4	11.1±6.0	8.5±8.1	18.1±13.1
-5°	15.2±5.4	10.7±6.3	8.1±4.7	15.0±3.3	9.4±5.0	8.5±5.0	21.8±11.7
-7.5°	15.6±5.8	11.7±6.4	8.6±4.8	15.5±3.0	10.7±6.5	9.8±5.8	23.0±13.3

Key: * / ** = significance where p<0.05 and < 0.01 respectively between the selected patient group to controls.
† / †† = significance where p<0.05 and < 0.01 respectively for the selected implant group between preoperative and postoperative.

Here, controls consistently showed larger peak angles to patients, and the activities where the largest peak angles were seen were stair navigation and decline walking. Smallest rotation was seen during level and uphill walking. Significance to controls was during overground level walking only, to the post-CRDD (p=0.005, mean difference ± SE: 9.0±2.4°) and post-UCR (p=0.021, mean difference ± SE: 8.8±2.8°) groups, the largest mean differences for this activity.

6. Group Results

Both the CRDD and UCR groups presented average intra-implant decreases in peak rotational angle for all activities. The UC group did not show the same and post-surgery values were higher than preoperative for all activities except stair descent where a decrease of 1° was seen. Significant intra-implant differences were only observed for the CRDD group during overground (p=0.008, 5.8±2.1°) and treadmill level walking (p=0.048, 5.4±2.6°). During stair descent and +5° walking, the CRDD group also showed clinically relevant decreases in peak rotational angles. On the whole the CRDD group showed larger postoperative decreases in peak rotational angle than the UCR group.

Lastly, the UC group showed a clinically relevant increase in knee rotation from pre-surgery during both decline walking tasks by 6.9° and no other intra-implant change was clinically relevant. The UC group also showed the largest postoperative peak rotation for all tasks. Compared to the post-CRDD group, the post-UCR group showed similar or lower peak knee rotational angles for five activities implying that the rotational mobile bearing advantages were not evident.

6.4.1.4. Peak Ankle Flexion

Table 6-19 shows peak ankle dorsiflexion angles for participants for each activity:

Table 6-19: Peak ankle dorsiflexion of all participants during each ADL.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	19.0±4.9	18.2±2.6	16.6±2.8	14.6±3.4	19.8±4.4	17.6±3.3	16.4±3.9
Stair Ascent	19.8±4.3	18.6±3.5	16.9±3.3	16.7±1.4	22.9±6.5	20.2±3.8	22.4±5.3
Stair Descent	30.1±6.7	30.3±5.6	31.9±1.5	27.3±2.9	36.9±5.8	34.9±6.5	36.4±8.6
0°	19.0±4.0	18.3±2.1	17.8±5.1	15.3±2.6	18.9±4.7	18.9±2.2	17.2±4.2
+5°	23.3±4.5	20.6±2.9	20.2±5.3	19.1±4.1	23.8±4.1	21.4±3.0	21.3±3.8
+7.5°	24.5±3.5	22.9±4.3	21.2±5.5	20.7±4.5	24.5±2.3	22.7±2.5	24.6±3.7
-5°	19.6±4.2	19.1±3.9	16.2±6.0	14.0±1.4	20.5±6.0	20.4±2.5	19.3±6.1
-7.5°	21.0±4.3	19.7±4.8	16.3±6.2	14.5±2.2	24.1±2.5	21.1±2.1	20.7±5.8

The greatest ankle dorsiflexion angles (between 27-37^{o}) were seen during stair descent and level walking showed the lowest dorsiflexion. Unlike other parameters, controls did not unanimously show greater dorsiflexion angles to patients. However, where clinically relevant differences occurred, the controls showed higher dorsiflexion (during stair navigation). There were no significant findings and the largest clinically relevant differences between controls and patients were during stair navigation for the CRDD and UC groups at both operative states and decline walking for the post-UC group. Most patients showed a non-clinically relevant lower peak dorsiflexion from preoperative for most activities. The only exception to this was the CRDD group which showed a near-negligible postoperative increase during stair descent. The largest decrease ± SE from preoperative was seen in the UC group also during stair descent of 4.6±5.0^{o} which is close to being clinically relevant.

6.4.1.5. Summary

This section displayed peak knee angles in all dimensions along with peak ankle dorsiflexion values. Peak pelvic and hip angles were omitted due to high dependence on accurate pelvic marker placement which could not be guaranteed for each participant. Generally, controls showed greater peak knee flexion and rotation and lower peak adduction during stance angles compared to patients and no clear trend was seen for peak ankle dorsiflexion. From all parameters, control peak knee flexion showed greater significance to patients and all differences were clinically relevant. Intra-implant differences showed mostly decreases in peak knee flexion implying reduced function at one year postoperative, particularly for the UC and UCR implants.

As stated, the UCR implant is a mobile bearing implant which allows some axial rotation to occur. It could be theorised that peak knee rotation should be higher compared to the other fixed bearing implants. This was not actualised as the post-UCR group generally demonstrated the lowest peak rotation angles compared to the other postoperative implants. However, only peak values were presented and since some external rotation also occurs during gait, analysing ROM may show more rotational function in the MB group.

For peak kinematics, no significant inter-implant differences were seen, although some clinically relevant differences were observed. More statistical differences were seen to controls than pre-surgery. Postoperative peak knee flexion frequently showed a greater degree of significance to controls (where p<0.01 or 0.001) and preoperative values were less significant to controls (where p<0.05 or 0.01). Investigating knee flexion ROM may clarify whether function was lost or perhaps gained in form of improved knee extension capabilities.

6.4.2. Range of Motion

The range of motion during the ADLs are given below, this is the difference (excursion) between the highest and lowest joint angle observed during gait. Sagittal pelvis, hip, knee, and ankle ROM is given, as well as non-sagittal knee ROM.

6.4.2.1. Pelvic Tilt ROM

Sagittal pelvic ROM is given in Table 6-20:

Table 6-20: Pelvic tilt ROM of all participants for each ADL.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	3.5±1.6	3.3±1	3.9±1.0	3.5±0.7	2.7±1.0	3.1±1.0	3.1±0.9
Stair Ascent	7.2±2.2	**5.1±2.2[2,33]**	11.0±4.0	9.7±4.4	**11.3±5.3***	**10.4±4.1***	6.0±1.7
Stair Descent	6.2±2.9	4.6±1.8	9.2±5.4	5.8±1.1	5.9±3.1	5.3±1.6	4.0±1.3
0°	3.4±1.0	3.3±0.9	3.5±1.2	3.1±0.9	2.4±1.3	2.9±0.9	3.4±1.1
+5°	3.7±1.2	4.0±2.5	2.6±0.9	2.7±0.8	2.9±1.8	2.6±1.1	3.0±1.1
+7.5°	3.6±1.6	4.3±2.8	2.5±1.1	2.6±1.3	3.9±2.5	2.5±0.8	2.7±1.2
-5°	3.8±1.1	3.5±0.6	3.4±0.8	4.0±0.6	3.3±1.4	2.9±1.3	4.2±1.2
-7.5°	3.6±1.0	3.4±0.7	3.1±0.7	3.6±0.9	2.7±0.8	3.2±0.4	4.2±1.5

Key: *** / ** / ***** = significance where p<0.05, < 0.01 and < 0.001 respectively between the selected patient group to controls.

[1,2,3] / [11,22,33] = significance where p<0.05, < 0.01 respectively between the selected group to CRDD (1), UC (2) or UCR (3) implant at the same operative state.

Apart from stair navigation, all activities showed comparable pelvic tilt ROM with findings ranging between 2–4°. Participants showed a higher ROM during stair ascent (peak values of 11°) than descent (peak values of 9°). Controls showed significantly

higher ROM compared to patients at two instances: to the pre- (p=0.028, mean difference ± SE: 5.3±1.7°) and post-UCR group (p=0.044, mean difference ± SE: 4.4±1.5°). Another notable finding was that the pre-UC group showed the only other clinically relevant mean differences to controls of ~5° during stair navigation, although not significant findings (p=0.073 and 0.12 for ascent and descent respectively). During stair navigation, the post-UC pelvic ROM decreased such that the difference to controls were no longer clinically relevant which may be considered functional improvement. On average all implants showed a non-clinically relevant reduction in pelvic ROM between operative states.

One significant inter-implant difference was seen during stair ascent where the post-CRDD group showed significantly lower pelvic ROM to the other groups. The mean differences ± SE between the CRDD group to the UC and UCR groups were 4.5±1.6° (p=0.038) and 5.2±1.5° (p=0.0072) respectively.

6.4.2.2. Hip Flexion ROM

The range of hip flexion during all activities is shown in Table 6-21:

Table 6-21: Hip flexion ROM of all participants for each ADL.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	41.6±7.2	44.8±4.6	**36.7±3.6***	41.4±6.2	**37.1±6.4***	40.5±8.0	47.2±4.2
Stair Ascent	**55.0±6***	**59.0±3.5†**	**52.6±3.1***	58.0±4.7	57.3±5.7	57.2±3.4	61.5±2.6
Stair Descent	27.5±10.1	24.3±4.4	25.5±1.7	26.6±1.6	23.1±8.7	**22.1±5.7***	30.7±7.0
0°	37.6±6.1	**45.2±5.5†**	**32.1±7***	40.5±9.6	**28.3±11.0****	**37.5±7.6†**	43.1±6.4
+5°	42.6±9.5	**51.7±5.6†**	**33.9±9.7***	44.4±7.8	**36.4±10.3***	41.3±9.1	51.1±7.2
+7.5°	43.2±9.6	**52.4±6.2†**	34.8±8.7	**46.7±9†**	40.4±3.2	46.6±4.3	50.6±10.8
-5°	33.1±8.1	38.7±7.8	**26.4±10***	34.2±9.4	31.0±8.0	**27.9±6.5***	40.3±4.7
-7.5°	29.3±6.3	36.5±9	**23.8±11.7***	28.5±6.1	23.9±4.2	27.0±7.9	38.5±6.1

Key: ***/ **/** = significance where p<0.05 and < 0.01 respectively between the selected patient group to controls.

† = significance where p<0.05 for the selected implant group between preoperative and postoperative.

Similar to pelvic flexion ROM, peak hip flexion ROM were seen during stair ascent (average across all participants: ~57°.) The lowest hip flexion ROM were seen during stair descent of ~26°. Overground and treadmill level walking were mostly comparable, with slightly larger ROM seen during overground walking in some groups. Similar to stair navigation, participants showed higher ROM when walking uphill than downhill.

Controls showed higher hip flexion ROM to all groups and activities aside from the post-CRDD group during treadmill level and uphill walking where up to 2° less ROM was seen. The post-CRDD hip flexion ROM for those activities were also significantly higher than preoperative ($p<0.05$) and clinically higher than the other implants. Clinically relevant differences to controls at pre-surgery were seen in all activities for the UC group, and for seven activities each for the CRDD and UCR groups. At postoperative the number of activities where a clinically relevant difference in patients to controls decreased to one, four and six activities for the CRDD, UC and UCR groups respectively. All patient significance to controls were mostly where $p<0.05$, except for treadmill level walking for the pre-UCR group where $p=0.0047$.

Each implant group showed higher intra-implant flexion ROM at post-surgery for all activities excluding four instances where non-clinically relevant decreases was seen. The post-UC group was the only group that gave an increase from pre-surgery for all activities, five activities were by a clinically relevant amount, the largest during uphill walking (~11°). The CRDD group also showed clinically relevant intra-implant changes for five activities, whereas the UCR group showed the same for two activities. Whilst the CRDD group showed significance for four of these activities, the UC group (which often showed greater mean changes than the CRDD group) showed only one significant finding, during +7.5° walking.

6.4.2.3. Knee Flexion ROM

Participants' knee flexion ROM for all activities is displayed in Table 6-22:

Table 6-22: Knee flexion ROM of all participants for each ADL.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	**51.6±5.3*** **	**56.8±4.6†**	**49.7±4.3****	**52.1±4.4****	**49.3±8.2*** **	**53.2±6.6****	62.6±3.2
Stair Ascent	**73.3±11.5***	**77.9±5.6***	79.5±8.4	81.6±9.3	83.2±6.6	**69.1±11.5*** **,†**	88.2±7.2
Stair Descent	**77.4±9.2***	**79.3±6.1***	86.0±5.6	80.9±6.5	81.4±4.0	**73.9±11.3****	89.2±7.2
0°	**49.8±8.2***	57.2±6.3	**43.8±5.8****	**50.3±6***	**39.7±15.4*** **	**48.4±7.0*** **	62.5±4.0
+5°	**42.6±9.2*** **	**52.7±9*,3,†**	**39.6±7.2*** **	49.0±7.6	**38.8±8.3*** **	44.0±6.0	61.0±6.9
+7.5°	**41.7±9.5*** **	**52.9±9.3††**	**41.4±6.9****	50.5±7.0	**43.9±3.8***	**44.2±2.5***	61.7±8.5
-5°	**54±11.3***	**62.7±7.2[2,3]**	**44.1±10.6****	**51.6±3.9****	50.2±11.3	**53.0±6.8*** **	67.7±2.2
-7.5°	**56.7±11***	65.7±9.2	**47.6±12.4****	**54.6±5.8***	50.7±12.9	**56.7±7.8***	70.4±2.9

Key: ***/ ** / ***** = significance where p<0.05, < 0.01 and < 0.001 respectively between the selected patient group to controls.

† / †† / ††† = significance where p<0.05 and < 0.01 respectively for the selected implant group between preoperative and postoperative.

[1,2,3] = significance where p<0.05 between the selected group to CRDD (1), UC (2) or UCR (3) implant at the same operative state.

The largest knee flexion ROM was seen during stair navigation, with descent showing a slightly higher ROM than ascent. Lowest flexion ROM were seen during level walking (overground and treadmill) and uphill walking. Patients generally showed lower ROM during treadmill level walking compared to overground level walking, whereas this difference was negligible in controls. Control flexion ROM was higher than all patients for all activities and there were only four instances where this finding was not clinically relevant. These were for the pre-UC and pre-UCR groups during stair descent and ascent respectively and the post-CRDD group during both downhill walking inclinations. Collating each participant groups' mean differences across all ADLs, all preoperative patients showed larger differences to controls which decreased at post-surgery indicating functional improvement. The UCR group showed smaller ROM improvements compared to the CRDD and UC groups. Furthermore, the implant group and activity which gave the largest difference to controls was the pre-UCR group during level treadmill (22.7±4.6°, p<0.001) and +5° walking (22.2±4.5°, p<0.001).

6. Group Results

Only three inter-implant differences were significant, all to the post-CRDD group. Firstly, between the CRDD and UCR groups during +5° walking where the mean difference ± SE was 8.8±4° (p=0.013). The CRDD group also showed significantly greater ROM during -5° walking to both of the other implants where p=0.042 (11±3.7°) and 0.034 (9.7±3.1°) to the UC and UCR groups respectively. The CRDD group also showed a clinically relevant higher flexion ROM than the UCR group for all other activities excluding overground walking where the mean difference was 3.6±2.2°.

Significant intra-implant differences were seen in the CRDD and UCR groups only. For the UCR group this was due to a decrease from preoperative of 14.0±5.5° (p=0.014) during stair ascent. A postoperative reduction in ROM was also seen in the UCR group during descent, the difference ± SE (7.5±5.8°) was halved compared to stair ascent, although clinically relevant it was not a significant result (p=0.205). There was one other instance of a lowered flexion ROM from preoperative, seen in the UC group during stair descent (by 5.0±6.8°) which again was not significant (p=0.47) but was clinically relevant. The post-CRDD group showed increases in ROM for all activities, and except during stair navigation all increases were clinically relevant. Significance at $p<0.05$ was seen in the CRDD group during overground level and +5° walking, and at $p<0.01$ for +7.5° walking.

Comparing the frequency and degree of significant differences, the pre-CRDD group showed significantly lower flexion ROM to controls for all activities (of which, three of these showed a p value less than 0.001). The post-CRDD flexion ROM improved such that only three activities showed a significantly lower ROM to controls (all where $p<0.05$). The ROM during these three activities were also significantly improved from preoperative and one of these (+5° walking) was also significantly greater than the UC and UCR groups. Lastly, the non-significant intra-implant changes for the CRDD group were also clinically relevant. Based on this it may be considered that this implant showed the greatest improvement in function.

The post-UC group displayed good function similar to the CRDD group albeit not to the same degree. At preoperative, the ROM during six activities were significantly lower to controls, where $p>0.01$ apart from $+5^{o}$ incline walking where $p<0.001$. Following the surgery, the number of activities where a significantly lower ROM to controls was seen decreased to four, with p values either unchanged ($p<0.01$) or improved to $p<0.05$. A further finding was that the ROM seen during $+5^{o}$ incline walking changed from being extremely significantly different to controls at preoperative to statistically similar at post-surgery.

The UCR group displayed opposite trends to the other implants, preoperative ROM was significantly lower to controls for four activities, where $p<0.001$ apart from $+7.5^{o}$ walking where $p<0.05$. At postoperative, significance to controls were seen for more activities (seven altogether) and the degree of significance varied from $p<0.05$ ($\pm 7.5^{o}$ walking), $p>0.01$ (level walking and stair descent) and $p<0.001$ (stair ascent, treadmill level walking, and -5^{o} walking). Since some of the newly significant postoperative findings (namely decline walking) showed a greater ROM than the non-significant preoperative ROM the discrepancy in significance may be due to higher preoperative standard deviation. The pre-UCR group showed the greatest patient knee flexion ROM during the stair tasks, which subsequently became the lowest at postoperative. With this function loss in mind, and the high number of postoperative significant differences this group may be considered to have improved function the least during level and incline walking, and lost function during stair navigation.

6.4.2.4. Knee Adduction ROM

Again, since cross talk is more evident during swing, knee adduction ROM during stance only for all activities is given in Table 6-23:

Table 6-23: Knee adduction ROM of all participants for each ADL during stance.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	7.8±5.3	6.2±2.5	6.9±5.6	6.0±2.6	5.5±2.4	7.5±5.5	5.7±3.3
Stair Ascent	11.8±5.9	**21.0±6.5†**	11.8±5.4	17.3±6.5	14.1±6.3	20.1±14.6	12.5±3.9
Stair Descent	12.6±8.8	**24.3±9.6†**	7.3±1.9	21.0±12.1	13.4±12.4	20.5±18.1	10.9±4.1
0°	7.7±4.6	9.9±5.1	8.2±6.2	6.5±2.8	5.3±4.4	7.9±3.1	10.2±11.4
+5°	9.0±4.2	9.0±4.1	5.8±3.2	7.2±2.7	5.7±2.8	7.3±2.2	10.8±9.3
+7.5°	8.6±4.4	10.0±4.8	4.5±1.9	8.2±2.4	6.8±4.0	9.3±3.5	12.2±12.6
-5°	10.3±7	12.9±6.4	10.8±8.0	7.1±2.9	6.4±1.7	9.8±6.1	8.5±4.8
-7.5°	11.1±6.7	14.2±7.2	10.8±9.4	8.2±3.5	6.2±0.3	7.5±3.3	9.4±4.8

Key: † = significance where $p<0.05$ for the selected implant group between preoperative and postoperative.

Lowest ROM were seen for level walking, and peak values were observed during the stair tasks. Clinically relevant differences between patients and controls were lower control adduction ROM during the stair tasks, and larger control ROM during uphill walking. No significance to controls were seen, however, a near significant finding to the post-CRDD group during stair descent ($p=0.057$, mean difference ± SE: 13.4±4.8°) was observed. This was the second largest overall mean difference seen, the first was between pre and postoperative for the UC group during the same activity (mean difference ± SE :13.7±9.8°, $p=0.14$).

Intra-implant differences did not show consistent trends. The UCR group showed postoperative increases in stance phase ROM for all activities, and the CRDD and UC groups showed an increase in six and four activities, respectively. These intra-implant changes were clinically relevant during the stair tasks only. Intra-implant significance was only seen in the CRDD group during stair navigation. The mean differences ± SE were 9.2±3.4° ($p=0.01$) and 11.6±5.1° ($p=0.03$) for ascent and descent respectively.

Subtracting the stance ROM (Table 6-23) from the peak stance phase adduction given in Table 6-17 calculates the minimum knee adduction (or maximum abduction). The CRDD group remained in adduction throughout the activities at pre- and post-

surgery. The UC group showed some abduction for most tasks except +5° walking (preoperative) and stair descent at (postoperative). The same postoperative trend seen in the UC group was also seen for the UCR group, however the pre-UCR group remained in adduction for all activities.

6.4.2.5. Knee Rotational ROM

The range of longitudinal knee rotational motion of the participants is presented in Table 6-24:

Table 6-24: Knee longitudinal rotation ROM of all participants for ADL.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	16.2±5.0	**13.2±4.6***	**11.8±4.4***	15.3±6.7	15.4±5.8	13.5±5.4	20.6±6.1
Stair Ascent	13.3±4.5	12.4±4.7	10.9±3.9	11.4±3.0	15.0±5.4	12.3±8.3	16.7±6.0
Stair Descent	17.2±4.3	13.4±4.4	15.1±10.7	12.4±2.4	16.4±1.5	13.9±2.9	18.9±8.3
0°	15.3±5.9	**13.6±2.2***	**8.4±2.1***	**12.7±5.6***	**11.3±6.4***	**12.4±4.4****	22.8±7.7
+5°	12.1±4.9	12.1±2.0	**8.4±2.2***	10.7±3.9	10.1±4.7	10.7±3.5	16.9±6.2
+7.5°	9.9±3.4	11.7±2.4	7.4±2.4	11.1±3.7	11.5±4.9	11.5±6.3	17.5±6.7
-5°	17.4±6.9	16.1±3.3	**10.1±4.8***	13.5±5.0	13.5±5.4	14.2±4.7	25.9±10.5
-7.5°	16.6±5.9	16.1±4.1	**10.4±5.9***	14.1±4.7	15.7±6.8	14.2±3.4	26.1±11.7

Key: *** / ** / ***** = significance where p<0.05, < 0.01 and < 0.001 respectively between the selected patient group to controls.

Participants showed the smallest rotational ROM during uphill walking, and rotational ROM was similar across the other activities. Patient rotational ROM was consistently lower than controls. Differences in mean ROM between patients and controls were more pronounced during both decline walking activities, and treadmill level walking. These were all clinically relevant findings, reaching a peak difference of 16° seen in the pre-UC group during -5 and -7.5° walking (p=0.02 and 0.038 respectively). All significant differences were also between the control and patient groups, and from pre- to postoperative the CRDD group saw an increase in the number of activities where significance to controls was seen. The UCR group showed statistically similar ROM between operative states, where significance was seen

(during treadmill level walking), this was larger (or, closer to control figures) and the degree of significance is also higher ($p<0.01$ and $p<0.001$) than at preoperative. Postoperative standard deviations were lower than pre-surgery which may explain the increase in the degree of significance.

The post-UC group showed a decrease in the number of activities where a significantly lower rotational ROM to controls was seen. Preoperative significance was where $p>0.05$, and similar to the UCR group, postoperative ROM where significance to controls were seen were greater than at preoperative.

Intra-implant changes were all less than 5^o, however some trends were seen. The CRDD group showed a postoperative reduction in rotational ROM for all activities except during $+7.5^o$ walking where an increase of less than 2^o was seen. Conversely the UC group showed small postoperative increases in rotational ROM for all activities apart from stair descent where a reduction was observed. The UCR group presented a decline in ROM four activities, an increase in two, and no change for wo. The UCR group gave postoperative ROM higher than the other implants during stair descent only by a non-relevant amount. Postoperative inter-implant differences were less than 3^o so are not clinically relevant.

6.4.2.6. Ankle Dorsiflexion ROM

Table 6-25 displays the ankle dorsiflexion ROM of all participants during each ADL:

Table 6-25: Ankle dorsiflexion ROM of all participants for each ADL.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	28.1±6.5	27.2±3.8	22.0±4.7	22.6±5	23.2±5.2	23.1±4.7	23.2±2.9
Stair Ascent	29.7±6.6	29.9±5.9	29.8±6.8	33.7±8.9	33.2±4.3	28.2±5.9	32.8±5.7
Stair Descent	45.8±11.2	46.9±7.6	49.1±4.5	46.0±5.6	52.2±1.5	50.6±8.1	53.5±9.9
0°	23.1±3.8	25.8±3.2	17.9±6.5	**16.3±2.4*,11**	18.3±7.4	22.0±3.7	23.1±4.1
+5°	24.7±6.2	25.9±4.3	**14.2±4.3*,1**	18.5±5.2	20.5±9.5	22.0±6.2	24.8±4.3
+7.5°	22.1±6.3	26.2±4.0	14.9±3.8	19.9±2.9	22.7±7.5	18.3±8.4	23.6±9.3
-5°	23.0±3.7	24.5±3.7	19.5±6.2	17.7±2.9	25.3±2.7	23.9±2.8	23.0±4.1
-7.5°	26.2±4.2	25.7±5.5	22.2±5.8	19.9±3.3	27.7±4.0	25.7±2.1	22.7±5.8

Key: ***** = significance where p<0.05 between the selected patient group to controls.
1,2,3 / 11,22,33 = significance where p<0.05 and p<0.01 respectively between the selected group to CRDD (1), UC (2) or UCR (3) implant at the same operative state.

Like peak dorsiflexion, largest dorsiflexion ROM occurred during stair descent reaching 50°. Level and uphill walking gave the lowest ROM of around half of stair descent values. Control dorsiflexion ROM was consistently higher than all patients for stair descent only and this trend was not seen during other activities. Overall, few significant differences were presented, the pre-UC group showed significantly lower ($p<0.05$) dorsiflexion ROM to controls (mean difference ± SE: 10.6±3.4°) and the pre-CRDD group (mean difference ± SE: 10.5±3.3°). These are the largest differences in ROM seen. At post-surgery again only the UC group showed significance, during level treadmill walking only. The differences and standard errors in ROM during level treadmill walking between the post-UC group and controls was 6.7±2.2°, ($p=0.028$) and to the post-CRDD group: 9.5±2.3°, ($p=0.002$). The post-UC group also showed clinically lower ROM to both other implants for all sloped walking tasks. Other notable patient ROM where clinically relevant differences to controls was seen included the CRDD group (both operative states) and the post-UC group during stair descent, and the pre-UC group at +7.5° walking. The mean differences were greater than some statistically significant differences.

6. Group Results

Postoperative differences in ROM from preoperative were small for each implant, and only one clinically relevant finding was seen in the UCR group which showed 5° less ROM during stair ascent. Although some changes were negligible, some minor trends were observed. The CRDD group showed an overall postoperative increase in dorsiflexion ROM, the UCR group showed a decrease, and the UC group showed increases and decreases in half of the activities. Postoperative inter-implant differences showed that the CRDD group exhibited higher ROM than the other implants, and the UC implant showed lower ROM values.

6.4.2.7. Summary

Sagittal plane ROM for all lower limb joints, as well as knee adduction and longitudinal rotation ROM was presented here. In general, controls showed higher hip and knee flexion and knee rotational ROM values. Patient changes from preoperative were generally an increase in hip and knee flexion, and knee adduction ROM during stance. The other parameters showed mostly showed mix of increasing and decreasing in ROM.

Throughout all parameters and activities, the UCR group often showed the smallest changes between pre- and postoperative and least frequent occurrences of clinically relevant changes, which may not conclusively imply a gain nor loss of function. This was especially evident when knee flexion ROM was considered. The mean change from preoperative during all activities achieved by the CRDD and UC groups were around 7.3° and 4.9° respectively (seen in Table 6-26), these are clinically relevant or near clinically relevant findings. For the UCR group, the average postoperative change in knee flexion ROM from preoperative was 0.6°, however changes ranged from +6° to -14.1° spreading over 20° compared to 9° and 16° for the CRDD and UC groups respectively. The UCR group was also the only group that showed a postoperative decrease in flexion ROM for both stair activities, the most functionally demanding activity.

Whilst the mobile bearing UCR implant is theoretically capable of more longitudinal rotation compared to the CRDD and UC fixed bearing implants this was not observed. At post-surgery, across all activities the CRDD group showed the highest average rotational ROM values, followed by the UCR then UC groups. However, compared to preoperative, the CRDD and UCR groups showed an overall decline in rotational ROM whereas the UC group showed an average increase in ROM.

Like the other ROM parameters, ankle dorsiflexion ROM was greatest during the stair tasks, particularly stair descent. Action at the ankle (and indeed the hip) may reflect compensatory actions as a result of lower knee function. Across all activities the CRDD group showed an overall postoperative increase in ankle dorsiflexion ROM, the UCR group a decrease, and the UC group showed roughly no change. Aside from a postoperative 5^{o} decrease in ROM by the UCR group during stair ascent, no intra-implant changes were clinically relevant.

The average intra-implant changes and standard error in sagittal pelvis, hip, knee, and ankle ROM across all activities is summarised below:

Table 6-26: Intra-implant changes in sagittal lower limb ROM across all ADLs. Results are average±standard error.

Joint (^{o})	CRDD	UC	UCR
Pelvis	-0.4±0.9	-0.5±1.3	-0.3±0.7
Hip	5.3±4.1	6.8±3.5	2.8±4.1
Knee	7.3±3.1	4.9±4.9	0.7±7.7
Ankle	1.2±1.6	0.6±3.3	-1.2±2.9

On average, all implant groups showed small postoperative reductions in pelvic tilt ROM, and increased hip and knee flexion ROM. The increased hip and knee ROM may explain the decrease in pelvic ROM, albeit negligible. The CRDD and UC groups showed small increases in ankle ROM, and the UCR group contrasted this and showed a postoperative decrease. Although no joints in the UCR group showed a clinically relevant average change in dorsiflexion ROM, the small motions imply compensatory effects (as the hip and knee undergoes a greater ROM, the pelvis and ankle undergoes

less). The fixed bearing CRDD and UC groups seem to show additional function of all joints, the CRDD group showing greater improvement than the UC group.

6.4.3. Further Sagittal Knee Angles

Knee flexion at heel strike (HS) and toe off (TO), as well as maximum and minimum knee flexion angle during stance is given in this section. Where these occur in the gait cycle is outlined in Figure 6-8. Peak and minimum stance flexion were omitted for the stair tasks as no defined stance peak was observed during these activities. In addition, sagittal angles at HS and TO for the pelvis, hip and ankle were not found to show any relevant significant differences between the participant groups, nor between pre- and postoperative, and so are omitted from this chapter.

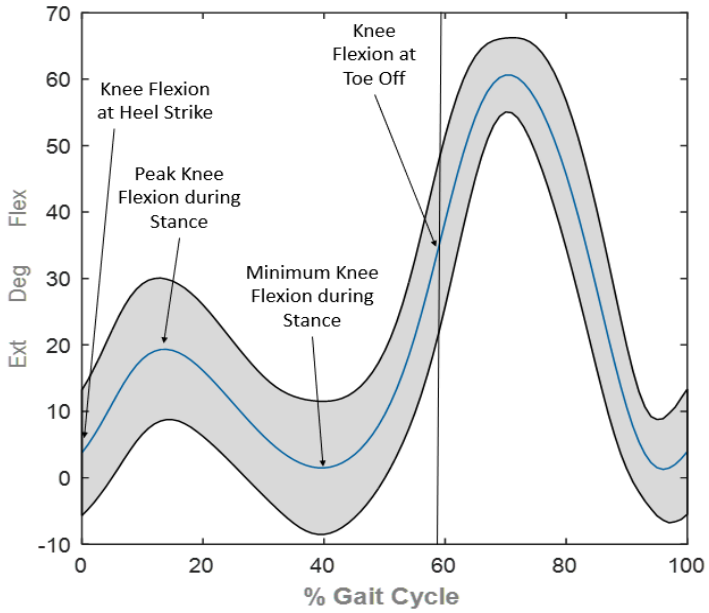

Figure 6-8: Knee flexion angle during gait and the locations of the additional sagittal parameters.

6.4.3.1. Knee Flexion at Heel Strike

The knee flexion angle at initial heel strike (KFHS) of all participants is presented in Table 6-27:

Table 6-27: KFHS of all participants for each ADL. Negative values infer joint extension.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	5.6±6.2	**-0.3±2.9††**	4.9±6.6	**-1.8±3.7†**	4.5±5.2	3.5±5.7	3.8±4.7
Stair Ascent	54.7±8.7	**49.3±6.8****	56.8±6.6	**48.8±6.0****	53.8±7.7	**47.1±11.3****	63.4±4.6
Stair Descent	7.2±8.1	**3.9±2.8***	1.8±0.3	**0.6±2.9****	3.8±7.1	3.5±7.5	10.1±4.1
0°	7.2±7.8	**-0.4±1.9†**	13.9±9.0	5.9±8.2	9.3±6.9	4.7±7.4	2.6±6.0
+5°	15.3±9.0	**10.0±4.3*****	23.0±10.0	17.9±7.3	15.9±7.8	**12.1±7.9****	25.8±6.4
+7.5°	23.9±7.8	**18.5±3. 6****	29.3±9.9	22.8±9	25.2±6.1	**19.2±10.4***	32.1±4.9
-5°	6.9±8.0	**-1.4±3.0†**	7.1±5.1	1.0±7.1	6.0±9.5	0.9±5.1	1.3±5.4
-7.5°	6±8.4	**-1.1±4.2†**	6.4±4.0	0.3±4.9	9.2±9.5	3.3±4.4	2.6±5.5

Key: *** / ** / ***** = significance where p<0.05, < 0.01 and < 0.001 respectively between the selected patient group to controls.

† / †† = significance where p<0.05 and < 0.01 respectively for the selected implant group between preoperative and postoperative.

Highest flexion angles at heel strike (KFHS) were seen during stair ascent reaching 57° in patients and 63° in controls. Uphill walking showed the next highest KFHS angles. Lowest KFHS were seen during level and decline walking. Between uphill walking inclinations, KFHS was consistently greater at +7.5° for all participants (range: 4.9° – 9.3°). Between downhill walking inclinations this difference was not held as the average difference was less than 1° (range: -0.9° – 3.2°) across all participants.

Control KFHS were consistently greater than patients' during the stair tasks and uphill walking. Aside from the pre-CRDD during stair descent and pre-UC groups during incline walking all differences were clinically relevant. Preoperative groups showed higher KFHS than controls for overground and level treadmill walking, and both decline walking tasks. However, the only clinically relevant differences were seen in the pre-CRDD and pre-UC groups in the treadmill level and -5° downhill walk, and the pre-UCR group for the treadmill level and -7.5° walking tasks. No significant difference between controls and OA groups was seen for any of the ADLs. Greater postoperative KFHS to controls was seen during treadmill level (for the UC and UCR groups) and -7.5° walking (UCR group) however not by clinically relevant amounts.

All post-surgery significance was to either the control or preoperative groups. During stair ascent, whilst no preoperative significance to controls were seen, all groups showed significantly lower postoperative KFHS ($p<0.01$). The largest mean difference ± SE to controls (16.3±3.8°) was carried out by the UCR group during stair ascent, and other postoperative groups exhibited mean differences of around 14° for the same task. During stair descent, all preoperative groups similarly showed no significance to controls, the KFHS were significantly lower than controls for the post-CRDD ($p=0.02$, mean difference ± SE: 6.2±1.9°) and post-UC groups ($p=0.007$, mean difference ± SE: 9.6±2.6°). Whilst the post-UCR group did not show a statistically lower KFHS to controls ($p=0.07$), a higher difference ± SE was seen (6.6±2.4°) than the post-CRDD group to controls which was a significant finding. Both the post-CRDD and post-UCR groups also gave significantly lower KFHS to controls for both uphill walking inclinations with the degree of significance decreasing with increasing inclination. This finding was not maintained by the UC group which displayed clinically lower KFHS to controls and no significance at both inclinations.

Intra-implant differences showed consistently lower KFHS at postoperative. The UCR group showed no significant change from preoperative however, clinically relevant decreases were observed during stair ascent, +7.5° walking and downhill walking at both inclinations. The CRDD and UC groups showed clinically relevant changes for all activities except stair descent, and the CRDD group showed a significantly lower KFHS during overground and treadmill walking, and both decline walking inclinations. These were where $p<0.05$ other than overground level walking where $p=0.008$. The UC group gave a significantly lower KFHS during overground walking only ($p=0.043$, mean difference ± SE: 6.7±3.2°). However, a greater intra-implant difference was seen in the UC group during stair ascent (-7.9±5.4°) which was not a significant finding ($p=0.15$).

6.4.3.2. Knee Flexion at Toe Off

Table 6-28 gives the knee flexion angle at the point of toe off (KFTO) seen during all ADLs:

Table 6-28: KFTO of all participants for each ADL.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	37.8±5.4	34.6±4.3	36.4±7.8	**29.9±4.5***	37.9±7.1	36.4±7.5	38.6±4.3
Stair Ascent	15.0±9.5	11.3±3.8	11.4±5.3	**3.1±8.7****	7.5±5.7	10.7±7.8	16.7±6.2
Stair Descent	82.3±7.4	79.2±6.6	87.1±5.4	78.6±2.7	81.3±7.7	74.7±13.3	86.9±6.1
0°	39.3±6.3	38.1±3.9	35.7±7.7	**30.0±3.9****	34.2±1	37.8±6	40.9±5.3
+5°	35.8±7.6	31.2±2.8	29.9±8.0	28.8±2.2	32.3±7.8	33.6±6.6	36.6±11.9
+7.5°	33.4±8.4	**28.3±5.9***	27.6±8.5	27.6±3.4	33.7±5.4	28.6±10.2	38.6±4.5
-5°	47.6±7.1	45.3±3.9	40.2±8.8	**35.6±4.4****	45.2±11.4	42.8±5	50.0±5.4
-7.5°	52.9±6.8	48.9±4.2	44.6±9.0	**39.0±6.8****	53.2±6.8	48.6±1.4	52.7±6

Key: ***/ **** = significance where p<0.05 and < 0.01 respectively between the selected patient group to controls.

Conversely to KFHS, greater KFTO angles were seen during stair descent ranging between 75°-87°, the lowest KFTO were seen during stair ascent (range: 3°-17°). As values peaked during stair descent, comparing between decline walking inclinations may show trends based on the similarity in downhill walking motion and descending stairs. Between -5° and -7.5° walking an increase in KFTO was seen for all participant groups, by 4.7° on average.

Control KFTO was mostly higher than all patients, except for three preoperative instances where increases of 0.5° or less was seen. All clinically relevant differences were where control KFTO was greater than patients. All postoperative data showed consistently lower KFTO to controls. This decrease was clinically relevant for all activities for the post-UC group, during both stair and uphill walking tasks in the post-CRDD group and, for the post-UCR group during both stair tasks, +7.5° and -5° walking. No significance to controls were seen from the post-UCR group, and one significant finding was seen in the post-CRDD group during +7.5° walking (p=0.043, mean difference ± SE: 10.3±3.4°). The mean difference was similar to other implants for the same activity where no significance was seen, however. This may be explained by the large standard deviations seen in the UCR group; however, the UC group at pre- and postoperative showed greater mean differences to controls and smaller

standard deviations than the CRDD group. Lastly, the UC group showed significantly lower KFTO to controls during overground ($p<0.05$) and treadmill level walking, stair ascent, and both downhill walking inclinations ($p<0.01$).

Intra-implant changes showed reductions in KFTO for the CRDD and UC implants for all activities, the UC group showing greater reductions. The UCR group showed mostly postoperative decreases (which were clinically relevant during stair descent and +7.5º walking) and non-relevant increases during stair ascent, treadmill level and +5º uphill walking. Overall, the UC group gave the lowest KFTO angles, and the CRDD and UCR groups showed similar post-surgery values.

6.4.3.3. Peak Knee Flexion during Stance

The peak knee flexion angles seen during stance (PKFS) are presented in Table 6-29. As this peak does not appear during stair navigation, results from this activity are not shown:

Table 6-29: PKFS of participants during level and sloped walking activities.

Activity (º)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	17.1±5.4	12.9±4.2	13.7±5.9	**7.6±7****	14.8±6.2	12.5±7.2	19.5±5.3
0º	15.3±7.5	11.4±4.2	19.1±5.1	11.2±8.3	15.8±5.4	14.4±7.8	18.2±7.1
+5º	22.3±8.9	**18.3±4****	25.3±8.5	20.2±7	22.2±6.6	21.1±7.4	30.3±6.7
+7.5º	27.5±8.5	**23.7±3.7***	30.2±9.1	24.3±8.2	28.3±4.3	25.3±9.2	34.5±5.5
-5º	21.5±5.0	**16.2±6.5***	**15.1±6.8***	**9.5±10.3****	19.8±7.4	**14.5±6.7***	28.1±6.6
-7.5º	**22.3±5.8***	**19.6±6.0***	**15.7±7.5****	**9.9±10.1*****	24.5±1.2	20.1±6.5	32.8±7.7

Key: *** / ** / ***** = significance where $p<0.05$, < 0.01 and < 0.001 respectively between the selected patient group to controls.

PKFS angle was greatest during +7.5º walking, ranging from 23.7º-34.5º across participants. Stance peaks at this inclination were, on average 5º higher than +5º walking. Lowest PKFS was seen during overground and treadmill level walking, the average difference between walking modalities less at around 1º. Aside from one non-relevant exception, controls showed a greater PKFS compared to all patients and

all activities. The pre-CRDD and pre-UCR groups showed clinically lower PKFS to controls during all sloped walking tasks, however only -7.5° walking in the pre-CRDD group was significantly lower than controls. The mean difference ± SE was 10.5±3.1° (p=0.016) which was the largest mean difference of all preoperative CRDD and UCR group differences.

Postoperative PKFS decreased from preoperative and, by extension, from controls. No significant intra-implant differences were seen in any groups, and all postoperative PKFS showed clinically relevant decreases to controls. The post-UC group showed a greater degree of significance to controls in both downhill walking tasks from preoperative. In addition, the post-UC group showed clinically relevant intra-implant decreases for all activities (greatest difference of 8°). The other groups showed a single clinically relevant intra-implant decrease during -5° walking by 5.3°.

6.4.3.4. Minimum Knee Flexion during Stance

The lowest knee flexion angle during stance (MKFS), following stance peak flexion and preceding swing phase is shown in Table 6-30. Again, since this parameter is not seen during the stair tasks, these activities have been omitted:

Table 6-30: MKFS of all participants for the level and incline walking activities. Negative values = extension angles.

Activity (°)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Level Walking	**10.1±7.2***	**4.0±4.2†**	9.4±6.0	**-1.0±6.2†**	9.0±7.1	5.0±6.2	1.3±5.1
0°	7.8±8.2	**0.6±5.1†**	8.4±7.0	**-3.0±6.9†**	10.2±5.2	5.2±6.1	3.1±7.2
+5°	9.4±9.2	**-1.0±5.8††**	7.7±6.7	**-3.3±6.4†**	9.5±5.5	4.9±6.1	2.3±6.9
+7.5°	9.1±9.0	**-1.4±6.6†**	6.1±4.3	-3.5±7	9.3±5.3	3.9±7.1	3.3±6.7
-5°	14.4±6.7	8.1±5.5	11.5±7.9	1.6±6.4	12.8±10.8	8.8±4.4	13.6±8.7
-7.5°	17.1±7	12.3±4.5	12.9±7.1	5.7±7.5	19.2±1.3	14.3±4	19.7±9.9

Key: ***** = significance where p<0.05 between the selected patient group to controls.
† / †† = significance where p<0.05 and < 0.01 respectively for the selected implant group between preoperative and postoperative.

6. Group Results

MKFS was greatest at -7.5° walking and lowest during level and uphill walking, with some hyperextension exhibited in the post-FB groups. Preoperative MKFS angles greater than controls by a clinically relevant amount was seen during level and uphill walking for all implant groups except the preoperative CRDD and UC groups during treadmill level and +7.5° walking respectively. The largest difference in pre-surgery MKFS to controls was observed in the CRDD group during overground level walking (mean difference ± SE: 8.7±2.9°). In addition, this was the only significant finding to controls (p=0.028).

All other significance were intra-implant changes, seen in the CRDD and UC groups only. Whilst all groups showed postoperative decreases in MKFS throughout all activities, for the UCR group the only clinically relevant decrease was seen during +7.5° walking. The CRDD and UC groups showed significantly lower MKFS for both level walking activities and +5° walking, the CRDD group also showed significance to preoperative during +7.5° uphill walking. The smallest mean difference ± SE seen for a significant finding was 6.0±2.5° (p=0.021, by the CRDD group during overground level walking) and other non-significant findings showed larger mean differences than this, particularly the UC group where larger standard deviations were also reported.

Comparing inter-implant postoperative differences, the UCR group showed the largest MKFS in all activities and, along with not showing significance to preoperative nor controls, may be considered to have improved function the least. The UC group showed the lowest MKFS for all activities which may be considered an improvement in function. This group also showed the largest average differences to controls which could imply this change is detrimental, although not significant. The CRDD group showed greater postoperative decreases in MKFS than the UCR group, and less than the UC group.

6.4.3.5. Summary

Further sagittal knee parameters were provided in this section, flexion at heel strike (KFHS) and toe off (KFTO) gives an indication of how straight the limb was at the start of stance and swing phase respectively for all activities. During the level and sloped walking tasks stance range of motion (PKFS and MKFS) was also reported. Broadly speaking, controls showed a more flexed knee at heel strike and toe off to patients, particularly to postoperative groups. A higher PKFS angle was also observed in controls, and conversely preoperative patients showed higher MKFS.

All postoperative parameters mostly decreased from preoperative or were higher by a non-clinically relevant amount. Regarding KFHS and KFTO this means postoperative patient limbs were more extended at heel strike and toe off than at preoperative. This is especially relevant for stair ascent and uphill walking, which are demanding activities and gave the highest KFHS. A more flexed knee during these tasks implies greater load bearing capabilities to be able to lift the body upwards and forwards. All patients showed a reduction in KFHS which may be considered a loss of function as a straighter limb may imply less confidence in the joint as more congruency and stability is seen compared to a flexed joint.

Whilst KFHS was lowest during stair descent and downhill walking, KFTO was greatest. Lowest KFTO were seen during stair ascent and uphill walking reflecting the nature of these activities. Controls showed greater KFTO for these activities indicating greater ground clearance prior to swing. Post-surgery patients showed reductions in KFTO from preoperative across all activities, except the post-UCR group that showed non-clinically relevant increases in three activities. The post-UC group also showed the lowest patient KFTO and the post-CRDD group showed the highest.

All patients showed a postoperative decrease in PKFS. Although the post-CRDD group showed significant reductions to controls for more activities, the post-UC group gave insignificant mean differences to controls that surpassed the CRDD group, particularly during downhill walking. The post-UCR group showed the smallest mean decrease in PKFS across all activities compared to controls and preoperative and often showed the largest PKFS at the inter-implant level.

MKFS was lowest in the level and uphill walking tasks and preoperative MKFS was higher than controls for all level and uphill walking tasks. However, only one significant finding to controls (pre-CRDD group during a level walk) was reported. Postoperative MKFS decreased for all implants and activities, and some hyperextension was also realised in the FB groups. This intra-implant decrease was greatest in the UC group which showed larger decreases than the other implant groups, although less significance. Postoperative MKFS in the UC group was also consistently lower than controls for all tasks. The post-CRDD group behaved similar to the UC group, albeit not to the same magnitude and the post-UCR group showed the smallest intra-implant decreases in MKFS, and four activities showed a MKFS higher than controls.

6.4.4. Joint Kinematics Summary

The kinematic data presented in this chapter included peak lower limb angles, joint ROM, and additional sagittal knee findings during the ADLs. Stair navigation was most functionally demanding as the highest peak and range of kinematics were seen here. Furthermore, highest knee flexion at heel strike and toe off were also seen during this activity. Whilst the scope of this project did not involve comparing the difference between level treadmill and overground walking, some small differences were observed. Across all participants, overground level walking showed 2^{o} and 3^{o} higher peak knee flexion and ROM respectively than treadmill walking. The same small difference was seen for maximum sagittal angles, sagittal ROM, and additional knee flexion during stance and at heel strike and toe off. This indicates participant kinematic profiles across both level walking types were near equivalent.

Between ±5^{o} and ±7.5^{o} sloped walking inclinations, some trends were noted as inclinations became steeper. From +5^{o} to +7.5^{o} increases in peak knee and ankle flexion was seen, as well as larger hip and knee flexion ROM. In addition, KFHS increased with steeper uphill walking and more considerably the PKFS also become greater. Compared to -5^{o}, walking at -7.5^{o} also showed larger peak and ROM for knee and ankle flexion. Conversely to the knee and ankle, hip flexion ROM decreased with

increasing downhill steepness. Stance flexion ROM decreased at -7.5° downhill walking, whilst PKFS increased across inclinations, MKFS also increased (became more positive) and so the net change resulted in a greater stance ROM during -5° walking. It is important to consider that this task was not completed for several participants either due to hardware issues or participant mobility, so the sample sizes is less for this task than others.

The range of knee flexion during stance was not calculated by PKFS less the overall lowest angle seen in stance. Instead, the minimum angle following the stance peak was used as the lowest stance angle. Healthy flexion traces commonly show two distinct flexion peaks and finding the extension maximum between these is more meaningful than finding the lowest angle during stance which may occurred at heel strike and would have been accounted in the overall ROM. OA patients commonly display some degree of flexion contracture that is often corrected with TKA. This correction was observed in all implant groups, with the UC group showing the greatest postoperative reduction in MKFS, then the CRDD group, then the UCR group.

Knee flexion data showed the largest variance of all joints. All post-surgery groups showed reductions in peak knee flexion, whilst ROM mostly showed an increase from preoperative, further implying improvements in joint extension was realised. Of all patients, the CRDD group showed the largest postoperative peak flexion for all activities (although ~11° less than controls on average). The UC group showed the lowest flexion angles for all activities excluding stair navigation where the UCR group showed considerably smaller peak flexion. Knee flexion ROM findings, however, do not complement this. Similar to peak flexion, the post-CRDD group showed the highest ROM for most activities apart from the stair tasks where the UC group showed the largest ROM. The UC group also showed the lowest ROM for level and both downhill walking inclinations, and the UCR group showed the lowest ROM for all other activities at post-surgery.

Reductions in peak knee flexion may be compensated by increases in peak flexion in other joints. This is unlikely to be at the ankle as intra-implant differences generally

showed non-clinically relevant decreases in peak dorsiflexion at postoperative. However, inter-implant differences showed that the post-UCR group demonstrated highest dorsiflexion for most of the activities, including the stair tasks where this group showed the exhibited lowest peak flexion and ROM. The post-UC group on the other hand, showed the lowest knee flexion for all tasks except stair navigation, and lowest dorsiflexion for all activities. Peak pelvic and hip angles were not available for comparison but may explain some of these findings.

High knee adduction during stance confirmed the presence of cross talk error, particularly in the CRDD group. As well as errors in the frontal plane, transverse plane motion is also known to be incorrectly reported. Peak rotation was highest in the post-UC group which also showed the lowest peak adduction of all implants for five activities. Taking values as they were, the smallest peak rotation was seen in the mobile bearing UCR group for five out of all ADLs.

Significance was mostly to controls or preoperative implying inter-implant differences were not significant, however clinically relevant findings may gauge improvements. Considering the greater sagittal peak and ROM angles displayed by the CRDD group it may be considered that they showed a better kinematic profile than the other implants at postoperative. However, since some function was lost by all implants (i.e., peak knee flexion during the whole gait cycle and during stance only) some improvements are required to regain function to preoperative levels, and eventually to controls. The UCR group perhaps shows reduced function compared to the other implants at postoperative, namely peak and range of knee flexion. The proceeding chapter presents changes in kinetic parameters which will further elucidate implant performance.

6.5. Joint Kinetics

The following chapter focusses on the hip, knee and ankle and their associated moments and powers. Only kinetic data during overground level walking and stair navigation were available since the force plates of the Motek CAREN system were not

synchronised with Vicon Nexus and once the system was at an incline the force readings were susceptible to an inertial shift.

Since the stair tasks showed smaller sample sizes than level walking (Table 6-5) as the incidence of the patient striking the instrumented step in the staircase with their affected limb was left to chance, only level walking kinetics will be presented as the statistical power is higher.

6.5.1. Joint Moments

The joint moments described are the external sagittal hip, knee, and ankle moments, as well as the knee adduction moment, a frontal plane moment. No rotational knee moments are reported as moments were small and similar between groups with no significant differences seen. Compound control moment traces for all parameters is presented (Figure 6-9) to describe the data points used for statistical analysis in Table 6-31. Stance phase is from 0% - 55% of the gait cycle graphs, this was the approximate percent of toe off for controls in this activity only and for analysis individual stance phase was used. For the hip, the peak flexion (HFM) and extension (HEM) moment occurred around heel strike and toe off, respectively. The knee flexion moment (KFM) shows two distinct flexion moment distinct peaks (in early (KFM1) and late (KFM2) stance) and an extension moment maximum (KEM) in midstance. The knee adduction moment (KAM) typically gives a bimodal pattern and the points analysed were the peak KAMs at early and late stance (KAM1 and KAM3) and a mid-stance adduction minimum (KAM2) between KAM1 and KAM3. In addition, the maximum abduction moment was reported. Lastly, peak ankle dorsiflexion moment (DFM) before toe-off was analysed in addition to the maximum plantarflexion moment (PFM).

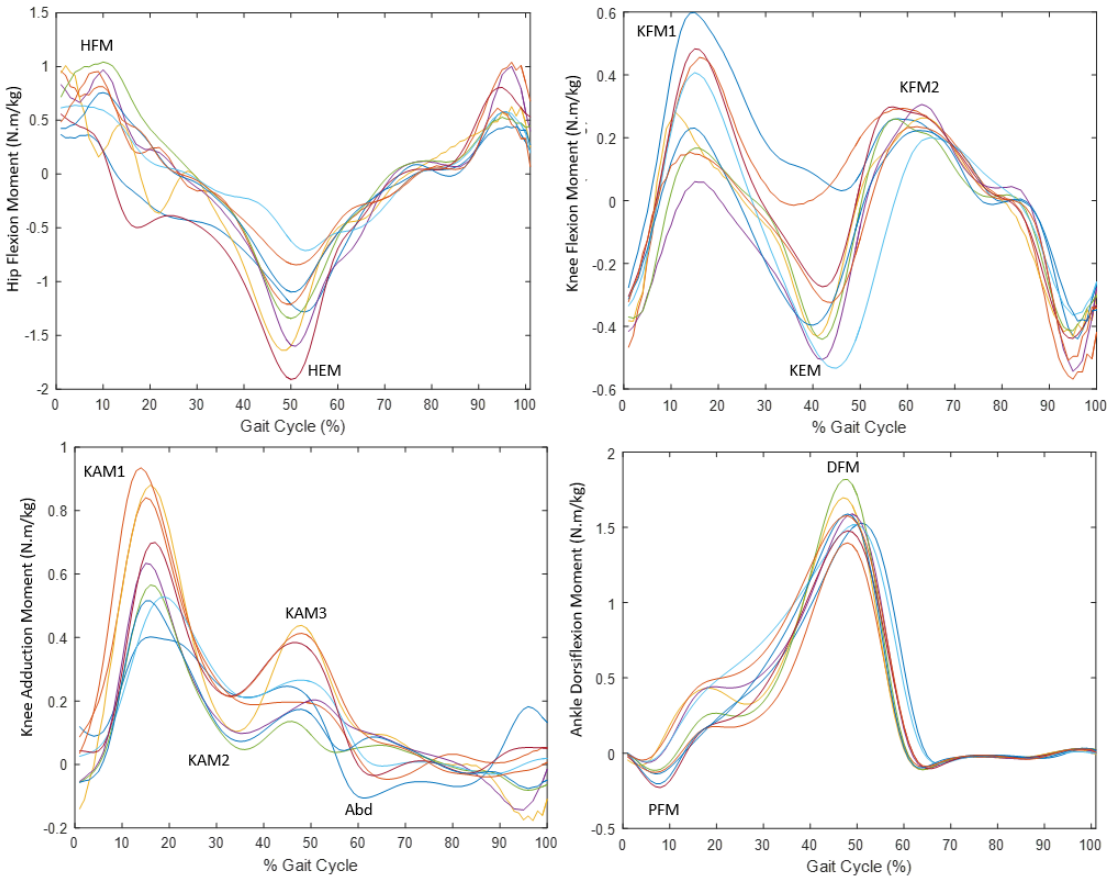

Figure 6-9: Compound control joint moment traces during overground level walking. Moments presented (clockwise from the top left) are hip flexion, knee flexion, ankle dorsiflexion and knee adduction.

6. Group Results

The maximum sagittal moments for all lower limb joints and frontal knee moments during the stance phase of the gait cycle is given in Table 6-31:

Table 6-31: Peak sagittal (all joints) and frontal knee moments during level walking.

Joint Moment (N.m/kg)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
HFM	**0.50±0.22***	**0.72±0.2†**	**0.48±0.17***	0.70±0.28	**0.45±0.14****	0.60±0.12	0.82±0.25
HEM	0.96±0.17	0.97±0.3	**0.86±0.3***	1.00±0.16	**0.90±0.32***	**0.83±0.27***	1.32±0.37
KFM1	0.36±0.26	**0.12±0.12**,†**	0.17±0.13	**-0.01±0.22****	0.27±0.24	0.19±0.2	0.42±0.21
KEM	0.01±0.26	**0.20±0.11†**	0.06±0.15	**0.37±0.16†**	-0.02±0.17	0.07±0.18	0.26±0.17
KFM2	0.26±0.16	**0.18±0.04*****	0.20±0.1	0.23±0.08	0.24±0.1	**0.18±0.04****	0.28±0.04
KAM1	0.55±0.21	0.48±0.15	0.47±0.3	0.46±0.09	0.61±0.12	0.49±0.15	0.67±0.19
KAM2	**0.35±0.15***	**0.21±0.13†**	0.29±0.26	0.23±0.13	**0.38±0.08***	0.27±0.16	0.15±0.07
KAM3	0.43±0.15	0.29±0.14	0.35±0.31	0.33±0.14	0.46±0.09	0.33±0.18	0.27±0.1
Knee Abduction	0.03±0.04	0.05±0.03	0.05±0.03	0.04±0.05	0.00±0.01	0.02±0.05	0.02±0.06
Ankle DFM	**1.33±0.19***	**1.51±0.16†**	**1.19±0.04****	1.36±0.11	**1.24±0.21****	**1.28±0.19[1,]****	1.57±0.11
Ankle PFM	0.09±0.06	0.06±0.06	0.04±0.03	0.05±0.05	0.07±0.03	0.09±0.07	0.13±0.06

Key: ***/ ** / ***** = significance where p<0.05, < 0.01 and < 0.001 respectively between the selected patient group to controls.

† = significance where p<0.05 for the selected implant group between preoperative and postoperative.

[1,2,3] = significance where p<0.05 between the selected group to CRDD (1), UC (2) or UCR (3) implant at the same operative state.

The ankle showed the largest moments during this activity and the hip showed the second greatest. Knee abduction and ankle plantarflexion moments were near negligible, and no significance was seen between any groups for these parameters. Controls showed mostly larger moments to patients except for KAM2 and KAM3. All peak sagittal knee moments were also plotted to illustrate the relationship between the moments and participant groups further (Figure 6-10):

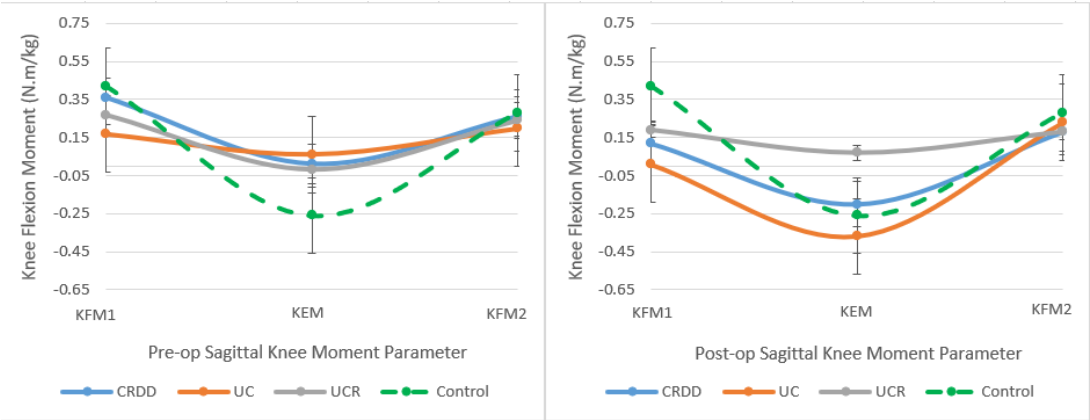

Figure 6-10: Participant KFM and standard deviation at early, mid, and late stance during level walking. Patient moments are given at preoperative (left) and postoperative (right).

Preoperative knee flexion moments (KFM1 and KFM2) were statistically similar to controls, following the surgery peak KFMs either declined or remained the same. Compared to controls, the post-CRDD and UC groups showed significantly lower KFM1 ($p<0.01$) and the post-CRDD and UCR groups gave significantly lower KFM2 ($p<0.01$ and 0.001 respectively). Although the post-UC group gave similar KFM2 as the other implants, no significance to controls was seen, likely due to the greater standard deviation seen. The largest knee flexion moment difference ± SE between controls to patients was seen to the post-UC (KFM1) group of 0.42±0.1 N.m/kg. Control peak knee extension moment (KEM) was higher than all patients except the post-UC group and no significance to controls was seen. All groups showed a near negligible KEM at preoperative. All postoperative groups showed an increase in KEM, the UCR group by a lesser, non-significant amount ($p=0.53$), then the CRDD group (0.19±0.08 N.m/kg, $p=0.03$) and the UC group by the largest amount (0.31±0.1N.m/kg, $p=0.04$).

In the frontal plane all patients showed higher KAM2 and KAM3 to controls. Preoperative KAM2 was higher than controls by 0.14-0.23 N.m/kg and this was significant in the pre-CRDD ($p=0.026$) and pre-UCR groups ($p=0.022$). Although showing a similar mean difference as the CRDD group, the pre-UC group showed no significance to controls ($p=0.543$) likely due to the larger standard deviation. No significance to controls was seen for the other KAM peaks. Whilst all implants showed

an intra-implant reduction in KAM2 by a similar amount, this was only a significant finding in the CRDD group ($p<0.05$).

All groups showed postoperative increases in peak sagittal hip and ankle moments. For both of these moments significance to controls was seen at preoperative ($p<0.05$ or $p<0.01$) which were mostly not seen at postoperative. The exception to this was the post-UCR group which maintained a significantly lower peak DFM to controls ($p=0.006$, 0.33±0.08 N.m/kg) as well as the CRDD group ($p=0.03$, mean difference±S.E: 0.23±0.07 N.m/kg). Regarding extension moments, the hip gave the largest moments followed by the knee. Controls showed higher hip extension moments (HEM) to all patients which was significant to the UC (preoperative) and UCR groups (both operative states) where $p<0.05$. The CRDD group showed a negligible change in peak HEM between operative states, and a postoperative increase for the UC group and reduction in the UCR group was seen. This postoperative decrease resulted in the largest mean difference in moments seen between controls and patients of 0.49±0.2N.m/kg ($p=0.015$).

The CRDD group showed significant intra-implant differences from preoperative for five parameters: peak HFM, KFM1, KEM, KAM2, and DFM and all changes were where $p<0.05$. This may imply this implant group showed the greatest improvement from preoperative. As the UC and UCR groups showed similar postoperative values, and the same mean difference as the CRDD group was seen in UC group for three parameters (peak HFM, KAM2, and DFM), and the UCR group for one parameter (KAM2) both of these may be considered to have improved function to somewhat of a clinically relevant extent and lesser extent than the CRDD group.

6.5.2. Joint Powers

Peak concentric (generative) and eccentric (absorptive) powers are shown for the hip, knee, and ankle during overground level walking. These are the scalar product of the joint moments multiplied by the angular velocity given in watts per kilogram (W/kg). To illustrate the data points extracted for analysis, compound control power traces

are plotted for the hip, knee, and ankle (Figure 6-11). Again, stance is from 0% - 55% of the gait cycle for this group for this activity. Peak concentric and eccentric powers for the hip (CHP, EHP) and ankle (CAP, EAP) occurred around toe off, and for the knee peak generative power (CKP) during midstance and absorptive power (EKP) at toe off were investigated.

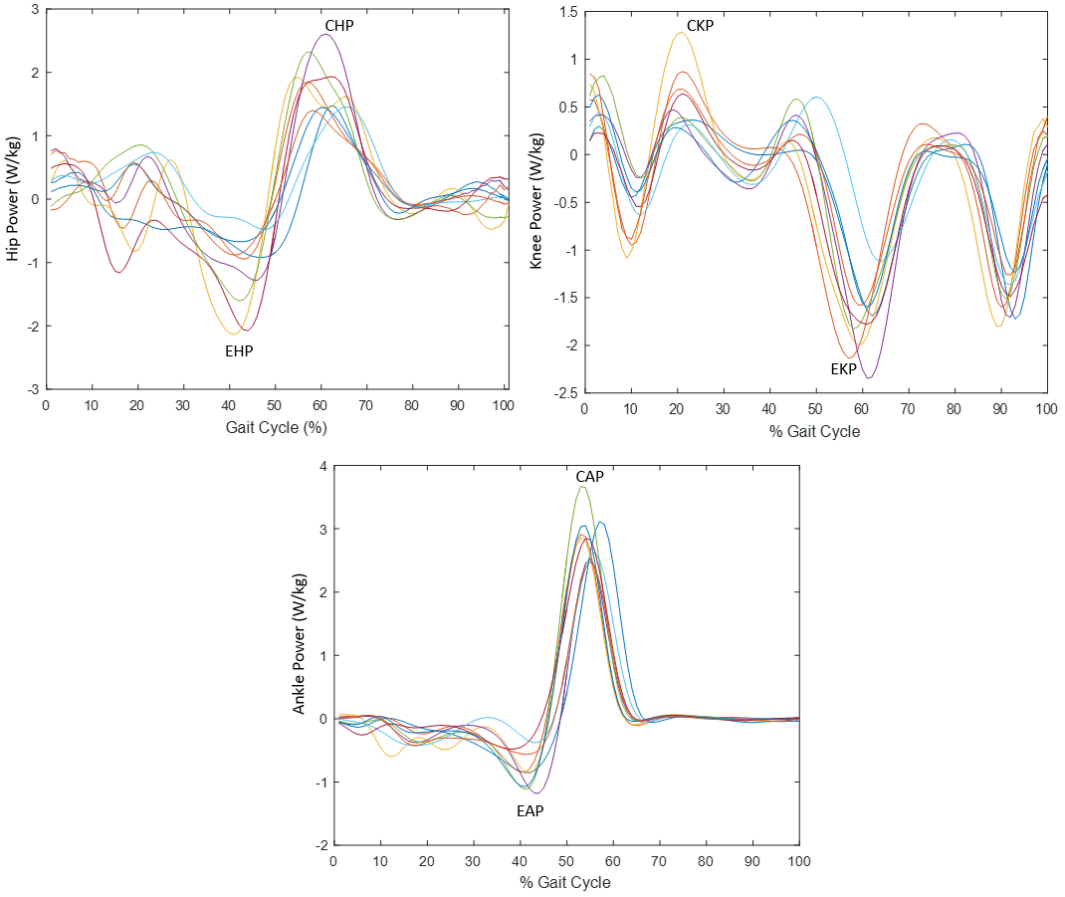

Figure 6-11: Compound control joint powers during overground level walking. The joints presented are the hip, knee, and ankle (clockwise from top left).

Table 6-32 displays peak concentric and eccentric joint powers of all participants during overground level walking:

Table 6-32: Peak concentric and absorptive joint powers during level walking.

Joint Power (W/kg)	CRDD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Concentric							
Hip	**0.84±0.27*** **	**0.92±0.21*** **	**0.72±0.27*** **	**0.95±0.18*** **	**0.86±0.43*** **	**0.88±0.26*** **	1.89±0.5
Knee	**0.18±0.13*** **	**0.20±0.12*** **	**0.12±0.08*** **	**0.20±0.15** **	**0.23±0.15** **	**0.20±0.12** **	0.71±0.37
Ankle	2.30±0.7	2.78±0.64	**1.68±0.26***	2.24±0.61	**1.89±0.84***	2.03±0.81	2.86±0.4
Eccentric							
Hip	0.96±0.52	0.86±0.53	0.67±0.53	0.50±0.27	1.06±0.34	0.90±0.31	1.15±0.57
Knee	**0.80±0.26*** **	**0.66±0.17*** **	**0.59±0.4*** **	**0.78±0.52*** **	**1.09±0.35*** **	**1.06±0.21*** **	1.58±0.47
Ankle	0.65±0.22	0.70±0.16	0.59±0.16	0.79±0.16	0.77±0.16	0.84±0.23	0.81±0.26

Key: *** / ** / ***** = significance where p<0.05, < 0.01 and < 0.001 respectively between the selected patient group to controls.

The ankle showed the highest concentric powers, and the hip and knee showed peak eccentric powers in the patient and control groups respectively which may reflect compensatory strategies in patients. Controls showed greater power magnitudes than patients for all joints: particularly the concentric hip and eccentric knee powers. In addition, all significance was to controls only. All patients' peak concentric hip powers (CHP) showed the same degree of significance to controls whereby p<0.001. All patients' peak concentric knee powers (CKP) were also significantly lower than controls where p<0.01 or 0.001 and all patients showed similar CKP's values, with the post-UC group showing the relatively large intra-implant change of +0.08W/kg for this parameter (over double the changes seen by the other implants). Patient peak concentric ankle power (CAP) were significantly lower than controls in the pre-UC and pre-UCR groups only (p=0.01, 1.18±0.3W/kg and p=0.03, 0.97±0.3W/kg respectively). At post-surgery, both the CRDD and UC groups showed large increases in peak CAP of 0.48W/kg and 0.56W/kg respectively, whereas the post-UCR group showed a smaller increase of 0.14W/kg from preoperative.

Peak eccentric hip and ankle powers were statistically similar to controls, despite some large mean differences. Namely the post-UC groups' EHP where this group

showed a larger or similar difference in EHP to controls as seen between the control and all patient groups' peak CKP which was significant. So, the difference in EHP between post-UC and controls (0.65±0.3W/kg, p=0.22) may be considered clinically relevant. Intra-implant differences showed that all implant groups' EHP became less eccentric at post-surgery. ad EAP increased close to control levels. The knee showed less eccentric power (EKP) at postoperative for the CRDD and UCR groups and an increase in EKP was seen in the UC group. Although patients at both operative states showed significantly less peak EKP than controls ($p<0.001$), the increase in power by the UC group for this and other parameters implies a greater improvement in function for this group. The CRDD group showed comparatively smaller improvements and the UCR group possibly showed the least improvement.

6.5.3. Joint Kinetics Summary

Hip, knee, and ankle kinetic parameters were presented here, including sagittal joint moments, frontal knee moments and all joint powers during overground level walking. Kinetic data were not available for sloped walking, nor was appropriate to be analysed for stair negotiation due to low power but is presented in Appendix 10: Stair Navigation Kinetic Data. All statistical analysis was carried out on data to a high precision and only reported to two decimal places here.

Across parameters some discrepancies in significance were seen, whereby some values were not found to be significantly different to controls or preoperative, despite other significant values giving the same or smaller mean differences. This is likely due to higher standard deviations seen in the unsignificant results. With kinematic parameters a 5^{o} threshold in mean difference could be applied to determine clinically relevant parameters which may not have flagged as statistically significant. A similar threshold could not be applied to kinetic parameters as none have been determined in the literature. Furthermore, clinically relevant kinetic thresholds may vary depending on the joint, parameter and activity so in this instance, overall mean differences were used for general comparison and differences similar to significant differences were considered clinically relevant.

6. Group Results

The largest moment and power values were given by the ankle, namely the dorsiflexion moment and concentric generative power. Controls generally showed greater sagittal moments and powers to all patients, with the exception of peak KEM which was greater in the post-UC group (albeit not significant). In addition, the pre-CRDD peak KFM equalled controls. In the frontal plane, controls showed larger peak early stance KAM1 than all patients, and all patients exhibited greater KAM2 and KAM3 peaks. At post-surgery, the difference in KAM2 and KAM3 to controls lessened.

From preoperative, all patients exhibited increased peak HFM and DFM and reduced peak KFM and KAM. In addition to less postoperative KFM, increases in peak KEM was seen. However contrasting changes were seen for the hip moments, whilst patient HFM increased from preoperative, peak HEM showed minimal intra-implant change in the CRDD group, an increase in the UC group and a reduction in the UCR group. The UC group also showed the greatest intra-implant increase in moment by 0.31N.m/kg in KEM, and this group generally showed the largest postoperative increases across all kinetics.

Patients showed considerably lower powers to controls and the largest difference between controls and patients were for the concentric hip and ankle powers up to 1.18W/kg. As joint powers are the dot product of joint moments and its angular velocity, the lower patient moments lend itself to lower powers. Although not all moments in all directions were analysed it may also be inferred that patient joint angular velocities were equal to or lower than controls. Patient concentric powers showed a high degree of significance to controls, and considerably smaller intra-implant increases in concentric hip and knee powers were seen compared to the ankle. All patients showed an intra-implant increase ranging between 0.05-0.2W/kg in eccentric ankle power to control levels. For the hip and knee, the CRDD and UCR groups showed reductions from preoperative eccentric power ranging 0.03-0.17W/kg, and the UC group conversely showed an increase in EKP by 0.19W/kg.

Reductions in patient knee moments may be complemented by increases seen in hip and ankle function to near control levels. Postoperative patient hip and ankle powers

tended more towards generation and knee powers tended more towards absorption, possibly linked to increases in hip and ankle flexion moment and reductions in knee flexion/adduction moments.

6.6. Self-reported Questionnaire Answers

All participants were asked how they found each activity including the degree of difficulty, pain, and tiredness experienced. Further task-specific questions were also asked to gauge how carrying out the ADL in the laboratory compared to outside of the laboratory setting. All statistical analysis was carried out using multiple Fischer's Exact test with a Bonferonni correction and all significant differences are marked with asterisks. A single asterisk denotes significance where $p<0.05$, a double asterisk denotes $p<0.01$, and a triple asterisk represents significance where $p<0.001$ between the selected groups.

6.6.1. Level Walking

The responses related to the level walking task are presented first. The percentage responses for how much difficulty each participant group experienced during this activity is presented in Figure 6-12:

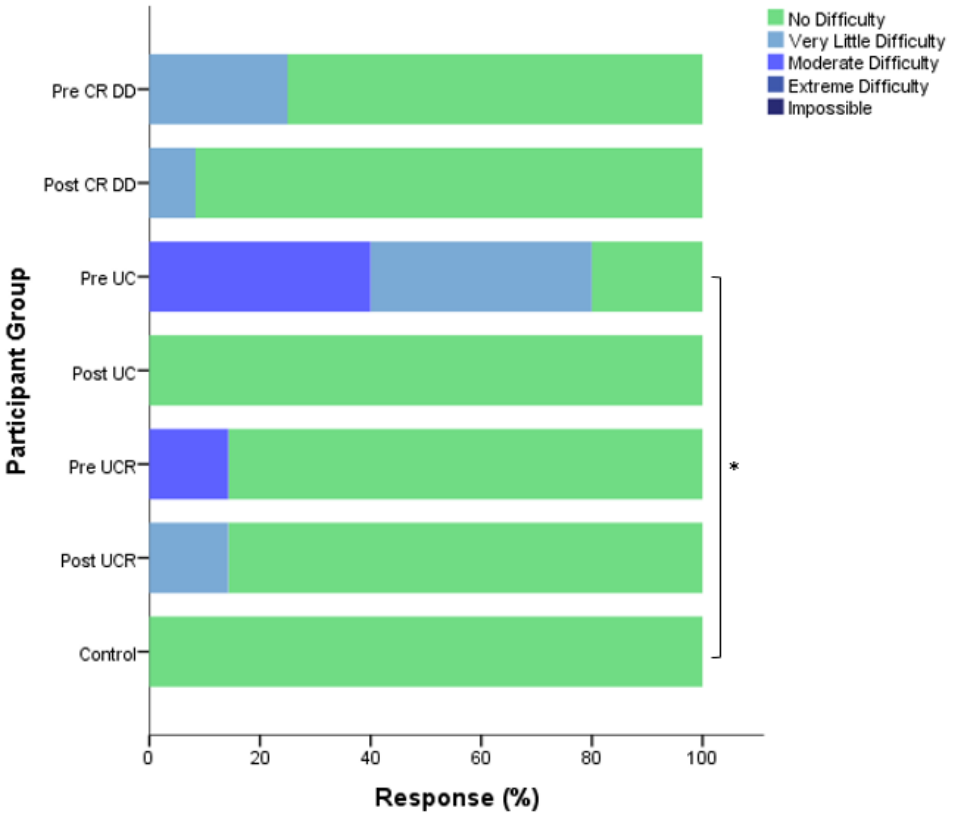

Figure 6-12: Questionnaire responses for how difficult each participant found level walking.

A higher portion of preoperative patients expressed difficulty or more severe difficulty which decreased at postoperative, whilst controls reported no difficulty. The UC group showed the greatest improvement as 80% of patients reported some level of difficulty (two people reported very little difficulty and two others reported moderate difficulty) at preoperative which fell to 0% at post-surgery. Furthermore, the pre-UC group showed the only significantly different responses to controls where p=0.03 which was not seen post-surgery. The UC group was also the only group where no patients reported any difficulty at postoperative.

6. Group Results

How much pain each participant group experienced during level walking is shown in Figure 6-13:

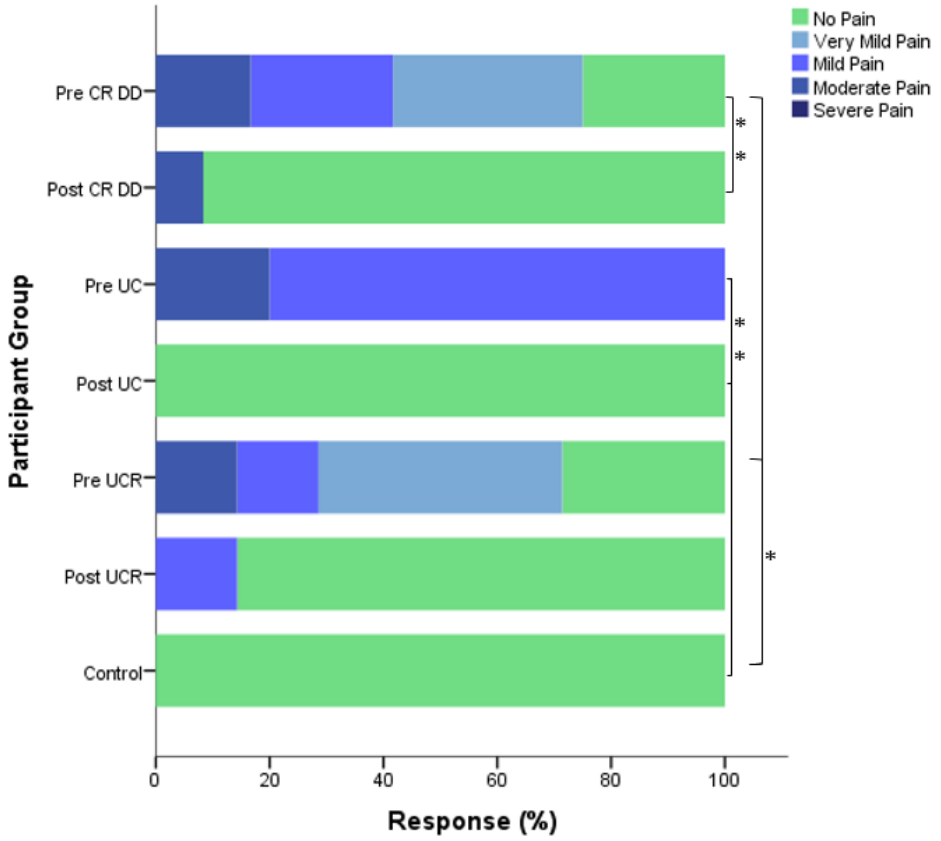

Figure 6-13: Questionnaire responses for how much pain each participant experienced during level walking.

Again, controls reported no pain during this task and pre-surgery implant groups reported the most pain, with the whole of the UC group reporting some degree of pain. This was significantly more than the controls (p=0.003) and significantly improved post-surgery where no pain was reported (p=0.008). The pre-CRDD and UCR groups also experienced significantly more pain than controls (p=0.014 and 0.029 respectively), which improved at postoperative as groups were statistically similar to controls. Intra-implant differences gave no significance between pre- and postoperative pain responses for the UCR group (p=0.086), and a significant improvement in reported pain by the CRDD group (p=0.002).

How tiring each participant found the activity is presented in Figure 6-14:

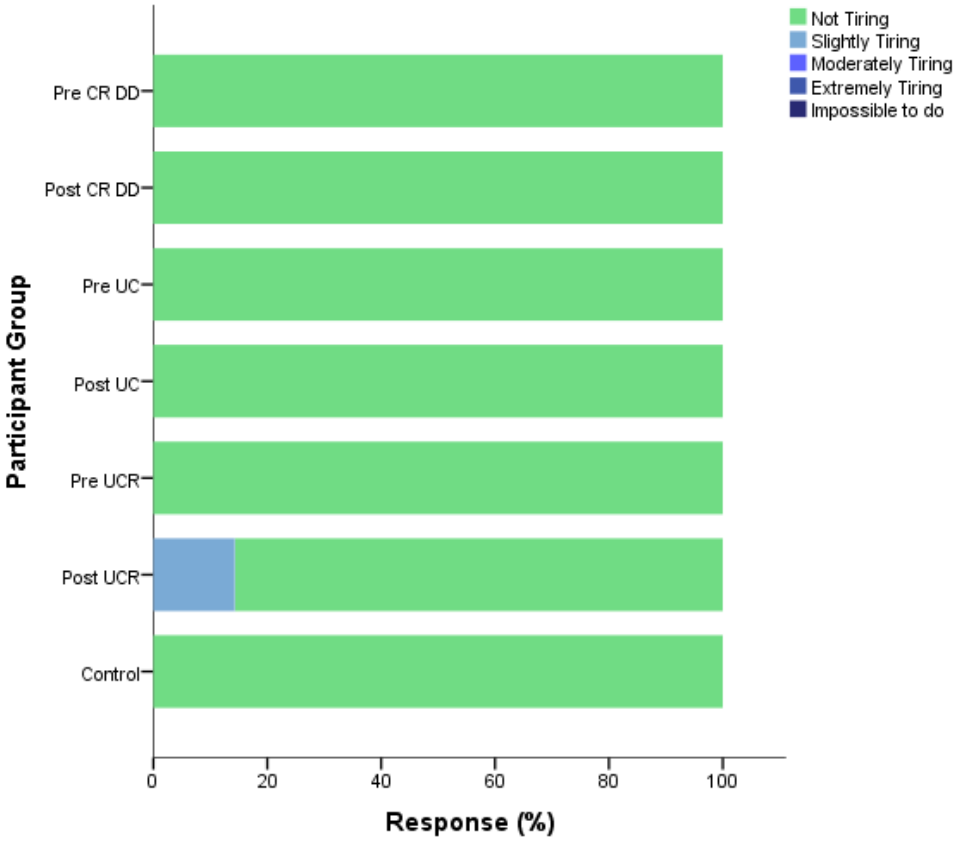

Figure 6-14: Questionnaire responses for how tiring each participant found level walking.

All groups aside from the post-UCR group reported no tiredness for this task. Although one participant in the post-UCR group expressed some slight tiredness at postoperative, this was not a significant finding to any other group.

Lastly, the responses for how much longer each participant felt they could continue walking is given in Figure 6-15:

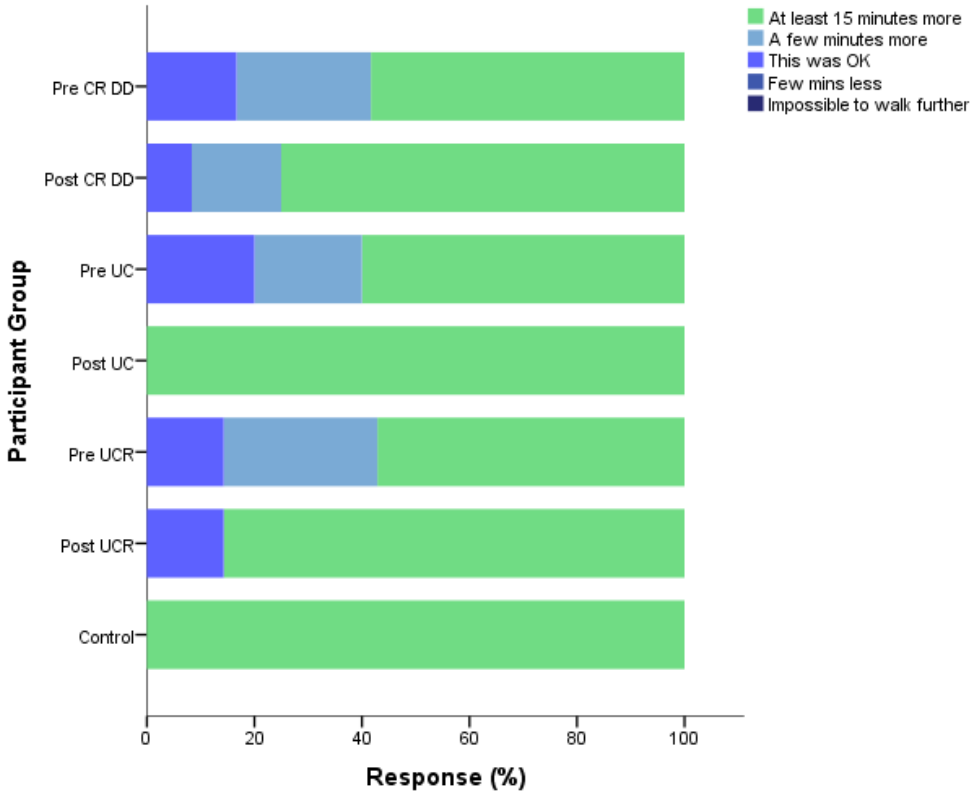

Figure 6-15: Questionnaire responses for how much further each participant felt they could continue walking following the level walk.

Patient answers ranged being able to walk 15 minutes more to no further walking would be possible. Whilst no significance was seen between any of the groups, some general trends were observed: 100% of the post-UC group stated they could walk for at least 15 minutes longer, the proportion of patients who felt they could not walk at least a few minutes longer decreased for the CRDD group, and in the UCR group the same proportion of patients at both operative states felt the amount walked in the laboratory was enough.

6.6.2. Stair Navigation

Questionnaire answers relating to stair navigation is presented here, and the sample sizes are equal to those seen in the level walking questionnaire responses as those who walked with a step-by-step strategy are included here. To gauge whether the laboratory staircase was appropriate to replicate this activity as an ADL all 33 control

and patient participants were asked whether they had stairs in their home. 28 participants (84.8% of the whole study) reported having a staircase at home and these were then asked if the laboratory staircase resembled their own. 23 of these 28 (82.1% of participants) agreed that the laboratory stairs showed a good resemblance: pertaining to physical characteristics such as step height, tread length and bannister height. These responses provide reassurance that the laboratory staircase was appropriate to use to replicate this ADL.

How difficult each participant found the stair ascent and descent tasks is displayed in Figure 6-16 and Figure 6-17 respectively:

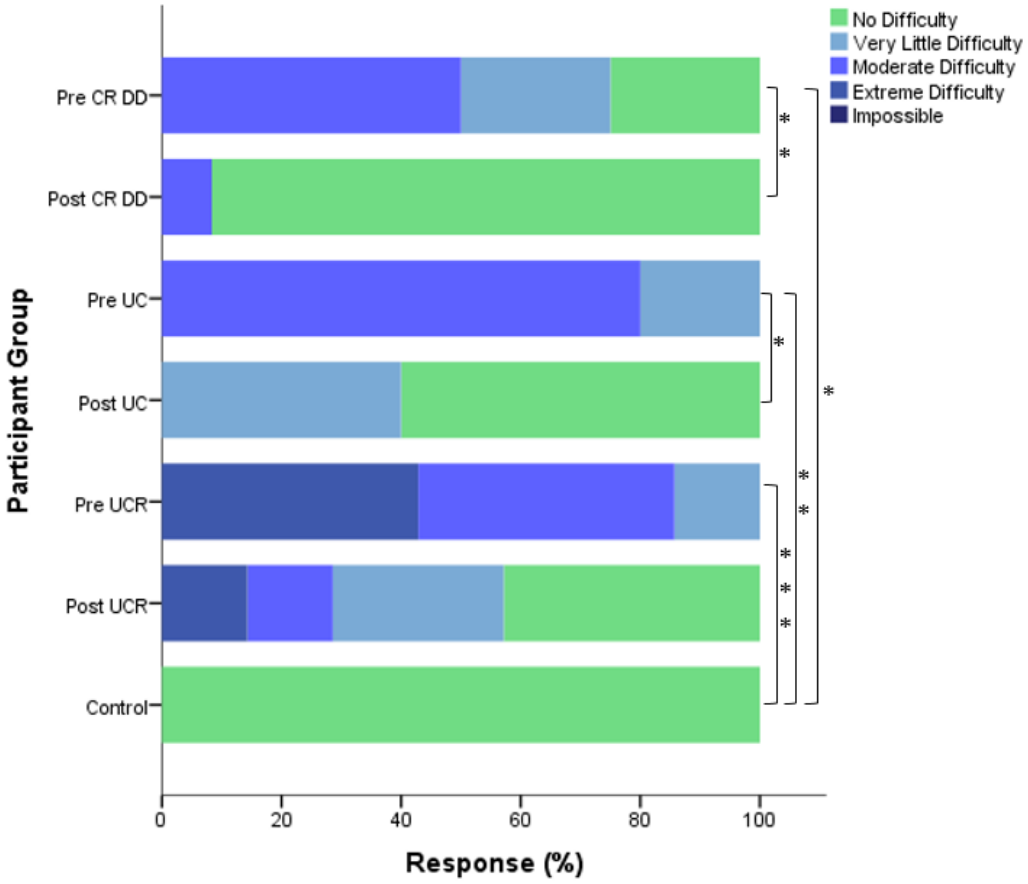

Figure 6-16: Questionnaire responses for how difficult each participant found ascending the staircase.

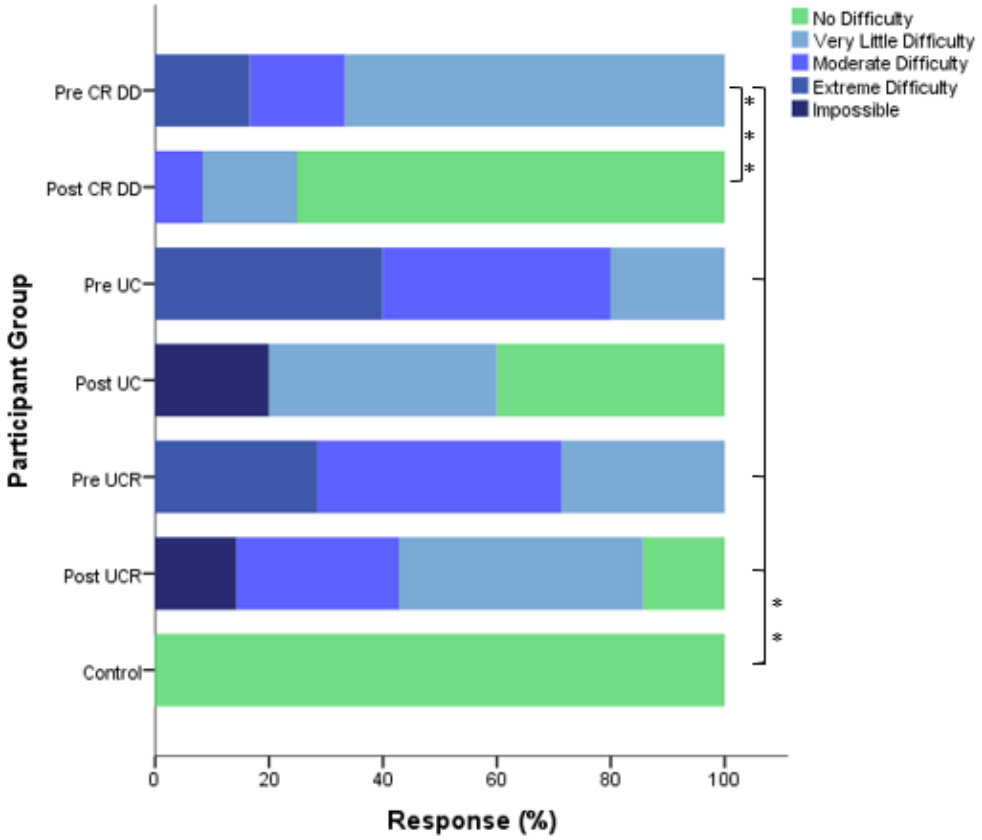

Figure 6-17: Questionnaire responses for how difficult each participant found descending the staircase.

For both tasks, controls reported no difficulty and no significant inter-implant differences were exhibited. All preoperative patients reported greater difficulty compared to postoperative and significantly more difficulty than controls. During stair ascent this was where p=0.015, 0.003, and < 0.001 between controls to the CRDD, UC, and UCR groups respectively. During descent, the p values to controls were p=0.001 for the pre-CRDD and pre-UCR groups, and p=0.003 to the pre-UC group. Intra-implant significance was only seen for the CRDD and UC groups whereby less difficulty was reported during stair ascent by the CRDD (p=0.004) and UC (p=0.048) groups, and the CRDD group also experienced significantly less difficulty (p<0.001) during descent.

Overall, postoperative function may be considered to have improved whereby nearly all implant groups (aside from the UCR group during stair descent) did not report significantly more difficulty than controls. Furthermore, greater proportions of patients who reported no difficulty was seen. Despite general improvements, the

UCR group showed no significant intra-implant improvement during both tasks and during descent still showed significantly greater difficulty to controls (p=0.0052). The UCR and UC groups also showed instances where this task was “impossible to do” (for one patient each) during stair descent which was not seen at preoperative and is a considerable loss of function.

The degree of pain experienced by all participants during stair navigation is reported in Figure 6-18:

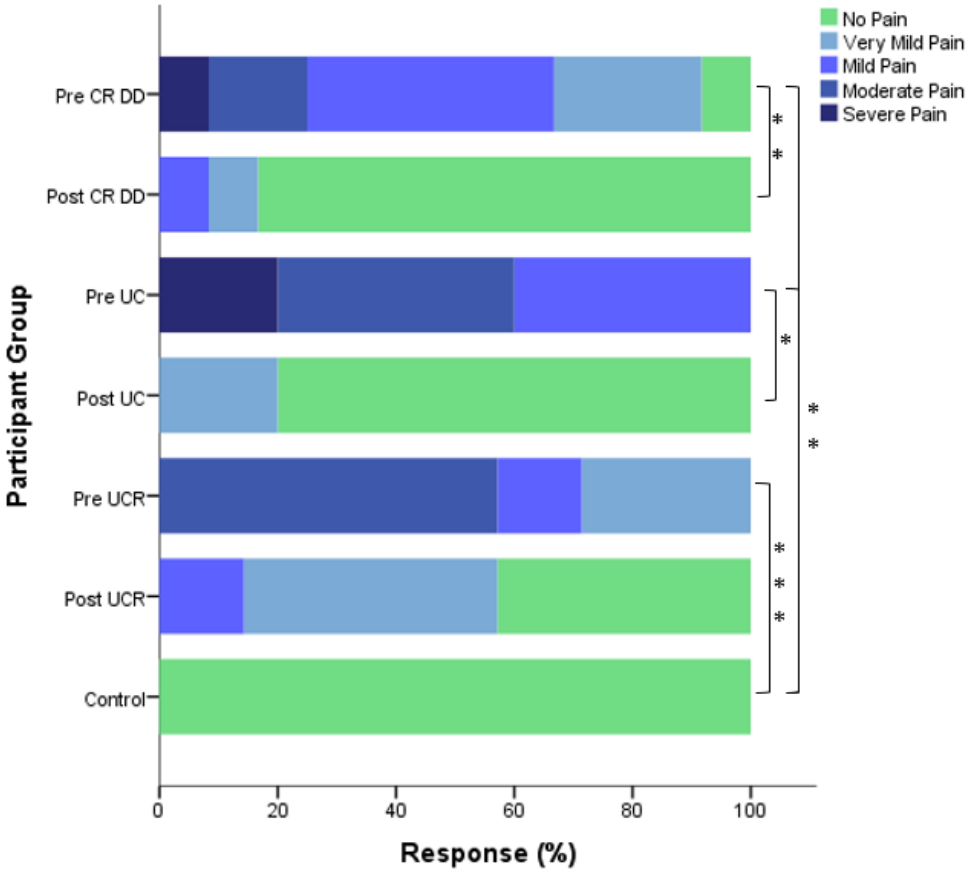

Figure 6-18: Questionnaire responses for how much pain each participant experienced during stair navigation.

Controls again reported no pain during this activity and more pain was experienced in preoperative patients which was relieved at post-surgery. All preoperative groups reported significantly more pain than controls where p=0.0014, 0.003 and <0.001 for the CRDD, UC and UCR groups respectively. At preoperative 100% of UC and UCR patients reported some degree of pain and one patient reported severe pain from the pre-CRDD and UC groups. The post-CRDD and post-UC groups reported

significantly less pain compared to pre-surgery (p=0.001 and 0.016 respectively) and the post-UCR group also showed an improvement but not to a significant level (p=0.073). Postoperative responses were statistically similar to controls implying functional improvement.

Figure 6-19 displays how tiring each participant found the stair navigation task:

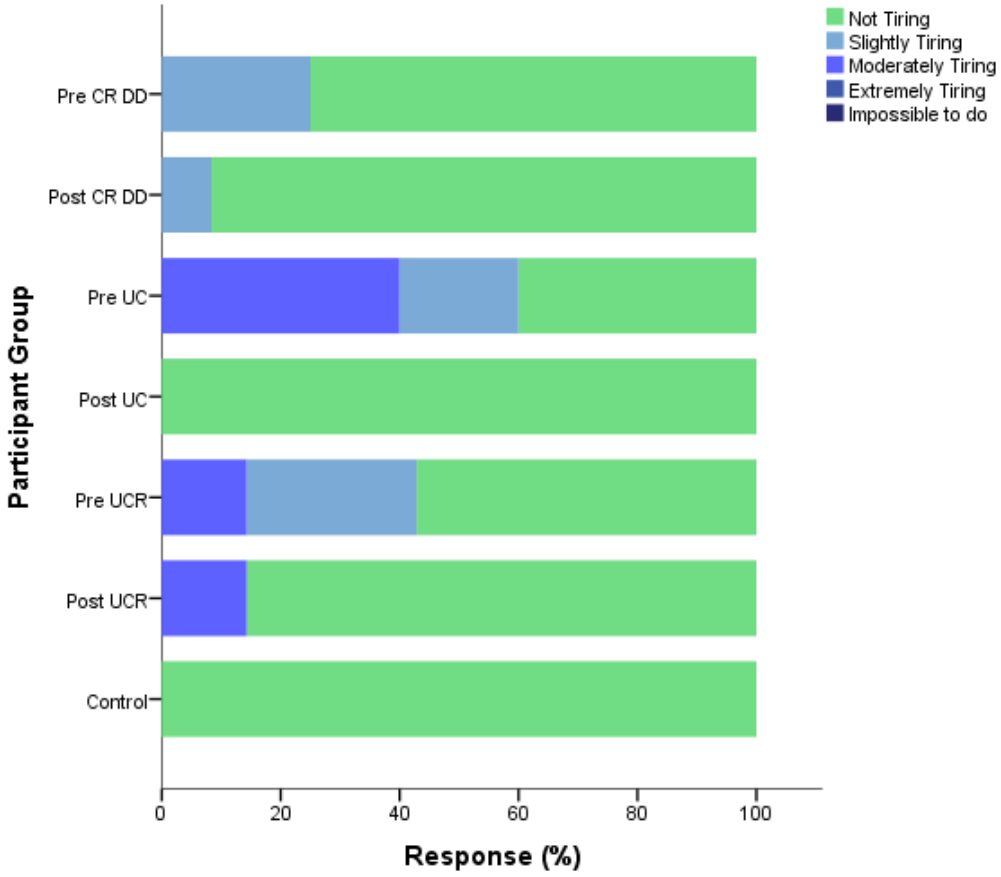

Figure 6-19: Questionnaire responses for how tiring each participant felt following stair navigation.

Controls reported no tiredness, and all preoperative groups reported some degree of tiredness although no significance was seen. A higher proportion of the pre-UC group reported tiredness than the other implants, and no tiredness was reported by the post-UC patients. At postoperative one patient each from the CRDD and UCR groups reported some slight and moderate tiredness respectively.

The responses to whether the participant could climb a flight of stairs without the use of the bannisters is presented in Figure 6-20.

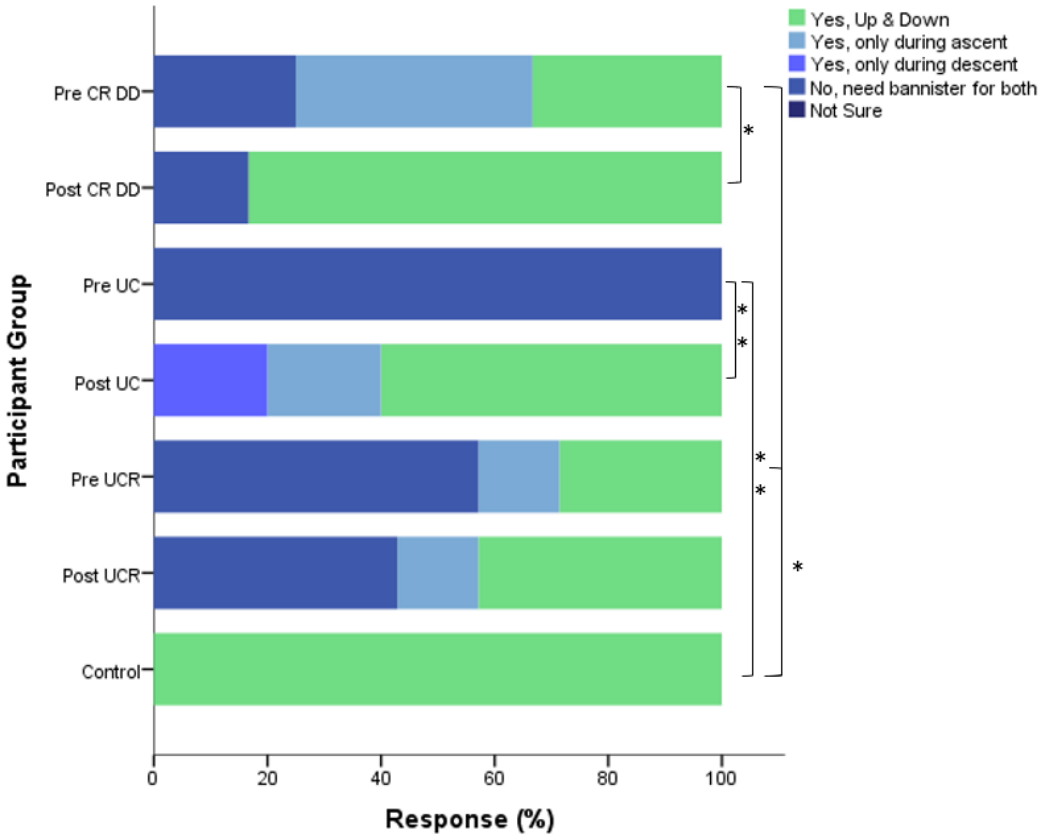

Figure 6-20: Questionnaire responses regarding normal bannister use in everyday stair navigation.

Controls reported no bannister use during everyday stair navigation and preoperative patients reported more bannister dependency (requiring a bannister for ascent and descent) than at postoperative. Across all patients, using the bannisters during both ascent and descent was the most common response, then ascending stairs without the use of a bannister (i.e., a bannister was required during descent), and one participant (in the post-UC group) reported requiring the bannister during ascent only. All preoperative answers significantly differed to controls where p=0.042, 0.0015, and 0.029 for the CRDD, UC and UCR implants respectively. No significantly different responses were seen between the postoperative groups and controls. Intra-implant differences gave significant changes in reported bannister use for the CRDD (p=0.02) and UC (p=0.008) groups only. 100% of the pre-UC group reported requiring a bannister during both ascent and decent which changed to less than half at postoperative which is a considerable improvement. The UCR group showed no significant intra-implant difference (p=1.00) in reported bannister use. Although

statistically similar to controls, the change in bannister use remains closer to preoperative so overall function may not be improved.

Lastly, participants were further asked to describe their usual bannister use for a full flight of stairs. Whether no bannisters, or a single, or double if available would normally be used during stair ascent (Figure 6-21) and descent (Figure 6-22) is displayed.

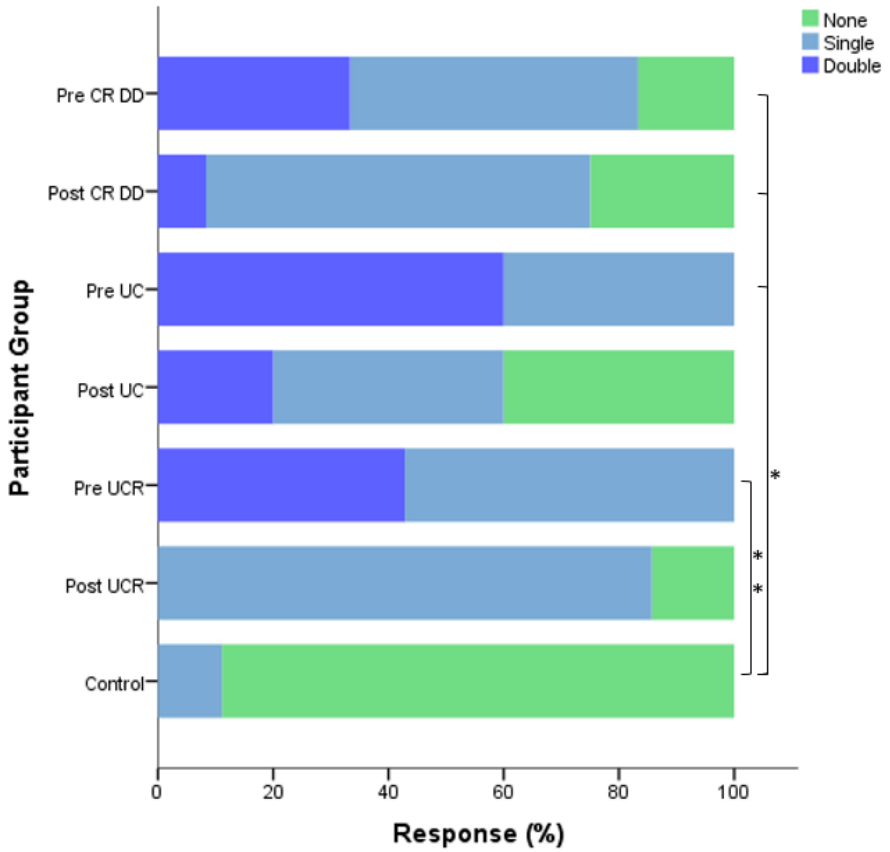

Figure 6-21: Questionnaire responses from those reporting bannister use during everyday stair ascent.

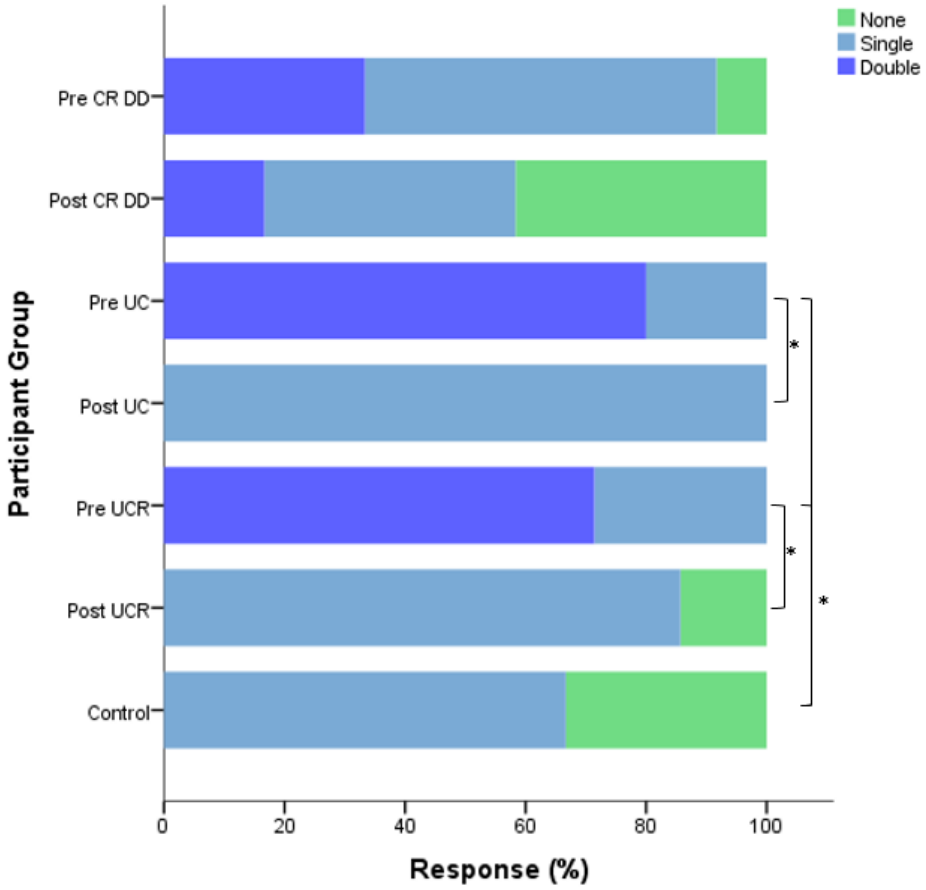

Figure 6-22: Questionnaire responses from those reporting bannister use during everyday stair descent.

During stair ascent and descent respectively, one and six control participants reported normally using a single bannister. All preoperative groups reported using up to two bannisters to carry out these activities and 100% of the pre-UC and pre-UCR groups reported some bannister use. During ascent, all preoperative responses were significantly different to controls (p=0.018, 0.018 and 0.0042 for CRDD, UC and UCR implants respectively). Whilst the post-CRDD group reported significantly more bannister use compared to controls (p=0.045), the same was not seen for the post-UC group (p=1.00) and the post-UCR group showed nearly significantly different bannister use to controls (p=0.052). No significant intra-implant differences in bannister use were seen, nonetheless the incidence of double bannister use generally decreased and reports of none or single bannister use increased.

During descent, the CRDD group presented significantly similar bannister use to controls at both operative states. Pre-UC and pre-UCR groups reported significantly

greater bannister use than controls (p=0.048 and 0.035 respectively) and postoperative (p=0.048 and 0.021 respectively). All postoperative patients reported statistically equivalent bannister use to controls. More of the post-CRDD group reported no or single bannister use during descent than ascent, the whole of the UC group reported using a single bannister during descent compared to a range of no to two bannisters during ascent and the UCR group showed roughly the same patterns between tasks.

6.6.3. Incline Walking

For participants who were able to complete all or part of the sloped walking task their responses to questions relating to this task were recorded. The sample sizes were slightly lower compared to other activities, the control group remained at 9, at pre- and postoperative the implant groups' respective sample sizes were: CRDD - n = 12 and 8, UC - n = 5 and 4, and for the UCR group - n = 6 and 7.

To determine whether the selected inclines were appropriate for this task, participants were asked how the slopes compared to everyday life. The majority of control participants (88.89%) felt the inclinations felt "about right", and one participant answered that the inclines were a little too steep. Approximately of 70% preoperative participants also reported the inclines were "about right" with 26% and 4% reporting "a little" and "extreme" steepness respectively. At postoperative ~79% of patients reported the inclines felt "about right" and no one reported extreme steepness.

6. Group Results

The difficulty experienced by participants throughout the task is presented in Figure 6-23:

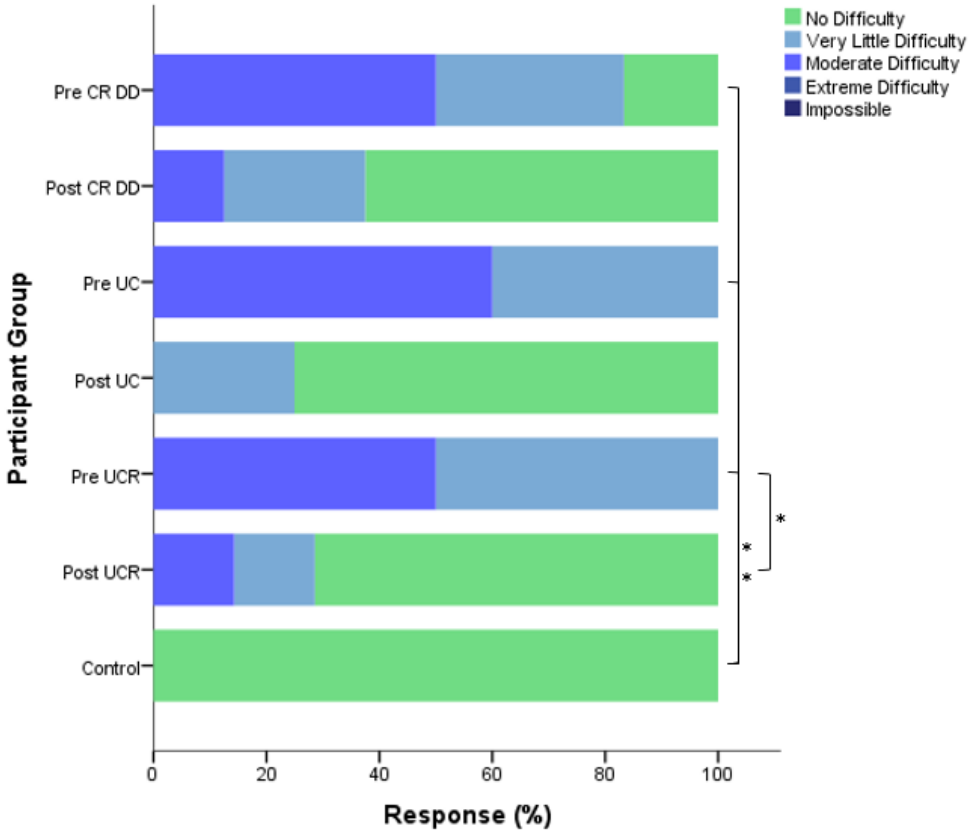

Figure 6-23: Questionnaire responses for how difficult each participant found the sloped walking activity.

Controls reported no difficulty and all preoperative groups comparatively reported significantly more difficulty (p=0.0013, 0.0030, and 0.0012 for the CRDD, UC and UCR groups respectively). 100% of the pre-UC and pre-UCR participants reported some level of difficulty during this task. Postoperative difficulty was markedly lower than preoperative and statistical similarity to controls was seen. All groups showed an intra-implant decrease in the number of patients reporting very little and moderate difficulty which was a significant improvement in the UCR group (p=0.033). Intra-implant p values for the CRDD and UC groups were 0.098 and 0.14 respectively.

Pain levels felt by all participants throughout this task is presented in Figure 6-24:

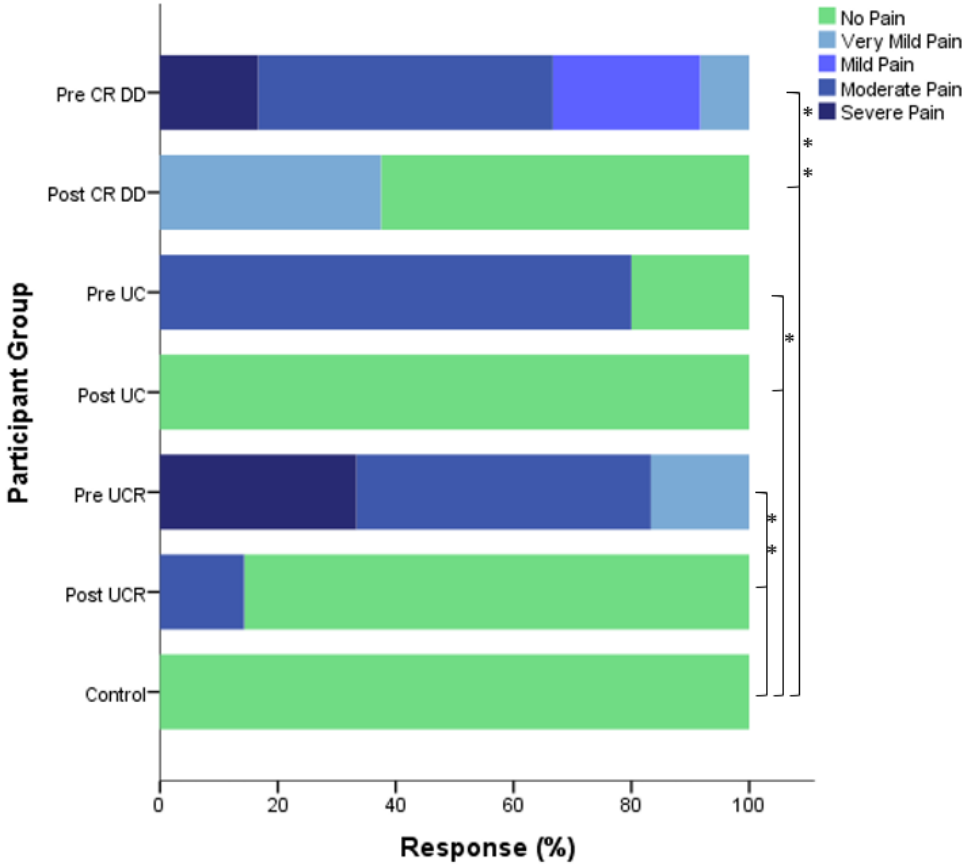

Figure 6-24: Questionnaire responses for how much pain each participant experienced during the sloped walking activity.

Controls reported no pain whereas preoperative groups reported significantly more pain compared to both controls and at post-surgery. All CRDD and UCR participants experienced some degree of pain at preoperative, even to the highest, severe levels. In the pre-UC group, one patient reported no pain, whilst all other participants reported moderate pain. The degree of significance seen for each implant was the same to controls and to postoperative, implying postoperative pain levels was equivalent to controls. The p values between controls to the pre-CRDD, UC and UCR groups were $p<0.001$, 0.030 and 0.0012 respectively, and between operative states the p values were $p<0.001$, 0.048 and 0.005 for the same implant groups. All post-UC patients reported no pain which is a considerable improvement. However, preoperative pain for this group was also the lowest from all implants. The CRDD and UCR groups showed possibly more improvement from pre-surgery as nine and five

patients respectively reported severe or moderate pain at pre-surgery. At postoperative this significantly changed to three CRDD patients that reported very mild pain, and one UCR patient reporting moderate pain.

Figure 6-25 shows the degree of tiredness experienced by all participant groups during this activity:

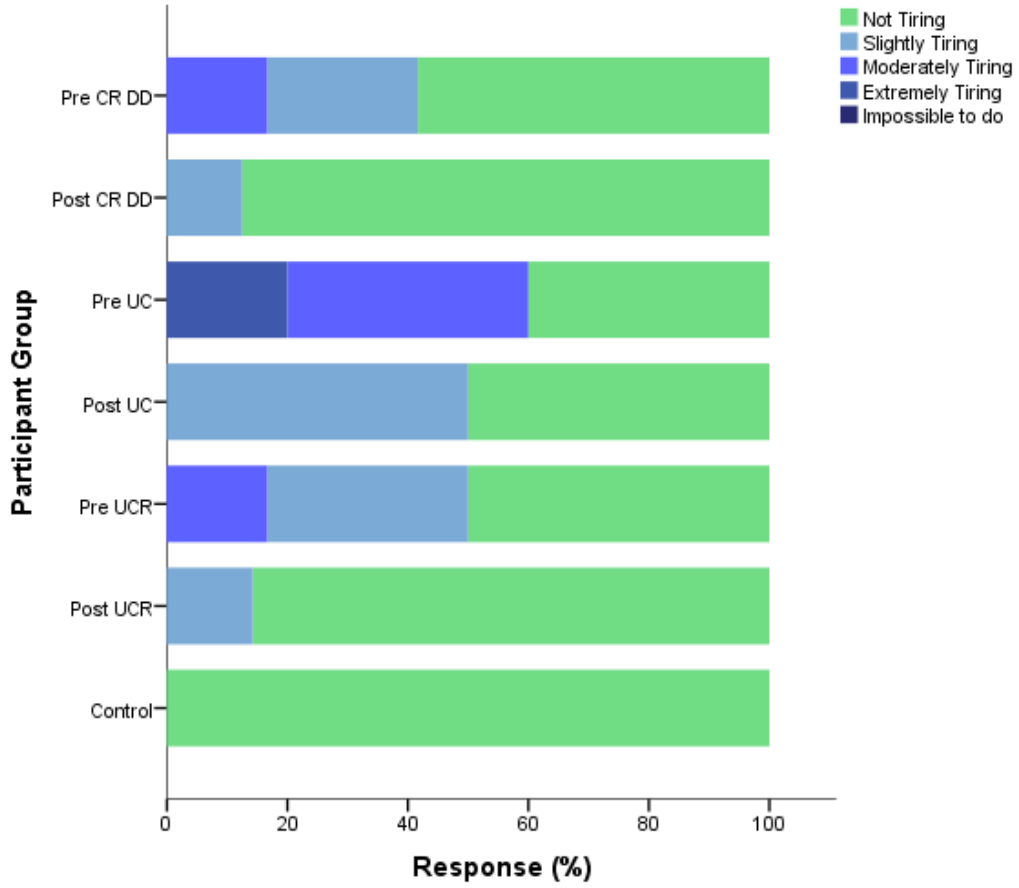

Figure 6-25: Questionnaire responses for how tiring each participant found the sloped walking activity.

Again, controls reported no tiredness during this activity. Preoperative groups reported more tiredness compared to postoperative and the pre-UC group exhibited greater degrees of tiredness (up to extremely tiring) whereas the other pre-surgery responses reached moderately tiring. Following the surgery, all groups still experienced some degree of tiredness with one patient each for the CRDD and UCR and two patients of the UC groups feeling the task was slightly tiring. No significant differences were found between any participant-reported tiredness.

6.6.4. Summary

This chapter presented all participant-reported measures relating to each ADL, for all activities, participants were asked how much difficulty and pain they experienced as well as how tiring they found the task. Controls consistently reported no difficulty, pain, or tiredness during the activities. To ensure laboratory set up was appropriate to replicate the ADL (particularly for the stairs and incline tasks) all participants were asked if the laboratory stairs physically resembled their own to which 82.1% of participants agreed. The inclinations during the sloped walking were also commented to be similar to the control group's everyday walking, and slightly too steep for patients. These findings provide confidence that laboratory set ups suitably replicate normal ADLs, which are as challenging for patients as they would be out of the laboratory, but not to the point of causing additional discomfort. Particularly for the sloped walking task if participants found the inclination they were walking at uncomfortable then they were encouraged to stop the task.

Patients often reported more difficulty, pain, and tiredness during the stair and incline walking tasks, particularly at preoperative. This usually improved post-surgery, where the number of patients reporting no difficulty/tiredness/pain would increase, or the preoperative severity would diminish slightly. No inter-implant significance was seen, instead all significance were either to preoperative or controls. From this it is possible to extrapolate improved or worsened inter-implant function. Regarding how tiring each participant found the activities, no significance was seen between any groups, however some notable trends were seen.

During level walking, most participants did not find this activity tiring apart from one person in the post-UCR group who reported slight tiredness. A higher proportion of the pre-UC group during the stairs and incline walking tasks found these tiring which decreased to 0% at postoperative during stair navigation. However, 50% of post-UC patients experienced slight tiredness during sloped walking which was more than the other two implants during the same activity. The UCR group reported the next highest amount of tiredness for these ADLs which slightly improved at postoperative,

however moderate tiredness was reported postoperatively for the stairs task, where the CRDD group showed slight tiredness for both activities.

Tiredness may relate to the difficulty and pain experienced. During level walking 0% of the post-UC group experienced tiredness, also no difficulty or pain was reported for this activity. For the same task, compared to the post-UC group, the post-CRDD group showed the next lowest instances of difficulty and pain, and the UCR group showed highest difficulty and pain levels. During incline walking, the post-UC group again reported the least amount of difficulty and no pain compared to the other implants, and a higher portion of the post-CRDD group experienced mild pain and difficulty than the post-UCR group.

Since the answers for difficulty during stair descent showed more severe responses than ascent these were used to compare performance along with the answers to the pain questions. The post-UCR group showed the smallest proportion of patients who experienced no difficulty during descent and also the most pain. Conversely, 60% of the post-UC group showed instances of “impossible” or “very little” difficulty, and ~60% of the group reported mild or very mild pain. 20% of the post-CRDD group reported very mild/mild pain during stair navigation as reflected by the difficulty as just over 20% of patients stated very little/moderate difficulty.

In terms of strategy, controls reported they would not ordinarily use a bannister when either ascending or descending a staircase, and preoperative patients mostly required the bannister either during both ascent and descent, or during descent only. A higher portion of patients were likely to use double bannisters where available. Postoperative patients reported less bannister use, and this was usually a single bannister. The post-UCR group showed the smallest change from preoperative in overall bannister use compared to the other implants, and the CRDD group showed least bannister dependency. The CRDD group were more likely to use none or a single bannister during stair descent, whereas the UC group showed the same trend for stair ascent. Postoperative strategies appeared to follow preoperative suggesting behaviours are habitual.

7. Discussion

This study compared spatiotemporal, kinematic, and kinetic parameters as well as participant reported questionnaire responses of participants carrying out the following ADL's: level walking, stair navigation and incline walking. Patients who received one of three randomly assigned B.Braun Columbus® knee implants were analysed as well as an age-matched asymptomatic control group. An initial case study was carried out (Chapter 5) where a full biomechanical profile at pre- and postoperative of a single patient carrying all ADLs was presented. Based on the findings from this patient and the literature, parameters for the group analysis were selected. Comparisons were made between all patient groups to controls, between pre and postoperative (intra-implant changes), between implant groups of the same operative state (inter-implant changes). The participant demographics seen in Table 6-1 indicate controls were similar in age to all patient groups and divide in sex to all patient groups except to the CRDD group which showed a significantly greater sex imbalance to all other groups. Controls also showed significantly lower BMI to all patients ($p<0.01$).

An overview of all observed significant biomechanical changes between the relevant participant groups was presented in Chapter 6.2. As significance in a parameter was observed for more activities this was indicated by a colour scale. Generally, patients showed more significant differences to controls, then to preoperative, and lastly to another implant. Also, the number of activities where a significantly different parameter was seen was higher compared to controls, then to preoperative and finally significant inter-implant differences were usually seen in no more than two activities. The number of parameters where significance to controls was seen was higher at preoperative than postoperative for the UC group, which was reversed for the CRDD and UCR groups. Intra-implant comparisons showed the CRDD group gave a significant change from preoperative for sixteen parameters, compared to seven and six parameters for the UC and UCR implants respectively which may be

improvements or worsening in function. All post-surgery inter-implant differences were between the CRDD to both the UC and UCR groups, the UC and UCR group showed no significant inter-implant differences at postoperative.

Whilst significant differences determine whether changes were actual or not, some significant changes may not be clinically relevant, or changes that were clinically relevant may not have flagged as significantly relevant. In addition, significant differences across samples with such different sample sizes will inherently be underpowered for the UC and UCR groups. For kinematic parameters, a change of 5^{o} may be considered to be clinically relevant, and thresholds for spatiotemporal or kinetic parameters have not been established. In these instances, clinically relevant differences were taken to be similar to the mean differences seen in significant results. Any clinically relevant values were not included in the overall comparison tables in Chapter 6.2 and so will be discussed further in the proceeding chapters.

7.1. Spatiotemporal Parameters

Spatiotemporal parameters during level walking in a healthy group is more reported than for other activities. Comparing healthy control STPs during this activity to other studies' trends gives confidence that the data recording and processing protocol was robust, and participants navigated each activity as expected. Healthy data during other activities is then discussed and finally comparisons between implant groups are given.

7.1.1. Control Comparison

In the literature, not all studies compared the same parameters within the same participant groups. A summary table of literature findings is shown (Table 7-1). Focus was made to studies which included elderly adults to provide an aged-matched comparator to data from this study. As level walking from two walking modes were presented, the same was included below:

Table 7-1: Summary of literature findings of mean and standard deviation spatiotemporal findings during level walking. SP treadmill = self-paced treadmill, FP treadmill = fixed pace treadmill. Values without a standard deviation were not reported by the paper and was calculated from known STP data.

Study	Parameter					
	Age (years)	Walking mode	Walking speed (m/s)	Cadence (steps/min)	Stride time (s)	Double support (s)
Present study	70.0±6.4	Overground	1.5±0.1	123.7±9.0	1.0±0.1	0.2±0
		SP treadmill	1.3±0.2	117.8±10.5	1.0±0.1	0.2±0
Urwin et al., 2014	60.5±7.0	Overground	1.29±0.11	120.4±14.7	1.0±0.11	n/a
Ouellet and Moffet, 2002	60.0±7.9	Overground	1.3±0.2	110.0±10.9	~1.08	~0.3
Kerrigan et al., 1998	72.7±5.5	Overground	1.19±0.13	119.0±9.0	~1.0	~0.3
Silder et al., 2008	72.4±5.0	Overground	1.32±0.13	115.0±7.0	n/a	n/a
Elboim-Gabyzon and Rotchild, 2017	76.7±7.7	Overground	1.0±0.2	107±1.3	1.06	~0.3
Sloot et al., 2014	29.2±5.0	FP treadmill	1.32±0.11	n/a	1.1±0.06	n/a
		SP treadmill	1.32±0.11	n/a	1.09±0.06	n/a
Wearing et al., 2013	21.6±3.0	Overground	1.3±0.1	115.6±6.4	1.04±0.06	~0.2
		FP treadmill	1.3±0.1	118.2±5.9	1.02±0.05	~0.2
Alton et al., 1998	25.7±5.3	Overground	n/a	117.0±6.0	~1.03	n/a
		FP treadmill	n/a	122.0±4.0	~0.98	n/a

Controls in this study walked with a faster speed and higher cadence compared to the literature. For comparative purposes studies comparing overground and treadmill walking were also included, despite these studies primarily using younger populations. The control self-paced treadmill speed and cadence showed more similarities to published findings, overground walking or otherwise (Ouellet and Moffet, 2002; Silder et al., 2008; Sloot et al., 2014; Urwin et al., 2014; Wearing et al., 2013). Where standard deviation was not reported for stride time, this was calculated either by the addition of reported swing and stance time or dividing stride length by walking speed. Where necessary, the period of double support (in seconds) was estimated by multiplying the % double support by stride time. The time used to complete a stride and the period of double support were also comparable to the literature (Elboim-Gabyzon and Rotchild, 2017; Kerrigan et al., 1998; Ouellet and Moffet, 2002; Urwin et al., 2014; Wearing et al., 2013). Overall control STP during

level walking showed good agreement to the literature, even when considering discrepancies in participant ages and walking modality.

STPs during other ADLs are less frequently reported, and a direct comparison of values would be more difficult since step height or inclination degree has a considerable effect on findings. A study investigating the lower limb biomechanics of healthy young adults (mean: 28 years) ascending and descending stairs of similar dimensions showed similarity to this study (Protopapadaki et al., 2007). Stride time was found to be greater during stair ascent (1.45±0.14s) than stair descent (1.32±0.13s), although the times were slightly longer than the findings here (Table 6-12) where control findings were 1.3±0.1s (ascent) and 1.1±0.1s (descent). An earlier study comparing different step heights also consistently found stair ascent stride times to be greater than descent (Riener et al., 2002). Walking speeds during ascent and descent were also alike: 0.49±0.05m/s for ascent and 0.56±0.6m/s for descent in Protopapadaki's study compared to 0.6±0.0m/s and 0.7±0.1m/s here (Table 6-9). Overall, stair descent showed faster walking speeds, also corroborated by an earlier study (Livingston et al., 1991), and discrepancies published in walking speed could be attributed to age differences.

Cadence was not exported in Protopapadaki's study, and percentage proportions of stride and stance during a gait cycle was not exported here so further comparison was not possible. Although, the step height was 3cm shorter and participants were younger adults in Livingston's study, a faster cadence throughout stair descent than stair ascent was seen in Livingston's and this study (Table 6-11). Livingston also divided participants according to height and found that whilst stride length is fixed based on the height of the step, stride time was affected by body height. Shorter participants showed slower stride times, and taller participants exhibited faster times. Although this was not found to be a significant finding.

Slope walking is less analysed compared to level walking, and less so in elderly or disabled populations. The inclinations recorded were ±5^{o} and ±7.5^{o} which may not directly translate to other studies. Studies analysing healthy adults walking uphill

have shown that as slopes increase, walking speed, cadence, and step length decreased, whereas step time, stance duration, and period of double support increased (Donath et al., 2016; Han et al., 2009; Kawamura et al., 1991). Some discrepancies concerning step or stride length were reported where step lengths increased with progressive steepness (Kawamura et al., 1991; Leroux et al., 2002), this may be explained due to the fact that Leroux's study analysed gait at inclines between 0% - 10% (up to 5^{o}) whereas other studies went beyond this. Since the incline walking occurred on a treadmill in the current study, spatial parameters were not comparable and so were not exported for analysis. The control group of this study did not show as consistently defined trends. Between uphill inclinations: cadence decreased (Table 6-11), stride time (Table 6-12) and period of double support (Table 6-15) either remained the same or increased, and walking speed (Table 6-9) decreased as expected. Since large standard deviations were present, and the change between inclines were small perhaps the study design was not sensitive enough to capture differences between inclines. This is acceptable since comparing between inclines was not the aim of this study.

Studies analysing the gait changes associated with downhill walking showed that with increasing downhill steepness: step length and time shortened and cadence and velocity increased to a maximum before slowing to a comfortable pace (Kawamura et al., 1991; McIntosh et al., 2006; Redfern and DiPasquale, 1997). This slowing is likely a compensatory mechanism allowing safe downhill ambulation as the effects of gravity increases speed up to faster and potentially unsafe limits. Control findings also correlate this, STPs between -5^{o} and -7.5^{o} walking gave an increase in walking speed (from 1.4±0.3m/s to 1.5±0.3m/s) and cadence (from 123±9.6steps/min to 129.6±11.6step/min). Average stride time also decreased (from 1.0±0.1s to 0.9±0.1s), however large standard deviations were seen possibly meaning overlap in parameter values between slope inclinations.

7.1.2. Patient Comparison

From the previous section there was a high degree of confidence that controls exhibited typical STPs during ADLs. Comparing studies with OA or TKA patients carrying out these activities would also confirm the same in these groups. Whilst the currently available studies may not strictly analyse the same implants as those in this study, it is possible to obtain an idea of patient behaviour especially when compared to controls. A meta-analysis comparing OA patients of varying severity carrying out level walking, found that severe OA groups (similar to those due to undergo a TKA) exhibited greater stride duration and lower cadence than controls (Mills et al., 2013). This was also seen in the present study (Table 6-12 and Table 6-11) where preoperative groups showed 0.2s longer stride times and 20 steps/mins less cadence than controls. Although conflicting evidence was seen for walking speed likely due to variation in disease progression, the patients in this study consistently showed a slower walking speed than controls, for all activities and operative states as seen in Table 6-9. OA patients also presented later point of opposite foot off, and longer periods of double support indicative of a reluctance to load the limb during contralateral swing (Brandes et al., 2008).

Comparing TKA gait has considerably more variables to consider such as: preoperative OA severity, time since surgery, and implant type used. And so, focus was to similar published studies to the present study. A study comparing level walking gait of patients with fixed or mobile bearing implants at nine months postoperative saw cadence and walking speed increase post-surgery, although not to control levels (Urwin et al., 2014). The same was mostly seen in patient groups in this study, with some exceptions for the sloped walking activity. Here, the post-CRDD and UCR groups presented decreases in cadence (Table 6-11) for all up and downhill inclinations (CRDD group) and -5^{o} walking only (UCR group). As walking speed increased (Table 6-9) in these instances, an increase in stride length may have occurred to facilitate these changes. Since incline walking was carried out on a treadmill, ipsilateral heel strike to heel strike distance would be a small value as the walking surface moves backwards during stance.

Another study analysed fixed and mobile bearing TKA gait during stair navigation at approximately 11 months (FB knee) and 20 months (MB) post-surgery. No comparison to preoperative was made, although an aged-matched control group was included (Catani et al., 2003). Controls in this study showed slower walking speeds (Table 6-9) and longer periods of double support (Table 6-15) during stair ascent than stair descent which was similar to the findings in younger healthy participants of other studies (Livingston et al., 1991; Protopapadaki et al., 2007). The MB group displayed the same trends, and the FB group showed the same trends for velocity and period of double support as controls, however, the percentage of stance was greater for stair ascent compared to descent.

The control group in Catani et al's study showed shorter periods of stance and double support, and faster walking speeds compared to patients during ascent and descent. One way ANOVA tests found this was a significant finding between the control and patient groups, but no significance was seen at the inter-implant level implying controls showed significantly different findings to either both or a single implant group. Although the percentage of the cycle spent in stance was not analysed in this project, comparisons to walking speed and period of double support can be made. Table 6-9 similarly shows faster control walking speeds to patients, and no significance was seen between postoperative bearing types. Table 6-15reiterates that participants showed a longer period of double support during stair ascent than descent, and patients maintained larger periods of double support than controls although the majority of significance was seen in the UCR, then CRDD groups.

One of the most significant effects on spatiotemporal parameters and resulting kinematic and kinetic finding is walking speed. Whilst patient speeds were statistically similar to each other a difference of 0.1-0.2m/s between groups is considered a minimal clinically important difference (Bohannon and Williams Andrews, 2011). At preoperative the UC and UCR groups walked at slower walking speeds than the CRDD group and differences to the CRDD group reached up to 0.4m/s (to the pre-UCR group during level treadmill walking). The subsequent effects of a slower walking speed on

other spatiotemporal parameters such as reduced cadence, increased stride time and period of double support were generally maintained at post-surgery, particularly in the UCR group. The effects of these on the lower limb kinematics and kinetics will be discussed further below.

7.1.3. Summary

Although the literature is scarce, control and patient STP data shows strong correlations to published studies where available and discrepancies can be accounted for. Overall, there little significance was observed at the intra- and inter-implant level and more significance was seen to controls. Patients showed improvements at postoperative, since significance were mostly to controls, this postoperative improvement was not yet to control levels. Changes seen in one parameter were often complimented by changes in other parameters: faster walking speeds is often matched with increased cadence, longer step and stride length and decreased step/stride time and period of double support as expected from the literature (Bovi et al., 2011; Fukuchi et al., 2019). Whilst neither step nor stride length were analysed (as these were not height normalised), patient groups generally followed this trend at post-surgery. No consistent or significant intra-implant decreases in stride time was seen and the period of double support was often unchanged compared to preoperative. Based on this, increases in walking speed and cadence may have been facilitated by potential decreases in step and stride length. Which, in turn correspond to altered kinematics and kinetics.

7.2. Kinematic Parameters

Peak joint angles, range of joint motion, and additional sagittal knee data from all ADLs were given in Chapter 6. To assess whether the study design provided expected results control group comparison to literature figures is initially presented. Studies analysing motion during level walking, then the other ADLs are described, followed by OA and TKA patient kinematics.

7.2.1. Control Comparison: Level Walking

To ensure the protocol for this study was reliable and gave expected results an initial comparison of the control group to the literature are shown in Table 7-2 and all reference below is made to this table. Although studies using an older population were preferred to be used, where this was not possible studies with younger participants were included to provide some idea of expected function.

Table 7-2: Literature kinematic findings of healthy controls during overground level walking. Values are mean ± standard deviation where available. Bold values indicate where data is taken from stance only. KFHS/KFTO = knee flexion at heel strike/toe off, P/MKFS = peak or minimum knee flexion during stance.

Study (continued on the next page)	Parameter (°)														
	Age	Pelvis Tilt ROM	Hip Flexion ROM	Peak Knee flexion	Knee Flexion ROM	Peak Knee Adduction	Knee Adduction ROM	Peak Knee Rotation	Knee Rotation ROM	KFHS	KFTO	PKFS	MKFS	Peak Ankle DF	Ankle DF ROM
Present study	70.0±6.4	3.1±0.9	47.2±4.2	61.2±3	62.6±3.2	**5.3±2.7**	**5.7±3.3**	17.3±8	20.6±6.1	3.8±4.7	38.6±4.3	19.5±5.3	1.3±5.1	16.4±3.9	23.2±2.9
Alice et al., 2015	64.8±8.3			54.4±5.9	54.9±5.2	2.2±1.7*				5.4±4.6	33.6±6.0		0.1±4.4		
Astephen et al., 2008	50.3±10.1		39.2±4.8	64.0±6.0	68.5±6.0							18.7±7.3			30.7±5.3
Benedetti et al., 2003	66.9±5.8			63.3±4.3	57.1± 5.4					1.5±5.6	38.2±5.3	16.7±5.6	6.1±5.0		
Benoit et al., 2007	26±4.7						**4.3±1.6**		**11.9±5.8**			ROM: 40.2±17.4			
Bytyqi et al., 2014	61.7±3.1			54.4±5.3	52.2±5.3		5.5±1.6		9.3±2.4	17.4±12.5		17.4±12.5	2.2±4.0		
Catani et al., 2012	27.9			64.5±5.5	64.6±4.1					4.9±3.4			4.2±3.9		
Fantozzi et al., 2015	27.0±3.4		36.0±3.9	56.8±4.4						−3.5±5.7					29.8±4.3
Fuchs et al., 2002	65.5±8			59.1±5.5	58.1±8.1							20.4±9.0	8.9±5.1		
Goujon-Pillet et al., 2008	44.3 (28 - 61)	8.0±5.0													
Heiden et al., 2009	64±6									1.7±4.3		14.3±7.1	2.6±4.7		
Jevsevar et al., 1993	> 60			61.5±9.1											
Kerrigan et al., 2001	73.2±5.6		40.7±4.4	58.3±4.9	59.9±5.2							16.5±6.6	1.7±3.9	8.6±3.6	23.3±6.5
Kerrigan et al., 1998	72.7±5.5		40.4±4.1	57.9±4.6	60.6±4.8							16.3±6.0	1.7±4.0		
Kirkwood et al., 2007	65.1±6.0		38.9	59.5	60.4				12.9						

Study	Parameter (°)														
	Age	Pelvis Tilt ROM	Hip Flexion ROM	Peak Knee flexion	Knee Flexion ROM	Peak Knee Adduction	Knee Adduction ROM	Peak Knee Rotation	Knee Rotation ROM	KFHS	KFTO	PKFS	MKFS	Peak Ankle Dorsiflexion	Ankle Dorsiflexion ROM
Present study	70.0±6.4	3.1±0.9	47.2±4.2	61.2±3	62.6±3.2	**5.3±2.7**	**5.7±3.3**	17.3±8	20.6±6.1	3.8±4.7	38.6±4.3	19.5±5.3	1.3±5.1	16.4±3.9	23.2±2.9
Komnik et al., 2016	57.6±6					2.4±1.6**	2.3±1.5**	**3.0±4.4**	**7.8±2.7**						
Lee and Hidler, 2008	~ 23 - 56		44.1±3.6	69.1±4.3	67.7±3.2									13.9±4.2	26.6±6.2
Lim and Lee, 2018	25 - 35	2.3±1.9													
McCarthy et al., 2013	71.3 ± 6.1				61.2±6.1							ROM: 18.0±4.0			
McClelland et al., 2011	69.6±8.3			55.5±4.4	56±4.3	**4.6±3.0**	**3.3±3.3**	**12.6±5.8**	**12.4±4.3**	7.1±4.0		17.3± 6.8	0.5±4.3		
Monda et al., 2015	60-69			62.9±3.1								ROM: 18.5±4.9			
Nagano et al., 2012	21.6±1.3									5.3±4.3					
Piazza and Delp, 1996	n/a										39.5±10.2				
Rahman et al., 2015	68.1±7.1				62.6±5.8							ROM: 19.8±4.9			
Rowe et al., 2000	67±8			67.4±6.3	64.5±5.9										
Saari et al., 2005	69.5 (50–87)		36	60	63	3	9								
Silder et al., 2008	72.3±6		49±5											20.0±5.0	28.0±5.0
Tadano et al., 2016	69.1±5.4			68.6±7.3	65.6±6.9										
Urwin et al., 2014	60.5±7.0			64.2±2.7	58.0±3.7	7.4±5.8	14.5±3.4								
Watt et al., 2010	70.3±4.8		48.0±5.5	49.1±11.0		31.8±15.7		5.2±6.5	12.9±9.1			14.4±6.8		15.5±4.6	26.1±6.5

** - value reported as the mean absolute value. ** - value given from the first 50% of stance phase.*

3. Discussion

Twenty-nine studies were found in which the gait of healthy participants was analysed. The participants of twenty of these studies were of a similar age to the control group shown here and where a standard deviation was not provided often a range was given. The technology used for motion capture was mostly three-dimensional motion capture, six studies used IMUs (Fantozzi et al., 2015; McCarthy et al., 2013; Monda et al., 2015; Rahman et al., 2015; Rowe et al., 2000; Tadano et al., 2016), one used accelerometers (Lim and Lee, 2018), another used two-dimensional motion capture (Fuchs et al., 2002), and a final study used modelling from EMGs and the equations of motion to give kinematic output (Piazza and Delp, 1996).

Nineteen studies reported maximum knee flexion and controls showed good likeness in peak knee flexion to twelve studies where mean differences less than 5^{o} were seen. The largest difference in peak knee flexion was where the controls in this study showed, on average, 12^{o} greater knee flexion to one paper (Watt et al., 2010). Since high peak knee adduction during the whole gait cycle was also observed in Watt's paper (31.8±15.7^{o}) some kinematic cross talk was evident. Knee flexion ROM was another highly reported measure and the difference between current study control ROM to twelve out of seventeen studies reporting this parameter was less than 5^{o}. Meaning the controls did not show a clinically relevant difference in knee flexion peak or ROM to the majority of studies. The largest difference in ROM was to a study by Bytyqi et al., (2014) where Bytyqi et al., showed 10^{o} less ROM than the controls of this study. In addition, Bytyqi et al., analysed treadmill walking using an exoskeleton KneeKG^{TM} system, and the gait trace presented did not look as expected for this task. Knee flexion at toe off (KFTO) and KFHS was higher than expected such that a PKFS was not visible. Whether the equipment, or the observer experience was a factor is to be clarified by the authors.

In this present study, average flexion ROM was ~1.4^{o} greater than peak flexion, indicating some hyperextension was present in controls. Fourteen studies reported both peak and range of knee flexion, and half of these also indicated hyperextension

was evident (Alice et al., 2015; Astephen et al., 2008; Catani et al., 2012; Kerrigan et al., 2001, 1998; Kirkwood et al., 2007; Saari et al., 2005). These studies showed similar differences between flexion peak and ROM as this study. Peak flexion was greater than ROM for the remaining studies (Benedetti et al., 2003; Bytyqi et al., 2014; Fuchs et al., 2002; Lee and Hidler, 2008; Rowe et al., 2000; Tadano et al., 2016; Urwin et al., 2014). And the difference between flexion peak and ROM were greater than the instances where the opposite was seen, peak flexion was higher than ROM by up to 6.2^{o} in two studies (Benedetti et al., 2003; Urwin et al., 2014) which may imply extension limitations.

Non-sagittal knee motion was reported by nine studies, with ROM parameters reported more frequently than peak adduction/rotational angles. A point of consideration is that this and other studies were not consistent with reporting non-sagittal kinematic values during the whole cycle or during stance only. Maximum adduction angles and ROM in this study were during stance only, whereas rotational peaks and ROM from the whole cycle was reported. Controls showed greater peak rotation than all studies reporting this parameter (Komnik et al., 2016; McClelland et al., 2011; Watt et al., 2010) with differences ranging from 4.7^{o} - 14.3^{o} (compared to McClelland et al., (2011) and Komnik et al., (2016) respectively). However, the studies by Komnik et al., (2016) and McClelland et al., (2011) reported rotation angles during stance only which accounts for this discrepancy.

Controls showed greater peak adduction to most studies (Alice et al., 2015; Komnik et al., 2016; McClelland et al., 2011; Saari et al., 2005) and lower peak adduction to two papers (Urwin et al., 2014; Watt et al., 2010). Although peak adduction during stance was reported, this was still higher than most of the studies giving peak adduction throughout the whole gait cycle. The difference in average peak adduction to the literature ranged from 0.8^{o} (McClelland et al., 2011) to 3.1^{o} (Alice et al., 2015) which are not clinically relevant. Furthermore, the values reported by Alice et al., (2015) were mean absolute values and the actual peak adduction may be lower than presented.

Much like peak values, adduction and rotational ROM in this study (5.7±3.3^{o} and 20.6±6.1^{o} respectively) were also greater than literature figures. As well as Komnik et al., (2016) and McClleland et al., (2011) one other paper reported also values during stance only (Benoit et al., 2007). These three papers showed comparatively lower adduction and rotational ROM by ~1.4 – 3.4^{o} and ~8^{o} – 12.8^{o} respectively. From studies reporting rotational ROM during the whole gait cycle differences to the current study ranged from 7.7^{o} (Kerrigan et al., 2001; Kirkwood et al., 2007) to 11.3^{o} (Bytyqi et al., 2014). Adduction ROM shown by Komnik et al., (2016) was the smallest adduction ROM from all studies, due to the fact that adduction values were taken during the first 50% of stance and as such, are not directly comparable.

Four studies reported peak and range of adduction, subtracting these revealed some abduction was present: 6^{o} (Saari et al., 2005) and 7.1^{o} (Urwin et al., 2014) during the whole gait cycle. No abduction was seen during the whole of stance (McClelland et al., 2011) nor the first 50% of stance (Komnik et al., 2016). This slightly agrees with the present study where an average of 0.7^{o} stance abduction was seen. Of the three studies reporting maximum and range of longitudinal rotation, it was calculated that 4.8^{o} and 7.7^{o} of external rotation was evident during the whole cycle and stance only (Komnik et al., 2016; Watt et al., 2010) and a near neutral figure during stance was reported by one study (McClelland et al., 2011). Overall, controls of this study agree with the literature as ~3.3^{o} of external knee rotation was seen.

KFHS and KFTO was exported to give an idea of joint behaviour at these gait events. The same data for the sagittal pelvis, hip, and ankle were also exported but were not included due to low significance and clinical relevance. In the literature, KFHS was more commonly reported (eight papers) than KFTO (three papers). Control average KFTO was comparable to papers reporting this parameter, with mean differences less than 1^{o} (Benedetti et al., 2003; Piazza and Delp, 1996) and 5^{o} to one paper (Alice et al., 2015). KFHS showed good similarity, with differences of less than 5^{o} seen to most papers (Alice et al., 2015; Benedetti et al., 2003; Catani et al., 2012; Heiden et al., 2009; McClelland et al., 2011; Nagano et al., 2012). Controls showed around 7.3^{o}

more KFHS than the participants of a study with younger adults (Fantozzi et al., 2015), and 13.6° less flexion at heel strike than another study (Bytyqi et al., 2014). The gait cycle trace seen in Bytyqi's study was earlier described as atypical and so the KFHS is likely to be incorrect.

During stance, a small flexion peak is expected as the contralateral limb undergoes swing. Eight studies reported peak (PKFS) and minimum (MKFS) flexion during stance. Two of these reported just the PKFS (Astephen et al., 2008; Watt et al., 2010), and one reported only the MKFS (Catani et al., 2012). Finally, four studies reported stance ROM only (Benoit et al., 2007; McCarthy et al., 2013; Monda et al., 2015; Rahman et al., 2015). The majority of studies showed similar or lower PKFS to the controls of this study, by no more than 5° demonstrating good agreement. The MFKS in this study was explicitly taken between the stance and swing flexion peaks and was seen to be near neutral with small flexion or hyperextension deviations. Of the studies reporting this parameter, seven showed similar values to the current control group where mean differences were not clinically relevant (Alice et al., 2015; Benedetti et al., 2003; Bytyqi et al., 2014; Catani et al., 2012; Heiden et al., 2009; Kerrigan et al., 2001, 1998; McClelland et al., 2011). One study showed a higher MFKS of near 9°, however, the walking modality was a treadmill as opposed to overground walking and the walking speed was equal for all participants of 0.56 m/s. This is considerably slower than the control group in the current study – by nearly three times which may result in unnatural walking patterns (Fuchs et al., 2002).

Calculating knee flexion ROM during stance from papers reporting both maximum and minimum flexion during stance, combined with four papers reporting stance ROM only gave a total of twelve stance ROMs available for comparison. Flexion ROM during stance in the current study came to ~18.2°, seven papers gave stance ROM figures within±5° of this (Bytyqi et al., 2014; Kerrigan et al., 2001, 1998; McCarthy et al., 2013; McClelland et al., 2011; Monda et al., 2015; Rahman et al., 2015). Three papers gave smaller stance ROM of ~11° (Benedetti et al., 2003; Fuchs et al., 2002; Heiden et al., 2009) and another study showed a much higher stance ROM 40.2±17.4°

(Benoit et al., 2007). Such large ROM values are likely a result of the using the largest flexion angle during the whole of stance as opposed to the apex of the primary flexion peak.

Peak and range of sagittal pelvis, hip and ankle motion were also analysed, although peak pelvic and hip flexion were not included. Pelvic tilt ROM was the least reported parameter in the literature, with two instances seen: 2.3±1.9° (Lim and Lee, 2018) and 8.0±5.0° (Goujon-Pillet et al., 2008). The pelvic tilt ROM seen in this study was within 5° of these, closer to that seen by Lim and Lee, (2018). Hip flexion ROM was reported in nine studies, two studies showed higher ROM than this study although not by a clinically relevant amount (Silder et al., 2008; Watt et al., 2010). The other studies showed comparatively lower ROM, one of these also not showing clinically a relevant difference (Lee and Hidler, 2008) and the largest difference to this control group was around 11° (Saari et al., 2005).

Four studies reported the peak and range of ankle dorsiflexion (DF) (Kerrigan et al., 2001; Lee and Hidler, 2008; Silder et al., 2008; Watt et al., 2010) and two papers presented dorsiflexion ROM only (Astephen et al., 2008; Fantozzi et al., 2015). Peak DF in the studies were mostly within ±5° of this study's control group, aside from the peak angle seen in Kerrigan et al., (2001)'s study which showed a decrease of nearly 8°. Ankle ROM in this study (23.2±2.9°) was lower than the values seen in the literature, where the difference in ankle DF ROM ranged from 0.1° (Kerrigan et al., 2001) to 7.5° (Astephen et al., 2008).

Controls in this study also walked on a self-paced treadmill at 0° incline as part of the sloped walking task. Good agreement between kinematic data between overground and level treadmill walking gives assurance to the sloped walking data. For all fourteen kinematic parameters, the difference in angle between treadmill and overground walking was consistently less than 5°. No statistical analysis was carried out as this was not within the scope of the project however some comment on mean differences can be made. The greatest difference was seen in the range of knee adduction during stance where controls showed 4.5° more ROM when treadmill

walking. Similar differences were seen for hip flexion ROM where controls showed 4.1° less ROM during treadmill walking. For other parameters, differences of 2.3° (KFTO) to -1.3° (PKFS) were seen between walking modes. Peak and range of knee flexion was respectively 0.3° higher and 0.1° lower during treadmill walking. Overall, differences in sagittal knee data between walking modalities in this study was small.

The above similarities to literature findings and walking modalities provide an excellent degree of confidence in the motion capture methods and data processing techniques. For the most commonly reported parameters: maximum knee flexion and flexion ROM, controls also showed mostly lower standard deviations and less variance than the papers. However, the controls' sample size was also smaller compared to the papers. Level walking is an easily comparable task as it is less challenging and has few variables affecting outcomes, however, for use to evaluate biomechanical performance has limitations.

7.2.2. Control Comparison: Other ADLs

Kinematic data during stair navigation and sloped walking of controls are discussed here. Direct comparison to literature papers is less accurate as different step heights and inclinations will generate altered biomechanical strategies to complete the task. Overall, these ADLs utilise greater peak joint angles and ROM compared to level walking and differences in angles between these tasks and level walking reached as high as 40 – 60° during the stair tasks in particular.

In this study, control group pelvic (Table 6-20) and hip ROM (Table 6-21) during stair ascent was greater than stair descent, pelvic ROM showed small increases of less than 5° and hip ROM between tasks was considerably larger by 27-35°. No papers were found that reported pelvic motion during stair navigation, and two studies similarly found higher hip ROM during ascent than descent (Livingston et al., 1991; Saari et al., 2004). The respective mean differences in hip ROM between stair tasks was ~18 and 31° (Livingston et al., 1991; Saari et al., 2004) the latter showing agreement to the present study. The peak and range of knee flexion angle (Table 6-16 and Table 6-22 respectively) was similar during both tasks and greater differences in the peak (Table

6-18) and range of ankle flexion (Table 6-25) was seen during stair descent reaching 21°. This implies that the proximal joints are utilised more than distal joints during stair ascent, and the opposite is seen for descent.

Similar to this study maximum knee flexion and/or ROM values during stair ascent and descent in the literature ranged between 80 – 100° (Bjerke et al., 2014; Fantozzi et al., 2015; Gao et al., 2012; Jevsevar et al., 1993; Livingston et al., 1991; Protopapadaki et al., 2007; Rowe et al., 2000; Saari et al., 2004). One instance of a lower ROM even less than that expected during level walking was seen where flexion peak and ROM during ascent reached 89° and 57° respectively (Catani et al., (2003), which the authors did not consider abnormal. Peak knee flexion was as expected for this task and subtracting the ROM implies a more flexed knee throughout this activity. Although not by a clinically relevant amount, controls in this study showed larger peak flexion during ascent, and a slightly higher flexion ROM during descent (Table 6-16). Differences in peak flexion between stair tasks was also not clinically significant in the studies reporting peak flexion in these activities. Furthermore, neither ascent nor descent consistently showed greater peak flexion. Flexion ROM was also not consistently higher in one activity more than the other, and only one paper reported a clinically relevant greater ROM, seen in stair descent (95°) than ascent (90°) (Livingston et al., 1991).

KFHS (Table 6-27) and KFTO (Table 6-28) was greatest during stair ascent and descent respectively. In order to reach the higher step to initiate stair ascent greater knee flexion is required and at toe off the limb is near neutral. Conversely during descent, to reach the proceeding lower step requires a near neutrally aligned limb and the knee is at near maximal flexion at toe off to progress the limb as the body has descended the step. No KFTO was not reported in the literature, four reported KFHS during ascent, and half of these reported KFHS during descent. Descent KFHS was highly similar to one study (Catani et al., 2003) with a mean difference less than 1°, and less than 5° different to another study (Hall et al., 2012). Controls' KFHS during stair ascent was also no more than 4.5° less than literature findings (Asay et al., 2009;

Catani et al., 2003; Fantozzi et al., 2015; Hall et al., 2012). Good agreement is seen despite small discrepancies between step height between the papers.

The peak and range of non-sagittal knee motion were not different by a clinically relevant amount between ascent and descent. Non-clinically relevant differences in adduction kinematics (Table 6-17 and Table 6-23) between ascent and descent was corroborated by the literature (Gao et al., 2012; Saari et al., 2004), although these published values were much lower compared to this study despite being taken from the whole gait cycle. In the current study and the one by Gao et al., the range of adduction was greater than peak adduction – indicating some abduction occurred during these activities, which was not seen in the paper by Saari et al., (2012). Peak (Table 6-18) and range of knee rotation (Table 6-24) were also much higher here than compared to Gao et al., (2012). Gao et al., presented peak rotational angles larger than the range, demonstrating internal and external rotation was evident, whereas in this study rotational ROM was less than peak rotation meaning only internal rotation was experienced.

The peak (Table 6-18) and range of ankle dorsiflexion (Table 6-25) was largest during stair descent compared to the other activities carried out in this study, including stair ascent. This finding gave the basis that distal joints are employed more than proximal joints during stair descent which has also been agreed by other papers (Andriacchi et al., 1980; Livingston et al., 1991; Protopapadaki et al., 2007). Of the papers presenting tabular results, the peak dorsiflexion for both stair ascent and descent in this current control group were greater than the literature (Livingston et al., 1991; Protopapadaki et al., 2007). In contrast, control dorsiflexion ROM in this study was less than literature findings meaning less plantar flexion was utilised during these tasks.

Sloped walking is similar to stair navigation in that the primary action is to translate the body forwards as well as upwards or downwards. Compared to stair navigation, the peak and range of lower limb joint angles were lower during the sloped walking. Although the inclinations of sloped walking in this study were ±5^{o} and ±7.5^{o} only the data from ±7.5^{o} is discussed here as most control participants walked at this incline.

3. Discussion

The inclination that participants walked in the literature did not equal ±7.5° but ranged between 5-12°. One study presented peak and range of knee flexion during both stair navigation and ±5° sloped walking and similarly lower values were seen during slope walking (Rowe et al., 2000). KFHS and KFTO were also markedly different between stair and ±7.5° slope walking in the present study. Controls showed ~30° and 8° more KFHS during stair ascent and descent respectively than to up and downhill walking (Table 6-27). During stair ascent, controls gave 22° less KFTO compared to uphill walking, and 34° more KFTO during descent than downhill walking (Table 6-28). Based on the step height (18.5cm) and tread (28.0cm) an equivalent slope of 33° is given, explaining the greater KFHS during stair ascent.

Pelvic tilt ROM during uphill walking in this study (Table 6-20) were around 0.4° greater compared to a study where participants walked at 10° (Kimel-Naor et al., 2017), a negligible difference. In addition, pelvic ROM during downhill walking equalled that reported by the same authors. Differences in hip flexion ROM during uphill walking was less than 5° compared to two studies that analysed uphill walking at 10° and 8.5° (Kimel-Naor et al., 2017; Lay et al., 2006). Hip flexion ROM during downhill walking was 10° greater in this study compared to the literature (Lay et al., 2006; Kuster et al., 1995). Since it has been suggested older participants prefer to employ proximal joint motion for stability compared to younger groups (Browne and Franz, 2018; DeVita and Hortobagyi, 2000) the age difference of the published studies (24 years (Lay et al., 2006) and ~27 years (Kuster et al., 1995)) to this one may account for this disparity.

Peak knee flexion (Table 6-16) and ROM (Table 6-22) in this study was higher during downhill walking than uphill walking by nearly 10° which is a clinically relevant finding. The same clinically relevant difference was seen in all papers reporting these parameters during uphill and downhill walking (Kimel-Naor et al., 2017; Lay et al., 2006; Rowe et al., 2000). The actual values seen were also comparable except compared to Kimel-Naor et al., (2017) which showed smaller peak and range of flexion values of 43 - 58°, compared to 61 – 72° here across both activities. Kimel-

Naor's participants may have compensated for this by increased motion at the hip and/or ankle.

KFHS was higher during uphill walking (32.1°) compared to downhill walking (Table 6-27), such that the knee was in a near neutral alignment when downhill walking (2.6°) which was also was seen another study (Lay et al., 2006). Two studies similarly showed high KFHS of 27.3° and 33° by younger participants walking at +8.5 and +10° respectively (Haggerty et al., 2014; McIntosh et al., 2006). Another paper presented data from +8° uphill walking, however KFHS was less than half of that seen here (Han et al., 2009). Han et al., also reported KFTO which was 12° less than the current study which again may be attributed to age differences where younger adults were analysed.

The stance flexion maximum (PKFS) and minimum (MKFS) showed no clinical relevance in PKFS (Table 6-29) between +7.5° and -7.5° walking (mean difference: ~1.7°), however MKFS was notably different between uphill (3°) and downhill (20°) – implying that the knee reached a larger stance ROM during uphill walking (Table 6-30). No studies reported both PKFS and MKFS during up- and downhill walking, however Lay et al., (2006) showed a near neutral MKFS during uphill walking of 4.7° similarly to this study. In addition, a higher MKFS was seen in the aforementioned study of 24.2° during downhill walking and compared to this study (19.7°) this is a near clinically similar result. A higher MKFS and comparable PKFS during downhill walking was reported by another paper (McIntosh et al., 2006). McIntosh et al., gave a higher PKFS during uphill walking compared to downhill, and no MKFS during uphill walking was presented. A final paper showed similar PKFS to this study during +8.5° incline walking (Haggerty et al., 2014) with a negligible mean difference of ~0.5°.

The stance peak (Table 6-17) and range of knee adduction (Table 6-23) during sloped walking was lower than during stair navigation in this study, although not by a clinically relevant amount. Furthermore, peak rotation (Table 6-18) was higher during stair navigation than slope walking by a non-clinically relevant amount also. Conversely, rotational ROM (Table 6-24) was higher during sloped walking than stair

navigation, and between decline walking and stair descent this difference was clinically relevant (7.2°). No papers compared non-sagittal motion during stair and sloped walking together. Comparisons to studies where these activities were carried out separately showed lower published values than this study, despite that in this study adduction kinematics are from stance (Gao et al., 2012; Komnik et al., 2016; Saari et al., 2004).

Compared to downhill walking, uphill walking presented higher peak and range of adduction – albeit not by clinically relevant amounts, and lower peak and range of rotation which were by clinically significant amounts. This finding was not corroborated by a paper also reporting non-sagittal motion during incline and decline walking (Komnik et al., 2016). Komnik et al., showed higher peak and range of both adduction and rotational values when uphill walking, by up to 3.8° which is not relevant. Peak adduction during uphill walking was similar between studies, whilst the angles reported by Komnik et al., were taken from the first 50% of stance, a visual inspection of the graphs showed that peak angles occurred in that period. Peak adduction during downhill walking in Komnik's study was 4° lower than uphill walking, unlike in this study which showed a 0.1° difference. Rotational angles in Komnik's paper were also ten times less than that seen in this study, possibly because values were taken from stance period only and peak rotation appeared during swing.

Sagittal peak (Table 6-18) and range of ankle dorsiflexion (Table 6-25) did not vary greatly between uphill and downhill walking, giving a range of 20.7 – 24.6°. Uphill walking gave marginally higher dorsiflexion peak and ROM compared to downhill walking which was also seen in the literature (Kimel-Naor et al., 2017; Lay et al., 2006). However, the differences in literature were considerably larger, for peak dorsiflexion differences of 10° and 23.3° was seen in these respective studies between slope directions, a higher degree of incline and younger participant group may account for this discrepancy. Lower peak dorsiflexion at a steeper incline with younger participants was also seen in another study (Kuster et al., 1995) so findings are inconclusive. Dorsiflexion ROM was 1° less than peak dorsiflexion for uphill

walking and 2° higher for downhill walking indicating a small amount of plantarflexion occurred during downhill walking. This is expected as to descend the COM to travel downslope would require more extension than for uphill walking.

7.2.3. Patient Comparison: Level Walking

As data from level walking is reported more frequently and easier to standardise than other tasks, implant comparison during this ADL is reported first. Patients walked overground and on a treadmill at 0°, giving two level walking modalities available for comparison. As overground walking showed full sample sizes than treadmill level walking, the kinematics of overground walking will be the main focus of this chapter. Assessing the similarity in kinematics between walking modalities provides confidence to the kinematics from sloped walking, so some comment will be made on this before the main implant comparison. Summing the absolute difference in average kinematic parameters between treadmill and overground walking of all patients gave a total of 120° of discrepancies at preoperative, which reduced to ~85° at postoperative. The largest preoperative discrepancies were for the hip, knee, and ankle flexion ROM and KFHS. All of these were considerably reduced at post-surgery and just the range of knee adduction showed high inconsistency between walking modes.

Compared to patients, controls generally showed greater hip flexion ROM, KFTO, PKFS, and peak and range of knee flexion and rotational angles during level walking. KFHS was also greater in controls compared to postoperative patients. On the whole, controls showed lower pelvic ROM, MKFS, peak and range of knee adduction during stance, and peak dorsiflexion compared to patients. Controls' sagittal pelvic ROM (Table 6-20) showed differences less than 1.0° to implant groups, and all patients showed little variation (range of 1.2°) implying that pelvic motion is less implicated post-TKA. This parameter was not found to be reported in the literature perhaps further confirming the low relevance of this parameter during level walking in TKA patients.

Control hip flexion ROM (Table 6-21) was greater than patients by a clinically relevant amount (except to the post-CRDD group where they exhibited a non-relevant increase) and this increase was significant compared to the preoperative UC and UCR groups. All patients showed intra-implant increases in hip flexion ROM and the greatest change was seen in the UC group. These increases were such that no postoperative hip flexion ROM was significantly lower than controls, however clinically relevant decreases to controls remained for the UC and UCR groups. This contrasts the literature where greater preoperative hip ROM was seen than postoperative for both bearing types (Tibesku et al., 2011). No comparison to controls (or bearings) was made by Tibesku et al., however, similar to this study the fixed bearing (FB) implant gave greater hip flexion ROMs than to the mobile bearing (MB) implant, albeit no significance was seen in both studies.

In this study, control peak and range of knee flexion was larger than all patient groups by a clinically relevant amount (Table 6-16 and Table 6-22 respectively). Preoperative peak knee flexion was not statistically different to controls, which changed at post-surgery where all implant groups showed significance to controls. Both FB implants showed non-clinically relevant intra-implant decreases in peak knee flexion, the UC group by ~3^{o} more than the CRDD group implying greater function loss here. The FB implants also showed the largest and smallest patient peak flexion (CRDD and UC respectively). Although the UCR group showed a non-significant nor relevant postoperative increase in peak knee flexion, the angle seen was less than the CRDD group. In the literature, implants of both bearing type showed increases in peak flexion from preoperative (Möckel et al., 2004; Sosio et al., 2008; Tibesku et al., 2011; Urwin et al., 2014; Zeng et al., 2019) whereas in the current study the same finding was only seen in the MB group.

Knee flexion ROM increased from preoperative for all patients, still not to control levels. Of the postoperative groups, the CRDD group presented the largest ROM and showed the only significant intra-implant finding. The post-CRDD group was the only postoperative group presenting a statistically similar ROM to controls, although the

mean difference was 5.8^{o} which is clinically relevant. The post-UC group presented the smallest intra-implant increase and inter-implant value. The MB UCR group gave an intra-implant change and ROM between the CRDD and UC groups implying bearing type is not relevant to this activity. Postoperative increases in ROM not to control levels was also noted in the literature, in some papers the FB implant was seen to increase ROM more than the MB implant (Möckel et al., 2004; Urwin et al., 2014) and another reported larger increases in ROM by the MB implant (Zeng et al., 2019). In contradiction to these papers and the current study, one study reported postoperative reductions in ROM more so by the MB group than FB (Tibesku et al., 2011). Although Tibesku et al., (2011) did not analyse inter-implant significance the difference in postoperative ROM for between bearing types was ~0.6^{o}, assumed to be statistically similar, intra-bearing differences were significant, however.

During stance, the contralateral limb begins swing and the ipsilateral limb undergoes a loading response phase, followed by single limb support, culminating to pre-swing in late stance (Perry and Burnfield, 2010). Instability is frequently noticed during stance where the limb is under load – which may be mitigated by maintaining smaller ROM to maintain joint conformity and stability (Farrokhi et al., 2015; Vaienti et al., 2017). From preoperative, all patients showed a consistent decrease in the additional sagittal knee flexion parameters (KFHS, KFTO, PKFS, MKFS), indicating a tendency towards extension. This was a clinically relevant decline for all parameters in the UC group, less so in the CRDD group, and the MB UCR group presented no clinically relevant reductions at postoperative.

The largest postoperative reductions in these parameters were seen for the MKFS (Table 6-30), the minimum between stance and swing flexion maxima which was a significant and clinically relevant change from preoperative for the FB groups ($p<0.05$). The post-UCR MB group also showed near clinically relevant decreases from preoperative indicating any flexion contracture or other pre-surgery impediments to extension were alleviated, typical of TKA (Tanzer and Miller, 1989; Tew and Forster, 1987). Most patients showed greater MKFS than controls (except for the post-UC

group) and only the pre-CRDD group showed significance to controls ($p<0.05$). The post-UCR group presented the highest MKFS from all groups, which was also corroborated by one study (Sosio et al., 2008) where significance was seen indicating limits to extension in the MB group than the FB. In contrast to this, a lower MB MKFS was observed in another paper (Tibesku et al., 2011), although no statistics were given between bearing types here. It should also be stated that these papers reported the maximum extension during stance which may refer to flexion at heel strike where extension was greatest. Some studies carried out radiographic evaluations between implant bearings and found similar clinically relevant decreases in flexion contracture at post-surgery, however inter-implant differences were not statistically relevant (Boldt et al., 2006; Chaudhry and Goyal, 2018; Kim et al., 2010, 2012; Woolson and Northrop, 2004) much like in this study.

Patient PKFS was lower than controls at both operative states (Table 6-29), by clinically relevant amounts to the pre-UC and all postoperative groups. Significance was reported between the post-UC group and controls only ($p<0.01$) although all groups showed lower PKFS to controls by clinically relevant amount, similar to the findings by Möckel et al., (2004). Another study showed no clear trends between control and patient findings: control PKFS was greater than the FB group, and lower than the MB group with no clinical or statistical significance seen (Sosio et al., 2008). In this current study, the post-UC group gave a clinically lower PKFS to the other implant groups and showed the only clinically relevant decline from preoperative. The MB UCR group exhibited the smallest post-surgery reduction in PKFS and MKFS by 2.3^{o} and 4^{o} respectively, implying some limits to extension. Greater PKFS in MB than FB groups were also presented by two studies (Möckel et al., 2004; Sosio et al., 2008) and greater FB PKFS was seen in another study (Tibesku et al., 2011) complimenting the unclear trends seen here. Only Tibesku et al., (2011) presented preoperative findings, and intra-implant differences showed contradictory postoperative increases in PKFS, with the FB group showing a larger increase than the MB group. As patients in the current study showed postoperative decreases in PKFS,

less weight-bearing capabilities in the operated limb is implied which may result in compensatory non-sagittal motion not presented in this study.

Knee flexion at the start (KFHS) and end (KFTO) of stance on the whole, were clinically and statistically similar to controls seen in Table 6-27 and Table 6-28 respectively. The UCR group showed the largest values (more flexion), and the UC group showed the lower, more extended angles at post-surgery. Only the post-UC group presented KFHS and KFTO angles lower than controls by a clinically relevant amount, and significance was seen for KFTO ($p<0.05$). As mentioned, all patients showed postoperative decreases in angles, such that controls consistently displayed greater KFTO to all patients. Preoperative KFHS was greater than controls by up to 2^{o}, which reduced at postoperative as between 0.3-5.6^{o} less flexion to controls was seen. This finding was not corroborated by Sosio et al., (2008), where controls presented lower KFHS to both the FB and MB implant groups. Sosio's MB group also did not show significantly greater KFHS compared to the FB group, a finding also seen here. Pre-surgery findings were not reported by Sosio et al., (2008) nor were these parameters reported by other papers, so comparison is limited. At post-surgery, both FB groups in this study showed the only significant and clinically relevant intra-implant decrease in KFHS. Despite the UC group displaying the greatest mean difference, the CRDD group showed larger degrees of significance between operative states likely due to the larger standard deviations in the UC group.

In the non-sagittal plane, controls generally presented smaller peak (Table 6-17) and range of knee adduction (Table 6-23) during stance to patients. Preoperative OA groups showed the largest peak adduction of nearly 10^{o} which lessened at postoperative to around 8^{o}, for the UC group this was to control levels. Post-surgery peak adduction in the other two implant groups were up to 2.6^{o} more than control values. Stance adduction ROM trends were more varied, the largest ROM was again seen in the preoperative groups as well as the post-UCR group. FB groups showed post-surgery decreases in ROM to still above control levels and conversely the UCR group showed a small postoperative increase. However, no significant nor clinically

relevant differences between participant groups were observed for either adduction parameter.

In the literature, adduction angles are often reported during the whole cycle, however some comparisons may be made as stance trends may translate to whole cycle. Greater preoperative/OA peak adduction than controls and/or postoperative was corroborated by the literature (Gök et al., 2002; Weidow et al., 2006; Zeng et al., 2019). One study conversely showed lower peak adduction in a pre-MB group than controls, whereas pre-FB groups gave the highest adduction (Urwin et al., 2014). Large standard deviations were seen in Urwin's participant groups and so are potentially susceptible to the effect of outliers. Maximum adduction in the literature were, on occasion, smaller than the stance peak angles seen in this study confirming cross talk error. The largest difference was compared to the findings by Gök et al., (2002) where Gök et al.,'s OA group showed ~7^{o} less adduction during stance than the preoperative groups here. Greater peak adduction reaching 15^{o} were reported by two studies (Weidow et al., 2006; Zeng et al., 2019), however these were taken from the whole gait cycle and are less comparable to this study.

Trends in adduction ROM in the literature was as varied as this study, OA ROM was presented to be: larger than controls (Bytyqi et al., 2014), equal to controls (Weidow et al., 2006) and lower than both controls and postoperative (Zeng et al., 2019). Much like peak adduction, the adduction ROM presented by Urwin et al., (2014) was greater than controls in the pre-FB group, and lower than controls in the pre-MB group. Whole cycle adduction ROM presented in three studies (Bytyqi et al., 2014; Weidow et al., 2006; Zeng et al., 2019) were smaller than the stance ROM presented in this study's participants further implying measurement error here. Whole cycle ROM was larger in one study than presented here with differences of up to 10^{o} seen in related participant groups (Urwin et al., 2014). As the same model was utilised by both studies, marker error could have occurred in both instances considering the lower adduction ROM reported in other studies in Table 2-1.

Intra-implant differences in the literature showed postoperative reductions in peak adduction of corresponding groups (Urwin et al., 2014; Zeng et al., 2019). However, ROM changes gave postoperative increases in both fixed and mobile bearing groups in the papers, whilst FB groups in the current study presented small post-surgery decreases. The magnitudes of intra-implant changes in the literature, particularly peak adduction, were greater than this study. Changes from preoperative were clinically relevant, and Zeng et al., (2019) reported statistical analysis between operative states and saw significance where $p<0.001$ for both implant bearings. Similar to this study, inter-implant changes in the literature showed no significance between bearing types for either peak or range of adduction motion.

In the transverse plane, controls exhibited larger peak (Table 6-18) and range of longitudinal angles (Table 6-24) than all patients by 3.2–9.0^{o}. This was a clinically relevant difference to all patients aside from the pre-CRDD group. Two studies also reported greater control peak and/or range of knee rotation to OA patients by 2-4^{o} (Bytyqi et al., 2014; Weidow et al., 2006) and significance was seen in Weidow et al., (2006)'s study. Another study showed significantly larger peak and insignificantly larger range of rotation in OA patients than controls (Zeng et al., 2019). As values from the whole gait cycle were presented, direct comparison to the literature is possible. On the whole, peak values in the current study were more than 10^{o} larger than published values (Weidow et al., 2006; Zeng et al., 2019). Some similarity was seen in postoperative groups to another study (Zürcher et al., 2014) with differences less than 1^{o}. Rotational ROM in this study ranged between 11.8–20.6^{o} which is slightly higher than literature values of 7.6–13.4^{o} (Bytyqi et al., 2014; Weidow et al., 2006; Zeng et al., 2019; Zürcher et al., 2014).

Post-CRDD and UCR patients showed diminished rotational kinematics from preoperative, this was a significant finding for the CRDD peak rotation ($p<0.01$). The UC group conversely showed post-surgery increases in rotational kinematics although not to a significant or clinically relevant extent. These trends are in contrast to the findings by Zeng et al., (2019) where the both the FB and MB groups showed a

postoperative increase in peak rotation and ROM. In the current study, the highest and lowest post-surgery rotational kinematics were respectively seen in the FB UC and CRDD groups. The MB UCR group presented rotation angles similar to the CRDD group, with differences no more than 0.3°. This finding is inconsistent with other studies where the MB rotational kinematics were larger than FB groups (Zeng et al., 2019; Zürcher et al., 2014). Only the MB peak rotation presented by Zeng et al., (2019) was significantly greater than the FB group.

In this study, the peak and range of ankle dorsiflexion during gait of all participants were similar, the lowest values were seen in the UC group for both parameters, and the pre-CRDD and UCR groups presented the greatest ROM and peak dorsiflexion, respectively (Table 6-25 and Table 6-19). No clinically relevant or significant differences were seen amongst relevant groups. Patients mostly showed higher maximum dorsiflexion than controls – aside from the post-UC group which showed smaller dorsiflexion angles. Control dorsiflexion ROM was smaller than the CRDD group, more than the UC group, and roughly equivalent to the UCR group at both operative states. Studies reporting these parameters showed lower control peak dorsiflexion compared to pre-surgery OA groups (Al-Zahrani and Bakheit, 2002; Levinger et al., 2013). Inconsistences were seen for dorsiflexion ROM in the literature as studies reported greater control ROM (Astephen et al., 2008; Levinger et al., 2013) or lower control values (Ko et al., 2011) than OA groups by non-relevant amounts. Sosio et al., (2008) reported control and postoperative patient dorsiflexion kinematics in FB and MB groups during stance, their control ROM was greater than both bearing types and peak angles were non-significantly greater than the FB group and less than the MB group. Whilst ROM was not explicitly presented by Sosio et al., significance was seen for peak plantarflexion angle between groups.

Intra-implant differences in the current study mostly gave non-clinically relevant decreases for both the peak and range of dorsiflexion, except for the UC group which exhibited a postoperative increase in dorsiflexion ROM by ~0.6°. This is in slight contrast to a study by Tibesku et al., (2011) where non-significant increases in peak

dorsiflexion angle was seen in both FB and MB groups. Dorsiflexion ROM also decreased in both bearing types in Tibesku's study which was not consistently observed here. Furthermore, in Tibesku's study postoperative inter-implant comparisons showed the largest peak and range of dorsiflexion was seen in the FB group, and greater intra-implant differences were seen in the MB group. Tibesku et al., did not carry out inter-group significance testing in their study, however intra-group analysis was carried out and no significance was seen much like in this study.

Preoperative groups showed no significant and, mostly, no clinically relevant differences to each other, implying groups are near equivalent during level walking. Exceptions to this were where the pre-CRDD group presented ~6.1^{o} more dorsiflexion ROM than the pre-UC group, and the pre-UC group showed up to 6.2^{o} less rotational ROM to the other groups. The parameters where all pre-surgery groups displayed a clinically relevant difference to controls were the range of hip and knee flexion, peak knee flexion, rotational ROM, and MKFS. The pre-UC and UCR groups also showed clinically relevant differences to controls for peak knee rotation. Controls showed larger values for these parameters except for the MKFS. At postoperative, the UC group exhibited smallest knee kinematic values, except for the peak and range of knee rotation where the largest values were presented. Smaller sagittal and frontal motion are perhaps complemented with larger transverse ROM for the UC group, and the opposite was seen in CRDD group which showed largest sagittal and frontal motion and smallest rotation angles, implying bearing had no effect here.

Whilst no statistically significant inter-implant differences were observed at postoperative, more clinically relevant increases to the UC group were seen. These were namely observed for peak knee flexion and PKFS (CRDD and UCR), and KFHS, KFTO, and MKFS (UCR group only). Clinically or statistically relevant post-surgery differences to controls persisted from preoperative for hip flexion ROM, peak and range of knee flexion and rotational angles, and no longer for the MKFS angle as TKA removed any indication of contracture. Usually, the direction of intra-implant change between the peak and range of a kinematic parameter would be the same, for

instance, an increase in peak adduction would be corroborated by an increase in adduction ROM. This was not observed for knee flexion where patients displayed a postoperative decrease in peak flexion and an increase in flexion ROM. The improvement in ROM was not to control levels, and although the CRDD group showed the largest ROM of all implants and was the only group showing no significance to controls, a clinically relevant difference to controls was still maintained.

Whilst neither clinically or statistically relevant, intra-implant changes in the FB group included decreases in pelvic tilt ROM, peak knee flexion, and stance adduction ROM. Whereas MB implant showed increases in the same parameters. Post-surgery inter-implant comparisons showed that the FB groups exhibited greater pelvic (< 0.5^{o}) and hip (1-4^{o}) ROM, and less adduction ROM (1.5^{o}), KFHS, KFTO and MKFS than the MB group. These were by a clinically relevant amount for the KFHS, KFTO and MKFS between the UC and UCR groups. For these same sagittal parameters, the CRDD and UC groups showed some degree of clinically or statistically relevant difference to preoperative and/or controls which was not observed in the UCR group. Whilst all groups showed postoperative decreases in additional sagittal parameters, this was by a lesser extent in the MB UCR group indicating this group is limited in limb extension.

The impacts of spatiotemporal parameters on joint kinematics are considerable. The pre-UCR group showed STP data typical associated with a slower walking speed, and similarly showed the smallest peak knee flexion angle, KFHS, and smallest sagittal pelvic and knee ROM. Conversely the pre-CRDD group showed greatest walking speeds, and greatest sagittal hip, knee (in all dimensions), and ankle kinematics. Greater kinematics at higher walking speeds is corroborated by the literature (Bejek et al., 2006; Fukuchi et al., 2019; Kwon et al., 2015; McClelland et al., 2011; Stoquart et al., 2008) . At postoperative similar trends persisted, the CRDD group maintained a faster walking speed and greater sagittal hip, knee, and ankle kinematics. However, the post-UC group who had a similar walking speed to the CRDD group showed lowest kinematics. Despite that it could be considered that the UCR group had the lowest

baseline at preoperative the magnitude of kinematics often surpassed the UC group, albeit by a small (<5^{o}) margin. Preoperative function is a considerable determinant of postoperative function confirming that intra-operative change is an insightful parameter to assess.

7.2.4. Patient Comparison: Other ADLs

Stair negotiation and sloped walking require controlled raising or lowering of the centre of mass repeatedly to safely traverse a staircase or uneven terrain. These activities are functionally demanding, and participants commonly exhibit greater kinematic profiles compared to level walking. The samples sizes for these tasks were fewer than during overground level walking, due to hardware and/or mobility issues. Participants were also excluded if they employed a step-by-step strategy when ascending or descending stairs. And participants either found it too difficult to walk at steeper slopes on the CAREN system, or hardware faults in the system made it not possible to complete the activity. Generally, more participants completed stair navigation than sloped walking meaning statistical power is slightly greater for the stair activity. Furthermore, as fewer studies complete motion analysis during these ADLs, especially with these participant groups, comparison to the literature is limited.

Whilst no significance testing was carried out between activities, some clinically relevant differences were seen between walking uphill and downhill, and between increasing gradients. At +7.5^{o} walking, participants showed up to 20^{o} more hip flexion ROM, up to 30^{o} more KFHS and a PKFS up to 15^{o} more than -7.5^{o} walking. Downhill walking presented higher values by a clinically relevant amount in more parameters than uphill walking. Namely these were in the peak and range of knee flexion (up to 12 and 15^{o} respectively), up to 20^{o} more KFTO, and up to 16^{o} greater MKFS. These findings are expected where the goal is to raise or lower the COM with each step. Between +5 ^{o} and +7.5^{o} walking, some clinically relevant differences included larger hip and knee flexion ROM, KFHS and PKFS. MKFS also decreased and peak dorsiflexion increased with steeper uphill inclines although not by a relevant amount. As downhill walking increased from -5^{o} to -7.5^{o}, clinically relevant observations included

reductions in hip ROM with steepness and in some participant groups, increases were seen in peak knee flexion, KFTO, PKFS and MKFS. Non-clinically relevant differences or even no change was also observed between inclines, which may be expected as the number of participants walking at ±5^{o} then ±7.5^{o} were not equal.

Sagittal pelvic ROM during the stair tasks ranged between 4.0-11.3^{o} and were larger than overground level walking ROM that ranged between 2.7-3.9^{o} (Table 6-20). Furthermore, a greater ROM was seen for stair ascent than descent. During sloped walking, no clinically or statistically relevant differences in ROM between participants were observed, and variability between groups was small (less than 2^{o}). Any clinically relevant differences in pelvic ROM to controls was seen during the stair tasks, where the pre-UC (both ascent and descent) and pre-UCR (ascent only) groups showed a larger pelvic tilt ROM likely to account for limited knee mobility (Asay et al., 2009; Linley et al., 2010). Significance was seen during stair ascent only where the UCR group exhibited a higher than control ROM (at both operative states), and the post-CRDD group displayed significantly lower ROM to both of the other implant groups. Reduced post-surgery sagittal pelvic ROM may be complemented by increased ROM in other joints and indeed the CRDD group presented the largest postoperative hip ROM, and peak knee flexion angle compared to the other implants as discussed later.

Hip flexion ROM during the ADLs were greatest during stair ascent (range: 52 - 61^{o}) and the next highest values were seen during uphill walking (Table 6-21). The lowest ROM was seen during stair descent, around half of ascent values. Higher ROM during stair ascent than uphill walking in OA patients was also reported by one study (Son and Kim, 2013). However, the preoperative group of this current study showed greater ROM values than Son and Kim's study which may be attributed to step height and slope inclination discrepancy or even OA severity. Near double the hip flexion ROM during stair ascent than descent was seen in another study (Saari et al., 2004) where postoperative TKA patients and controls were analysed. The controls in Saari's study and the current study showed comparable ROM of ~61^{o} and 30^{o} during stair ascent and descent, respectively. One discrepancy is that patients here primarily

showed lower ROM to controls, whereas ROM was larger in TKA patients than controls in Saari's study. As the step heights are similar between studies, assessing intra-implant changes would confirm whether patient trends in Saari's study are expected, however preoperative information was not provided by the authors. In the current study, most significance was between preoperative OA patients to controls, as postoperative patients mostly increased in ROM. The MB UCR group showed the smallest post-surgery ROM of all groups and was the only group still presenting significance to controls. The post-CRDD group gave the largest ROM, and all significant instances were to preoperative. Intra-implant differences for the UC group were comparable or greater than the CRDD group implying the FB groups showed more increases in hip flexion ROM than MB.

Peak knee flexion was largest during stair ascent and smallest during uphill walking, even less than level walking (Table 6-16). Controls showed higher peak flexion to all patients by a clinically relevant amount for all activities, and most significance was also to controls. During stair ascent controls reached a peak knee flexion of 100^{o}, and patient peak flexion reached ~89^{o} and ~85^{o} at pre- and postoperative, respectively. 100^{o} of peak control flexion during stair ascent was also seen by one study (Gonçalves et al., 2017), and peak preoperative flexion of the moderate OA group was 2^{o} less than controls, as opposed to ~11^{o} seen in this study. Stair descent gave the next highest peak knee flexion also reported by two studies (Hicks-Little et al., 2011; Myles et al., 2002), and a conversely higher flexion during descent than ascent was seen in one study (Saari et al., 2004). Intra-implant differences during the stair tasks showed reductions in flexion for all implant groups, by a clinically relevant amount for the UC and UCR implant groups and significance was seen for the UCR group during ascent. This was not corroborated by the literature where postoperative increases in peak flexion were reported (Myles et al., 2002; Okamoto et al., 2014). Okamoto et al, (2014) reported considerably greater peak flexion reaching 129-136^{o} during a step-up task using fluoroscopy, the disparity in findings may be related to the step height which was 7cm higher than the staircase used here. The CRDD group showed the largest peak flexion of all postoperative groups, and the UCR group gave the smallest

peak flexion for the stair tasks by clinically relevant amounts to the other implants. Greater FB flexion during the stair tasks was also seen in the literature (Catani et al., 2003; Fantozzi et al., 2003) although not by a clinically relevant amount. One study showed larger MB group flexion (Okamoto et al., 2014) again not by a clinically relevant amount.

Where studies reported peak flexion during both sloped walking and stair navigation, similar to the current study, flexion during sloped walking was also less than the stair tasks (Myles et al., 2002; Son and Kim, 2013). Peak preoperative flexion during +5^{o} walking in this study ranged between 47.1^{o} and 51.8^{o}, comparable to the findings by Myles et al., (2002) where peak flexion of 48.2^{o} was reported at the same incline. Son and Kim (2013) reported higher peak flexion values of 63.4^{o} and 58.6^{o} when walking uphill at 4.76^{o} and 7.13^{o}, dissimilar to the current study. Only Myles et al., (2002) reported postoperative peak flexion too, and intra-implant differences showed improvements of 10^{o} from pre-surgery. In this study, FB groups showed small intra-implant decreases, and the MB implant exhibited a post-surgery increase in peak flexion, all changes were less than 2^{o} contrasting Myles et al., (2002). Peak flexion during downhill walking only was reported by Myles et al., (2002) at -5^{o} and around 8.4^{o} more postoperative flexion than preoperative and 10.2^{o} less flexion than controls was reported. These are less similar to the current study where intra-implant differences were less than 2.1^{o} and in the CRDD, UC and UCR groups, no change, a postoperative increase, and a postoperative decrease respectively was seen.

Knee flexion ROM was primarily greatest during stair descent (Table 6-22) which was corroborated by the literature (Catani et al., 2003) although the published ROM were less than the present study. Two instances of a larger stair ascent ROM were seen in the post-UC and pre-UCR groups, however not by a clinically relevant amount. Flexion ROM during downhill walking was greater than uphill walking, in keeping with the trends of the stair tasks as well as the literature (Myles et al., 2002). Preoperative OA ROM during uphill and stair ascent in the current study was also comparable to published findings (Son and Kim, 2013). Myles et al, (2002) also presented OA ROM

during sloped walking and stair tasks, and in the current study pre-surgery sloped walking findings were similar and stair ascent ROM was at least 10^{o} lower in Myles's study. Much like peak knee flexion, controls in this study consistently showed higher flexion ROM to patients for each ADL. This was a clinically relevant finding for all patients except the pre-UC group during stair descent. Greater control values to patients' were also seen in the literature, however published ROM figures showed discrepancies to this study where, lower ROMs were seen (Catani et al., 2003; Wen et al., 2019), or higher (Saari et al., 2004), or comparable to this study (Myles et al., 2002) for slopes of similar gradients.

Unlike peak knee flexion, generally intra-implant increases in knee flexion ROM were exhibited, with the FB CRDD and UC groups presenting the largest changes. Particularly during sloped walking all FB group intra-implant increases were clinically relevant and the CRDD group exhibited the largest ROM for all sloped walking tasks. The CRDD group also showed significantly higher ROM than the UCR group for +5^{o} walking, and significance to both of the other implants during -5^{o} walking. Despite the post-UC group exhibiting clinically relevant decreases in ROM during stair descent, they also showed the largest ROM from all implants for the stair tasks. The MB UCR group showed larger decreases in ROM for both stair tasks (by 14^{o} and 8^{o} for ascent and descent respectively), and smaller increases during sloped walking. Most of the clinically relevant inter-implant differences was also to the post-UCR group. This was not corroborated by other studies where small postoperative increases were seen in both FB and MB implants during stair ascent (Okamoto et al., 2014). Myles et al., (2002) also reported increases in ROM from preoperative for all stair and incline activities however all patients received the same implant and so no bearing comparison was possible. In the literature greater MB ROM over FB groups were also reported, although differences were not significant (Catani et al., 2003; Okamoto et al., 2014).

Knee flexion at heel strike (KFHS) and toe off (KFTO) provides an idea of limb alignment at the beginning and end of stance phase. Less flexion may relate to a need

for stability and low confidence in the joint, which would be compensated by increased motion at other joints. KFHS was greatest during stair ascent, and smallest during decline walking to near extension levels (Table 6-27). KFTO was conversely lowest during stair ascent, and largest during stair descent close to overall peak values (Table 6-28). In this study, controls showed the largest KFHS for all activities other than decline walking where OA groups showed presented greater KFHS. Controls also presented the largest KFTO aside from stair descent and -7.5° walking where up to two OA groups gave greater KFTO per activity. No studies reported these parameters during sloped walking in TKA patients and controls, and studies corroborated greater control flexion to patients for stair navigation (Asay et al., 2009; Catani et al., 2003; Draganich et al., 2002; Fantozzi et al., 2003; Gonçalves et al., 2017; Hicks-Little et al., 2011). One study showed greater OA KFHS and KFTO than controls during stair navigation and whilst some parameters showed significance between OA and controls all differences were less than 5° (Hinman et al., 2002).

With increasing uphill inclination, greater KFHS and decreases in KFTO was seen in OA patients in this current study and in the literature (Son and Kim, 2013). Greater KFHS as inclines become steeper is expected: as the surface is progressively raised, the limb extends less to reach the ground than it would for level or decline walking. Similarly, during decline walking, KFTO increases with steepness as the surface is progressively closer to the COM. To achieve adequate ground clearance to commence swing greater flexion is required. At postoperative, all patients showed less KFHS and KFTO for all activities, except for an intra-implant increase in average KFTO of 3° by the UCR group during stair ascent and +5° incline walking. More clinically relevant decreases in these parameters were seen in the FB (UC then CRDD) then the MB group and any clinically relevant inter-implant difference was to the UC group. During +5° walking the other implants showed less KFHS and more KFTO during stair ascent and decline walking by more than 5°. KFTO during these ADLs in TKA patients were not reported in the literature, and KFHS findings showed some inconsistencies. During stair ascent, FB groups exhibited at least 5° greater flexion

than the MB group (Fantozzi et al., 2015), and other studies presented greater flexion from the MB group than FB groups (Catani et al., 2003; Draganich et al., 2002).

As established for level walking, during stance the knee reaches a maximum flexion (PKFS) during contralateral limb swing, proceeded by a maximum extension (MKFS) prior to ipsilateral limb swing. This is less seen during stair navigation since during stair ascent contralateral swing occurs whilst the ipsilateral limb is extending (Protopapadaki et al., 2007). A minor stance peak or plateau is seen during descent however it was not possible to robustly extract this data and so only data from sloped walking was analysed. Compared to level walking, PKFS was greater in all groups for all activities (Table 6-29), and level walking MKFS were between up and downhill walking, with downhill walking presenting higher MKFS (Table 6-30). For both PKFS and MKFS, all participants showed greater flexion at ±7.5^{o} than ±5^{o} walking. Whilst controls showed the largest PKFS of all groups, preoperative patients frequently presented largest MKFS - typical for OA patients. Following TKA all patients exhibited reductions (or more extension) in PKFS and MKFS with the FB implants showing larger, more clinically relevant and some significant intra-implant decreases than the MB group. This confirms impediments to extension are evident in the MB group (as was discussed during level walking) perhaps linked to reduced quadriceps function (Sosio et al., 2008). Pre-existing conditions may contribute to this lower postoperative function, however the pre-CRDD showed similar MKFS values to the pre-UCR group and the CRDD group showed considerably larger improvements.

In the non-sagittal planes, the peak (Table 6-17) and range of stance phase knee adduction (Table 6-23) during the ADLs were greater than level walking, and whole cycle rotational peak (Table 6-18) and ROM (Table 6-24) were smaller compared to level walking. Stair navigation (particularly descent) gave the largest non-sagittal values in participants, which was often over double compared to sloped walking predominantly for stance adduction parameters. One exemption to this is that decline walking rotational ROM was equal or higher than stair navigation values albeit by a non-clinically relevant amount. Whilst no single study reported non-sagittal

parameters during both stair and sloped walking, greater stair adduction values were similarly reported by Saari et al., (2004) than those reported during sloped walking in other studies (Komnik et al., 2016; Wen et al., 2019). The published adduction values were taken from the whole cycle, ranging from 2.3° to 11°, which are considerably smaller than stance only adduction seen in this study (range: 2.2° to 28.6°). Rotation during stair negotiation in TKA patients were not reported in the literature, and non-sagittal stance kinematics during sloped walking were presented by Komnik et al, (2016) which were far smaller compared to the current study. The slope inclination of ~2.6° was smaller than in this study however, Komnik et al. also reported findings from level walking which were also considerably less than the stance adduction presented displayed by participants here. Komnik et al. used the Anybody™ Modelling System (AnyBody Technology, Aalborg, DK) to process data which is possibly less validated than PiG however, it may reduce cross talk errors associated with PiG. This is because the AnyBody Technology system can produce an optimised biomechanical model, utilising medial markers, and inverse dynamics is based upon scaled cadaver datasets to the participant-specific anatomical landmark scaled model.

Despite the likelihood of error in the current study, if assumed to apply similarly for each participant some trends may be described. Group comparisons showed that controls primarily exhibited smaller peak stance adduction for all tasks, and smaller stance adduction ROM for all tasks aside from uphill walking where controls showed the largest ROM. Peak control adduction showed significance to only the post-CRDD group during the stair tasks, and the CRDD group showed the overall peak largest adduction figures of all groups, and largest adduction ROM of all patients. Control rotation were consistently larger than all patients and some significance to patients was seen during the sloped walking tasks. The only clinically relevant intra-implant changes were increases during both stair tasks in peak adduction (both FB groups) and adduction ROM (all groups). No clinically relevant intra-implant changes were seen in rotation ROM, the post-CRDD group showed a clinically relevant decline in peak rotation, and the post-UC group showed an increase during decline walking,

contradictory to the other implants. Although no significant inter-implant differences were reported, a number of clinically relevant differences were seen mainly to the CRDD group for the adduction parameters.

Participants' peak ankle dorsiflexion was greater during all ADLs than level walking (Table 6-18), and stair dorsiflexion ROM was greater than level walking whilst sloped walking showed smaller a ROM (Table 6-25). The largest dorsiflexion was observed during stair descent, where ROM was between 18.8^{o} – 29.8^{o} larger than level walking reaching 52^{o} in patients. Literature dorsiflexion ROM during uphill walking and stair ascent were found from one study only, from OA participants (Son and Kim, 2013). Dorsiflexion ROM during uphill walking were comparable to this study, however stair ascent ROM in this study was between 9.1-13.1^{o} higher than in Son and Kim's study. This study's step height was 7.5cm higher in Son and Kim's study, and decreased hip and knee flexion ROM as well as less KFHS and double the KFTO was seen which may be attributed to the step height discrepancy. Peak dorsiflexion was more frequently reported in the literature although findings are from controls and OA participants with no TKA results, all studies showed smaller peak dorsiflexion compared to this study (Gonçalves et al., 2017; Hicks-Little et al., 2011; Son and Kim, 2013). The step heights of Gonçalves's and Hicks-Little's study were similar to the step height in this current study, and peak knee flexion in both papers was larger than the participants of this study. In addition, hip kinematics were omitted which may further explain the lower peak dorsiflexion.

Most clinically relevant differences between patients and controls were where controls exhibited larger dorsiflexion to either FB implant, and one instance where the pre-UCR group gave a greater than control dorsiflexion ROM during -7.5^{o} walking. No clinically or statistically relevant intra-implant differences were seen and as no studies reported TKA values during these tasks it is not possible to gauge whether values were in expected ranges. No significance between groups was seen in peak dorsiflexion and any clinically relevant inter-implant differences were to the post-UC group which often showed the smallest dorsiflexion peak and ROM. The UCR group

gave the greatest peak dorsiflexion of all post-surgery groups, and the CRDD group generally exhibited the largest ROM.

Overall, stair navigation and sloped walking are functionally demanding tasks that, as well as showing greater kinematics than level walking, also showed larger variability between groups. This evidenced the demanding nature of these tasks, and highlighted mobility limitations in patients. Whilst comparing postoperative inter-implant differences gives an idea of function at that operative state, the intra-implant change from preoperative provides a better insight into implant behaviour. For instance, the highest postoperative knee flexion ROM during stair descent was seen in the UC group of 80^{o}, however this was a 6^{o} decrease from pre-surgery. The next highest ROM was seen in the CRDD group of ~79^{o}, which was 2^{o} greater than preoperative. Although not a clinically relevant increase, to remain around preoperative levels may be preferred to losing function. Furthermore, a reduction in angle is not necessarily a loss of function, decreases in either the pelvic, hip and ankle sagittal angles are acceptable if accompanied by an increase in knee angle. A reduction in MKFS is also an improvement in function as more pre-swing extension is realised, which is commonly reduced in OA as some contracture is evident.

There were no parameters where the FB and MB groups showed trend disparities for all activities. A notable instance for a single activity includes knee flexion ROM where FB groups showed non-clinically relevant postoperative increases in ROM during stair ascent, however the MB group showed substantial decreases in ROM of ~14^{o}. For the same activity, KFTO in the FB groups showed decreases (more extension at toe off), and the UCR group showed non-clinically relevant increases in flexion. Conversely during sloped walking all implant groups exhibited postoperative increases in knee flexion ROM (albeit to a lesser extent in the post-UCR group) as well as decreases in KFTO. This may suggest the UCR implant is less suitable for carrying out stair ascent, though findings are likely more influenced by unequal sample sizes across activities.

7.2.5. Summary

3. Discussion

Control and patient kinematics during overground level walking, and stair and sloped walking tasks were presented here. Control data were initially presented as this group is asymptomatic and comparison to literature should be slightly more valid as there are fewer variables affecting data. During level walking, controls showed a good likeness to literature data aside from knee adduction and rotation which were larger in this study. Greater kinematic profiles were seen during stair and sloped walking than level walking, particularly the stair tasks required more function. Control findings during the stair and sloped ADLs were again comparable to the literature aside from peak rotation which were considerably larger in this study. More similarity was seen for adduction however literature findings were from the whole gait cycle, whereas values from stance was reported here.

To give a comprehensive idea of implant performance it is necessary to compare to controls, preoperative and between implants at the same operative state. Since differences between groups may be statistically significant yet not clinically relevant (or vice versa), differences greater than 5^{o} were considered clinically relevant. Level walking in TKA patients is widely reported, and whilst values were mostly similar (aside from non-sagittal knee kinematics) to published findings, trends of whether a particular implant type was expected to give higher or lower values to another were less consistent.

In general, for all ADLs, patients showed smaller kinematics to controls as corroborated by the literature. Exceptions to this were preoperative MKFS which was more flexed than controls, and all patients showed larger peak and range of adduction than controls. Controls showed a statistically or clinically relevant greater hip flexion ROM, peak and range of knee flexion and rotation, KFTO and PKFS to patients. A notable trend seen across all ADLs was that patient peak knee flexion decreased from preoperative, with the MB group showing the largest reduction. Knee flexion ROM conversely increased at postoperative, with the MB group showing the smallest increases, and large decreases for the stair tasks. Although kinematic cross talk error was certainly present making non-sagittal knee changes difficult to

quantify, there were assumed to affect participants equally. Despite the MB design purportedly providing greater rotational motion, this was not evident as the MB group often showed postoperative reductions in peak and range of rotation. In addition, the MB group often showed smallest rotational kinematics from all implants. The FB implants showed the largest peak rotation (UC) and ROM (CRDD), and the UC group particularly may have gained function here.

No intra-implant trends immediately confirm whether a particular implant is best suited for a certain activity over another. Currently, clinically relevant trends in a parameter seen for one activity, were generally observed for all activities for all implant groups. One exception to this was knee flexion ROM where the UC and UCR groups showed decreases during the stair tasks and increases during sloped walking. This could be attributed to different sample sizes during these tasks due to participant disability or hardware errors. Where the UC group presented the greatest postoperative knee flexion ROM during stair descent of all patients despite declining from preoperative, preoperative function and the intra-implant change is perhaps more meaningful than postoperative inter-implant comparisons. Again, no consistent trends were seen between FB and MB groups across parameters and activities, potentially a further effect of low power in this study.

7.3. Kinetic Parameters

Peak joint moments and powers during overground level walking was presented in Chapter 6.5. Like the kinematics discussion, healthy control kinetics will be discussed first, proceeded by patient kinetics. Unlike kinematic data, minimum clinically relevant thresholds have not been determined for kinetic parameters and so, non-significant differences of the same magnitude of significant findings of each parameter were considered clinically relevant. Generally, fewer papers were available for comparison as parameters may be normalised by weight, height or mass or a combination of these. The moments and powers in this study were mass normalised, given in N.m/kg and W/kg respectively.

7.3.1. Control Comparison: Level Walking

The peak mass normalised external moment and power findings of healthy participants during level walking in this study and the literature is presented in Table 7-3 and Table 7-4 respectively. Twenty-six studies were included, nine reported both moments and powers, and the remaining seventeen reported either moments or powers only. One slight discrepancy to the papers are the findings from Watt et al., (2010) where results were given in Nm/kg m and W/kg m (mass and height normalised). These are still presented to give an idea of relationships between parameters.

Table 7-3: Literature moment findings of healthy participants during level walking. Values are mean ± standard deviation where available. Values in bold indicate moments given in N.m/kg m (mass and height normalised).

Study	Age	Peak Moments (N.m/kg)									
		Hip Flexion	Hip Extension	KFM1	KEM	KFM2	KAM1	KAM2	KAM3	Knee Abduction	Dorsi-flexor
Present Study	70.0±6.4	0.82±0.2	1.32±0.4	0.42±0.2	0.26±0.2	0.28±0.0	0.67±0.2	0.15±0.1	0.27±0.1	0.02±0.1	1.57±0.1
Alnahdi et al., 2011	62.7±0.9						0.7±0.2				
Astephen et al., 2008	50.3±0.1			0.5±0.3	0.4±0.1			0.2±0.1			1.4±0.1
Bovi et al., 2011	43.1±15.4	0.4±0.2	0.7±0.2	0.4±0.2	0.1±0.2	0.2±0.1	0.4±0.1	0.3±0.1	0.3±0.1	0.01±0.05	1.3±0.1
Keenan et al., 2011	34.0±11.0	0.6±0.2	0.8±0.2	0.2±0.2			0.5±0.1		0.3±0.1		1.7±0.2
Kerrigan et al., 1998	72.7±5.5	0.7±0.2	0.9±0.3	0.5±0.2	0.3±0.2	0.4±0.1					1.3±0.1
Komnik et al., 2016	57.6±6						0.3±0.1				
Kulmala et al., 2014	78.0±4.0	1.7±0.5	1.3±0.2			1.2±0.3					1.5±0.2
Kuster et al., 1995*	27.9	1.0±0.6*		1.2±0.5*		0.6±0.4*					1.1±0.2
Lay et al., 2006	24.0±3.0	**0.5±0.2**		0.4±0.2	0.4±0.2						1.6±0.1
Lee et al., 2007	65.4±6.2	0.7±0.2**	0.7±0.3**	0.3±0.3**	0.4±0.2**						1.2±0.2**
Lee and Hidler, 2008	70.3±4.8	0.4±0.2	0.8±0.3	0.6±0.3	0.1±0.2	0.3±0.2					1.4±0.2
McGibbon and Krebs, 2004	71.1±8.2	0.6	0.7	0.7	0.2	0.4					1.3
Messier et al., 2005	73.0±1.6	1.0±0.1	0.5±0.0	0.3±0.1	0.5±0.1						1.4±0.1
Moisio et al., 2003	31.0±12.0	0.9±0.3	0.7±0.2	0.5±0.2	0.6±0.1		0.5±0.1				1.6±0.2
Pinzone et al., 2014	10.6±3.4	0.7±0.4	0.6±0.2	0.4±0.2	0.2±0.1	0.1±0.1	0.3±0.1	0.1±0.1	0.2±0.1	0.05±0.1	1.2±0.2
Riley et al., 2001	23.9±4.4	0.7±0.2	0.8±0.2	0.5±0.01	0.3±0.2						1.6±0.2
Saari et al., 2005	62.5±5.9	0.6	1.3	0.6	0.3		0.6			0.3	
Schmitt and Rudolph, 2007	58.5			0.5	0.4		0.5		0.4		
Sims et al., 2009 (f)	60.4±9.8						0.3±0.1				
Sims et al., 20n09 (m)	63.7±9.5						0.4±0.2				
Smith et al., 2004	67.0±7.0			0.3±0.1	0.2±0.1	0.2±0.1					
Stoquart et al., 2008	23.0±2.0	0.6±0.2	0.6±0.1	0.7±0.3		0.9±0.1					0.7±0.2
Urwin et al., 2014	60.5±7.0			1.0±0.3	0.4±0.05		0.5±0.1			0.1±0.04	
Watt et al., 2010	70.3±4.8	**0.8±0.3**	**0.5±0.2**		**0.3±0.1**	**0.1±0.1**	**0.4±0.1**		**0.2±0.1**		

* - given as a net value. ** - units not specified

Controls presented greater hip extension moments (HEM) than hip flexion moments (HFM), and of the fourteen papers reporting both parameters a similarly greater extension moment was seen in half of these. The range of hip moment of this study's controls (~2.1N.m/kg) was larger than all the studies reporting both parameters aside for one paper where a range of 3.0N.m/kg was seen (Kulmala et al., 2014). Kulmala's study showed a larger peak HFM and equivalent peak HEM values to this study. The hip flexion and extension moment peaks here were similar to those reported by Saari et al., (2005). Two other studies showed peak HFM greater than this study (Moisio et al., 2003 and Watt et al., 2010) . Although Watt et al., gave a peak HFM that equalled this study's findings, the parameters were normalised by both mass and height. Based on the heights given in other studies, peak HFM may increase by a magnitude of at least 1.5 times, confirming that peak HFM given by Watt et al., is larger than in this study.

Early stance knee flexion moment (KFM) peaks were greater than late stance peaks in this study and for nearly all papers reporting both parameters. The only exception to this was seen in Stoquart et al., (2008)'s study where the second KFM peak was ~0.2N.m/kg larger than the first peak. Stoquart compared the effects of walking speed on lower limb biomechanics using a treadmill which could account for this discrepancy. Of the fifteen studies that reported both peak KFM and KEM: ten of these showed a greater range in sagittal knee moment compared to this study, two showed an equal range (Lee et al., 2007; Lee and Hidler, 2008), and three presented a smaller range (Bovi et al., 2011; Pinzone et al., 2014; Smith et al., 2004). Whilst the participant group in the study by Pinzone et al., were much younger than other studies, their reported knee flexion moment range surpassed the other two studies which showed lower moment ranges than this study. The sum of the moments between the first flexion to extension peak, and extension peak to second flexion peak came to 1.22N.m/kg in this study. Six studies similarly reported all three peaks and the sum of the moment ranges were greater (1.5N.m/kg) in two of these (Kerrigan et al., 1998; McGibbon and Krebs, 2004) and less (0.9 – 1.1N.m/kg) in the

remaining four (Bovi et al., 2011; Lee and Hidler, 2008; Pinzone et al., 2014; Smith et al., 2004) implying normal function here.

Two knee adduction moment maxima with a minimum between them is a characteristic of healthy gait (Hurwitz et al., 2002; Kutzner et al., 2013; Rutherford et al., 2008; Teichtahl et al., 2003). Thirteen studies reported at least one of the key adduction peaks – of these, twelve studies reported peak KAM and most gave a lower peak KAM than this study. The only exception was where the same peak KAM was seen (Alnahdi et al., 2011). Five studies reported both early and late stance peak KAM, and in all instances KAM1 was larger than KAM3 (Bovi et al., 2011; Keenan et al., 2011; Pinzone et al., 2014; Schmitt and Rudolph, 2007; Watt et al., 2010) similarly to this study. The magnitude of KAM3 was also similar to two studies (Bovi et al., 2011; Keenan et al., 2011) less than one study (Schmitt and Rudolph, 2007) and more than Pinzone et al., (2014)'s study – likely explained by the fact that their participants were children. The minimum between KAM1 and KAM3 maxima (KAM2) were found in three studies and the difference of this minimum to both peaks provides an idea of peak distinction. Of two studies' supplementary data all three KAM parameters were found (Bovi et al., 2011; Pinzone et al., 2014) and the total difference between peaks in this study (0.79N.m/kg) was much larger than that seen in both studies (0.1N.m/kg and 0.3N.m/kg respectively). Lastly, the peak abduction moment was found in four studies, and two reported near negligible moments like in this study (Bovi et al., 2011; Pinzone et al., 2014). The remaining studies showed higher abduction moments of 0.3N.m/kg (Saari et al., 2005) and 0.1N.m/kg (Urwin et al., 2014). Whilst this finding is perhaps less relevant in Urwin's study as this abduction moment finding is still low, the abduction moment range of 0.2–0.7N.m/kg reported in Saari et al.,'s paper is a considerably a large range.

The greatest moments in this study were ankle dorsiflexor (DF) moments, and in the majority of the fifteen studies reporting this parameter. The peak DF moments seen in three papers were lower than a peak hip or knee flexion moment (Kulmala et al., 2014; Kuster et al., 1995; Stoquart et al., 2008). Only net moments were presented

by Kuster et al., (1995) which is likely to explain this finding. Kulmala et al., (2014) showed a larger peak hip flexion moment and inferred this was because older participants are more likely to utilise greater muscular efforts at proximal joints. Whilst not unanimously seen in the other studies with older participants, the average participant age in Kulmala et al.,'s study was higher than the other groups' participants. Both knee flexion moment peaks were larger than the peak DF moment in the study by Stoquart et al., (2008), and as mentioned earlier, since a treadmill was used this may account for the lower ankle moments in this study. Overall, DF moments were comparable between the majority of studies to the present study: one paper presented greater DF moments (Keenan et al., 2011), three papers showed the same DF moment (Lay et al., 2006; Moisio et al., 2003; Riley et al., 2001) and the remaining papers showed smaller peak DF moments. Excluding the differences mentioned, the next largest difference between peak DF moments in this study to another was ~0.4 N.m/kg (Lee et al., 2007; Pinzone et al., 2014). Considering this parameter has the largest magnitude this is a small proportional difference, again demonstrating good agreement between this study to the literature.

Twelve papers reporting mass normalised lower limb joint powers during level walking is shown in Table 7-4:

Table 7-4: Literature joint power findings of healthy participants during level walking. Values are mean±standard deviation where available.

Study	Age	Peak Power (W/kg)					
		Concentric Hip	Eccentric Hip	Concentric Knee	Eccentric Knee	Concentric Ankle	Eccentric Ankle
Present Study	70.0±6.4	1.89±0.5	1.15±0.6	0.71±0.4	1.58±0.3	2.86±0.4	0.81±0.3
Bovi et al., 2011	43.1±15	0.8±0.3	0.7±0.4	0.3±0.3	0.8±0.3	3.3±0.9	0.6±0.5
Cofré et al., 2011	66.8±5.4	2.2±0.6	0.8±0.3	0.7±0.3	1.7±0.6	3.8±1.0	0.9±0.3
Kerrigan et al., 1998	72.7±5.5	1.6±0.4	0.7±0.4	0.6±0.2	2.2±0.6	2.9±0.4	0.7±0.2
Kulmala et al., 2014	78.0±4.0	1.8±0.9	1.3±0.4	1.7±0.5	3.5±1.4	3.2±0.	1.3±0.4
Kuster et al., 1995*	27.9	1.3±1.0*			2.8±2.0*	3.1±1.2*	
Lee and Hidler, 2008	70.3±4.8	1.0±0.4**	0.6±0.3**	0.6±0.4**	1.7±0.8**	3.1±1.0**	0.8±0.4**
McGibbon and Krebs, 2004	71.1±8.2	0.9**	0.5**	0.7**	1.8**	3.4**	0.8**
Pinzone et al., 2014	10.6±3.4	0.8±0.4	0.4±0.3	0.7±0.6	0.6±0.6	2.6±1.0	0.3±0.4
Riley et al., 2001	23.9±4.4	1.6±0.4	0.6±0.2	0.8±0.3	1.1±0.2	3.3±0.9	1.1±0.2
Silder et al., 2008	72.5±5.0	1.5±0.4	0.7±0.2			3.1±0.7	
Stoquart et al., 2008	23.0±2.0	1.1±0.3			2.1±0.5	2.3±0.9	
Watt et al., 2010	70.3±4.8	1.9±0.6	0.7±0.2	0.7±0.3	1.6±0.6	3.5±0.7	1.2±0.3

** - given as a net value. ** - mentions powers are sagittal.*

Some trends were seen where both generative concentric powers and absorptive eccentric powers were reported in the literature. Peak hip and ankle concentric powers were greater than eccentric powers in all studies and for the knee, eccentric powers were greater than concentric powers in all papers except one (Pinzone et al., 2014). There, the difference in the magnitude of peak concentric and eccentric knee powers was 0.1 W/kg which is low and could be expected as children made up this participant group.

The range between peak concentric and eccentric powers describes the relationship between both values as a single figure. For the hip, the power range in this study was 3.04W/kg, and greater than all studies reporting both figures, except in one instance where the range was equal (Kulmala et al., 2014). As peak concentric and eccentric hip powers in this study was greater than ten and nine of the papers respectively this explains the larger range seen presently. Exceptions to this were where the peak

concentric hip power was similar to that reported by Watt et al., (2010) and less than Cofré et al., (2011), and peak eccentric hip power was smaller than that reported by Kulmala et al., (2014).

Controls' range in knee power seen in this study was approximately 2.29W/kg. Nine studies included both knee power values, and this study's range was considerably compared to two studies (Kerrigan et al., 1998; Kulmala et al., 2014). The peak concentric knee power value was similar between this and the study by Kerrigan et al., (1998), and the eccentric power reported by Kerrigan et al., (1998) was considerably greater. The peak concentric and eccentric knee powers reported by Kulmala et al., (2014) were larger than this study by ~1.0 and ~1.9W/kg respectively. The range of knee power of six studies were smaller than this study, however three of these were only slightly so (Cofré et al., 2011; Lee and Hidler, 2008; Watt et al., 2010). The remaining papers showed greater reductions, attributed to smaller eccentric knee powers (Bovi et al., 2011; Pinzone et al., 2014; Riley et al., 2001). One paper with similarly aged participants to this study also showed the same peak power values, and thus, the same range (McGibbon and Krebs, 2004).

Ankle powers gave the largest power range of all joints in this study, and all nine papers reporting concentric and eccentric ankle powers showed mostly comparable ranges. The range of knee power (5.2W/kg) reported by Kulmala et al., (2014) was greater than their range of ankle power (4.5W/kg) and their knee power values seems more of an anomaly than the ankle power which may be attributed to the authors' conclusion that elderly people utilise proximal musculature more than distal. This present study's range of ankle power was greater than two papers (Kerrigan et al., 1998; Pinzone et al., 2014) – the difference to Kerrigan et al., (1998) is negligible, and age differences could contribute to the discrepancy to Pinzone et al., (2014)'s study. The remaining studies showed larger ankle power ranges, mostly attributed to higher concentric ankle power (Bovi et al., 2011; Cofré et al., 2011; Kulmala et al., 2014; Lee and Hidler, 2008; McGibbon and Krebs, 2004; Riley et al., 2001; Watt et al., 2010).

Much like the kinematic findings, literature figures of healthy participants' kinetics during level walking also corresponded well with the control group in this study. Although the number of papers available for comparison were fewer based on the different normalising methods used as many available papers as possible were included here. Discrepancies to the literature could be a result of the differences in participant group ages, walking modality, and data recording processes. As these were relatively small there is a high confidence of this study's findings.

7.3.2. Patient Comparison: Level Walking

Patient kinetics during overground level walking was presented in Chapter 6.5, and discrete maxima and minima of parameters during the gait cycle was analysed. Sagittal moments generally reflect the ability to propel or decelerate appropriately during gait, whilst frontal plane moments indicate stabilisation of the lower limb, particularly key during contralateral swing phase (Sloot and van der Krogt, 2016). Abnormalities in moments may come about due to an alteration in the moment arm and potentially indicate muscle or loading problems, resulting in either higher or smaller moments than asymptomatic groups. Joint powers reveal how a participant propels or stabilises the body during gait, and irregular powers may reveal where energy is dissipated ineffectively, potentially affecting a person's ability to walk. The ankle then the hip joints are the main contributors to propulsive power and deficits in one joint is often compensated for by another (Chen et al., 1997). The ankle is vital to swing initiation and the hip controls forward acceleration of the COM particularly during pre-swing where the limb is pulled up and forwards. The knee also controls loading during stance, and primarily acts to smooth gait movement by transferring energy to the ankle or hip via biarticulate muscles (Sadeghi et al., 2001). Of all non-sagittal moments only the frontal knee moment was presented, as the other moments showed no significant differences, and may be considered equivalent between groups.

As moments may be normalised to height, weight, or mass, direct comparison to the literature to this study is limited. In this chapter, reference is made to Table 6-31 and

Table 6-32 for moment and power data respectively. Overall, controls showed the largest hip, knee, and ankle flexion moments, hip extension and ankle plantarflexion moments, and greatest power values compared to all patients. Controls also showed the largest early stance knee adduction peak (KAM1) and lowest mid-stance adduction minimum (KAM2) and late-stance adduction (KAM3) to all patients. Altogether indicating some improvement is required for patients to exhibit asymptomatic kinetics.

At the hip, greater control than patient kinetics during gait was also seen in a number of studies (Levinger et al., 2013; Meinders et al., 2019; Saari et al., 2005). Peak extension moments (HEM) was almost double the peak flexion moments (HFM) which was also seen in one other study (Saari et al., 2005). However, the patients and controls of two studies showed a larger HFM than HEM (Braito et al., 2016; Meinders et al., 2019). Peak HFM in this current study (0.45-0.82N.m/kg) were comparable to the literature, and peak HEM (0.83-1.32Nm/kg) was similar to the findings by Saari et al., (2005) and larger than other studies findings (Braito et al., 2016; Meinders et al., 2019). As Meinders et al., did not describe data collection and all studies used different implants these factors may be the cause of the inconsistencies seen. In this study, all patient groups showed postoperative increases in HFM and behaved differently for the HEM: the FB CRDD and UC groups showed no change and a small increase in HEM respectively, and the UCR group showed a decline. Only Braito et al., (2016) reported pre- and postoperative sagittal hip moments and a small (>0.05N.m/kg) increase in HFM and a decrease in HEM was seen between operative states similar to the UCR group.

Patient concentric and eccentric hip powers did not show much variation across groups (ranging between 0.50–1.06W/kg), the lower findings showing similarity to another study reporting these parameters in OA and TKA patients (Levinger et al., 2013). Control powers here were considerably larger than patients', with concentric (CHP) and eccentric (EHP) hip powers reaching 1.89W/kg and 1.15W/kg respectively. This is much larger than the powers reported by Levinger et al., (2013), where the

largest value was the EHP of 0.89W/kg. Both studies used the same model (PiG), however the controls in Levinger's study walked around 0.2m/s slower than this study's controls which may contribute to this result. Patient intra-implant changes generally exhibited increases in CHP (with no change for UCR group) and reductions in EHP (no change for the UC group) by at least 0.1W/kg. This is larger than the intra-group differences in Levinger's study which were no more than 0.05W/kg.

Three sagittal knee moment peaks were analysed, flexion maxima during early (KFM1) and late stance (KFM2), and a mid-stance extension maximum (KEM). The first flexion moment peak occurs when the limb reaches a peak flexion angle during stance and the extension maximum occurs when the MKFS is seen. KFM2 is usually smaller than KFM1 and occurs when the contralateral limb makes initial contact. Of the participant groups, only the controls, pre-CRDD, and pre-UCR groups showed larger KFM1 to KFM2. This is similar to the controls in one study reporting both flexion moment peaks (Smith et al., 2004), and the same studies' postoperative group also showed a greater KFM1 peak than KFM2. This was not maintained at preoperative in Smith et al.,'s paper. In the current study controls displayed the largest flexion and extension moments, postoperative patients mostly gave decreases in flexion moment and increases in extension moment from preoperative. Increased postoperative extension moment was also seen in the literature (Smith et al., 2004; Urwin et al., 2014; Xu et al., 2010). However flexion moment trends in the same studies followed extension moment trends where postoperative moments were larger than preoperative, or unchanged between preoperative and six month post-surgery (Worsley et al., 2013). Extension moments from papers reporting TKA only findings were comparable to this study (Braito et al., 2016; Hyodo et al., 2020; Saari et al., 2005; Smith et al., 2006, 2004; Urwin et al., 2014) except for one paper which showed near double the extension moment seen here (Xu et al., 2010). Flexion moments were less consistent, with some studies presenting larger TKA moments (Hyodo et al., 2020; Nishizawa et al., 2020; Saari et al., 2005; Urwin et al., 2014) and others presenting comparable peak flexion moments to this study (Meinders et al., 2019; Smith et al., 2006, 2004; Sun et al., 2020; Xu et al., 2010).

A primarily greater flexion moment is indicative of quadriceps overuse gait, and conversely, a mostly extension moment implies quadriceps avoidance. As discrete values were analysed rather than whole waveforms, comparing KFM1, KEM, and KFM2 gives some estimation to quadriceps function as seen in Figure 6-10. KFM2 was similar amongst groups and gave low variability, so the difference between patient KFM1 and KEM to controls better determined quadriceps function. At preoperative, KFM1 reached control levels and ranged to being half of control values with no distinction between bearing types. Peak KEM was near negligible for all preoperative patients compared to 0.26N.m/kg in controls confirming quadriceps overuse patterns as expected in OA groups where flexion contracture or stiff knee gait is frequently exhibited. Postoperative patients' KFM1 was distinctly lower than controls (particularly the FB groups), and KEM was closer to controls. Instances of positive flexion and extension moments similar to control traces implies that quadriceps avoidance pattern was diminished in the CRDD and UCR groups. The UC group showed no positive KFM1 and perhaps showed a quadriceps avoidance gait. However, it should be noted that these discrete values are of net moments, of which the role of individual muscles cannot be derived from as instances of co-contraction are neglected.

Frontal knee adduction moments (KAM) are considered a surrogate measure of loading across the medial compartment of the knee. A KAM is seen during stance only to stabilise the joint during contralateral swing phase and commonly shows two peaks during early (KAM1) and late stance (KAM3), along with a midstance adduction minimum (KAM2). Increased KAM and less distinction between peaks are linked to varus malalignments and medial OA progression (Foroughi et al., 2009; Rutherford et al., 2008; Teichtahl et al., 2003). Controls in this study however presented greater peak KAM1 than all OA groups, which was also seen in one study (Urwin et al., 2014). Other studies described lower control KAM to medial OA patients and higher than lateral OA patients (Turcot et al., 2013; Weidow et al., 2006). Controls showing lower or comparable peak KAM to OA patients were reported more frequently (Gök et al., 2002; Ko et al., 2011; Lewek et al., 2004; Worsley et al., 2013). In the current study,

smaller postoperative reductions in KAM1 for the FB group was seen, whereas a larger decrease was given by the UCR group. All post-surgery findings were greater than most published results (Abdel et al., 2014; Alnahdi et al., 2011; Braito et al., 2016; Komnik et al., 2016; Nishizawa et al., 2020; Smith et al., 2006; Sun et al., 2020; Urwin et al., 2014) except for one study where moments were equivalent (Hyodo et al., 2020). Discrepancies may be a result of the various time periods of postoperative sessions, and some studies compared surgical technique instead of implant type. Only three papers reported both pre- and postoperative KAM and intra-implant changes were around 0.1N.m/kg(Braito et al., 2016; Urwin et al., 2014; Worsley et al., 2013). This is slightly larger compared to this study where postoperative reductions of 0.07N.m/kg, 0.01N.m/kg, and 0.012N.m/kg were observed in the CRDD, UC and UCR groups respectively.

No papers described both KAM2 and KAM3, and in the current study significance was seen for KAM2 only. All patients showed postoperative reductions in KAM2 which was a significant finding for just the CRDD group, this intra-implant change may be considered clinically relevant for the UCR group. A decline in KAM2 by itself would increase the distinction between KAM1 and KAM3 peaks, however reductions were seen for all implants for KAM1 and KAM3. Taking the difference between peaks, both FB groups showed greater peak distinction at post-surgery, by 0.07N.m/kg (CRDD) and 0.09N.m/kg (UC) and are considered to have improved function (although not to control levels). The MB UCR group presents a worsened peak distinction from preoperative by 0.03N.m/kg at post-surgery. Finally the knee abduction moment in this study showed no significance between groups and was near negligible (<=0.05N.m/kg), agreeing that the frontal adduction moment is primarily positive during stance (Abdel et al., 2014; Hyodo et al., 2020). Some studies presented knee abduction moments larger than 0.1N.m/kg in one or more groups (Saari et al., 2005; Urwin et al., 2014; Weidow et al., 2006) which may imply a slight valgus alignment post-TKA.

The peak sagittal and frontal moments in TKA groups of different bearing types was reported by one study only (Urwin et al., 2014),. Both bearing types showed postoperative increases in flexion and extension moments in Urwin's study which was not seen here. Their MB group exhibited greater intra-implant increases in sagittal moments than the FB group, however it is not known if this was significant as statistical analysis was not carried out between operative states. Most pre- and postoperative sagittal moments in this study were significantly lower than the controls in Urwin's study which was partly seen here where postoperative groups showed significance to controls. More in line with this study, frontal patient moments in Urwin's study showed reductions from preoperative, both bearing types decreased by the same amount, and showed no significant inter-bearing differences. Significance in frontal moments in Urwin's study was between postoperative groups and controls, and the mean differences were alike to the findings of this study which were not found to be significant here.

Knee power is traditionally multi-phasic as the musculature generates and absorbs power to control weight acceptance, extend the limb and maintain stability during gait. The largest concentric (CKP) and eccentric (EKP) powers were exported as these are expected to show the largest differences, and where these occur in the gait cycle are assumed to be in the same location. Higher peak eccentric powers than concentric were seen in this study and in the literature (Levinger et al., 2013). However eccentric powers were between three and six times larger than concentric powers here, whereas Levinger presented larger differences in the magnitude of six and fourteen times between the peak values. This disparity is attributed to the larger CKP seen in this study. Peak patient CKP in this study was ranged between 0.12-0.23W/kg, this is alike to published findings of postoperative patients where the same powers were seen (Abdel et al., 2014). Controls showed significantly higher CKP and EKP to all patient groups in this study, and inter-implant comparisons showed no relevant or significant differences between groups. From preoperative, the post-UC group showed small increases in EKP and CKP, and both of the other implants

presented no change in CKP or small decreases in EKP so all implants may be considered to have equivalent postoperative knee power capabilities.

Ankle powers and sagittal moments reflect increasing dorsiflexion post-initial contact, to plantarflexion at toe off. During loading response energy is absorbed and a small plantarflexion moment is seen. As the shank progresses over the foot an increasing dorsiflexion moment (DFM) occurs, and energy is absorbed until late stance where concentric contraction of the plantarflexors occurs to push the foot off the ground. This is the key source of energy generation during gait and propels the limb forwards and upwards into swing (van der Krogt et al., 2012). Controls showed greatest ankle kinetics showing patients were not yet to control levels. Patients showed negligible differences in external plantarflexion moments and no study reporting this parameter was found. Control DFM was significantly larger than all preoperative patients. All patients showed a post-surgery increase in DFM which was also seen in one study (Braito et al., 2016). Despite this increase, the post-UCR group still showed a significantly lower DFM to controls and to the post-CRDD group (the only significant inter-implant difference seen for this parameter). The intra-implant improvement in DFM by the CRDD group was also significant, and the change from preoperative was similar to the UC group, which was not statistically relevant but may be considered clinically relevant.

Control and patient peak eccentric ankle powers (EAP) were comparable to published findings (Levinger et al., 2013) and peak concentric powers (CAP) were similar to one study (Abdel et al., 2014) and lower than the findings by Levinger et al., (2013). Walking speed may be a factor for the patient groups, however Levinger's control group walked slower and showed larger peak CAP to this study's control group so the reason for the inconsistency is not clear. All patients showed postoperative increases in ankle powers much like in Levinger's study, with the FB groups presenting the greatest increase in CAP of ~0.5W/kg. The MB UCR group showed smaller increases of 0.14W/kg more power than at pre-surgery. The post-UCR group also showed the smallest peak CAP of all groups, and the post-CRDD group showed the highest. Intra-

implant increases in maximum EAP ranged between 0.05-0.2W/kg and results were equivalent between all participant groups.

Much like for kinematics, spatiotemporal parameters also have profound effects on kinetic data. It is well described in the literature that walking with a faster walking speed increases the magnitude of lower limb moments and powers (Chen et al., 1997; Landry et al., 2007; Stoquart et al., 2008). At preoperative this was indeed the case in the CRDD group which showed faster walking speeds to the pre-UC and UCR groups by 0.1 and 0.2m/s respectively. Greatest sagittal hip, knee and ankle moments were seen in this group compared to the others. The pre-UCR group walked the slowest yet showed the largest power values at preoperative. In addition, the pre-UC group which walked at the middle fastest speeds presented the lowest moments and powers. At postoperative the FB implants walked at the same speed with the UCR group walking 0.1m/s slower. Sagittal hip, knee, and ankle moments were greatest in the post-FB groups, and the UCR group showed the greatest KAM magnitudes. Larger hip and ankle concentric powers were seen in the post-FB groups, with the post-UCR group showing greater eccentric powers for all joints. This could be a compensatory mechanism that remained from preoperative where the UCR group also showed the greatest eccentric joint powers.

7.3.3. Summary

Control and patient kinetics were discussed in this section during overground level walking only. Sagittal hip, knee and ankle moments were analysed, along with knee adduction moments and all joint powers. Other parameters such as hip adduction moment and knee rotation moment are relevant during gait, however, these were omitted from the results due to lack of significance and clinical relevance.

Overall peak discrete values were exported, however when these occur in the gait cycle would have been further beneficial to understanding motion. For instance, generative power peaks during early stance may reveal ineffective traits whereby the body is accelerated upwards instead of forward propulsion should the generative power occur at the end of stance. Even more insightful would have been waveform

analysis of unnormalized data, however this was beyond the scope of this study, and a less demanding ADL such as level walking may not have revealed much more than presented. Whilst significant differences were described, in the absence of literature thresholds, similarly large insignificant differences were also described as these were considered clinically relevant. As clinically relevant differences altered considerably based on the joint and parameter, clinically relevant differences for one parameter would not necessarily be relevant another.

Initially, healthy kinetics were discussed and compared to the literature. Since kinetics may be normalised by different properties, studies with any adult group were included to provide a larger dataset to refer to. Most control-only kinetic findings agreed well to published findings except for peak hip extension moment, KAM1 and eccentric hip power where this study's controls showed smaller values. Discrepancies are likely due to differences in participant group ages, walking modality, or data capture processes. Comparing patient data to the literature was also challenging, in addition to the various normalisation methods that limited the number of available papers to compare with, generally less studies are available that analyse the kinetics of OA and TKA groups. Fewer papers also investigate implant bearing function.

Multiple knee moment peaks were exported to gain an idea of moments over the gait cycle, which is a key predictor of loading and alignment. In the literature, data around adduction moment peaks were particularly lacking for comparison. Whilst control levels were frequently not met, postoperative improvements were noted. On the whole, controls exhibited the largest flexion moments and joint powers, and OA patients gave the highest KAM2 and KAM3 and smaller KEM associated with flexion contracture. These improved with TKA, particularly the FB groups improved KAM peak distinction more than the UCR group. Other key inter-implant trends were seen for joint powers where the FB groups showed greater improvements in generative ankle and hip powers compared to the MB group however these were not statistically relevant.

7.4. Questionnaire Answers

Participant responses to standardised questionnaires were presented in Chapter 6.6. These recorded difficulty, tiredness, and pain levels for all activities as well as comments on the similarity of the laboratory activity to participants daily life. Generally, there was good agreement that the tasks replicated the ADLs which provided reassurance that the tasks were not overly challenging and recorded natural motion. Since these questions regarded this study specifically and not, say, a clinical score, comparison to the literature is not possible, however these responses give an insight into patient satisfaction which is a key outcome measure of a TKA procedure. Although direct satisfaction was not recorded, it may be inferred from the levels of pain, tiredness, or difficulty experienced.

TKA is generally considered to be a successful procedure as pain is successfully reduced. It is reported that between 82-89% of patients are satisfied, leaving as much as 20% of patients dissatisfied with their procedure (Baker et al., 2007; Bourne et al., 2010). One third of patients from a large sample of 8,050 also reported limited postoperative functional improvement (Franklin et al., 2008) so a link between function and satisfaction may well be evident. Other sources of dissatisfaction stem from unmet expectations and may include residual pain, postoperative complications, comorbidities, or even health care received. Links between the change in passive ROM were also found to increase satisfaction, as opposed to absolute values indicating the importance of intra-patient improvement (Dhurve et al., 2017).

Overall, controls reported no difficulty, pain, nor tiredness for any of the ADLs and pre-surgery OA groups experienced more of these than at postoperative. Responses to level walking showed the least severe answers, with patients generally finding the stairs tasks more difficult (descent more so than ascent) and painful, and the sloped walking task to be more tiring. Although no significance between participant groups were seen for the tiredness answers, some moderate and extreme tiredness was reported during sloped walking. Since this ADL was carried out last, and uphill and downhill walking was recorded consecutively with no break between changing slope direction this could contribute to the increased tiredness. Sloped walking was also

the only activity where the sample sizes between operative states were not equal, and so intra-implant comparisons are also limited.

Only difficulty levels for the stair task were divided between ascent and descent, ideally to do the same for pain and tiredness, and to divide between incline and decline walking would have been beneficial to understanding patient satisfaction. It was also not specified whether the difficulty from carrying out the sloped walking task was due to the physical steepness of the incline, or due to unfamiliarity with walking on a self-paced treadmill. Anecdotally speaking, participants seemed to walk more naturally once the treadmill was set at an incline or decline, and any difficulty reported was associated to the functionally demanding nature of the task.

It was frequently seen that the postoperative FB groups' responses to task difficulty (Figure 6-12, Figure 6-16, Figure 6-17 and Figure 6-23) and pain (Figure 6-13, Figure 6-18 and Figure 6-24) were significantly improved from preoperative and similar to controls. Much like the FB groups, pre-UCR group responses were significantly worse than controls, however fewer post-surgery improvements were seen, either as a significant improvement from preoperative, or statistical similarity to controls. The MB group also generally presented the lowest hip and knee flexion ROM and largest MKFS and ankle DF ROM, which implies increased ankle mobility to compensate for proximal joint action. During level walking the MB group also exhibited the largest KAM2 minimum implying loading is not to the same point as FB groups. Sagittal moments and generative joint powers were also generally lower in the MB group which may be a result of high postoperative pain limiting function.

The post-CRDD group presented the largest peak and range of knee flexion (and greatest improvement from preoperative) as well as largest concentric ankle powers during level walking which suggests function and muscle strength is recovered. Although no postoperative instances of 100% "No difficulty" or "No pain" was reported in the post-CRDD group, the sample size is more than double the UC group increasing the propensity to not show 100% satisfaction. Whilst all participant's questionnaire responses were collated, for the stairs task particularly, trials were

excluded from biomechanical analysis if a step-by-step strategy was seen. This makes comparison between questionnaire answers and stair kinematics less accurate as the average values are taken from different participants.

As well as difficulty, pain, and tiredness, bannister use may also indicate functional improvement during the stair task. Bannisters provide stability and lessen the load on the lower limb through increased upper body loading and so, increased bannister dependency may be linked to lower knee function and confidence. Although controls answered they do not require a bannister when ascending or descending stairs, when asked about everyday bannister use, some single bannister use was reported, particularly during stair descent. At preoperative more double or single bannister use is evident which shifted mostly to single or no bannister use post-surgery. Although individual changes were not presented, it is likely pre- surgery habits influence post-surgery patterns. It is expected that participants have learned behaviour from years of disability and assessing function at one-year post-operative may be too soon to see a reversal of these habits. Further rehabilitation would improve outcomes to control levels if necessary (Bandholm et al., 2018).

Postoperative improvements were seen across all groups overall. Stair descent looked to be the most difficult activity, and two instances of worsened postoperative answers were seen where “Impossible” was answered in the post-UC and UCR groups. All groups consistently showed increases in the proportion of participants answering “None” to difficulty, pain, tiredness, and bannister use so relative improvements were seen. Linking biomechanical findings showed patient perception of function is somewhat accurate to actual function despite the subjective nature of questionnaire answering. This is in agreement with one study where improved peak flexion was strongly associated with patient satisfaction (Bonnefoy-Mazure et al., 2020). The post-UCR group who possibly showed less ability, also reported greater difficulty and pain. Although the FB groups showed better questionnaire responses, the large discrepancy in sample sizes and exclusion of participants’ biomechanical data who walked with a step-by-step strategy limits interpretation.

7.5. Study Limitations and Future Work

The present study has a number of limitations, some of which may be used to form the basis for future work or are to be considered when interpreting results. Limitations arose either from the study protocol, biomechanical model, or the intrinsic demographics of the participants. Whilst steps were taken to reduce the effects of these it was not possible to account for them completely.

As stated, the Vicon motion analysis system has been established as the "gold standard" for motion capture (Deltombe et al., 2017; Desloovere et al., 2010; Kruk and Reijne, 2018; Müller et al., 2017). As repeated gait measurements often shows differences in findings, it may be assumed some error is present. This error relates to reliability - the extent to which repeated measurements are consistent, or free from variation. Lower error gives confidence to results and suggests any change in results before and after an intervention are real. High error could lead to an over-interpretation of results where a patient could be seen to be highly improved or worsened after an intervention where this was not the case.

One form of variation may arise from the inherent variation within a participant. Whilst this occurs naturally and cannot be minimised, it gives a baseline and differences arising from an intervention should be reflected. Together with the markers and subject anthropometrics, estimations for joint centres are made which are used to calculate anatomical reference systems. And a further source of error is soft tissue motion between a skin-surface marker relative to the underlying bone which may cause a soft tissue artefact (STA) affecting the estimation of skeletal kinematics. Biomechanical models consider segments to be rigid bodies, which STA does not compliment. The degree of STA can differ depending on the task undertaken and filtering techniques to remove STA may cause a loss of real information. Overall, STA is regarded as one of the most critical sources of intrinsic error in human movement analysis (Leardini et al., 2005).

Errors may also arise from the system due to an inaccurate calibration or software errors. Such errors are extrinsic and often simple to rectify. System calibration establishes relationships between the test volume to the cameras. Dynamic calibration involves moving a wand with markers on it at a known, fixed distance apart within the test volume and provides the precise relationship between each camera. A static calibration gives the orientation of the cameras and the room. If the cameras move post-calibration, then data will not be accurate. To reduce systematic errors standard protocols were used when completing data capture.

Another form of extrinsic variability is inter-observer variability that arises from the mixed skill of assessors, which often can only be improved with experience (McGinley et al., 2009). Intra-observer variation is exhibited when an observer records data from the same person more than once. This variation is inevitable since placing markers in precisely the same position as previously is unlikely, especially over areas where anatomical landmarks are difficult to palpate (Peters et al., 2009). In this study to mitigate the effects of inexperience particularly during initial preoperative sessions, a pilot study under the supervision of an experienced motion analyst was carried out. Markers were thoroughly palpated prior to placement and recalibration occurred should markers become detached during the session.

There is high scope for variability when carrying out motion analysis. To ensure the data produced is valid, it is crucial to record anthropometric data and affix markers correctly. Similar to how STA results in marker movement such that it no longer represents the bony landmark it was placed on, incorrect marker placement is a large extrinsic cause of error during motion capture (Kadaba et al., 1989; Schwartz et al., 2004). This error is caused by three main factors: the fact that palpable anatomical landmarks are surfaces (sometimes large and irregular) rather than points, soft tissue covering the landmarks are of variable thickness and composition, and identification of landmarks depends on the palpation method used (Della Croce et al., 2005). Both STA and marker placement error have huge effects on kinematic data as the anatomical axes calculated from these marker positions are affected and shifted.

As mentioned, the Plug in Gait (PiG) model is a variant of the conventional gait model (CGM), the most extensively validated model in current use (Baker et al., 2018). Despite the work done validating the model, it is susceptible to kinematic cross talk error as seen in this thesis. However, cross talk presents minimal effect on flexion angles and so future work to improve kinematic data in all planes will improve acceptability among researchers and clinicians. In this study it is unlikely that the non-sagittal kinematics are accurate whilst small marker or KAD misplacement causes high errors. However, it is expected that more experienced assessors would see less error (Davis et al., 2000; de Vet et al., 2003; McGinley et al., 2009). Also, the use of medial markers during calibration could also be explored. Kinetics from PiG are less affected by marker misplacement since kinetics are calculated using inverse dynamics based on segment accelerations and inertial parameters, although it is noted the axis alignment must be accurate so as not to conflate out of plane motions incorrectly.

An alternative model to PiG may have been considered, particularly a cluster-based model (CAST) which is becoming more widely used. By using clusters to represent and track a lower limb segment relative to anatomical landmarks defined during a calibration this considerably improves data quality and reduces processing time. Studies have found improvements in non-sagittal motion compared to PiG and another advantage to CAST is theoretically the clusters may be placed in areas less prone to soft tissue artefact (STA) to give more accurate kinematics. Since the project was due to be completed at a second research centre and PiG was the model of choice there it was decided by the project leads to proceed with PiG and the KAD to improve the reliability of non-sagittal findings. In addition, upper body outputs were required for the second doctoral researcher's work which this model provides.

As opposed to recording a static calibration and using regression equations to calculate joint centres, utilising functional techniques may improve joint centre determination (Ehrig et al., 2006). By moving segments in a predictable way, participant specific joint centres and axes can be calculated, for instance the centre of thigh circumduction will give the hip joint centre. Functional methods have been

shown to be precise and robust (Kornaropoulos et al., 2010; Kratzenstein et al., 2012). However functional calibration techniques relies participants to have a healthy ROM and so may not be suited for those with mobility difficulties (Sangeux et al., 2011). Where functional measures are not appropriate, utilising medial malleolus and femoral condyle calibration markers may improve the accuracy of joint centre calculation. This reduces errors associated with thigh and shank marker placement required to define the flexion plane that gives the cross talk phenomenon (Manal et al., 2002; Schache et al., 2006). Stief et al., (2013) compared PiG to an advanced protocol using medial markers (MA). For the MA, no alignment devices or anthropometric measurements were necessary, and the authors found lower error rates using MA than PiG in non-sagittal angles and moments. This suggests the MA model has improved accuracy of knee axis alignment than PiG alone which could have been utilised in this study.

For all tasks, the identification of the beginning (HS), toe offs (TO), and end of the gait cycle differed slightly between activities depending on the data available. For overground level walking the Nexus software able to detect gait events using the force plates and auto correlate across to gait cycles off the force plates. For sloped walking the HS and TO detection algorithms described by (Zeni et al., 2008) were used. For stair ascent, algorithms based on the toe marker acceleration (HS), or maximum vertical difference between toe and pelvic markers were used (TO). For descent, HS was taken as the minima of vertical velocity of the whole-body CoM and TO was taken where contralateral knee flexion was at its peak. No suitable papers provided robust stair gait cycle identification in an OA/TKR group, and such these four methods were employed. On occasion the algorithms would display additional gait cycle instances which required to be deleted and at the same time the correct position of the gait event was manually verified. To further confirm correct gait event detection compound plots were plotted of knee flexion angle and moment (where available) of both limbs for each participant and activity as given in Appendix 9.4.1. By visually inspecting these graphs the final quality of the data were assessed for any gaps in marker data, or whether gait cycles were out of phase. Any trials which looked

incorrect were double checked and corrected and reexported if necessary. Whilst this process was quite time-consuming, it was deemed necessary when using the PiG model in a less mobile group.

Whilst three-dimensional motion capture gives a net calculation for joint moments, these give estimations for internal joint function, particularly muscle function. This has limits in that any co-contraction or muscle spasticity may be disguised. Whilst utilising EMG gives direct muscle activity during the ADLs, EMGs were not used in this study to reduce data processing time and could be a consideration for future work. As well as muscle activity during the ADLs, maximal muscle strength testing could give an idea of postoperative improvement, especially during loading which requires greater strength. Strength testing was not carried out to reduce participant burden but also could be a future area of research.

Unfortunately, the kinetic data for the stairs and incline walking tasks were not suitable for analysis. Limitation in the stair tasks were the low incidence of patients striking the instrumented step with their operated limb and bannister use may have affected findings (Reeves et al., 2008). By not directing participants as natural motion as possible was encouraged, however adjustments to starting position could increase the chance of the affected limb striking the instrumented step. Using stairs with two instrumented steps would also have been beneficial and eliminates the need to alter starting position. Towards the end of the task, it could have been possible to ask the patient to not use the bannister if able, however this would increase the risk of additional pain and discomfort to participants. For incline walking, two main factors affected kinetic data recording, the main issue being that a drift in force plate readings was seen when the platform was set at an incline. These inertial effects of the platform required considerable work to rectify and considering the force readings were not synchronised with Vicon Nexus at the time, it would not have been possible to calculate kinetics even if there were no inertial effects.

Hardware issues in the Motek CAREN system were also seen, particularly for one-year postoperative patient sessions where six patients' data were not able to be

recorded. At this time internal safety precautions would cause the treadmill to stop and return the platform to neutral and errors were shown describing that the dual treadmill belts were translating to unacceptable levels. The CAREN system was due for an annual maintenance around this period and since it was not possible to be sure when the error and the safety mechanism would be activated it was decided to stop recording once the error was seen. This meant that the number of available datasets were even lower for these activities, further reducing statistical power.

Some participants' stair task data were also excluded from analysis as they navigated the stairs with a step-by-step (SBS) strategy rather than step-over-step (SOS). A SBS strategy presents a considerably larger stance period in the affected limb, as the contralateral leading limb undergoes swing then stance before the ipsilateral limb. The leading limb is responsible for forward COM translation, and the trailing limb completes the action to continue moving along. This is in contrast to a SOS strategy where both limbs undergo periods of stance and swing to progress the body onwards. As the gait cycle between SOS and SBS walkers are dissimilar it was not possible to include both when combining and averaging implant data. By excluding SBS trials from analysis an inherently lower performing group are not accounted for who stand to gain the most from a TKA procedure. It was observed that most preoperative SBS walkers improved their strategy at postoperative for other ADLs in line with two studies by the same authors as this study (Komaris et al., 2020, 2018) and to quantify the biomechanical differences between SBS and SOS walkers could be a further area of research.

In this study, level walking was recorded whilst walking overground or on a treadmill as a precursor to the sloped walking tasks. Treadmill walking has benefits in that it is possible to collect data from many more gait cycles than overground walking. As the goal of rehabilitation is to walk overground safely, comparing data from both modalities can gauge similarity between them. As participants were to walk on an incline using a treadmill, confirming level treadmill biomechanics was correct gives confidence to sloped walking data. Compared to level walking, a slower walking

speed, shorter stride length, and higher cadence is generally seen when treadmill walking (Alton et al., 1998; Wearing et al., 2013). Differences in joint kinematics or kinetics were generally smaller than measurement error in healthy subjects (Lee and Hidler, 2008; Riley et al., 2007) and generally agree both walking modalities are acceptably similar, particularly following a familiarisation period (Matsas et al., 2000). In TKA patients, significant differences have been reported between walking modalities (Guan et al., 2017) which future studies could verify.

As well as conventional treadmill and overground walking differences, another factor affecting data is that a self-paced treadmill was used in the Motek CAREN system. Allowing the treadmill speed to adjust accordingly, particularly at an incline, theoretically produces more natural kinematics as the walking speed is automatically adjusted to comfortable levels. One study found biomechanical parameters were not affected by the treadmill mode, although a healthy participant group was used (Sloot et al., 2014). Whilst a large familiarisation period of up to fifteen minutes in an older group is recommended to maintain natural walking patterns (Wass et al., 2005), in a less mobile group a long familiarisation period would certainly accelerate fatigue. This certainly raises questions as to whether this system is helpful to assess those with a lower limb disability. To encourage natural walking participants initially walked at a slow fixed pace which gradually increased to a natural speed. Self-paced mode was then initiated and ten second recordings were taken once walking speed looked stable. For reassurance, a researcher was alongside each participant to also remind participants to walk forwards which helped with data recording. By utilising the self-paced mode any overground walking could theoretically be replicated on the CAREN system. What appeared to be the largest barrier to walking naturally in this study was that this was potentially the first time some participants had ever walked on a treadmill. The considerable amount of hardware may be overwhelming to elderly participants and the system may feel more natural to others who may have had more prior experience on a treadmill and isn't uncomfortable around unfamiliar equipment. To be able to record multiple gait cycles under controlled conditions is certainly beneficial in general.

Regarding the parameters selected for analysis, key parameters from other studies analysing TKA gait were referred to and parameters where significant differences were seen were reported here. Additional low significance parameters that were analysed and not reported were the peak and range of hip adduction angle, hip adduction moment, knee rotational moment, and sagittal pelvis, hip and ankle angles at heel strike and toe off, and foot progression angle. Whilst little to no significance was seen, some clinical relevance may have been present which could be explored further. Other parameters mentioned in the literature and possible areas for future analysis were knee velocity at toe off, trunk kinematics, moment impulse data, and knee flexion angle at maximum flexion moment. Active and passive knee flexion ROM (commonly associated with clinical scores), or other gait scores and deviation indices were also not exported which may provide an insight into functional implant performance. Lastly links between non-gait related data and biomechanical parameters, for instance alignment from radiographs, could be investigated by correlating alignment and function.

Although ADLs carried out in the laboratories have benefits in that conditions are the same for every participant, these may not reflect the same ADL completed in a natural setting. Laboratories may feel unnatural for the participant, for instance when recording gait, depending on the size of the room it may be possible to just record a few (~ five) complete gait cycles. The participant is unlikely to have fallen into their natural stride by the time they reach the capture volume which would not accurately reflect their gait (Brodie et al., 2016). In addition, the tasks recorded do not reflect all activities experienced by an individual. Whilst car ingress/egress and sit to stand/sit to walk tasks in this same population group were analysed by another researcher, other ADLs that could be analysed to give an idea of function include walking around a corner (where the operated and non-operated limb as used as pivots), single leg balance, lunging and getting in and out of a bath. Out of laboratory activity monitoring would allow motion to be as realistic, although less information would be available. Finally, analysing the motion of the contralateral non-operated limb could indicate any compensatory mechanisms related to lower TKA function. Although,

contralateral limb biomechanics was exported in this study, as the extent of any disability was not known no analysis was carried out.

To gather data points for analysis multiple gait traces for a single activity (per participant) were averaged, and discrete parameter values from the average activity trace was found. This was combined with other participants receiving the same implant to give an average value per group. The difference of this value to another group (for instance to preoperative) may not accurately reflect the intra-implant change seen. Possibly finding the average of each patient's change from preoperative baseline per implant type may better reveal differences between groups. Whilst comparison to controls would not be possible, reporting mean differences as seen here could be possible. A limitation to analysing mean discrete values is that only information at a single point is presented and may ignore other discrepancies between whole waveforms of different groups. Whilst this was mitigated by picking multiple points for analysis, particularly for the knee flexion angle and knee moments, this was only for a small number of parameters. Waveform analysis such as PCA is increasingly used to compare gait traces, as it is possible to analyse un-normalised data PCA also includes temporal analysis lost with usual gait analysis (Hatfield et al., 2011; Young-Shand et al., 2020). Furthermore, waveform analysis would have been more effective than analysing discrete data points at clarifying whether the mobile bearing knee provides enhanced rotational properties. Internal/external rotation would be evident between 30-90^{o} of flexion (Zarins et al., 1983), typically where swing occurs during gait, and the collateral ligaments are lax.

There are clear concerns around the small sample sizes seen underpowering statistical analysis in this study, despite maximum effort given to recruiting and analysing patients within the timeframe. Smaller sample sizes are susceptible to the effects of outliers, showing a large variance which gives less likelihood of observing a significant difference, and larger chance of type two errors. Repeated tests across different participant groups also gives a considerably larger chance of a type one error. As some p values were multiplied by six due to the Bonferroni correction during

ANOVA testing, to reduce the chance of seeing a type one error could be to lower the alpha level. However, 0.05 is a widely accepted alpha value and keeping a conservative alpha was considered acceptable albeit likely over-conservative. Despite the conservative statistics, the low and unequal number of participants per implant group (and differences in sample size per repeated measure) is not representative of a whole population, and further work with more participants is required to confirm the conclusions highlighted here.

Other factors that may influence findings relate to participant demographics. Whilst controls were aged matched, they were not matched by BMI and control BMI was significantly less than patients'. A higher BMI associated with obesity is a risk factor for OA, and so it may be that asymptomatic controls would inherently be expected to have a lower BMI. Studies comparing gait in overweight and lighter participants showed some altered biomechanics between groups which may have influenced findings on top of any mobility difficulties in patients. Obese participants were found to walk with reduced walking speeds (likely to lessen joint loading) and greater ground reaction forces. Since similar frontal knee moment magnitudes (Lai et al., 2008; Runhaar et al., 2011) the compensatory mechanisms may successfully reduce knee loading however may adversely affect other joints. Certainly, obese patients show larger hip adduction and ankle eversion which could eventually alter the loading axis to about accelerated wear and begin the cycle of osteoarthritis progression.

Another key demographic was the participant sexes, where the CRDD group showed a significantly greater difference in sex split than other implants with eleven males and one female. The other groups had more females, with the greatest ratio of one male to four females (UC group). A number of studies have described sex differences in various participant groups during gait. Female OA patients purportedly present lower knee adduction moments than men (Sims et al., 2009) and whilst reductions in male adduction moments were seen post-TKA, females did not show the same (Paterson et al., 2018). Female TKA patients showed small or no differences to female controls, than male patients who showed more differences to male controls

(McClelland et al., 2018; Paterson et al., 2018). By maintaining mixed-sex cohorts female biomechanical outcome may be underestimated (and vice versa for males) and future work could assess outcomes from males and females independently.

As well as bearing differences between the implants, whether the PCL was retained or sacrificed is a considerable factor that may have affected function interpretation. The CRDD implant is a PCL retaining implant whereas the UC and UCR implants are PCL sacrificing. The CRDD and UC implants are both fixed bearing and differences in function between these implants could be due to the PCL. Some meta-analyses present significantly improved knee flexion ROM in the PCL-sacrificing group (Bercik et al., 2013; Jacobs et al., 2005; Jiang et al., 2016), however findings were not consistent amongst studies. To account for bearing and PCL differences would require much larger sample sizes in this study however, which was not possible within the timeframe.

Clinically relevant kinematic thresholds described differences that may not have flagged as statistically significant. Combined with kinematics, kinetic and spatiotemporal parameters all provide a key insight into joint function, and future research could also determine clinically significant thresholds for kinetics and STPs. Based on the significant mean differences, and the chain-link nature of inverse dynamics, clinically relevant kinetic thresholds could vary considerably based on the joint, dimension analysed and perhaps even the activity. Finally, one-year postoperative is generally considered an acceptable recovery period, where the replaced joint is alleged to be healed. Longer-term studies provide an insight into loosening and wear and general survivorship brought about by stresses at the point of bone fixation, or lack of congruency to the polyethylene insert (Poirier et al., 2015) which could be researched further. Regarding gait biomechanics, finding long-term function will give further confidence as to whether functional advantages exist between implant types, or even if differences between implants narrow over time.

8. Conclusions

This study used quantitative methods to provide biomechanical information during various ADLs of TKA patients and healthy age-matched controls to determine whether functional advantages between different B.Braun Columbus® knee implants exist. The aim was to see whether a mobile bearing implant showed benefits over two fixed-bearing knees with a high and low-congruency polyethylene insert. Comparisons from preoperative, between implants, and to controls was carried out and statistical and clinical relevance was analysed where possible.

Across all ADLs, patient function improved (although not to control levels) for key spatiotemporal parameters such as walking speed, cadence, and double support period. Kinematic improvements (again not to control levels) were seen for range of hip and knee flexion, and minimum flexion during stance (MKFS). During level walking, patient sagittal moments and all powers were less than control values. Whilst controls showed a greater peak adduction moment, more distinction between adduction moment peaks was also seen compared to patients. Intra- and inter-implant function were mostly similar in terms of significance, the CRDD group showed more significant differences to preoperative, and any significant postoperative inter-implant findings were also to the CRDD group. For kinematics, using a 5^{o} threshold for clinical relevance found that the UC group showed the largest clinically relevant changes from preoperative, and to other implants at postoperative.

Intra-implant changes may better provide an idea of function gain or loss, particularly during more challenging activities. During stair navigation, the UC and UCR groups showed clinically relevant decreases in peak and range of knee flexion suggesting function is limited during challenging activities. The CRDD group showed non-clinically relevant increases from preoperative for this task. Generally, decreases in peak flexion and increases in ROM were exhibited during the remaining tasks implying more extension was realised. From preoperative, the MB group showed greater reductions in knee flexion ROM during stair navigation, and smaller ROM

increases during other activities than the FB implants. In addition, postoperative decreases in MKFS by the MB group were less than other groups and the MB group gave higher MKFS to the other groups indicating some limits to extension.

Non-sagittal kinematics were considerably affected by kinematic cross talk, if assumed to affect participants equally then no theorised rotational advantages were seen in the MB group as they presented mostly non-clinically relevant decreases from preoperative. The CRDD group showed also showed postoperative declines in rotational kinematics across the tasks, to a greater extent than the MB group. The UC group showed postoperative increases in peak and rotational ROM.

During level walking the FB groups presented the greatest walking speeds and sagittal hip and ankle kinetics – the largest contributors to propulsive power. OA groups presented larger adduction minimum between maxima moments and secondary adduction moment peaks to controls and postoperative. Following TKA, the FB groups showed no change in initial KAM peak and decreases for other KAM parameters resulting in greater peak distinction. The MB group showed decreases for all KAM peaks and less peak distinction at postoperative implying loading may have worsened. The MB group also reported more difficulty, pain, and tiredness than the FB groups indicating the MB group is attune to their limitations. The MB group also exhibited stair strategies that were least similar to controls, however these were likely influenced by preoperative strategies and additional rehabilitation is maybe required to improve further.

Whilst the current gold-standard methodology was employed for analysis, limitations to interpretations are evident particularly small and uneven sample sizes which reduced further during more challenging tasks. Using a conservative p value gave confidence that type two errors were less seen. Effects of retaining or resecting the PCL is also a considerable variable to patient function and data interpretation. Utilising clinically relevant thresholds for spatiotemporal and kinetic parameters for all activities would also better determine whether functional advantages exist between implant bearing types.

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Appendices

Appendix 1: Plug in Gait Biomechanical Model

An example of a biomechanical model available for motion capture analysis is the Vicon Plug-in Gait (PiG) model as shown in Figure 0-1 which is a variation of the Conventional Gait Model (CGM), The full body PiG version divides the human body into eleven segments: the head, torso, both arms, pelvis, and both thighs, shanks and feet. These segments are linked together with joints that have three rotational degrees of freedom.

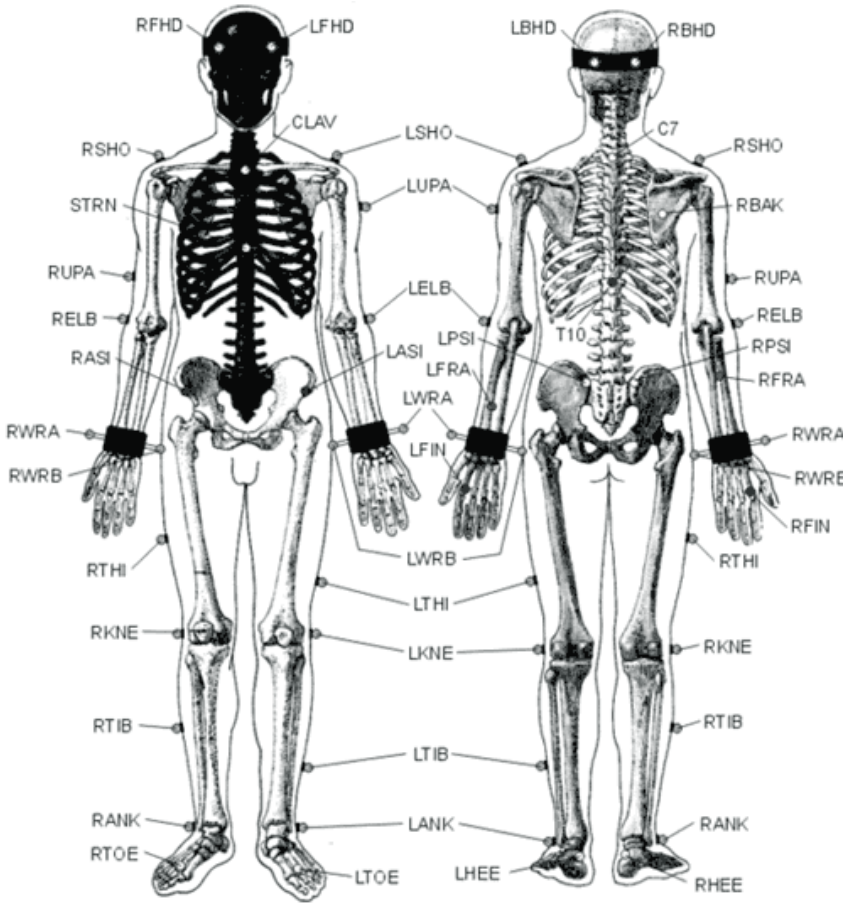


Figure 0-1: The marker positions of the full body PiG model.

This model requires 35 markers to be affixed to various anatomical landmarks at the following locations (Table 0-1):

Table 0-1: The markers used in the full body PiG model and their anatomical location.

Segment	Marker	Location

Head	LFHD	Approximately over left temple
	RFHD	Approximately over right temple
	LBHD	At back of the head, in a horizontal plane of front head markers
	RBHD	At back of the head, in a horizontal plane of front head markers
Torso	C7	Spinious process of the 7th cervical vertebrae
	T10	Spinious process of the 10th thoracic vertebrae
	CLAV	Jugular notch where the clavicles meet the sternum
	STRN	Xiphoid process of the sternum
	RBAK	Placed in the middle of right scapula- for asymmetry
Arm	L/RSHO	On the acromio-clavicular joint on both sides
	L/RELB	On lateral epicondyle of elbow joint
	L/RWRA	On the wrist in line with the thumb
	L/RWRB	On the wrist in line with the little finger
	L/RFIN	Placed on the dorsum of the hand, below the head of second metacarpal
Pelvis	LASI	Over left anterior superior iliac spine
	RASI	Over right anterior superior iliac spine
	LPSI	Over left posterior superior iliac spine
	RPSI	Over right posterior superior iliac spine
Thigh	LTHI	On the proximal third of thigh, along invisible line between the greater trochanter and lateral epicondyle
	RTHI	On the distal third of thigh, along invisible line between the greater trochanter and lateral epicondyle
	LKNE	On lateral epicondyle of left knee
	RKNE	On lateral epicondyle of right knee
Shank	LTIB	On distal third of shank along the invisible line between the lateral epicondyle and lateral malleolus

	RTIB	On proximal third of shank along the invisible line between the lateral epicondyle and lateral malleolus
	LANK	On the lateral malleolus
	RANK	On the lateral malleolus
Foot	LTOE	Over the second metatarsal head, on the midfoot side of the equinus break between forefoot and midfoot
	RTOE	Over the second metatarsal head, on the midfoot side of the equinus break between forefoot and midfoot
	LHEE	On the calcaneous at the same vertical height as the toe marker
	RHEE	On the calcaneous at the same vertical height as the toe marker

To approximate joint centres and segment masses the following subject anthropometrics are also required as given in Table 0-2:

Table 0-2: The subject anthropometrics required for the full body PiG model.

Anthropometric Parameter	Measurements Required
Body mass	Mass in kg of participant
Height	Height in mm of participant
Shoulder offset	Vertical distance in mm of acromio-clavicular joint to centre of glenohumeral joint
Elbow width	Distance in mm between the medial and lateral epicondyles of the humerus
Wrist width	Distance in mm between ulnar and radial styloids
Hand thickness	Distance in mm between dorsal and palmar surfaces of the hand
Interasis distance	Distance in mm between LASI and RASI

ASIS- trochanter distance	Vertical distance in mm in sagittal plane between ASIS and the greater trochanter. Can be calculated from the regression equation: $(0.1288 * leg\ length) - 48.56$
Leg length	Distance in mm between ASIS to medial malleolus
Knee width	Distance in mm between lateral and medial epicondyle
Ankle width	Distance in mm between lateral and medial malleolus

When using PiG, the modelling stage gives four interdependent models: a kinematic lower body, kinematic upper body, and upper and lower kinetic bodies. The kinematic models describe the relationships of the rigid body segments, and the kinetic models apply masses and moments of inertia to the segments. The origin of each segment is defined in the global (laboratory) reference system and each segment then assigned a local embedded coordinate system within which allows its orientation in space to be described with respect to one another.

Any segment is defined by at least three marker points. Generally, the axis system is defined using two directions derived from marker data, the dominant direction is used to define one of the segment axes. A secondary direction subordinate to the first is used to define a plane. A third axis is created which is perpendicular to this plane and a final orthogonal axis perpendicular to the first and third axes completes the right-handed system. By combining marker position with subject anthropometrics, the software produces a customised Plug-in Gait model for the participant, and gives biomechanical data as required. Detail here will be provided about the lower limb kinematic and kinetic models only as these are relevant to this thesis.

1.1.1. Modelled Joint Calculations

For some parameters, joint centres are required for further calculations. A static calibration process generates the virtual positions of some joint centres depending on marker data and subject anthropometrics. Reference here is made to Vicon documentation (Vicon Motion Systems, 2017a).

- Hip Joint Centre

Each hip joint centre (HJC) is defined in the pelvic coordinate system using the pelvis size (inter-ASIS distance) and leg length as scaling factors. A value C is calculated from the mean leg length using the following formula:

$$C = MeanLegLength * 0.115 - 15.3$$

C is then used to calculate the mean offset vectors for the left and right HJCs along with two other variables: aa (half the inter-ASIS distance) and mm (the marker radius):

$$\mathrm{LHJC_x} = \mathrm{C} \times \cos(\vartheta) \times \sin(\beta) - (\mathrm{AsisTrocDist} + \mathrm{mm}) \times \cos(\beta)$$
$$\mathrm{LHJC_y} = -(\mathrm{C} \times \sin(\vartheta) - \mathrm{aa})$$
$$\mathrm{LHJC_z} = -\mathrm{C} \times \cos(\vartheta) \times \cos(\beta) - (\mathrm{AsisTrocDist} + \mathrm{mm}) \times \sin(\beta)$$

Where:

ϑ = 0.5rad

β = 0.314rad

For the right HJC, the Y offset is negated as Y is in the lateral direction for the pelvis embedded system, so the left HJCy is made equal to the right HJCy.

- Knee Joint Centre

The Knee Joint Centre (KJC) ordinarily depends on the thigh and shank marker positions to define the sagittal plane. It is also possible to use a Knee Alignment Device (KAD) to make locating the KJC more accurate. A KAD is a clamp attached around the knee that contains three orthogonal markers as seen in Figure 0-2. This device is placed on the participant during a static calibration to give the plane of the KJC. Of the three orthogonal markers, the most lateral marker is commonly labelled as a KAX marker which ought to be positioned to be in line with the knee flexion axis. Once the static calibration is recorded, the KAD is removed and a marker is placed in the centre of where the lateral KAD pad was for dynamic trials.

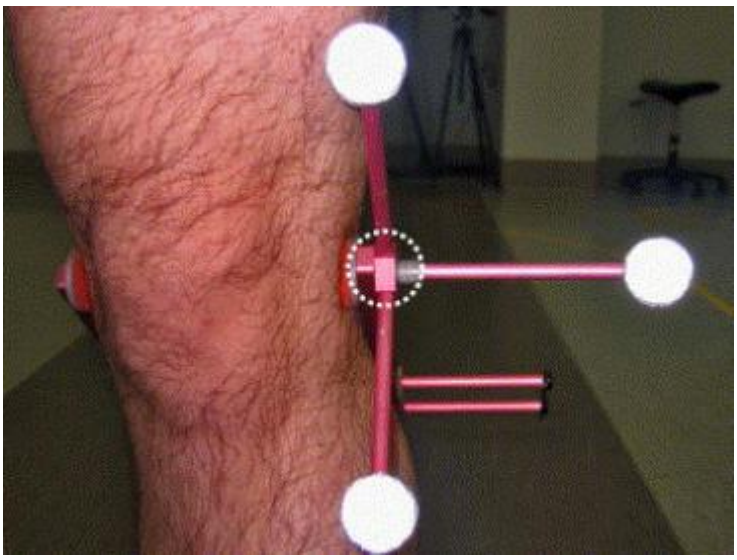


Figure 0-2: The Knee Alignment Device (KAD) as worn around the knee for calibration. The KAX marker is positioned laterally with respect to the other markers. Image from "Knee Alignment Device - Visual3D Wiki Documentation," n.d.

A virtual KNE marker is firstly determined by finding the point equidistant to the orthogonal markers on the KAD. A chord function (Figure 0-3) is then used to give the KJC using the HJC, virtual KNE and KAX marker. This is such that the line between the HJC and KJC is perpendicular to the line between the KJC and virtual KNE marker, and the length of the KJC-KNE line is equal to the knee offset (KO). During static calibration a thigh rotation offset angle (Θ) is calculated based on the position of the orthogonal KAD markers relative to the THI marker, and this angle is used to give the KJC position in dynamic trials without the KAD. This is calculated by projecting the THI marker position onto a plane perpendicular to the HJC – KJC line, for instance inferiorly downwards. A positive thigh rotation offset indicates the thigh marker plane is externally rotated with respect to the plane of the knee axis and vice versa.

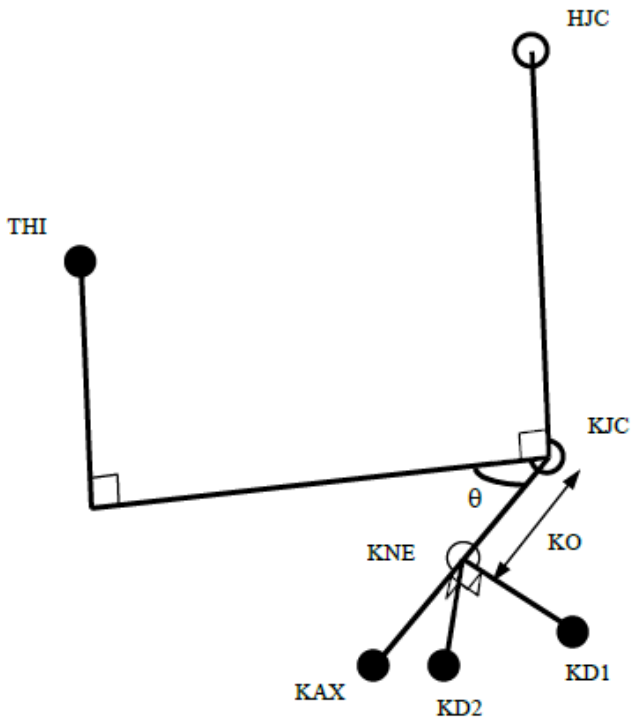


Figure 0-3: The chord function used to calculate the KJC. The thigh rotation offset is measured out or within the page.

- Ankle Joint Centre

When using a KAD, the KAX marker defines knee flexion axis which the ankle flexion axis is assumed to be parallel to. The ankle joint centre (AJC) is then calculated using a modified chord function (as depicted in Figure 0-4) whereby it has a distance equal to the ankle offset (AO) with respect to the ankle marker (ANK). The AJC is also given so that the line between the ANK marker and the AJC forms an angle equal to the tibial torsion (τ) to the plane defined by the KAX. The tibial torsion may be entered manually and rotates the plane the AJC lies in relative to the KAX marker. Also, when a KAD is used, a value for the shank rotation offset (Θ) is calculated based on the TIB marker position when projected to the plane perpendicular to the KJC-AJC line (inferiorly down). A positive shank rotation offset value denotes an externally rotated shank marker plane with respect to the ankle axis plane.

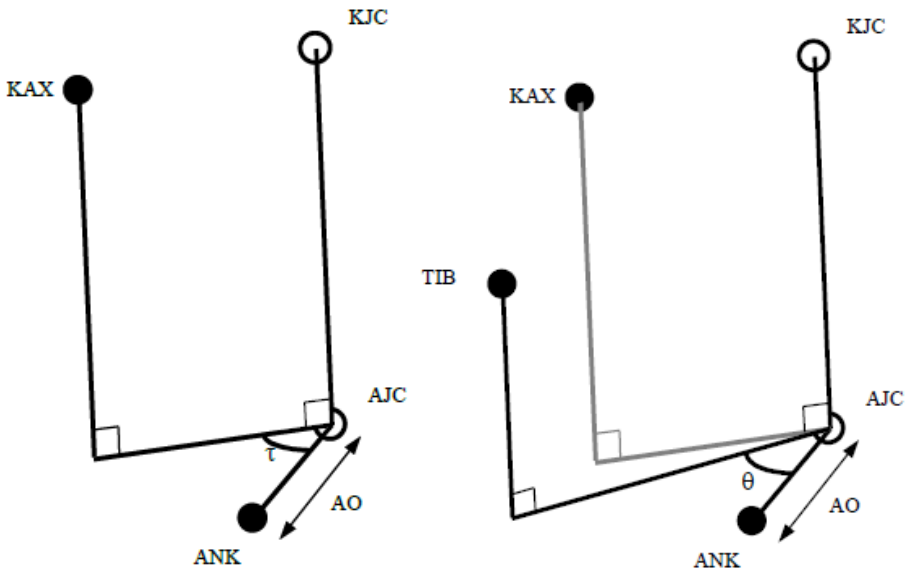
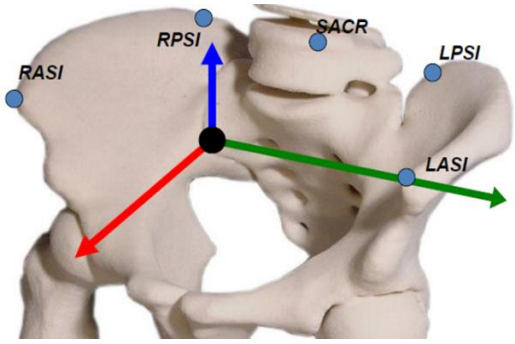
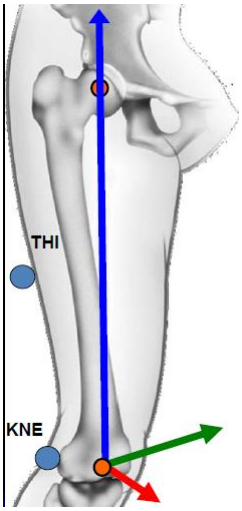
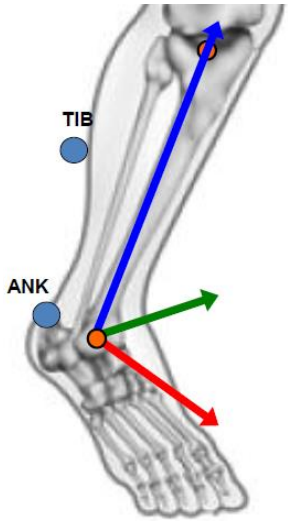
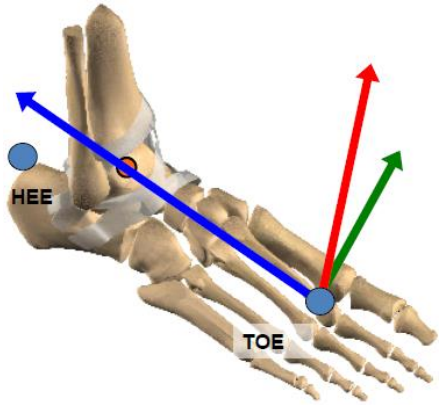


Figure 0-4: Chord functions to calculate the AJC with respect to the KAX marker. The angle τ signifies the rotation of the AJC plane with respect to the KAX marker. And the angle Θ signifies the shank rotation offset.

1.1.2. Anatomical Frame Calculations

As well as the joint centres, anatomical reference frames are required to provide biomechanical data of a person. The definition of local segment anatomical reference frames are given below, reference here is made to Vicon Motion Systems, 2017. Where shown the axis colour coding is as so: X axis = Red, Y axis = Green and Z axis = Blue.

The Pelvis	
	Origin: $\frac{LASI+RASI}{2}$ X axis: Perpendicular to Y and Z axes, directed anteriorly. Secondary axis: Anterior facing axis, calculated by: $\frac{LASI+RASI}{2} - (\frac{LPSI+RPSI}{2})$. Y axis: mediolateral axis Left- $RASI \rightarrow LASI$. Z axis: Vertically upwards, perpendicular to Secondary and Y axes.

The Femur 	Origin: KJC (midpoint of epicondyles) X axis: Directed anterior to the knee-perpendicular to secondary knee flexion axis (K_F) and the Z axis. Secondary axis (K_F): parallel to line from KJC to knee marker. Y axis: Directed left for both knees, perpendicular to X and Z axes. Z axis: Longitudinal axis between $KJC \rightarrow HJC$ (centre of sphere fitted to femoral head).
The shank 	Origin: AJC (midpoint of epicondyles) X axis: Perpendicular to K_F and Z axes facing anteriorly. Secondary axis: K_F (this is thought to be parallel to the ankle flexion axis). Y axis: Cross product of X and Z axes. Z axis: Longitudinal axis between $AJC \rightarrow KJC$.
The foot 	Origin: TOE marker X axis: Directed superiorly upwards, perpendicular to secondary ankle flexion axis (A_F) and Z axes. Secondary axis: Ankle flexion axis (A_F) Y axis: Directed left. Perpendicular X and Z axes. Z axis: $TOE \rightarrow HEE$

1.1.3. Kinematic Modelling

For all joints, the kinematic outputs are calculated from the YXZ Euler angles. Euler angles represent the set of sequential rotations along an orthogonal axis system that would rotate a segment from its actual position to a related segment or to a laboratory reference system (Baker et al., 2018). Here, these represent the absolute rotations of the pelvis and foot segments, and relative rotations at the hip, knee, and ankle joints. Relative rotations here meaning: the knee angles are calculated from the femoral and tibial segments and the ankle angles are calculated from the tibial and foot segments. Absolute angles meaning the relationship between the pelvis segment or foot segment relative to the global (or laboratory) axis system.

The kinematic variables are depicted in Figure 0-5 and are described as follows. Reference here has been made to Vicon Motion Systems, 2017b and Baker et al., 2018. Where axes are mentioned, they are described as the following, sagittal axes pass medially from one side of the body to the other, frontal axes pass from the back of the body to the front, and transverse axes pass in the direction from the centre of the body to the top of the head:

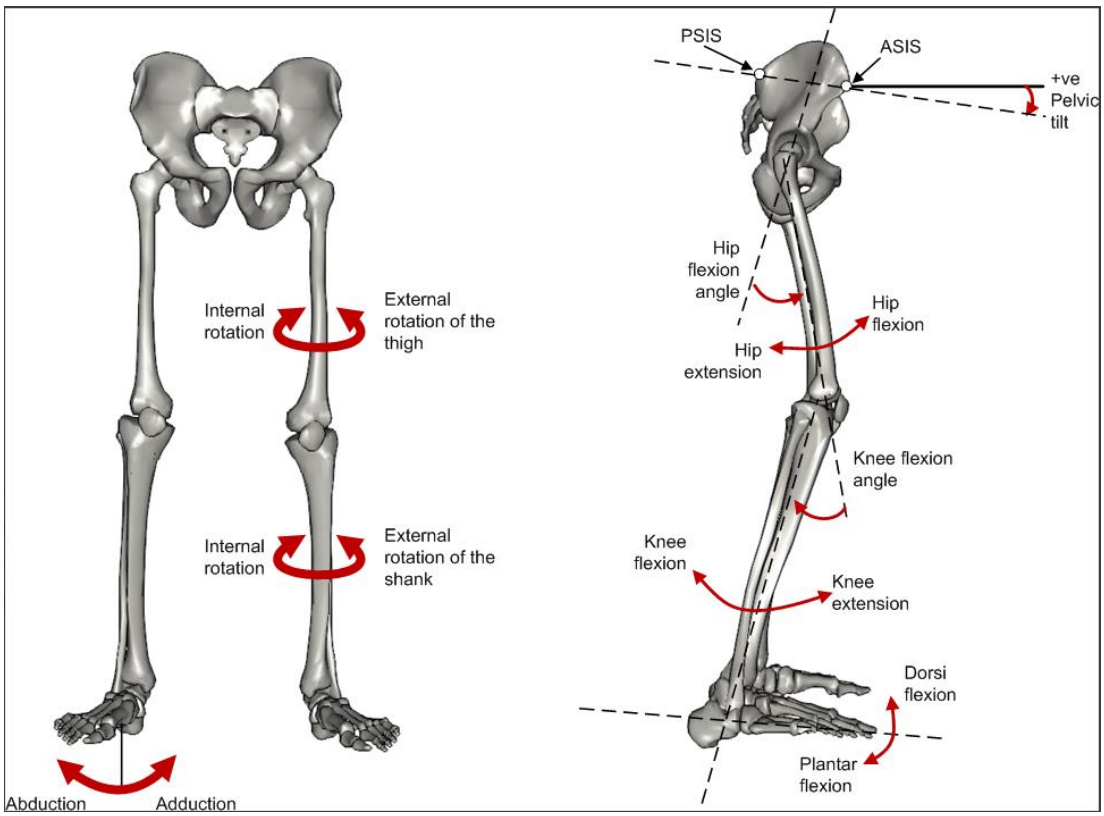


Figure 0-5: Motion of the lower limb in three-dimensions. Left: transverse (rotation) and coronal (ab/adduction) planes. Right: sagittal (flexion/extension) plane.

- **Pelvis** - These are with respect to the global coordinate system:

  Anterior/posterior tilt: rotation around the mediolateral (Y) axis around the laboratory sagittal or front axis depending on the direction of progression.

  Internal/external rotation: rotation of the mediolateral (Y) axis about the vertical transverse (Z) axis of the pelvis embedded coordinate system.

  Obliquity (up/down): rotation of the mediolateral axis out of the horizontal plane.

- **Hip** - thigh with respect to pelvis:

  Flexion/extension: rotation of the proximal-distal (Z) axis of the thigh about the medio-lateral (Y) axis of the pelvis.

  Ad/abduction: rotation of the thigh Z axis about the transverse (Z) pelvic axis.

  Internal/external rotation: rotation around the thigh Z axis.

- **Knee** - shank with respect to thigh:

  Flexion/extension: rotation of the shank proximal-distal (Z) axis about the mediolateral (Y) axis of the thigh.

  Ad/abduction: angle between the shank Z axis and the thigh Z axis.

Internal/external rotation: rotation around the shank Z axis of the frontal (X) shank and thigh axes.

- **Ankle** - with respect to shank:

  Dorsi/plantarflexion: rotation of the foot proximal-distal (Z) axis about the ankle medio-lateral (Y) axis

  Inversion/eversion: angle between the foot vector and mediolateral shank axis projected onto the foot transverse plane

  Foot progression (in/out): foot with respect to global system, the absolute angle between the foot vector when projected into the laboratory's transverse plane.

1.1.4. Kinetic Modelling

To model kinetic parameters, the model assigns masses and radii of gyration to the segments in the kinematic model. An estimate of the centre of mass (CoM) of each segment is also required, which is defined as a point at a given proportion along a line connecting the distal and proximal joint centres. For each segment, the mass is calculated as a proportion of whole-body mass. The principle axes moments of inertia are calculated from mass normalised radii of gyration displayed in Table 0-3.

Table 0-3: Ratios required to estimate segment centre of mass position from the distal end, total mass, and radius of gyration. Data from Winter, 1990.

Segment	Definition	Centre of Mass/Segment length	Segment Mass/Total Mass	Radius of Gyration
Thigh	Proximal: greater trochanter Distal: femoral condyles	0.567	0.1	0.323
Shank	Proximal: femoral condyles Distal: medial malleolus	0.567	0.0465	0.302
Foot	Proximal: lateral malleolus Distal: second metatarsal head	0.5	0.0145	0.475

Kinetic modelling is then carried out using inverse dynamics. The value of external forces applied to the body is also required which is obtained from ground reaction force (GRF) data is collected from force plates. Inverse dynamics uses velocities and accelerations of the kinematic segments to produce joint moments and powers. A hierarchy of kinetic data works upwards from the force plates to both feet which links

to both tibias, both femurs and lastly the pelvis. From the GRF data, two equations are used to calculate linear and angular kinetic data:

1) $F = m * a$ - where *F* = Resultant Force, *m* = mass, and *a* = linear acceleration
2) $M = I * \alpha$- where *M* = Resultant Moment (or Torque), *I* = Mass moment of inertia, and α = angular acceleration

The convention used here is “external forces” convention, whereby a GRF which would result in joint flexion produces a positive flexion moment, i.e., active hamstrings would give a positive flexion moment. Conversely, for a positive internal joint moment, an equal and opposite negative external moment would be an extensor moment. Moments are commonly measured in Newton metres (N.m), and often normalised to body mass reported in N.m/kg. Joint power is the scalar product between joint moments and their angular velocities given in watts per kilogram (W/kg).

Appendix 2: NHS Research Protocol

Research Protocol

Clinical investigation of the functional outcomes of high congruency versus low congruency knee bearings

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Reference: Ortho 14-12

Version: 1, 20/3/2015

IRAS ID: 165718

1. Introduction

Total knee arthroplasty (TKA) procedures are carried out as a last resort to relieve joint pain and disability in those with degenerative knee conditions. Due to the aging population and surgery being offered to younger patients, the number affected will increase. The younger and more active older patients will create a demand for longer lasting joint replacement implants with improved functional performance and lower revision surgery rates.

In the natural knee the femoral condyles carry out a rolling and gliding motion relative to the proximal tibia during flexion and extension rather than a simple hinge movement and when flexed some axial rotation between the two is also possible. This is a difficult motion to replicate in a prosthetic design. Joint implants used in TKA have a femoral component which replaces the distal end of the femur and a tibial component which replaces the proximal tibia. A polyethylene insert is added between the femoral and tibial components to provide a bearing surface between them. These bearings can be of a fixed bearing or mobile bearing design, which provide different approximations to the movement of the natural knee.

TKA using fixed bearings, where the polyethylene insert is locked to the tibial plate, is well established with successful long term follow up for patients with osteoarthritis. In contrast, mobile bearing implants allow some movement of the bearing in an attempt to replicate the movements found in the natural knee. In the case of the rotating platform type mobile bearings, planar rotation about the vertical axis of the tibia is permitted (Catani et al., 2003; McEwen et al., 2001) and dual surface articulation promotes load sharing between the relative displacements of the femoral and tibial components. This allows dissipation of knee moments and shear forces to the soft tissues surrounding the knee in a similar manner to the normal knee (Callaghan et al., 2001). Despite the theoretical benefits of the mobile bearing design, many are yet to be substantiated and many authors have documented no improvement in outcomes compared to fixed bearing designs (Mahoney et al., 2012; Jacobs et al., 2011; Post et al., 2010). However, most of the studies have only reported clinical outcomes.

Another consideration in bearing choice is the degree of congruency between the femoral component and the bearing. Lower congruency bearings have a shallower profile providing less constraint between the two components. Such bearings allow the femoral component some anterior/posterior translation and internal/external rotation to mimic the movements found in the natural knee. However, the small

contact area may increase the wear rate of the bearing material (McEwen et al., 2005). This issue can be addressed by increasing the degree of congruency between the components allowing for a greater contact area. The disadvantage of this is that it increases the constraint between the femoral component and the bearing thus reducing the translation and rotation found in lower congruency bearings. However, when combined with a mobile bearing design, these movements are able to occur.

The success of an individual TKA may be quantified by a clinical knee scoring system such as the Oxford Knee Score (OKS), which is widely used throughout the United Kingdom. However, there is evidence to suggest that the OKS is insensitive to post-operative differences between individual patients or groups of patients undergoing different TKA interventions (Jenny & Diesinger, 2012). This may increase the chance of a type II error whereby it could be concluded that no difference exists when one does exist. To more accurately assess the functional outcome of a TKA, multidimensional gait analysis methods in motion capture laboratories can be used. Such equipment allows small post-operative differences in biomechanics between intervention groups to be identified and quantified.

The purpose of this study is to evaluate and compare the function of three different knee prostheses combinations in mobile and fixed bearing variations with high and low congruency compared to the performance of healthy participants across a range of activities of daily living (ADLs). Only one study was found which used current movement analysis techniques to investigate functional differences between bearings with high and low congruency (Saari et al., 2005). They found no differences in biomechanics between the two bearings. However, they restricted their analysis to level walking, which is one of the easy ADLs and therefore may not require the higher ranges of knee flexion and rotation which may highlight any differences between the bearings. Movement analysis has been used a little more frequently for mobile versus fixed bearing investigations of a few ADLs, (Catani et al., 2003; Okamoto et al., 2014; Urwin et al., 2014; Zürcher et al., 2014). The result of this lack of research is that the function of particular designs in many ADLs is still not known, particularly the more challenging ADLs which may benefit from the increased stability of the high congruency bearings or the freedom of movement of the lower congruency bearings. By comparing 3-dimensional (3D) kinematic (movement) and joint load (kinetic) parameters during a range of common ADLs (including high demand ADLs), we hope to determine whether functional advantages exist between three bearing combinations: a low congruency fixed bearing, a high congruency fixed bearing and a high congruency mobile bearing.

2. Study Objectives and Outcomes

The primary objective is to determine whether any functional advantages exist between the three bearing combinations used in total knee replacement that are being investigated.
The study outcomes will be:
1. To determine if differences (clinically, biomechanically, radiologically and in patient reported outcome measures (PROMs)) exist between three patient groups (randomised to one of the three prosthesis combinations being studied),
2. To determine and quantify any functional differences in the three patient groups compared to a healthy control group,
3. The study will form part of an educational PhD degree in Biomedical Engineering by Dimitrios Sokratis Komaris and Cheral Govind within the University of Strathclyde.

3. Study Design

This will be a clinical double blind, randomised control study intended to investigate the functional outcomes of a high congruency knee prostheses in fixed (HCF) and mobile (HCM) configurations and a low congruency bearing in a fixed configuration (LCF).

3.1 Recruitment

Ninety consecutive patients from four surgeons at the Golden Jubilee National Hospital (GJNH) scheduled for TKA surgery who meet the inclusion/exclusion criteria and agree to take part will be randomly assigned to one of the three groups (HCF, HCM, LCF). Suitable patients for inclusion will be identified by a member of the direct care team who is also a member of the research team. These patients will be sent a copy of the patient information sheet (PIS) with a covering letter asking them to read the PIS and consider taking part in the study. The PIS will give full details of the study and explains how and why they have been approached. It will inform them that if they decide to take part, persons out with the direct care team will have access to parts of their medical and clinical records, but that no identifiable data will be taken outside the research team.

Since there are two patient routes to surgery; allocated (who are aware that they are to have TKA) and see and treat (who have been referred for assessment for TKA), the covering letter sent with the PIS will be appropriate to each route. On their subsequent visit to the GJNH they will be approached to seek their consent to take part in the study by a researcher in the direct care team. This will be at least 24 hours after they would have received the PIS, although it is likely to be longer. After consent, patients will be randomised into one of the three study groups.

Thirty healthy individuals will be recruited separately as a matched control group. These will be recruited through the University of Strathclyde using university links with external community groups, through advertising and through members of the university staff and student population. Potential control participants will be given a control group specific PIS and given a minimum of 24 hours to decide whether or not to take part. No member of the control group will be recruited through the GJNH or any other NHS location. The University of Strathclyde Departmental Ethics Committee has already given approval for this part of the study (REFERENCES?)

Patient recruitment is expected to last for up to 12 months from the recruitment of the first patient.

3.2 Methods

This project will compare three knee prostheses configurations from the Columbus® Knee System range (Aesculap AG, Tuttlingen, Germany) that are currently used in TKA surgery. These are a high congruency bearing in mobile (HCM) and fixed (HCF) configurations and a low congruency bearing in a fixed configuration (LCF). Once recruited, patient and control participants will be invited to attend functional testing sessions at the university or at the GJNH. Patient participants will attend on three occasions: 1) before the surgery (within four weeks before the surgery), 2) 4-8 weeks after the surgery and, 3) 1 year after the surgery. All testing sessions will be led by the two PhD students who will be blinded to which group the patients are allocated to. Control participants will be invited to attend the testing session on a single occasion.

The activities that will be carried out will be:

1) Sit to stand to sit
2) Sit to stand to walk
3) Level walking in a straight line
4) Level walking along a curved pathway
5) Single leg balance
6) Walking on an incline and decline
7) Ascending and descending stairs
8) Getting into and out of a car

These activities will be carried out in motion capture laboratories either at the university or at the GJNH. Reflective markers will be attached to anatomical landmarks on the participants' skin, or to tight fitting clothing, which a computerised infrared camera system will identify and track in 3D during task performance. Force

plates in the floor of the motion capture laboratories will also be utilised to record ground reaction forces. The set-up of the activities will follow the generic protocol and standard practice for the use of each laboratory. In general, each activity will be performed three times to average out variability in performances, however, to make sure that enough high quality data are captured participants may be asked to carry out a repeat of an activity if the task was not completed successfully or the data properly capture e.g. if when carrying out a walking trial the participant does not cleanly strike the force plate with the limb that is being assessed. The researcher will make every effort to keep the burden on the participants as low as possible and it will be made clear to the participants that they can at any time stop, rest or decide that they do not wish to carry out that task again. The processing of this data allows joint movements (kinematics) to be recorded and joint loadings (kinetics) of the participants during the activities to be calculated. At the end of each task participants will be asked a number of short questions about how they found performing that task. In addition, for the tasks with stairs and the getting into and out of a car there will be questions about how well the laboratory set up mimics real life.

Other clinical outcome measures will be recorded for patient participants including the Oxford Knee Score (OKS), patient satisfaction, joint ranges of motion and post-operative complications by staff at the GJNH as part of the standard review procedure. Lower limb alignment and component placement data will be collected from standing hip-knee-ankle and lateral knee radiographs and from the computer navigation system used during surgery.

3.3 Timescales

From the start of the study (posting of first letters to patients) it is expected to take around three weeks to first recruitment and a further two weeks to first operation. Overall, it is expected to take around 12 months to recruit the 90 patients so recruitment will close approximately 13 months after the study starts. Each patient will undergo one year follow up so the final data should be collected approximately 25 months after the study starts. Therefore, if the study starts by 1st July 2015 this will be within the three-year funding timescales (section 10).

3.4 Power calculation and statistical analysis

Given the paucity of biomechanical data for the implant combinations and ADLs in our study, the sample size per group was calculated using published data for knee flexion during stair ascent of fixed bearings (Okamoto et al., 2014). Using a knee flexion of 76.6° for the fixed bearings, an expected increase of 6.2° (half of the

difference between the study group and the control group) for the mobile bearings, a power of 80% and an alpha value of 0.05 (two-tailed) the sample size per group was 27. This was increased to 30 to factor in a dropout rate of 10%.

Statistical analysis will be carried out using an appropriate statistical package to determine if any significant differences exist in the data between the three patient groups and between the patient groups and the control group. All data will be tested for parametricity. Parametric data will be analysed using one-way analysis of variance (ANOVA) tests with post hoc Bonferroni correction. If the data is found to be non-parametric, Kruskal-Wallis tests will be used with Bonferroni correction. A 95% confidence level will be used throughout.

4. Selection Criteria

The inclusion criteria for patient participants are:

1) Primary unilateral total knee arthroplasty
2) Suitable to have any of the three study implants
3) Over 35 years of age
4) Willing to take part
5) From one of the following NHS Scotland Health Boards:

Ayrshire & Arran, Forth Valley, Greater Glasgow & Clyde, Highland, Lanarkshire or Lothian

6) Able to return for follow up sessions

The exclusion criteria for patient participants are:

1) Previous hip or knee replacement procedure if carried out in the previous six months
2) Unable to give written consent
3) Unable to attend the movement analysis sessions
4) Journey time from home to the university in excess of two hours
5) Previous ankle surgery

The inclusion criteria for control participants are:

1) No pre-existing condition or injury likely to influence performance of test activities
2) Over 35 years of age
3) Willing to take part

The exclusion criteria for control participants are:

1) Previous lower limb joint replacement procedure
2) Unable to give written consent

5. Treatment of Subjects

Participating patients will be randomised to receive one of three different variants of the Columbus® knee prosthesis (models CR DD (LCF group), UC (HCF group) or UCR (HCM group)). Two of these variants (UC and UCR) are ultra-congruent designs, while the other (CR DD) has a lower congruency. Both of the ultra-congruent variants are posterior stabilised, but the CR DD is a cruciate retaining design. Posterior stabilised designs require the posterior cruciate ligament (PCL) to be resected. One of the variants (UCR) is a rotating platform bearing, while the other two are fixed designs. No patient will be recruited to the study if randomisation to receive any of the variants would leave them clinically or functionally disadvantaged.

All patients will undergo TKA following the standard operating practice of the participating GJNH surgeons, and the implant will be used in accordance with the manufacturer's recommendations. All patients, including those approached who decline to take part in the study, will be given the same standard of care as all GJNH TKA patients. The Columbus® knee prosthesis is used by all the consultant orthopaedic surgeons involved in the study as part of their standard practice. The fixed bearing variants are currently used within the GJNH, but the mobile bearing variant is not. Therefore 30 of the patients in the study will receive a variant not available to other patients. However as stated above patients will only be recruited to the study if they are suitable to receive any of the three prosthesis designs.

6. Assessment of Safety

For the TKA replacement the risks of operation will be the same as for patients outside the study undergoing TKA. All three implant variants being used are CE marked commercially available devices.

For the study assessments, as the patients are TKA candidates, they are likely to have some degree of mobility limitation. Although the functional tasks carried out would not be considered strenuous to a healthy individual, these tasks may be difficult for a person about to undergo, or recovering from, a TKA. All patients recruited into this study will have been identified as candidates by their consultant who will have deemed them suitable to take part in the functional testing for this study safely. To minimise any potential pain or discomfort, breaks will be scheduled between activities and as required during them at the participant's request for as long as

necessary. If any participant feels unable to complete the activity, the test will be stopped.

These activities will be carried out in university laboratories or at the GJNH and each session may last up to 4 hours. During activities with any risk of tripping or falling, support will be provided by a safety harness. For all tasks, participants will be asked to perform the activity in a natural and consistent manner. In the activities that require walking, the participants will be encouraged to use their natural walking speed.

Risk assessments have, or will have, been carried out for the laboratories in which testing will be carried out. The university laboratories have extensive experience of carrying out motion capture analysis on many different groups of people under a wide variety of conditions and therefore are very used to the precautions that must be taken with older persons and patients undergoing lower limb joint replacement.

7. Quality Control and Quality Assurance

The study will be carried out in accordance with the standards of Good Clinical Practice and the Declaration of Helsinki. All members of the research team will have up to date GCP training. The study will be monitored by the GJNH R&D department. The study will be overseen on a day-to-day basis by the Orthopaedic Audit and Research Coordinator.

8. Ethical Considerations

A. Patient consent

Following selection of suitable patients, a member of the direct care team will meet the patient when they attend the GJNH to ensure that they have received a copy of the PIS, to answer any questions they may have related to it and to seek their written consent. A minimum of 24 hours will be given for the potential participant to decide whether or not to take part. In reality, the period would be longer since suitable patients will have been identified and letters sent to them prior to attendance at GJNH clinics. The original consent form will be kept in the patient case notes and a copy will be given to the patient and another copy will be filed in the study folder.

Patients will be entitled to withdraw consent up to the point of the completion of the analysis of the dataset for the student thesis without giving a reason for doing so. If a patient requests that their data is withdrawn from the study, it can be removed from the study dataset up until the analysis is completed. Once the analysis is

completed it would not be possible to remove the data as this would change the summary results which will already be presented in the students' theses.

B. Ethical Issues

Patients will have to agree to being randomised to receive one of the three study implants. No patient will be recruited into the study that would not be suitable to receive either of the study implants.

In addition to their standard pre-operative assessments and post-operative reviews, study patients will also be required to attend three motion analysis sessions at the university lasting up to four hours. The first of these will occur prior to surgery with the other two being 4 – 6 weeks and one year post-operatively. It is unlikely that these sessions will be scheduled to coincide with their GJNH appointments due to scheduling issues within both the GJNH and the university. Return transportation from the patients' home to the university may be provided by the university. During the motion analysis session, patients will be given rest breaks and refreshments. Patients will not be asked to perform task which they have difficulty or discomfort with. Patients will be allowed to end the session at any time.

One of the study bearings (HCM group) is not currently used within the GJNH; therefore, the three surgeons who will be involved in the study will not have the same experience of using this bearing as the other two at the start of the study. The surgeons will have had training in the use of the UCR, the tools required and the implantation procedure prior to the study commencing. In addition is it a variation of the implant they currently use as standard practice, not a completely new implant.

Both fixed and mobile bearing designs are generally used for TKA although fixed designs are more widely used. The differences in designs may lead to different outcomes, better or worse. The design of the mobile bearing is hoped to give a more natural function but may also show no significant advantages over the fixed bearing knees.

C. Ethical Approval

Approval from an NHS Research Ethics Committee and the University Ethics Committee will be required prior to the commencement of the study in addition to approval by GJNH R&D department.

Separate ethical approval has been sought through the University Ethics Committee for the collection of control data since this does not involve NHS patients, staff or premises.

9. Data Handling and Record Keeping

The study ID plus initial and year of birth will be used to identify all data, electronic or hard copy, collected as part of the study. Due to many datasets being recorded at different times in different locations it is felt that some cross-reference apart from a study ID is required to ensure that all data are allocated to the correct study patient. These data will not allow anyone outside the research team to identify the participants.

A. GJNH

Research specific personal data will be treated in the same way as clinical records for the patient following the NHS code of practice on protecting patient confidentiality (NHS Scotland). Patients who agree to take part in the study will be assigned a unique study ID number. All documents such as consent forms linking the patient participants' personal details to the study ID number will be stored separately as a hard copy in a locked filing cabinet in a secure room in the GJNH. All electronic data linking patient participants' details to the study ID number will be stored on the GJNH server which requires a user ID and password to access and is restricted to members of the Orthopaedic department.

All electronic data will be stored on the GJNH server accessible only to members of the Orthopaedic department by user ID and password. Any electronic data transfers between the GJNH and the university will be in a pseudo anonymous manner using the study identification only. Paper records will be stored in locked cabinets in a secure room within the Orthopaedic department. No identifiable data will be taken outside of the research team (data transferred to the university containing only the study ID will be identifiable to the researchers at the university as they will know which patient has which ID). All data generated at the GJNH will be under the supervision of the department's Audit and Research coordinator, Dr Angela Deakin.

The summary data and statistical analysis will be retained on GJNH computers to allow the publication of the study results in peer-reviewed journals. Non-identifiable patient data will be stored for 5 years after the end of the study on the GJNH server area accessible only by members of the Orthopaedic department. A master file to identify patients from the data will be kept for the same time period on a password protected account on a GJNH server.

B. University of Strathclyde

All data collected at the university will be identified by the study identification alone. This data will be in hard copy and electronic format. Hard copy data will be kept in a locked cabinet accessible by members of the research team only. Electronic data collected during the functional testing sessions will be stored on a stand-alone computer with password secured user accounts accessible only by members of the research team. Electronic data may include video recordings which would be used for analysis and educational purposes. Other study related electronic data will be stored on university servers in the personal accounts of research team members and accessible only by that individual by a username and password. All data generated at the university will be under the supervision of the students' supervisor, Dr Andrew Murphy.

In addition to the students, their academic supervisors, Dr Andrew Murphy and Dr Phil Riches, will review the analysis of pseudo-anonymised data as part of their supervision.

Once the students have submitted and passed their theses, anonymised data will be deleted from the university computer. The students will inform the chief investigator of the destruction of the data. Provisions will be made to store the data elsewhere as part of an anonymised database at Strathclyde University. This will be made clear to all participants on the consent form.

C. General

No personal data will be published. Only summary data will be transferred outside the GJNH. Care will be taken if description of particular cases is necessary that patients will not be able to be identified from this. No video footage will be published, but still shots from the video may be published or used for educational purposes. In this event, consent will be obtained from the participant prior to use and measures will be taken to obscure the identity of the patient.

10. Financing and Insurance

Financial support has been provided by Aesculap AG, Am Aesculap Platz 1, 78532 Tuttlingen, Germany. Finance is being provided over a three-year time scale from October 2014.

As the study sponsor, the hospital will cover any insurance liability during any elements of the study carried out at the hospital. During the functional testing, which will be carried out at the University of Strathclyde, the university will cover any insurance liability.

11. Publication Policy

This study will be published as two PhD theses. It is intended to publish the results of this study in peer-review journals and to present them at appropriate conferences. The eight tasks will be divided to four and each PhD candidate will analyse and write up data from their own tasks and collate the information for journals/conferences.

12. References

1. Callaghan JJ, Insall JN, Greenwald AS, Dennis DA, Komistek RD, Murray DW, et al. Mobile-bearing knee replacement: concepts and results. Instr Course Lect 2001; 50:431–49.
2. Catani F, Benedetti MG, De Felice R, Buzzi R, Giannini S, Aglietti P. Mobile and fixed bearing total knee prosthesis functional comparison during stair climbing. Clin Biomech 2003; 18:410–8.
3. Jacobs WC, Christen B, Wymenga AB, Schuster A, van der Schaaf DB, Ten Ham A, et al. Functional performance of mobile versus fixed bearing total knee prostheses: a randomised controlled trial. Knee Surg Sports Traumatol Arthrosc 2011; 20:1450–5.
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5. Mahoney OM, Kinsey TL, D'Errico TJ, Shen J. The John Insall Award: no functional advantage of a mobile bearing posterior stabilized TKA. Clin Orthop Relat Res 2012; 470:33–44.
6. McEwen HMJ, Fisher J, Goldsmith AAJ, Auger DD, Hardaker C, Stone MH. Wear of fixed bearing and rotating platform mobile bearing knees subjected to high levels of internal and external tibial rotation. J Mater Sci Mater Med 2001; 12:1049-52.
7. McEwen HMJ, Barnett PI, Bell CJ, Farrar R, Auger DD, Stone MH, et al. The influence of design, materials and kinematics on the in vitro wear of total knee replacements. J Biomech 2005; 38:357-65.
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9. Post ZD, Matar WY, van de Leur T, Grossman EL, Austin MS. Mobile-bearing total knee arthroplasty: better than a fixed-bearing? J Arthroplasty 2010; 25:998–1003.

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11. Urwin SG, Kader DF, Caplan N, St Clair Gibson A, Stewart S. Gait analysis of fixed bearing and mobile bearing total knee prostheses during walking: do mobile bearings offer functional advantages? Knee 2014;21:391-5.
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Appendix 3: Patient Information Sheet

Golden Jubilee National Hospital

NHS National Waiting Times Centre

Chair Jeane Freeman
Chief Executive Jill Young

Agamemnon Street
Clydebank G81 4DY
Scotland
Telephone 0141 951 5000
Fax 0141 951 5500

Clinical investigation of the functional outcomes of high congruency versus low congruency knee bearings.

We would like to invite you to take part in our research study. The study is being organised by the Golden Jubilee National Hospital, the B. Braun Medical Ltd (who manufacture the Columbus knee implant you will be receiving) and the University of Strathclyde.

Before you decide if you would like to be part of this study, it is important for you to understand why the research is being done and what you would need to do.

Please take the time to read the following information carefully and discuss it with others in order to decide whether or not you would like to take part.

You are free to choose whether to not to take part. If you decide not to take part this will not affect the care you get.

There are two main implant types used in total knee replacement (TKR) surgery: the mobile and the fixed bearing prosthesis. The fixed bearing prostheses may also be subdivided into high and low congruent implants which refer to how well the implant components fit together. It is thought that the mobile bearing prosthesis better replicates the natural knee however current studies are yet to show which implant type patients prefer.

This study will involve using these three implant types (mobile high congruency, fixed high congruency and fixed low congruency) in TKR patients and using tests that aren't already commonly used to find which implant is better at restoring natural function. This will give us new information about which implant should be used in future surgeries.

Why have I been invited?

You are being asked to take part because you have been identified as a suitable candidate for both the TKR and the testing at Strathclyde University and would be a suitable candidate for any of the three implants being tested.

Do I have to take part?

No, participation is voluntary and it is up to you to decide. Use this information sheet to help you decide. Discuss it with family and friends or speak to a member of the research team for more information.

***What will I have to do*?**

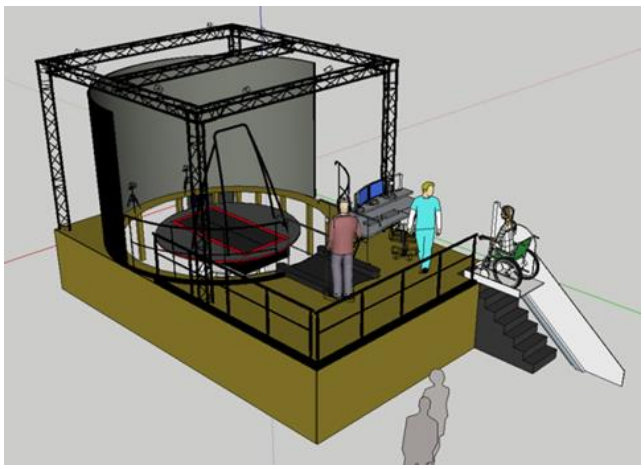
If you decide to take part in this study, you will be randomly assigned to receive one of the three implant variants of the knee in the study. The two variants with the more conforming shape require the posterior cruciate ligament in the knee to be removed. If you are assigned to receive one of these, you will have this ligament removed. In all instances you will receive the standard surgical practice for the knee you will be receiving.

You will be asked to attend three sessions (location given below) for no longer than 4 hours (1 hour of which will be solely dedicated to activities of daily living such as walking, single leg balance and ascending stairs and the rest of the time will be used for resting between activities, and ensuring that you are ready for testing.). The sessions will take place before your operation, 4-8 weeks after the operation and one year after the operation.

You will need to wear appropriate clothing so that accurate motion of the body while moving can be recorded (appropriate clothing will be provided by the department, if necessary). Male participants will be required to wear tight cycling type shorts with no top on. Female participants will be required to wear tight cycling type shorts and a tight-fitting crop type top. You will be required to bring sports type shoes (without reflective flashes if possible). Reflective markers that will be detected by the cameras will be placed on you (as depicted below) using medical grade non-allergic tape and you will be asked to perform a set of functional tasks as so:

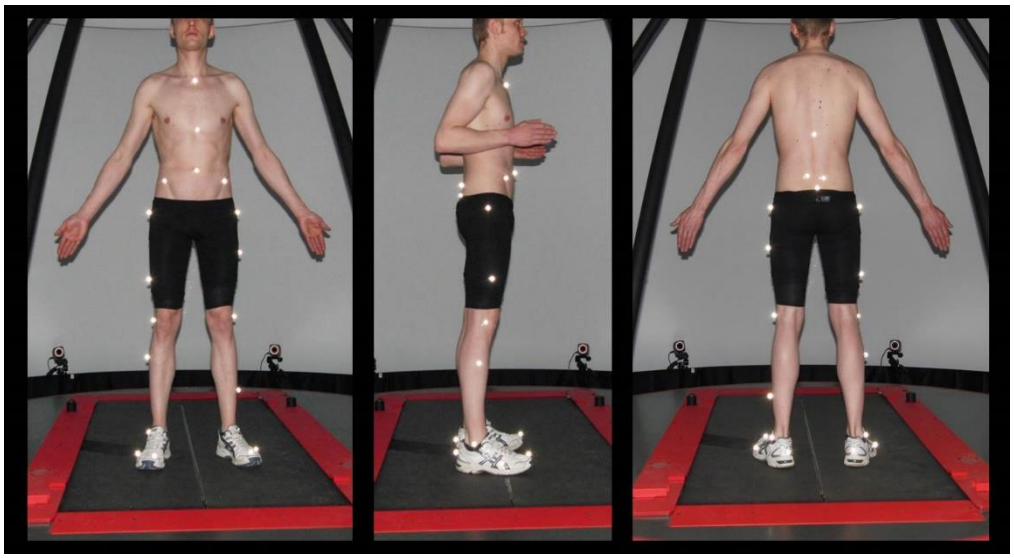
1. Sit-to-stand
2. Sit-to-stand-walk
3. Getting into and out of a car
4. Ascending and descending stairs
5. Walking on an incline
6. Level walking in a straight line
7. Level walking along an S-shaped pathway
8. Single leg balance

All sessions will be conducted and supervised by appropriately qualified and certified members of the research team. Some tasks will be held in our Motek Laboratory. The Motek lab consists of a motion platform, a treadmill, a motion capture system, and a large diameter 180° projection screen for displaying virtual reality environments.



The Motek Lab

Tasks 1-4 will be held in a standard motion capture laboratory with a number of infrared cameras installed in it. Task 5 will be held in the Motek Lab where the participant will be asked to walk on a treadmill with a virtual reality aid. Finally, tasks 6-8 will be held in both laboratories.



A male participant affixed with reflective markers while standing on the treadmill in the Motek Laboratory.

After every task (and during if required), you will be allowed to rest if necessary. There could be times when video recording or photographs will be taken, but only if you agree to this beforehand. This experiment offers no incentives for application or reimbursements to potential participants. The laboratory session will take place in the following location:

The University of Strathclyde,

Department of Biomedical Engineering,

Biomechanics laboratory 2,

Level 1,

Wolfson Centre,

106 Rottenrow,

G4 0NW.

What are the possible disadvantages and risks of taking part?

Risk assessments have been carried out to ensure that no harm will come to you or the researchers. Appropriate safety harnesses will be provided where necessary and during the tasks you are encouraged to carry them out at your own leisure and will be able to take breaks between and during the tasks if you wish. The test is estimated to last 4-6 hours.

If you are not used to walking very much you may experience mild aches and pains after the walking sessions. We hope to minimise this possibility by only increasing the speed and duration of walking according to your levels of comfort and then only by small amounts.

There is a possible risk of tripping or losing balance. To prevent this loss of balance we will make sure that there is nothing on the floor that you might trip over and when you are on our treadmill you will wear a safety harness.

We will attach some sticky markers to your skin, occasionally this can cause a mild irritation similar to having sellotape attached to your skin. This should only

be a temporary irritation since the markers will only be in place for a short time and we will be very careful when we take the markers off you.

What are the possible benefits of taking part?

The main benefit of you talking part is that you will gain an understanding of how well your surgery went as you will be able to compare your pre-operative data to post-operative.

What happens when the research study stops?

You will continue to receive the same care as all orthopaedic patients treated at the Golden Jubilee National Hospital.

If you are considering taking part, please read the additional information in Part 2 before making your final decision.

Part 2 – Further information about the study

What will happen if I don't want to carry on with the study?

You can withdraw from the study at any time by speaking to a member of the research team or by writing to us. You do not have to give a reason for not wanting to carry on with the study and the care you receive will not be affected because of your decision.

The data for the study will be written up by two PhD students as part of their degree. If you decide you want to stop being in the study before the data has been analysed, it can be removed from the study. If you decide after this point it will not be possible to take your data out of the study.

If the study is stopped for any reason, you will be told why. Your care will not be affected.

What if there is a problem?

If you have a concern about any aspect of this study, you should ask to speak to the researchers who will do their best to answer your questions. You can contact them on 0141 951 5966.

If you remain unhappy and wish to complain formally, you can do this through the NHS Complaints Procedure. Details can be obtained from the hospital by contacting:

The Complaints Officer
Golden Jubilee National Hospital
Agamemnon Street
Clydebank
G81 4DY
Telephone 0141 951 5440

In the event that something goes wrong, and you are harmed during the research due to someone's negligence then you may have grounds for legal action or compensation. You may have to pay legal costs. The normal National Health Service complaints mechanisms will still be available to you.

Will my study data be kept confidential?

All information that is collected about you during the course of this study will be kept strictly confidential and secure in line with Caldicott principles and the Data Protection Act 1998. If you join the study, data collected will be kept securely within the Golden Jubilee National Hospital. Electronic data will be kept on password protected NHS computers. No data that allows you to be identified from it will leave the hospital. However, copies of the data will be taken to the University of Strathclyde. These data will be identified with a number so that the researchers will know who the data belong to but no-one else will be able to tell.7

***Will my GP be notified of my participation in the study*?**

We will tell your GP that you are taking part in this study. This is because they are responsible for your health so need to know what is happening to you. We will ask you to agree to this on the consent form for the study.

Who is organising and funding the research?

The study is organised by the Orthopaedic Research team at the Golden Jubilee National Hospital working with the Biomedical Engineering Department at the University of Strathclyde.

What will happen to the results of the research study?

The results of the study will be written up as part of the students' degree work. They may also be published as papers in medical journals. If you would like to know about the results of the study, please indicate this on the consent form and we will send you a summary of the result. This will also let you know how to ask for copies of any of the papers. The study data will be kept on a secure database and may be used to follow up the long-term outcomes (up to ten years) of your operation. You will not be identified in any report or publication unless we have asked you if this okay and you have agreed to it. Also, a summary of the result will be sent to your GP to discuss with you.

Who has reviewed the study?

All research in the NHS is looked at by an independent group of people, called a Research Ethics Committee, to protect your interests. This study has been reviewed and approved by the West of Scotland REC 5 Research Ethics Committee.

Thank you for taking the time to read this information. If you would like any more information on this study, please contact a member of the research team who will answer any of your questions.

For specific information about this study, contact:

Dr Alistair Ewen
Orthopaedic Researcher
Manager
Department of Orthopaedics
Hospital
Golden Jubilee National Hospital
Agamemnon Street
Clydebank
G81 4DY
Tel: 0141 951 5966

For independent general information about research, contact:

Dr Catherine Sinclair
Research and Development

Golden Jubilee National

Clydebank
G81 4DY
Tel: 0141 951 5440

Appendix 4: Patient Consent Form

Golden Jubilee National Hospital

NHS National Waiting Times Centre

Chair Jeane Freeman
Chief Executive Jill Young

Agamemnon Street
Clydebank G81 DY
Scotland
Telephone 0141 951 5000

Patient Identification Number for this trial:

CONSENT FORM

Title of Project: **Clinical investigation of the functional outcomes of high congruency versus low congruency knee bearings**

		Please initial all boxes
1.	I confirm that I have read and understand the information sheet dated 8/12/16 (version 4) 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 is voluntary and that I am free to withdraw at any time without giving any reason, without my medical care or legal rights being affected.	☐
3.	I understand that relevant sections of my medical notes and data collected during the study may be looked at by individuals from regulatory authorities or from the NHS Trust and/or the University of Strathclyde, where it is relevant to my taking part in this research. I give permission for these individuals to have access to my records.	☐
4.	I consent to having video images recorded during parts of the study as per the information sheet dated XX/X/XX (version xx) for data analysis purposes.	☐
5.	I give my consent for the Principle Investigator to use images recorded during the study in scientific publications and conference presentations where their use would be in aid to better understanding of the results.	☐
6.	I give my consent for my data stored at the University of Strathclyde for long term storage as part of an anonymised database.	☐
7.	I agree to my GP being informed of my participation in the study.	☐
8.	I agree to take part in the above study.	☐

_ Name of Participant

Date Signature

_ Name of Person taking consent

Date Signature

Appendix 5: Departmental Ethics Protocol

Ethics Application Form

Please answer all questions

1. Title of the investigation
Clinical investigation of the functional outcomes of high congruency versus low congruency knee bearings
Please state the title on the PIS and Consent Form, if different:

2. Chief Investigator (must be at least a Grade 7 member of staff or equivalent)
Name: Dr Andrew J Murphy ☒ Research Fellow (Grade 7) ☐ Professor ☐ Reader ☐ Senior Lecturer ☐ Lecturer ☐ Senior Teaching Fellow ☐ Teaching Fellow Department: Biomedical Engineering Telephone: 01415482855 E-mail: andrew.j.murphy@strath.ac.uk

3. Other Strathclyde investigator(s)
Name: Dr Philip Riches Status (e.g., lecturer, post-/undergraduate): Lecturer Department: Biomedical Engineering Telephone: 01415485703 E-mail: philip.riches@strath.ac.uk Name: Mr Dimitrios Socratis Komaris Status: Postgraduate research student Department/Institution: Biomedical Engineering Name of supervisor: Andrew J Murphy Telephone: +447873563209 E-mail: dimitrios.komaris.2013@uni.strath.ac.uk Name: Miss Cheral Govind Status: Postgraduate research student Department/Institution: Biomedical Engineering Name of supervisor: Andrew J Murphy

Telephone: +447599214067
E-mail: cheral.govind@strath.ac.uk

4. Non-Strathclyde collaborating investigator(s) (where applicable)

Name:
Status (e.g. lecturer, post-/undergraduate):
Department/Institution:
If student(s), name of supervisor:
Telephone:
E-mail:
Please provide details for all investigators involved in the study:

5. Overseas Supervisor(s) (where applicable)

Name(s): N/A
Status: N/A
Department/Institution: N/A
Telephone: N/A
Email: N/A
I can confirm that the local supervisor has obtained a copy of the Code of Practice: Yes ☐ No ☐
Please provide details for all supervisors involved in the study: N/A

6. Location of the investigation

At what place(s) will the investigation be conducted:

The University of Strathclyde
Department of Biomedical Engineering
Biomechanics laboratory 2
Level 1
Wolfson Centre
106 Rottenrow
G4 0NW

If this is not on University of Strathclyde premises, how have you satisfied yourself that adequate Health and Safety arrangements are in place to prevent injury or harm?
N/A

7. Duration of the investigation

Duration(years/months) : 3 months

Start date (expected): / / Completion date (expected): 27 / 02 / 2014

8. Sponsor
Please note that this is not the funder; refer to Section C and Annexes 1 and 3 of the Code of Practice for a definition and the key responsibilities of the sponsor.

Will the sponsor be the University of Strathclyde: Yes ☒ No ☐
If not, please specify who is the sponsor: N/A

9. Funding body or proposed funding body (if applicable)

Name of funding body: B.Braun
Status of proposal – if seeking funding (please click appropriate box):
☐ In preparation
☐ Submitted
☒ Accepted
Date of submission of proposal: / / Date of start of funding: / /

10. Ethical issues

Describe the main ethical issues and how you propose to address them:

The participants of this study are healthy, non-vulnerable adults and so any issues regarding the Protection of Vulnerable Groups are not relevant to this proposal.

The main ethical issue is that the testing sessions will be carried out in an unfamiliar environment to many of the participants. During the recruitment process, clear and comprehensive information will be provided to the potential participants fully explaining the nature of the session. The contact information of Mr Komaris, Miss Govind and Dr Murphy will be provided to the potential participants to answer any questions To aid the understanding of the potential participants, information regarding the scope of the project will be available in written form via posters/flyers as well as visual aids such as photographs and videos. Once a participant expresses their interest in the project, they will have the opportunity to visit the laboratory prior to their own session to see a live demonstration. Participants will also have the right to withdraw from the study at any time.

11. Objectives of investigation (including the academic rationale and justification for the investigation) Please use plain English.

There are different types of knee implant that may be used in total knee replacement (TKR). The two main types are fixed bearing and mobile bearing. The fixed bearing implants may be further divided into low and high congruency implants whereas all mobile implants are considered to have a high congruency. Theoretically, TKR patients using mobile implants ought to offer closer-to natural biomechanics compared to fixed bearing however, a number of studies have shown that outcome in patients with both

implant types shows no significant differences (Kim et al., 2007; Geiger et al., 2008;). However, the tests used to measure outcome include using subjective Knee Score tests or questionnaires, and radiological and clinical assessment. This study aims to compare between the implants using gait analysis in motion capture laboratories to quantify functional performance during a series of tasks.

The primary objective will be to determine if the high congruency mobile bearing knee implant provides closer to normal postoperative function during activities of daily living (ADL) compared to the two fixed bearing configurations.

The secondary objectives are:
To identify functional differences between the three patient groups compared to the control group
To determine the improvement in function post-operatively compared to pre-operative function in the three patient groups
To compare clinical outcomes between the three patient groups
To compare Oxford Knee Scores post-operatively between groups
To compare patient satisfaction post-operatively between groups

References:

Geiger F, Mau H, Kruger M, Thomsen M (2008) Comparison of a new mobile-bearing total knee prosthesis with a fixed-bearing prosthesis: a matched pair analysis. Arch Orthop Trauma Surg 128:285–291.

Kim YH, Yoon SH, Kim JS (2007) The long-term results of simultaneous fixed-bearing and mobile-bearing total knee replacements performed in the same patient. J Bone Joint Surg Br 89:1317–132.

12. Participants

Please detail the nature of the participants:
The initial participants will include 10 able bodied adults from the department of Biomedical Engineering to be used as a pilot study to ensure that data from these individuals has integrity. 30 older adults will be then recruited as a control group to the study

Summarise the number and age (range) of each group of participants:
Number: 30 Age (range) 35-7

Please detail any inclusion/exclusion criteria and any further screening procedures to be used:

The inclusion criteria:

- Able bodied
- 5’2” to 6’2’ in height.
- Normal lower limb function

- Knowledge of using a treadmill
- 20/20 vision (with or without visual aid)

The exclusion criteria:

- Musculoskeletal, neurological or sensory deficit
- Pregnancy
- Previous hip or knee replacement procedure
- Unable to give written consent
- Unable to attend the gait analysis sessions
- Previous ankle surgery

No further screening procedures will be required.

13. Nature of the participants

Please note that investigations governed by the Code of Practice that involve any of the types of participants listed in B1(b) must be submitted to the University Ethics Committee (UEC) rather than DEC/SEC for approval.

Do any of the participants fall into a category listed in Section B1(b) (participant considerations) applicable in this investigation?: Yes ☐ No ☒
If yes, please detail which category (and submit this application to the UEC):
N/A

14. Method of recruitment

Describe the method of recruitment (see section B4 of the Code of Practice), providing information on any payments, expenses or other incentives.

A recruitment poster will firstly be displayed in the Department of Biomedical Engineering and then to various church groups/Glasgow Life. The poster will outline the project background, the inclusion and exclusion criteria and the contact details of the investigator and student researchers so the interested parties are able to request a participant information sheet. After due consideration, if the interested party decides to participate in the project then they will sign and return the consent form to the researcher. The participant will then be contacted to arrange an appropriate time to attend the session.

15. Participant consent

Please state the groups from whom consent/assent will be sought (please refer to the Guidance Document). The PIS and Consent Form(s) to be used should be attached to this application form.

Consent will be sought from all participants and will be asked to consider whether they wish to provide consent for the following:

- Consent to being photographed and video recorded as part of the project.
- Consent for unidentifiable photographs and video recordings to be used in publications or teaching materials.

However, these aspects are not an essential requirement to be included as a participant in the study.

16. Methodology

Investigations governed by the Code of Practice which involve any of the types of projects listed in B1(a) must be submitted to the University Ethics Committee rather than DEC/SEC for approval.

Are any of the categories mentioned in the Code of Practice Section B1(a) (project considerations) applicable in this investigation? ☐ Yes ☐ No
If 'yes' please detail:

ii. an extensive degree or duration of exercise or physical exertion beyond that to which all the participants are habitually accustomed

This activity does not require an extensive degree or duration of exercise or physical exertion as the tasks are carried out at a self-selected speed. However, the activity takes place in an environment (Motek is a virtual reality system) which participants will be unfamiliar with.

Describe the research methodology and procedure, providing a timeline of activities where possible. Please use plain English.

Each test session will take place in Level 1 of the Wolfson Building and should take no longer than 4 hours. The following functional tasks will be undertaken by each participant:

- Sit-to-stand (V)
- Sit-to-stand-walk (V)
- Level walking in a straight line (VM)
- Level walking along an S-shaped pathway(VM)
- Single leg balance (V)
- Walking on a decline (M)
- Ascending and descending stairs. (V)

The performance of the participants during these functional tasks will be referred to as Biomechanical performance. The biomechanical performance of each participant will be assessed via fully instrumented three-dimensional human movement analysis in either the conventional gait VICON laboratory or the new MOTEK Caren system. The letter in the parenthesis indicates which lab the activity will be carried out in, where V= Vicon; M= Motek and VM= both labs. Established biomechanical models, data processing and outcome measures will be used.

Throughout the sessions, participants will be required to wear appropriate clothing that will allow accurate recording of the body whilst in motion. Male participants will be

required to wear tight cycling-type shorts and to be topless. Female participants will be also required to wear tight cycling-type shorts as well as a tight-fitting cropped t-shirt. All participants will be required to bring appropriate sports shoes. Appropriate clothing would be provided to participants who aren't able to supply their own clothes.

Following this, reflective spherical markers will be attached to the skin in several areas on the legs and pelvis with medical grade non-allergenic tape. Once this is completed, model calibrations will be carried out and the functional tasks will be carried out on the ground or the treadmill (depending on the test). Participants will initially walk on the treadmill to derive an individualised normal walking speed in accordance with the protocol developed at the Motek installation at the VU Medical Centre in Amsterdam. This method is non-invasive and non-maximal.

All sessions will be conducted and supervised by Dr Andrew Murphy. Dimitrios Komaris and Cheral Govind will (the student investigators) conduct as much of both tests as deemed appropriate whilst under the supervision of the named supervisor. Data analysis will be carried out by the student investigators.

What specific techniques will be employed and what exactly is asked of the participants? Please identify any non-validated scale or measure and include any scale and measures charts as an Appendix to this application. Please include questionnaires, interview schedules or any other non-standardised method of data collection as appendices to this application.

Where an independent reviewer is not used, then the UEC, DEC or SEC reserves the right to scrutinise the methodology. Has this methodology been subject to independent scrutiny? Yes ☐ No ☒
If yes, please provide the name and contact details of the independent reviewer:
N/A

17. Previous experience of the investigator(s) with the procedures involved.
Experience should demonstrate an ability to carry out the proposed research in accordance with the written methodology.

Dr Andrew Murphy has 14 years of experience studying and working in human movement science. Current appointment at the University of Strathclyde as a Research Fellowship under the University's Strategic Appointment and Investment Scheme and the Scottish Centre of Excellence in Rehabilitation Research. He holds an honorary contract with NHS GGC where, with clinical colleagues he helps to run a mobility service, primarily for paediatric Cerebral Palsy patients, as well as some other conditions that impair individuals' ability to walk. Sports Bio mechanist to the GB Olympic Rowing team for 4.5 years. Dr Murphy was awarded "2008-2010 Best Medical Engineering PhD" by the Institution of Mechanical Engineers, he sits on the Executive Council of the European Society of Movement Analysis and has published and presented his work extensively and internationally. He is also a certified CAREN operator.

Phil Riches (B.Eng., M.Sc., M.Sc., Ph.D., C.Eng. (MIMechE) has been a lecturer in Biomechanics, research methodology and applied statistics at the University of Strathclyde for over 14 years. In this time, he has supervised around 40 undergraduate

and postgraduate student projects in the field of clinical and sports biomechanics using motion analysis and associated techniques. Phil is currently a PI and CI in various experimental projects, including clinical trials, involving orthopaedic implants and surgical techniques, and especially in the discipline of computer-assisted and robotic orthopaedic surgery. A member of both the International Society of Biomechanics (ISB) and the European Society of Biomechanics, Phil is currently the Scientific Officer for the forthcoming ISB congress in Glasgow in 2015.

Both Dimitrios and Cheral have completed an MSc and MRes respectively in Biomedical Engineering at the University of Strathclyde. This involved completing a Biomechanics module where data from the Vicon system was analysed. As well as this, both investigators have been participants for a previous study comparing the Vicon and Motek systems. Both students will hold a current GCP certificate as well as attending and completing upcoming training sessions and online courses prior to the commencement of the study.

18. Data collection, storage and security

How and where are data handled? Please specify whether it will be fully anonymous (i.e., the identity unknown even to the researchers) or pseudo-anonymised (i.e. the raw data is anonymised and given a code name, with the key for code names being stored in a separate location from the raw data) - if neither please justify.

-Identifiable information

The consent forms from the participants will be kept confidential and stored for 5 years in a locked cabinet in the Department of Biomedical Engineering. The forms will be available for those named in this application only and will be destroyed after more than 5 years after completion of the study.

If consent is given by the participants, videos will also be taken. Participants will be identifiable from this but these will be stored on password protected non-networked hard drives with secure access only by the named researchers. Additionally, all the information will be saved as a backup in a password protected folder on password protected University of Strathclyde computers and external hard drives. If consent is given all videos will be kept indefinitely.

-Pseudo-anonymous data

A unique ID code will link the collected data to the participant. The code list will be stored in a locked cabinet in the Department of Biomedical Engineering. The coded list will only be available for those named in the application and will be destroyed 5 years after completion of the study. Thereby the pseudo-anonymous data will become anonymous.

All experimental data will be stored pseudo-anonymously, coded with an ID-number. Any videos will be coded with the same ID-number. All experimental data will be kept indefinitely, but will become fully anonymous 5 years after completion of the study; when the master file associating participants' names with their ID number is destroyed.

Explain how and where it will be stored, who has access to it, how long it will be stored and whether it will be securely destroyed after use:

Data will be securely stored and its access and destruction will be in accordance with the University of Strathclyde Data Protection Policy.

All computing systems holding electronic data, and all hard data will be stored within lock & key, and/or, magnetic swipe card security access enabled offices and laboratories within the Department of Biomedical Engineering of the University of Strathclyde. Videos and all other experimental data will be stored on password protected hard drives with secure access only by the named researchers

Will anyone other than the named investigators have access to the data? Yes ☐ No ☒
If 'yes' please explain:
N/A

19. Potential risks or hazards

Describe the potential risks and hazards associated with the investigation:
Please see attached risk assessments.

The Motek system may present health and safety risks to the participants if used incorrectly, for this reason the Motek system will only be operated by a certified user and in accordance with the system and safety manual. (Available upon request)

Has a specific Risk Assessment been completed for the research in accordance with the University's Risk Management Framework (Risk Management Framework)? Yes ☒ No ☐
If yes, please attach risk form (S20) to your ethics application. If 'no', please explain why not:

20. What method will you use to communicate the outcomes and any additional relevant details of the study to the participants?

The outcomes will not be debriefed to the participants, participants may contact the researchers following completion of the project for a brief report.

21. How will the outcomes of the study be disseminated (e.g., will you seek to publish the results and, if relevant, how will you protect the identities of your participants in said dissemination)?

This work is intended to be published as part of a PhD thesis, and presented on various poster presentation at University of Strathclyde research days.

The group aims to publish the results in peer-reviewed journals and present the results at national/international academic seminars and conferences.

8. Appendix

The researchers will seek consent from participants to record videos of the tests and to use these videos in presentations at scientific conferences, for teaching undergraduate and postgraduate students, and/or for University marketing collateral.

NB All participants will be given the opportunity to withhold consent regarding the use of video data as mentioned above whilst still permitting data to be collected for use in the analysis of experimental data (within the confines of this project).

Checklist	**Enclosed**	**N/A**
Participant Information Sheet(s)	☒	☐
Consent Form(s)	☒	☐
Sample questionnaire(s)	☐	☒
Sample interview format(s)	☐	☒
Sample advertisement(s)	☐	☐
Any other documents (please specify below)		
Risk Assessments	☒	☐
	☐	☐
	☐	☐
	☐	☐
	☐	☐

22. Chief Investigator and Head of Department Declaration
Please note that unsigned applications will not be accepted and both signatures are required

I have read the University's Code of Practice on Investigations involving Human Beings and have completed this application accordingly. By signing below, I acknowledge that I am aware of and accept my responsibilities as Chief Investigator under Clauses 3.11 – 3.13 of the Research Governance Framework and that this investigation cannot proceed before all approvals required have been obtained.

Signature of Chief Investigator 

Please also type name here: Andrew J Murphy

I confirm I have read this application, I am happy that the study is consistent with departmental strategy, that the staff and/or students involved have the appropriate expertise to undertake the study and that adequate arrangements are in place to supervise any students that might be acting as investigators, that the study has access to the resources needed to conduct the proposed research successfully, and that there are no other departmental-specific issues relating to the study of which I am aware.

Signature of Head of Department

Please also type name here

Date: / /

23. Only for University sponsored projects under the remit of the DEC/SEC, with no external funding and no NHS involvement

Head of Department statement on Sponsorship
This application requires the University to sponsor the investigation. This is done by the Head of Department for all DEC applications with exception of those that are externally funded and those which are connected to the NHS (those exceptions should be submitted to R&KES). I am aware of the implications of University sponsorship of the investigation and have assessed this investigation with respect to sponsorship and management risk. As this particular investigation is within the remit of the DEC and has no external funding and no NHS involvement, I agree on behalf of the University that the University is the appropriate sponsor of the investigation and there are no management risks posed by the investigation.

If not applicable, tick here ☐

Signature of Head of Department	
Please also type name here	
Date:	/ /

For applications to the University Ethics Committee, the completed form should be sent to ethics@strath.ac.uk with the relevant electronic signatures.

24. Insurance
The questionnaire below must be completed and included in your submission to the UEC/DEC/SEC:

Is the proposed research an investigation or series of investigations conducted on any person for a Medicinal Purpose? Medicinal Purpose means: ▪ treating or preventing disease or diagnosing disease or ▪ ascertaining the existence degree of or extent of a physiological condition or ▪ assisting with or altering in any way the process of conception or ▪ investigating or participating in methods of contraception or ▪ inducing anaesthesia or ▪ otherwise preventing or interfering with the normal operation of a physiological function or ▪ altering the administration of prescribed medication.	No

If "**Yes**" please go to **Section A (Clinical Trials)** – all questions must be completed

If "**No**" please go to **Section B (Public Liability)** – all questions must be completed

Section A (Clinical Trials)

Does the proposed research involve subjects who are either: i. under the age of 5 years at the time of the trial; ii. known to be pregnant at the time of the trial	N/A

*If "**Yes**" the UEC should refer to Finance*

Is the proposed research limited to: iii. Questionnaires, interviews, psychological activity including CBT; iv. Venepuncture (withdrawal of blood); v. Muscle biopsy; vi. Measurements or monitoring of physiological processes including scanning; vii. Collections of body secretions by non-invasive methods; viii. Intake of foods or nutrients or variation of diet (excluding administration of drugs).	N/A

*If "**No**" the UEC should refer to Finance*

Will the proposed research take place within the UK?	N/A

*If "**No**" the UEC should refer to Finance*

8. Appendix

Title of Research	N/A
Chief Investigator	N/A
Sponsoring Organisation	N/A

Does the proposed research involve:	
a) investigating or participating in methods of contraception?	N/A
b) assisting with or altering the process of conception?	N/A
c) the use of drugs?	N/A
d) the use of surgery (other than biopsy)?	N/A
e) genetic engineering?	N/A
f) participants under 5 years of age(other than activities i-vi above)?	N/A
g) participants known to be pregnant (other than activities i-vi above)?	N/A
h) pharmaceutical product/appliance designed or manufactured by the institution?	N/A
i) work outside the United Kingdom?	N/A

If **"YES"** to **any** of the questions a-i please also complete the **Employee Activity Form** (attached).

If **"YES"** to **any** of the questions a-i, and this is a follow-on phase, please provide details of SUSARs on a separate sheet.

*If "**Yes**" to any of the questions a-i then the UEC/DEC/SEC should refer to Finance (aileen.stevenson@strath.ac.uk).*

Section B (Public Liability)	
Does the proposed research involve :	
a) aircraft or any aerial device	No
b) hovercraft or any water borne craft	No
c) ionising radiation	No
d) asbestos	No
e) participants under 5 years of age	No

f) participants known to be pregnant	No
g) pharmaceutical product/appliance designed or manufactured by the institution?	No
h) work outside the United Kingdom?	No

*If **"YES"** to any of the questions the UEC/DEC/SEC should refer to Finance(aileen.stevenson@strath.ac.uk).*

For NHS applications only - Employee Activity Form

Has NHS Indemnity been provided?	Yes / No
Are Medical Practitioners involved in the project?	Yes / No
If YES, will Medical Practitioners be covered by the MDU or other body?	Yes / No

This section aims to identify the staff involved, their employment contract and the extent of their involvement in the research (in some cases it may be more appropriate to refer to a group of persons rather than individuals).

Chief Investigator		
Name	**Employer**	**NHS Honorary Contract?**
Dr Andrew J. Murphy		No
Others		
Name	**Employer**	**NHS Honorary Contract?**
Dr Philip Riches		No
Miss Cheral Govind		No
Mr Dimitrios Komaris		No
Dr Craig Childs		No

Please provide any further relevant information here:

N/A

Appendix 6: Departmental Participant Information Sheet

Participant Information Sheet

Name of department: Biomedical Engineering
Title of the study: Clinical investigation of the functional outcomes of high congruency versus low congruency knee bearings.

Introduction

Before you decide if you would like to be part of this study, it is important for you to understand why the research is being done and what you would need to do.

Please take the time to read the following information carefully and discuss it with others in order to decide whether or not you would like to take part.

You are free to chose whether to not to take part. If you decide not to take part this will not affect your relationship with the University of Strathclyde or any of its members.
Please contact us if there is anything that is not clear or you would like more information:

Chief Investigator: Dr Andrew Murphy
Status: Research Fellow
Department: Biomedical Engineering/Strathclyde University
Telephone: 01415482855
E-mail: andrew.j.murphy@strath.ac.uk

Research Student: Mr Dimitrios Sokratis Komaris
Status: PhD student
Department/Institution: Biomedical Engineering/Strathclyde University
Name of supervisor: Andrew J Murphy
Telephone: 07873563209
E-mail: dimitrios.komaris@strath.ac.uk

Research Student: Miss Cheral Govind
Status: PhD student
Department/Institution: Biomedical Engineering/Strathclyde University
Name of supervisor: Andrew J Murphy
Telephone: 07599214067
E-mail: cheral.govind@strath.ac.uk

What is the purpose of this investigation?

This study will determine any functional differences during activities of daily living, between a group of healthy individuals and patient groups who have undergone a total knee replacement surgery. The purpose of this study is to compare and evaluate the function of your knee prosthesis compared to the performance of healthy participants.

Do you have to take part?

No, participation is voluntary and it is up to you to decide. Use this information sheet to help you decide. Discuss it with family and friends or speak to a member of the research team for more information.

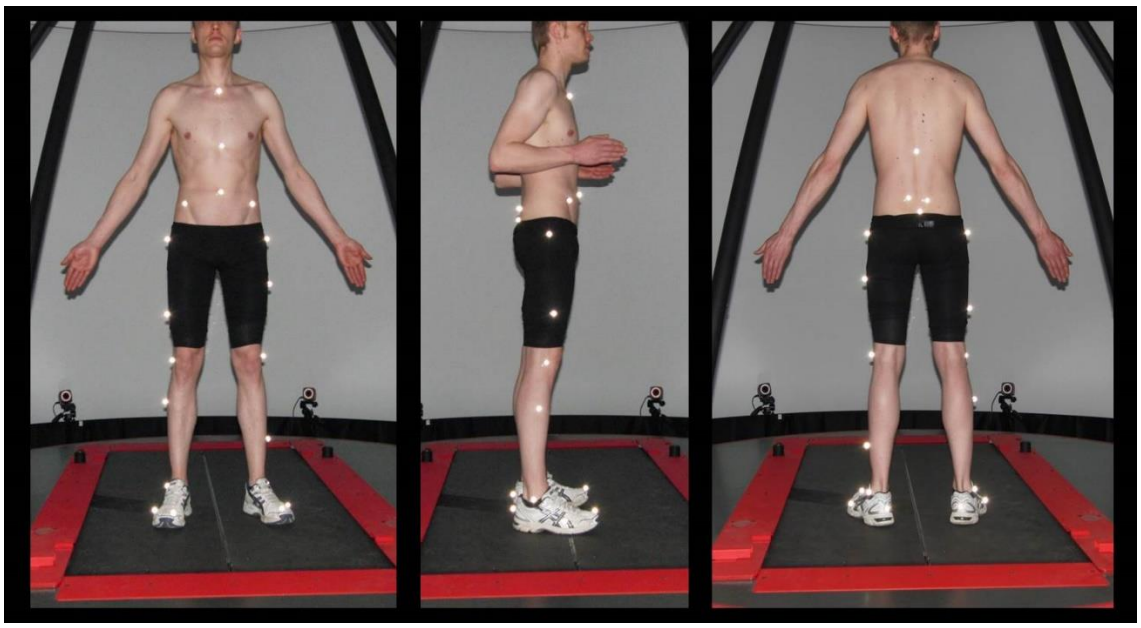
You will be expected to take part in this investigation voluntarily and it is up to you if you decide to refuse to participate before or during the investigation itself, if you decide to withdraw from the study (which you don't need to give a reason for), your data records will be removed. This will not in any way affect your relationship with the University of Strathclyde or any of its members.

What will you do in the project?

You, as a healthy volunteer, will be asked to attend a session (location given below) for no longer than 4 hours (1 hour of which will be solely dedicated to activities of daily living such as walking, single leg balance and ascending stairs). You will need to wear appropriate clothing so that accurate motion of the body while moving can be recorded (appropriate clothing will be provided by the department, if necessary). Male participants will be required to wear tight cycling type shorts with no top on. Female participants will be required to wear tight cycling type shorts and a tight fitting crop type top. You will be required to bring sports type shoes. Reflective markers that will be detected by the cameras will be stuck on you (as depicted below) using medical grade non-allergic tape and you will be asked to perform a set of functional tasks as so:

9. sit-to-stand
10. sit-to-stand-walk
11. ascending and descending stairs
12. walking on an incline
13. level walking in a straight line
14. level walking along an S-shaped pathway
15. single leg balance

Tasks 1-3 will be held in a motion capture laboratory which resembles a common room with a number of motion tracking cameras installed in it. Task 4 will be held in the Motek Laboratory where the participant will be asked to walk on a treadmill with a virtually reality aid. Finally, tasks 5-6 will be held in both laboratories.



A male participant affixed with reflective markers while standing on the treadmill found in the Motek Laboratory.

After every task, you will be allowed to rest if necessary. There could be times when you will be video recorded and photographed, but only if you agree to this beforehand. This experiment offers no incentives for application or reimbursements to potential participants. The laboratory session will take place in the following location:

The University of Strathclyde,
Department of Biomedical Engineering,
Biomechanics laboratory 2,
Level 1,
Wolfson Centre,
106 Rottenrow,
G4 0NW.

Why have you been invited to take part?

We are recruiting healthy adults over the age of 18, falling under the following criteria:

Inclusion criteria	Exclusion criteria
• Able bodied • 5'2" to 6'2' in height. • Normal lower limb function	• Musculoskeletal, neurological or sensory deficit • Previous ankle surgery

- Knowledge of using a treadmill
- 20/20 vision (with or without visual aid)
- Previous hip or knee replacement procedure
- Pregnancy
- Unable to give written consent
- Unable to attend the gait analysis sessions

What are the potential risks to you in taking part?
This study offers no potential risks to you.

What happens to the information in the project?
You will be given time to decide whether you'd like to be considered for participation in the study. Furthermore, you will be asked to consider whether you'd wish to provide consent for the following:

•Consent to being photographed and video recorded as part of the project.
•Consent for unidentifiable photographs and video recordings to be used in publications or teaching materials.

Any identifiable information: the consent form will be kept confidential, stored for 5 years in a locked cabinet in the Department of Biomedical Engineering. These will be available for those named in this application and will be destroyed on more than 5 years after completion of the study. As previously mentioned, if you give consent video recording may also be taken. You may be identifiable from this but these videos will be stored on password protected non-networked hard drives with secure access only by the named researchers. Additionally, all the information will be saved as a backup in a password protected folder on password protected University of Strathclyde computers and external hard drives. If consent is given, all videos will be kept indefinitely.

Any pseudo-anonymous data (anonymised raw data and given a code name, with the key for code names being stored in a separate location from the raw data): an ID code will link the collected data to the participant. The code list will be stored in a locked cabinet in the Department of Biomedical Engineering. The coded list will only be available for those named in the application and will be destroyed 5 years after completion of the study. Thereby the pseudo-anonymous data will become anonymous. All experimental data will be stored pseudo-anonymously, coded with an ID-number. All videos will be coded with the same ID-number. All experimental data will be kept indefinitely, but will become fully anonymous 5 year after completion of the study; when the master file associating participants' names with their ID number is destroyed. Data will be securely stored and its access and destruction will be in accordance with the University of Strathclyde Data Protection Policy. All computing

systems holding electronic data, and all hard data will be stored within lock & key, and/or, magnetic swipe card security access enabled offices and laboratories within the Department of Biomedical Engineering of the University of Strathclyde. Videos and all other experimental data will be stored on password protected hard drives with secure access only by the named researchers.

The University of Strathclyde is registered with the Information Commissioner's Office who implements the Data Protection Act 1998. All personal data on participants will be processed in accordance with the provisions of the Data Protection Act 1998.

What happens next?

Once you agree with the information given above and you decide if you would like to participate in this research study, all that will have to be done is for you to read and sign the consent form. This should be then handed to any of the investigators/researchers mentioned in the following section.

If you do not wish to be involved in the project, then the investigators of this study would like to take the opportunity to thank you for taking interest in this research project.

If you would like to receive feedback about the progress of the study post-testing then you are encouraged to contact any of the investigators on the contact details given below. If any of the results from this study will be published you will be informed beforehand.

Researcher contact details:

Research Student: Mr Dimitrios Sokratis Komaris
Status: PhD student
Department/Institution: Biomedical Engineering/Strathclyde University
Name of supervisor: Andrew J Murphy
Telephone: 07873563209
E-mail: dimitrios.komaris@strath.ac.uk

Research Student: Miss Cheral Govind
Status: PhD student
Department/Institution: Biomedical Engineering/Strathclyde University
Name of supervisor: Andrew J Murphy
Telephone: 07599214067
E-mail: cheral.govind@strath.ac.uk

Chief Investigator details:

Full Name: Dr Andrew Murphy
Status: Research Fellow
Department: Biomedical Engineering/Strathclyde University
Telephone: 01415482855
E-mail: andrew.j.murphy@strath.ac.uk

Thank you for reading this information – please ask any questions if you are unsure about what is written here. This investigation was granted ethical approval by the University of Strathclyde Ethics Committee. If you have any further questions/concerns, during or after the investigation, or wish to contact an independent person to whom any questions may be directed or further information may be sought from, please contact:

Linda Gilmour
Secretary to the Departmental Ethics Committee
National Centre for Prosthetics and Orthotics
Department of Biomedical Engineering
Curran Building, 131 St James Road
Glasgow G4 0LS
Tel: 0141 548 3298 E-mail: linda.gilmour@strath.ac.uk

Appendix 7: Control Participant Consent Form

DEPARTMENT OF BIOMEDICAl ENGINEERING

University of Strathclyde Engineering

Consent Form for Participants

Name of department: Biomedical Engineering
Title of the study: Clinical investigation of the functional outcomes of high congruency versus low congruency knee bearings.

- I confirm that I have read and understood the information sheet for the above project and the researcher has answered any queries to my satisfaction.
- I understand that my participation is voluntary and that I am free to withdraw from the project at any time, without having to give a reason and without any consequences.
- I understand that I can withdraw my data from the study at any time without giving reason.
- I understand that any information recorded in the investigation will remain confidential and no information that identifies me will be made publicly available.
- I understand that whether I participate in the project or not will in no way affect my standing within the University of Strathclyde.
- I confirm that I meet the inclusion/exclusion criteria.
- I consent to being a participant in the project and for the collection, documentation and usage of data gathered during the experiment.
- I understand that incentives/reimbursements will not be offered for participation.

Optional:

- I consent to the use of unidentifiable audio and video data recorded as part of the project for educational purposes Yes/No
- I consent to the use of unidentifiable audio and video data recorded as part of the project in future publications [delete which is not being used] Yes/ No

Full Name of Participant:	
Signature of Participant:	Date:

Appendix 8: Case Report Form

Case Report Form

Max Camera error:				
Force plate error:	**1)**	**2)**	**3)**	**4)**

VISIT: (TICK APPROPRIATE BOX)	**CONTROL**	**PRE-OP**	**POST-OP**	**1-YEAR-POST**

Age at surgery:	**Patient code** (IIYYYY)**:**
Affected leg:	**Session Code** (DDMONTHYY)**:**
	Visit date:

Has consent been obtained? **YES** ☐ **NO** ☐

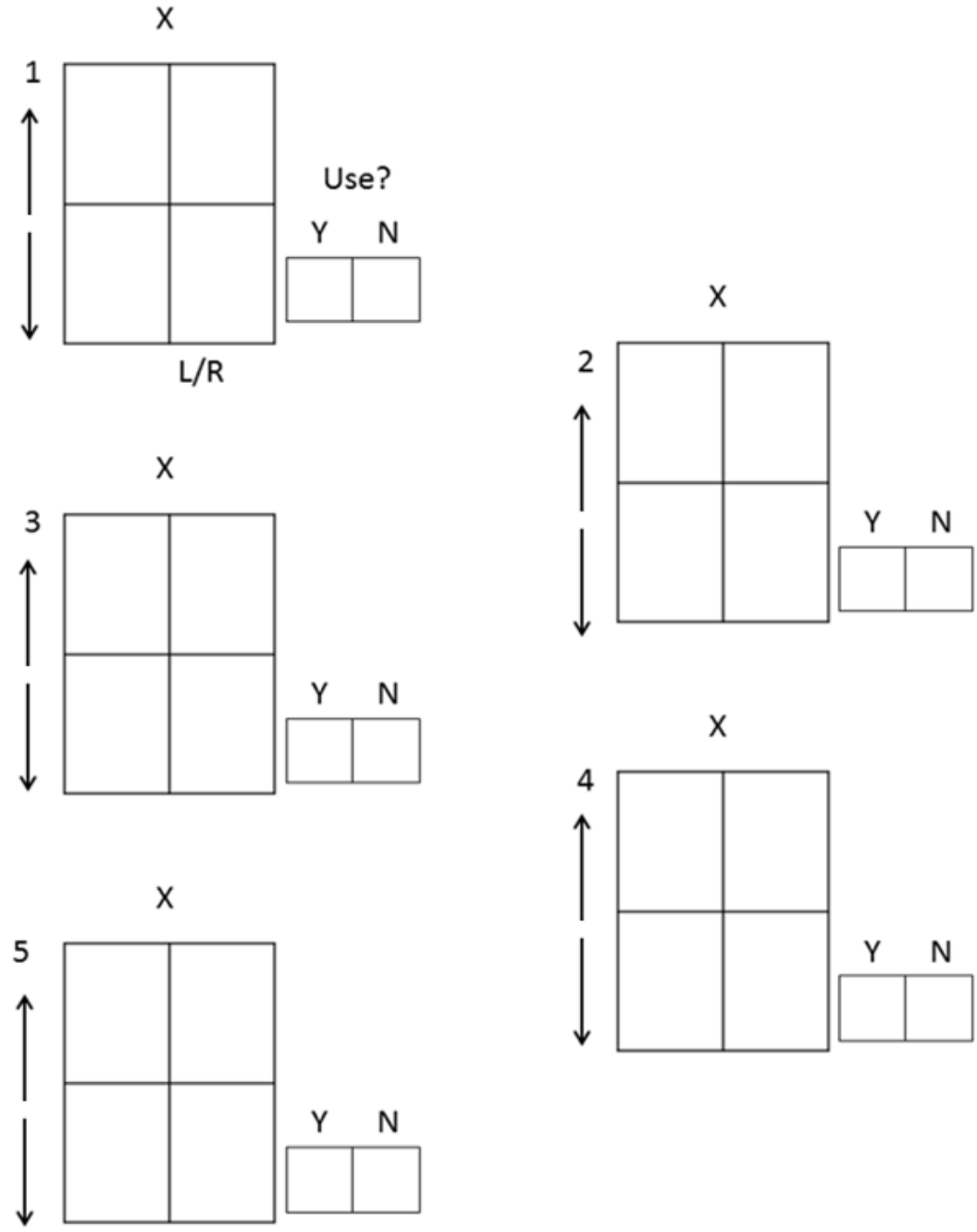
Subject details

Sex:	**Inter-ASIS distance:**
Age:	**Knee Height (mm):**
Height (mm):	
Weight (kg):	

LEFT	RIGHT
Leg Length (mm):	**Leg Length (mm):**
Knee Width (mm):	**Knee Width (mm):**
Ankle Width (mm):	**Ankle Width (mm):**
Shoulder offset (mm):	**Shoulder offset (mm):**
Elbow Width (mm):	**Elbow Width (mm):**
Wrist width (mm):	**Wrist width (mm):**
Hand thickness (mm):	**Hand thickness (mm):**

Weeks after surgery:	

Overground level walking: trial 1 to 5 / 6 to 10



Average walking speed: (m/s)

Overground level walking questionnaire

1) Did you have any difficulty during the task?				
No difficulty at all	Very little difficulty	Moderate difficulty	Extreme difficulty	Impossible to do

2) How would you describe any pain felt during the task?				
None	Very mild	Mild	Moderate	Severe

3) Was the task tiring for you?				
Not tiring at all	Slightly tiring	Moderately tiring	Extremely tiring	Impossible to do

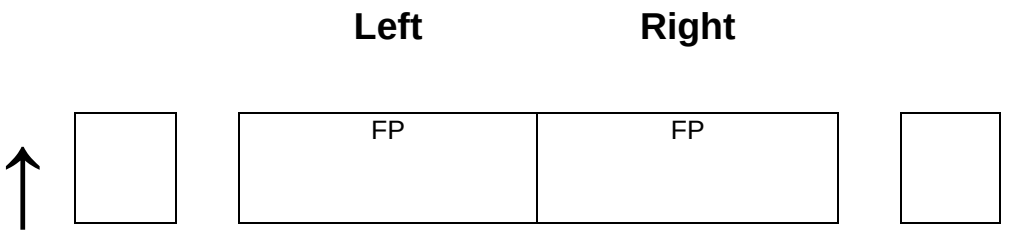
4) Altogether for this task, how do you feel about the length of time you walked?				
It was fine	It was a little too long	It was too long, but it was tolerable	It was too long and difficult	It is impossible for me to walk that much

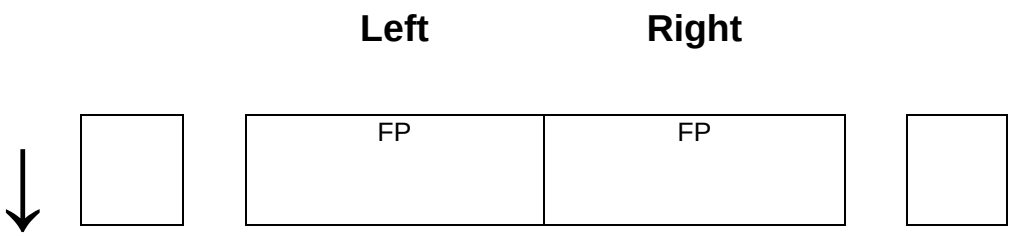
5) Could you usually walk more or less than this in one go without a break (with or without a walking aid)?				
I could walk at least 15 minutes longer with no break	I could walk a few minutes longer without a break	It was ok	I could walk a few minutes less without a break	It is impossible for me to walk that much without a break

Do you have any other comments?

Stair ascent/descent trial 1- 5

- (Check markers).





BANNISTER	
NONE	
SINGLE	
DOUBLE	

STRATEGY	
Step-over-step (1 foot per step)	
Step-by-step (both feet per step)	
Other	

Comments:

Stairs ascent/descent questionnaire

1) Did you have any difficulty during the task?				
No difficulty at all	Very little difficulty	Moderate difficulty	Extreme difficulty	Impossible to do

2) How would you describe any pain felt during the task?				
None	Very mild	Mild	Moderate	Severe

3) Was the task tiring for you?				
Not tiring at all	Slightly tiring	Moderately tiring	Extremely tiring	Impossible to do

4) Could you normally walk up a flight of stairs?				
Yes, easily	Yes, but with a little difficulty	Yes, but with moderate difficulty	Maybe, but with extreme difficulty	No, it is impossible to do

5) Could you normally walk down a flight of stairs?				
Yes, easily	Yes, but with a little difficulty	Yes, but with moderate difficulty	Maybe, but with extreme difficulty	No, it is impossible to do

6) Could you have performed this task without the banister?				
Yes when going up AND down the stairs	Yes, only on the way up	Yes, only on the way down	No, I need the banister on the way up and down	Not sure

7) Do you have stairs in your home?	
Yes	No

8) Other than the appearance of the stairs, do our stairs resemble yours? (Regarding step height/stair width/bannisters etc?)"	
Yes	No

If not, why?

9) Would you normally use a banister when ascending stairs?		
No banisters	Single banister	Two banisters

10) Would you normally use a banister when descending stairs?		
No banisters	Single banister	Two banisters

Do you have any other comments?

Walking on an incline

Maximum Camera error: _____

- Randomise order after level > ascending/descending

Level Walking

Trial 1	-7.5 -5 -2.5 **0** 2.5 5 7.5
Comments:	

Ascending/descending

Trial 4	-7.5 -5 -2.5 0 2.5 5 7.5
Comments:	

Trial 5	-7.5 -5 -2.5 0 2.5 5 7.5
Comments:	

Trial 6	-7.5 -5 -2.5 0 2.5 5 7.5
Comments:	

Ascending/descending

Trial 7	-7.5 -5 -2.5 0 2.5 5 7.5
Comments:	

Trial 8	-7.5 -5 -2.5 0 2.5 5 7.5
Comments:	

Trial 9	-7.5 -5 -2.5 0 2.5 5 7.5
Comments:	

Walking on an incline questionnaire

1) Did you have any difficulty during the task?				
No difficulty at all	Very little difficulty	Moderate difficulty	Extreme difficulty	Impossible to do

2) How would you describe any pain felt during the task?				
None	Very mild	Mild	Moderate	Severe

3) Was the task tiring for you?				
Not tiring at all	Slightly tiring	Moderately tiring	Extremely tiring	Impossible to do

4) Did the harness limit your mobility in any way?				
No, not at all	Yes, but just a little bit	Not sure	Yes, but it was tolerable	Yes, a lot

5) Did the harness make you feel safe during the task?				
Yes, very safe	Yes, fairly safe	Not sure	No, a little unsafe	No, very unsafe

6) How did you feel about the inclinations compared to everyday life?				
Extremely steep	A little too steep	About right	A little too shallow	Extremely shallow

Do you have any other comments?

Appendix 9: Custom MATLAB Scripts

9.1. Overground Level Walk

9.1.1. Create Events

```
% Script to be ran as a Vicon Nexus pipeline to create heel strike events if
% autodetect doesn't work well- have to create Toe off manually/via autocorelate
% events in Nexus. Algorithms from Zeni 2008 - C.Govind

clc
clear
vicon = ViconNexus;
```

```
SubjectName = vicon.GetSubjectNames;
[path, TrialName]=vicon.GetTrialName;

Sides={'Left','Right'}; %j

% check for heel strikes
% preallocate
Jar = NaN(2,2);
HS = NaN(1,2);

for j = 1:2
 Jar = vicon.GetEvents(SubjectName{1}, Sides{j},'Foot Strike'); %top = left, bottom =
right

 cmp1 = num2str(size(Jar,2));
 IsTwo = strcmp(cmp1,'2'); %if jar has 2 columns, 2 feet strikes
 if str2double(cmp1) >= 2
  IsTwo = strcmp(cmp1,cmp1);
 end

 IsOne = strcmp(cmp1,'1');
 IsZero = strcmp(cmp1,'0');

 if IsTwo == 0 && IsOne == 0 && IsZero == 1 %there are no strikes on that sides
  HS(j) = 2;
 elseif IsTwo == 0 && IsOne == 1 && IsZero == 0 %there is one strike on that side
  HS(j) = 1;
 elseif IsTwo == 1 && IsOne == 0 && IsZero == 0 %there are two strikes on that side
  HS(j) = 0;
 end
 clearvars IsOne IsTwo IsZero
end

%HS(1) = number of heel strikes needed in LEFT side
%HS(2) = number of heel strikes needed in RIGHT side

clearvars Jar cmp1 j

% to give sides
if HS(1) ~= 0 && HS(2) ~= 0
 Sides = {'Left','Right'};
elseif HS(1) ~= 0 && HS(2) == 0
 Sides = {'Left'};
elseif HS(1) == 0 && HS(2) ~= 0
 Sides = {'Right'};
 % elseif HS(1) == 0 && HS(2) == 0 %neither side needs any HS
 %  Sides = {'Left','Right'};
end

MarkerNames = vicon.GetMarkerNames(SubjectName{1});
MarkerNames2= strfind(MarkerNames, 'HEE'); %import heel marker names
MarkerNames3 = find(not(cellfun('isempty',MarkerNames2)));
Heels = MarkerNames(MarkerNames3); %i
MarkerNames4= strfind(MarkerNames, 'TOE'); %import heel marker names
MarkerNames5 = find(not(cellfun('isempty',MarkerNames4)));
Toes = MarkerNames(MarkerNames5); %i
MarkerNames6= strfind(MarkerNames, 'PSI'); %import heel marker names
MarkerNames7 = find(not(cellfun('isempty',MarkerNames6)));
Pelv = MarkerNames(MarkerNames7); %i
clearvars MarkerNames MarkerNames2 MarkerNames3 MarkerNames4 MarkerNames5...
 MarkerNames6 MarkerNames7

clearvars MarkerNames MarkerNames2 MarkerNames3
```

```
[Start, End]= vicon.GetTrialRegionOfInterest;

%get Heel marker for the required side

IsLR = strcmp(Sides,{'Left','Right'});

if IsLR (1) == 1 && IsLR (2) == 0
 Heels = cellstr(Heels{1});
elseif IsLR (1) == 0 && IsLR (2) == 1
 Heels = cellstr(Heels{2});
elseif IsLR (1) == 1 && IsLR (2) == 1
 Heels = Heels;
end

clearvars IsLR

[LPSIx,~,~,~] = vicon.GetTrajectory((SubjectName{1}),(Pelv{1}));
[RPSIx,~,~,~] = vicon.GetTrajectory((SubjectName{1}),(Pelv{2}));
LPSIx = LPSIx(Start:End);
RPSIx = RPSIx(Start:End);
AvPSI = (LPSIx + RPSIx)/2;
clearvars LPSIx RPSIx
```

```
% create HS
for i = 1:size(Heels,2) %check size is right if 2 heels in loop
 [x1, ~, ~, ~] = vicon.GetTrajectory((SubjectName{1}),(Heels{i}));
 x1 = x1(Start:End); %get HeelX coordinates within region of interest
 RelHeel2 = x1 - AvPSI;
 VelX = diff(RelHeel2)/0.01; % v = d (/dt)differenciate traj to find velocity
 VelX = circshift(VelX,[1,1]); %shift all values across 1
 VelX(1)= 1; %ignore first value
 VelX(VelX<0) = -1; %less than 0, make 0
 VelX(VelX>0) = 1; %more than 0, make 1
 for a = 1:(size(VelX,2)-1)
  b(a) = VelX(a) - VelX(a+1); %all subtractions
 end
 c = b == -2; %find where -2
 c = find(c);
 for d = 1:size(c,2)
  vicon.CreateAnEvent((SubjectName{1}),(Sides{i}),'Foot Strike', (c(d)+Start), 0)
 end
 clearvars a b c d VelX

end


% create TO
for i = 1:2
 [x2, ~, ~, ~] = vicon.GetTrajectory((SubjectName{1}),(Toes{i}));
 x2 = x2(Start:End); %get ToeX coordinates within region of interest
 RelToes = x2 - AvPSI;

 VelX = diff(RelToes)/0.01; % v = d (/dt)differenciate traj to find velocity
 VelX = circshift(VelX,[1,1]); %shift all values across 1
 VelX(1)= 1; %ignore first value
 VelX(VelX<0) = -1; %less than 0, make 0
 VelX(VelX>0) = 1; %more than 0, make 1
 for a = 1:(size(VelX,2)-1)
  b(a) = VelX(a) - VelX(a+1); %all subtractions
 end
 c = b == 2; %find where 2- where velocity turns negative = toe off
 c = find(c);
 for d = 1:size(c,2)
  vicon.CreateAnEvent((SubjectName{1}),(Sides{i}),'Foot Off', (c(d)+Start), 0)
 end
end
```

9.1.2. Export Data

```
% Script to be ran as a pipeline in Vicon Nexus, exports and normalises
% kinematic data and kinetic data where available. Data normalised is
% between two consecutive heel strikes, any toe off events outside of heel
% strike is ignored. To export heel strikes, select the number of the heel
% strike event in nexus in a pop up.Impulse also calculated as the area
% under Moment/time graph- C.Govind

clc
clear
```

```
vicon = ViconNexus;
```

```
SubjectName = vicon.GetSubjectNames;
[path, TrialName]=vicon.GetTrialName;
TrialName=strrep(TrialName,' ','');

newpath = 'H:\Raw Data\NexusOutput\BBraunMasterMotionCaptureDatabase\';

% load master output Struct
load([newpath 'Output.mat'])

splitStr = regexp(path,'\','split'); %splits up path where "\" is < to retrieve
participant group
[Start, End]= vicon.GetTrialRegionOfInterest;
```

```
% export data to Output structure

Sides={'Left','Right'}; %j
Joint={'Ankle','Knee','Hip','Pelvis'}; %i
TrialType = {'LevelWalking',...
 'StairsAscent','StairsDescent'};  %2, 3

splitStr{7} = TrialType{1};

%preallocate
ToNorm = zeros (1);
Norm = zeros (1);
AvAng = NaN (101,3);
str1 = TrialName;


for j= 1:2 %L and Rtoe
 [HSF,~] = vicon.GetEvents(SubjectName{1}, Sides{j}, 'Foot Strike');
 for i= 1:4 %joints
  TempVar=vicon.GetModelOutput(SubjectName{1}, [Sides{j}(1) Joint{i} 'Angles'])';

  for l = 1:length(HSF)

   if l <= length(HSF)-1
    ToNorm(:,:) = TempVar((HSF(l)):(HSF(l+1)),:);
   elseif l == length(HSF)
    continue
   end

   for m = 1:3
    Norm(:,m) = interpft(ToNorm(:,m),101);
   end
   str2 = num2str(l);
   OutputTrialName = strcat(str1,str2);

Output.(splitStr{7}).(splitStr{5}).(splitStr{6}).(OutputTrialName).(Sides{j}).([Sides
{j}(1) Joint{i} 'Angles'])=Norm ; %save in output
   clearvars ToNorm Norm
  end
 end
 clearvars TempVar
end
clearvars j m l
save([newpath 'Output.mat'],'Output')
```

```
%Get TO's as percentage of cycle
PercentTO = NaN(1);

[path, TrialName]=vicon.GetTrialName;
TrialName=strrep(TrialName,' ','');

for j = 1:2
 HSF = vicon.GetEvents(SubjectName{1}, Sides{j},'Foot Strike'); %top = left, bottom =
right
 TOF = vicon.GetEvents(SubjectName{1}, Sides{j},'Foot Off');

 if size(HSF,2) == 1
  continue
 elseif HSF(1) < TOF(1) %check got less TOF than HSF
  for a = 1:(size(HSF,2)-1)
   PercentTO(a) =(((TOF(a))- (HSF(a)))*100) /(HSF(a+1)- HSF(a));
  end
 else HSF(1) > TOF(1);
  for a = 1:(size(HSF,2)-1)
```

```
    PercentTO(a) =(((TOF(a+1))- (HSF(a)))*100) /(HSF(a+1)- HSF(a));
   end
  end

 Output.(splitStr{7}).(splitStr{5}).(splitStr{6}).ToeOffs.([TrialName Sides{j}(1)
'ToeOff'])=PercentTO;
 clearvars PercentTO
 save([newpath 'Output.mat'],'Output')
end


clearvars AvTO j TOF a StdDevTO TOStdDev j
```

```
% Export Kinetics

Joint = {'Ankle','Knee','Hip'};
Parameters = {'Moment','Power'}; %n


prompt = {'Which left heel strike?', 'Which right heel strike?'};
dlg_title = 'Input';
num_lines = 1;
defaultans = {'0','0'};
answer = inputdlg(prompt,dlg_title,num_lines,defaultans);

S = sprintf('%s*', answer{:});
HSNo = sscanf(S, '%f*');

if answer{1} == '0' && answer {2} == '0'
 disp ('No clean strikes');
 return
end
if answer{1} == '0' && answer{2} ~= '0' || answer{1} ~= '0' && answer{2} == '0'
 disp ('Single clean strike');
 x = find(HSNo);
 if x == 1
  Sides = {'Left'};
 elseif x == 2
  Sides = {'Right'};
 end
end

HSNo = HSNo(HSNo~=0);

str1 = TrialName;

for j = 1:length(Sides)
 exist x
 if ans == 1
  str2 = num2str(HSNo);
 else
  str2 = char(answer(j));
 end
 for i = 1:3 %for ankle knee and hip
  for n = 1:2 %get moment and power
   [HSF,~] = vicon.GetEvents(SubjectName{1}, Sides{j}, 'Foot Strike');
   TempVar=vicon.GetModelOutput(SubjectName{1}, [Sides{j}(1) Joint{i} Parameters{n}]
)';
   ToNorm = TempVar(HSF((HSNo(j))):HSF((HSNo(j))+1),:);
   %if gap in kinetics fill gap with average of point below and after
   for k = 1:length(ToNorm)
    if sum(ToNorm(k,:)) == 0 && k ~= length(ToNorm)
     MeanX = (ToNorm(k-1,1)+ToNorm(k+1,1))/2;
     MeanY = (ToNorm(k-1,2)+ ToNorm(k+1,2))/2;
     MeanZ = (ToNorm(k-1,3)+ ToNorm(k+1,3))/2;
     ToNorm(k,:) = [MeanX, MeanY, MeanZ];
    end
   end

   Norm = interpft(ToNorm,101)/1000;
   OutputTrialName = strcat(str1,str2);

Output.(splitStr{7}).(splitStr{5}).(splitStr{6}).(OutputTrialName).(Sides{j}).([Sides
{j}(1) Joint{i} Parameters{n}])=Norm;
  end
 end
 clearvars i n

 %calculate impulse during stance
 for i = 1:3

  [HSF,~] = vicon.GetEvents(SubjectName{1}, Sides{j}, 'Foot Strike');
```

```
  [TOF,~] = vicon.GetEvents(SubjectName{1}, Sides{j}, 'Foot Off');

  TempVar = vicon.GetModelOutput(SubjectName{1}, [Sides{j}(1) Joint{i} Parameters{1}]
)';

  if HSF((HSNo(j))) > TOF((HSNo(j)))
   TOF = TOF((HSNo(j)+1));
   ToImpulse = TempVar(HSF(HSNo(j)):TOF(1),:);
  else
   ToImpulse = TempVar(HSF(HSNo(j)):TOF(HSNo(j)),:);
  end

  Impulse = trapz(ToImpulse)/1000;
  OutputTrialName = strcat(str1,str2);
  Output.(splitStr{7}).(splitStr{5}).(splitStr{6}).(OutputTrialName).(Sides{j})...
   .([Sides{j}(1) Joint{i} 'Impulse'])= Impulse; %gives X Y Z impulses
 end

 disp ('Kinetics exported')
 save([newpath 'Output.mat'],'Output')
end


% save
save([newpath 'Output.mat'],'Output')
disp('Output saved')
```

9.2. Stair Navigation

9.2.1. Create Events

```
% Script to be ran as a Vicon Nexus Pipeline to detect Heel strike and Toe
% off events in Stair ascent or descent activities in participants who use
% a step over step strategy. Perhaps need to manually delete heel strike
% event close to a toe off. Works best with filtered trajectory/model
% output data. C. Govind


clc
clear

vicon = ViconNexus;
```

```
SubjectName = vicon.GetSubjectNames;
[path, TrialName]=vicon.GetTrialName;
[Start, End]= vicon.GetTrialRegionOfInterest;

Sides={'Left','Right'}; %j
Stairs = {'StairsAscent','StairsDescent'};

[~, ~, z, ~] = vicon.GetTrajectory((SubjectName{1}),'RKNE'); %Get vertical trajectory
of toe
z = z(Start:End);

if z(1) > z(end)
 Stairs = Stairs{2};
elseif z(1) < z(end)
 Stairs = Stairs{1};
end


% check for heel strikes
% preallocate
Jar = NaN(2,2);
HS = NaN(1,2);

for j = 1:2
 Jar = vicon.GetEvents(SubjectName{1}, Sides{j},'Foot Strike'); %top = left, bottom =
right

 cmp1 = num2str(size(Jar,2));
 IsTwo = strcmp(cmp1,'2'); %if jar has 2 columns, 2 feet strikes
 if str2double(cmp1) >= 2
  IsTwo = strcmp(cmp1,cmp1);
 end
```

```
 clearvars Jar

 IsOne = strcmp(cmp1,'1');
 IsZero = strcmp(cmp1,'0');
 clearvars cmp1

 if IsTwo == 0 && IsOne == 0 && IsZero == 1 %there are no strikes on that sides
  HS(j) = 2;
 elseif IsTwo == 0 && IsOne == 1 && IsZero == 0 %there is one strike on that side
  HS(j) = 1;
 elseif IsTwo == 1 && IsOne == 0 && IsZero == 0 %there are two strikes on that side
  HS(j) = 0;
 end
 clearvars IsOne IsTwo IsZero
end

%HS(1) = number of heel strikes needed in LEFT side
%HS(2) = number of heel strikes needed in RIGHT side

clearvars Jar cmp1 j

% to give sides
if HS(1) ~= 0 && HS(2) ~= 0
 Sides = {'Left','Right'};
elseif HS(1) ~= 0 && HS(2) == 0
 Sides = {'Left'};
elseif HS(1) == 0 && HS(2) ~= 0
 Sides = {'Right'};
 % elseif HS(1) == 0 && HS(2) == 0 %neither side needs any HS
 %  Sides = {'Left','Right'};
end

MarkerNames = vicon.GetMarkerNames(SubjectName{1});
MarkerNames2= strfind(MarkerNames, 'HEE'); %import heel marker names
MarkerNames3 = find(not(cellfun('isempty',MarkerNames2)));
Heels = MarkerNames(MarkerNames3); %i
MarkerNames4= strfind(MarkerNames, 'TOE'); %import heel marker names
MarkerNames5 = find(not(cellfun('isempty',MarkerNames4)));
Toes = MarkerNames(MarkerNames5); %i
MarkerNames6= strfind(MarkerNames, 'PSI'); %import heel marker names
MarkerNames7 = find(not(cellfun('isempty',MarkerNames6)));
Pelv = MarkerNames(MarkerNames7); %i
clearvars MarkerNames MarkerNames2 MarkerNames3 MarkerNames4 MarkerNames5...
 MarkerNames6 MarkerNames7

[Start, End]= vicon.GetTrialRegionOfInterest;

%get Heel marker for the required side

IsLR = strcmp(Sides,{'Left','Right'});

if IsLR (1) == 1 && IsLR (2) == 0
 Heels = cellstr(Heels{1});
elseif IsLR (1) == 0 && IsLR (2) == 1
 Heels = cellstr(Heels{2});
elseif IsLR (1) == 1 && IsLR (2) == 1
 Heels = Heels;
end

clearvars IsLR
```

```
%create HS- different for stair ascent + descent
[~,~,LPSIz,~] = vicon.GetTrajectory((SubjectName{1}),(Pelv{1}));
[~,~,RPSIz,~] = vicon.GetTrajectory((SubjectName{1}),(Pelv{2}));
LPSIz = LPSIz(Start:End);
RPSIz = RPSIz(Start:End);
AvPSI = (LPSIz + RPSIz)/2;

if strcmp(Stairs,'StairsAscent')
 % HS = naxima in vertical ank/toe acceleration - FIND REF
 for i = 1:size(Heels,2) %check size is right if 2 heels in loop
  [~, ~, ToeZ, ~] = vicon.GetTrajectory((SubjectName{1}),(Toes{i}));
  ToeZ = ToeZ(Start:End);
  VelToe = diff(ToeZ)/0.01;
  VelToe= circshift(VelToe,[1,1]); %shift all values across 1
  VelToe(1)= NaN;
  AccToe = diff(VelToe)/0.01;
  AccToe= circshift(AccToe,[1,1]);
  AccToe(1)= NaN;
  [~,WhereMax] = findpeaks(AccToe,'MinPeakProminence',7000);
  TOF = WhereMax + double(Start) - 1;
  for d = 1:length(TOF)
   vicon.CreateAnEvent((SubjectName{1}),(Sides{i}),'Foot Strike', TOF(d), 0)
```

```
   end
  end

  % TO = local maxima in vertical displacement between the toe and pelvis.
  for i = 1:size(Heels,2) %check size is right if 2 heels in loop
   [~, ~, ToeZ,~] = vicon.GetTrajectory((SubjectName{1}),(Toes{i}));
   ToeZ = ToeZ(Start:End);
   DiffPelvToe = AvPSI - ToeZ;
   [~,WhereDiff] = findpeaks(DiffPelvToe,'MinPeakDistance',150); %make pop up? use
MinPeakHeight, 900?
   TOF = WhereDiff + double(Start) - 1;
   for d = 1:length(TOF)
    vicon.CreateAnEvent((SubjectName{1}),(Sides{i}),'Foot Off', TOF(d), 0)
   end
  end


elseif strcmp(Stairs,'StairsDescent')
  % HS = vertical velocity minima of the whole body centre of mass was used to define
touch down

  for i = 1:size(Heels,2) %check size is right if 2 heels in loop
   [~, ~, ToeZ, ~] = vicon.GetTrajectory((SubjectName{1}),(Toes{i}));
   ToeZ = ToeZ(Start:End);
   VelToe = diff(ToeZ)/0.01;
   VelToe= circshift(VelToe,[1,1]); %shift all values across 1
   VelToe(1)= NaN;
   AccToe = diff(VelToe)/0.01;
   AccToe= circshift(AccToe,[1,1]);
   AccToe(1)= NaN;
   [~,WhereMax] = findpeaks(AccToe,'MinPeakProminence',7000);
   TOF = WhereMax + double(Start) - 1;
   for d = 1:length(TOF)
    vicon.CreateAnEvent((SubjectName{1}),(Sides{i}),'Foot Strike', TOF(d), 0)
   end
  end

  %Toe off = max flexion of same side.
  for i = 1:size(Heels,2)
   KneeAngles = vicon.GetModelOutput(SubjectName{1}, [Sides{i}(1) 'KneeAngles'])';
   KneeFlexion = KneeAngles(Start:End)';
   [~,WhereMax] = findpeaks(KneeFlexion,'MinPeakProminence',50);
   TOF = WhereMax + double(Start) - 1;
   for d = 1:length(TOF)
    vicon.CreateAnEvent((SubjectName{1}),(Sides{i}),'Foot Off', TOF(d), 0)
   end
  end
  % OR TO = Foot acceleration/foot velocity algorithms (accel - maximum foot
horizontal accel
end

clearvars End h i Start VelZ z
```

9.2.2. Export Data

```
% Script to be ran as a pipeline in Vicon Nexus, exports and normalises
% kinematic data and kinetic data where available. Data normalised is
% between two consecutive heel strikes, if kinetics are clean then select so to
% export kinetics - or select no to export kinematics only. Impulse also calculated
% as the area under Moment/time graph- C.Govind.


clear
clc

vicon = ViconNexus;
```

```
SubjectName = vicon.GetSubjectNames;
[path, TrialName]=vicon.GetTrialName;
TrialName=strrep(TrialName,' ','');
[Start, End]= vicon.GetTrialRegionOfInterest;

newpath = 'H:\Raw Data\NexusOutput\BBraunMasterMotionCaptureDatabase\';


% load master output Struct
load([newpath 'Output.mat'])

expression = '\';
```

```
splitStr = regexp(path,expression,'split'); %splits up path where "\" is < to
retrieve participant group
```

```
Sides={'Left','Right'}; %jj
TrialName=strrep(TrialName,' ','');
Joint={'Ankle','Knee','Hip','Pelvis'}; %k
Stairs = {'StairsAscent','StairsDescent'};

%determine if ascending or descending trial
[~, ~, z, ~] = vicon.GetTrajectory((SubjectName{1}),'RKNE'); %Get vertical trajectory
of toe
z = z(Start:End);

if z(1) > z(end)
 Stairs = Stairs{2};
elseif z(1) < z(end)
 Stairs = Stairs{1};
end
clearvars z
IsEvents = {0,0};

%get L and R events
for jj = 1:2
 IsEvents{jj} = vicon.GetEvents(SubjectName{1}, Sides{jj}, 'Foot Strike');
end

ix=cellfun(@isempty,IsEvents);
if ix(1) == 0 && ix(2) == 0
 Sides = {'Left', 'Right'};
elseif ix (1) == 0 && ix(2) == 1
 Sides = {'Left'};
elseif ix(1) == 1 && ix(2) == 0
 Sides ={'Right'};
end

%get kinematics
for jj= 1:length(Sides) %L and Rtoe
 for k= 1:4 %joints
  [HSF,~] = vicon.GetEvents(SubjectName{1}, Sides{jj}, 'Foot Strike');
  TempVar=vicon.GetModelOutput(SubjectName{1}, [Sides{jj}(1) Joint{k} 'Angles'])';
  ToNorm = TempVar(HSF(1,1):HSF(1,2),:); %get angles to norm between toe off to toe
off
  Norm = interpft(ToNorm,101); %normalise angles to 100%
  Output.(Stairs).(splitStr{5}).(splitStr{6}).(TrialName).(Sides{jj}).([Sides{jj}(1)
Joint{k} 'Angles'])=Norm; %save in output
 end
end
clearvars jj IsEvents ix k
```

```
%Get TO's as percentage of cycle
PercentTO = NaN(1);
for j = 1:length(Sides)
 HSF = vicon.GetEvents(SubjectName{1}, Sides{j},'Foot Strike'); %top = left, bottom =
right
 HSF = HSF(HSF >= Start);

 TOF = vicon.GetEvents(SubjectName{1}, Sides{j},'Foot Off');
 TOF = TOF(TOF > Start);

 %check got 1 less TOF than HSF
 if size(HSF,2) == 1
  continue
 elseif HSF(1) < TOF(1) %check got less TOF than HSF
  for a = 1:(size(HSF,2)-1)
   PercentTO(a) =(((TOF(a))- (HSF(a)))*100) /(HSF(a+1)- HSF(a));
  end
 elseif HSF(1) > TOF(1);
  for a = 1:(size(HSF,2)-1)
   PercentTO(a) =(((TOF(a+1))- (HSF(a)))*100) /(HSF(a+1)- HSF(a));
  end
 end


 Output.(Stairs).(splitStr{5}).(splitStr{6}).(TrialName).(Sides{j}).([Sides{j}(1)
'ToeOff'])=PercentTO;
 clearvars HSF TOF PercentTO AvTO StDevTO TOStdDev
end

clearvars a StdDevTO j

save([newpath 'Output.mat'],'Output')
```

```
%export kinetics

Joint={'Ankle','Knee','Hip'}; %k
Parameters = {'Moment','Power'}; %n

choice = questdlg('Clean strikes?', ...
 'Stairs', ...
 'Yes','No','Yes');
% Handle response
switch choice
 case 'Yes'
  if length(Sides) == 2
   for j = 1:2
    HSF(j,:) = vicon.GetEvents(SubjectName{1}, Sides{j}, 'Foot Strike');
   end

   if HSF(1,1) > HSF(2,1) %if first Left HS is greater than R
    Sides ={'Left'};
    clearvars HSF
    [HSF,~] = vicon.GetEvents(SubjectName{1}, Sides{1}, 'Foot Strike');
   elseif HSF(1,1) < HSF(2,1)
    Sides ={'Right'};
    clearvars HSF
    [HSF,~] = vicon.GetEvents(SubjectName{1}, Sides{1}, 'Foot Strike');
   end

  elseif length(Sides) == 1

   for j = 1
    [HSF,~] = vicon.GetEvents(SubjectName{1}, Sides{j}, 'Foot Strike');
   end
  end

  for i = 1:3 %for ankle knee and hip
   for n = 1:2 %get moment and power
    TempVar=vicon.GetModelOutput(SubjectName{1}, [Sides{1}(1) Joint{i} Parameters{n}]
)';
    ToNorm = TempVar(HSF(1):HSF(2),:);
    %if gap in kinetics fill gap with average of point below and after
    for k = 1:length(ToNorm)
     if sum(ToNorm(k,:)) == 0 && k ~= 1 && k ~= length(ToNorm)
      MeanX = (ToNorm(k-1,1)+ToNorm(k+1,1))/2;
      MeanY = (ToNorm(k-1,2)+ ToNorm(k+1,2))/2;
      MeanZ = (ToNorm(k-1,3)+ ToNorm(k+1,3))/2;
      ToNorm(k,:) = [MeanX, MeanY, MeanZ];
     end
    end

    Norm = interpft(ToNorm,101)/1000;
    Output.(Stairs).(splitStr{5}).(splitStr{6}).(TrialName).(Sides{1}).([Sides{1}(1)
Joint{i} Parameters{n}])=Norm;
    save([newpath 'Output.mat'],'Output')
    clearvars TempVar
   end

   % calculate impulse
   [HSF,~] = vicon.GetEvents(SubjectName{1}, Sides{1}, 'Foot Strike');
   [TOF,~] = vicon.GetEvents(SubjectName{1}, Sides{1}, 'Foot Off');
   TempVar=vicon.GetModelOutput(SubjectName{1}, [Sides{1}(1) Joint{i} Parameters{1}]
)';
   if HSF(1) > TOF(1)
    TOF = TOF(2);
   end
   ToImpulse = TempVar((HSF(1)):TOF(1),:);
   for k = 2:length(ToImpulse)
     if sum(ToNorm(k,:)) == 0 && k ~= length(ToImpulse)
      MeanX = (ToImpulse(k-1,1)+ToImpulse(k+1,1))/2;
      MeanY = (ToImpulse(k-1,2)+ ToImpulse(k+1,2))/2;
      MeanZ = (ToImpulse(k-1,3)+ ToImpulse(k+1,3))/2;
      ToImpulse(k,:) = [MeanX, MeanY, MeanZ];
     end
   end

   Impulse = trapz(ToImpulse)/1000;
    Output.(Stairs).(splitStr{5}).(splitStr{6}).(TrialName).(Sides{1}).([Sides{1}(1)
Joint{i} 'Impulse'])=Impulse;
    save([newpath 'Output.mat'],'Output')

  end
  disp ('Impulse exported')

 case 'No'
  disp('No kinetic data');
```

```
end


save([newpath 'Output.mat'],'Output')
clearvars choice End expression i j jj Joint k n Norm Parameters Start TempVar ToNorm
a StdDevTO
```

9.3. Incline walking

9.3.1. Import Motek files

```
% Script to be run within Matlab to retrieve frame numbers corresponding to
% the start of 10 seconds of incline walking. Navigate to and multiselect
% relevant Dflow files. Select which file is the start of the new nexus
% trial and the relevant frame numbers will be given in the FrameNumber
% variable. C.Govind

clc
clear
```

```
%Get first timestamp of each recording

cd ('H:\Raw Data\DFlowOutput\BBraunMasterMotionCaptureDatabase')


[FileName,PathName] = uigetfile('*','MultiSelect','on');
numfiles = size(FileName,2);

incline = NaN(1,7);
for ii = 1:numfiles
 Fid =fullfile(PathName,FileName{ii});
 delimiter = '\t';
 startRow = 2;
 endRow = 2;
 formatSpec =
'%f%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%
*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*
s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s
%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%
*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*
s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%[^\n\r]';
 fileID = fopen(Fid,'r');
 dataArray = textscan(fileID, formatSpec, endRow-startRow+1, 'Delimiter', delimiter,
'HeaderLines', startRow-1, 'ReturnOnError', false);
 fclose(fileID);
 incline(:,ii) = [dataArray{1:end-1}];
end

clearvars filename delimiter startRow endRow formatSpec fileID dataArray ans Fid
numfiles PathName;
```

```
%Get matching Nexus frame

prompt = {'Enter trial number of new baseline:'};
dlg_title = 'Incline';
num_lines = 1;
defaultans = {'5'};
NextIncl = str2double(inputdlg(prompt,dlg_title,num_lines,defaultans));


FrameNumber = NaN(1,8);
%first nexus trial
for jj = 1:(NextIncl-1)
 FrameNumber(:,jj) = ((incline(jj)) - (incline(1)))*100; %mutiply by frequency to
convert to nexus frame number
end

%2nd nexus trial
for jj = NextIncl:length(incline)
 FrameNumber(:,jj) = (incline(jj) - (incline(NextIncl)))*100;
end

clearvars defaultans dlg_title NextIncl num_lines prompt jj FileName incline ii FN
```

9.3.2. Create Events

```
%% Script to be ran as a Vicon Nexus pipeline to create heel strike events if
% autodetect doesn't work well- have to create Toe off manually/via autocorelate
% events in Nexus. Algorithms from Zeni 2008 - C.Govind

clc
clear

vicon = ViconNexus;

SubjectName = vicon.GetSubjectNames;
[path, TrialName]=vicon.GetTrialName;
[Start, End]= vicon.GetTrialRegionOfInterest;
Sides={'Left','Right'}; %i

TrialName=strrep(TrialName,' ','');
MarkerNames = vicon.GetMarkerNames(SubjectName{1});
MarkerNames2= strfind(MarkerNames, 'HEE');
MarkerNames3 = find(not(cellfun('isempty',MarkerNames2)));
Heels = MarkerNames(MarkerNames3); %i
MarkerNames4= strfind(MarkerNames, 'TOE');
MarkerNames5= find(not(cellfun('isempty',MarkerNames4)));
Toes = MarkerNames(MarkerNames5);
clearvars MarkerNames MarkerNames2 MarkerNames3 MarkerNames4 MarkerNames5


b = zeros (1,10);

% Create heel strike
for i = 1:2 %check size is right if 2 heels in loop
 [~, y, ~, ~] = vicon.GetTrajectory((SubjectName{1}),(Heels{i}));
 y = y(Start:End); %get HeelY coordinates within region of interest
 VelY = diff(y)/0.01; % v = d (/dt)differenciate traj to find velocity
 VelY = circshift(VelY,[1,1]); %shift all values across 1
 VelY(1)= 1; %ignore first value
 VelY(VelY<0) = -1; %less than 0, make 0
 VelY(VelY>0) = 1; %more than 0, make 1
 for a = 1:(size(VelY,2)-1)
  b(a) = VelY(a) - VelY(a+1); %all subtractions
 end
 c = b == -2; %find where -2, where velocity turns positive = heel strike
 c = find(c);
 for d = 1:size(c,2)
  vicon.CreateAnEvent((SubjectName{1}),(Sides{i}),'Foot Strike', (c(d)+Start), 0)
 end
end

%create toe off- manually delete extra toe offs
for i = 1:2 %check size is right if 2 heels in loop
 [~, y, ~, ~] = vicon.GetTrajectory((SubjectName{1}),(Toes{i}));
 y = y(Start:End); %get ToeY coordinates within region of interest
 VelY = diff(y)/0.01; % v = d (/dt)differenciate traj to find velocity
 VelY = circshift(VelY,[1,1]); %shift all values across 1
 VelY(1)= 1; %ignore first value
 VelY(VelY<0) = -1; %less than 0, make 0
 VelY(VelY>0) = 1; %more than 0, make 1
 for a = 1:(size(VelY,2)-1)
  b(a) = VelY(a) - VelY(a+1); %all subtractions
 end
 c = b == 2; %find where 2- where velocity turns negative = toe off
 c = find(c);
 for d = 1:size(c,2)
  vicon.CreateAnEvent((SubjectName{1}),(Sides{i}),'Foot Off', (c(d)+Start), 0)
 end
end

clearvars a b c d IsLeft
```

9.3.3. Export Data

```
% Script to be ran as a pipeline in Vicon Nexus, exports and normalises
% kinematic data. Data normalised is between two consecutive heel strikes, any toe
% off events outside of heel strike is ignored. As 10second recording are taken, all
% kinematics are ordered into a 3D matrix and then averaged and saved into a
% structure - C.Govind
clc
clear
```

```
vicon = ViconNexus;
```

```
%% Reminder for incline slope labels:
%Incline1 = level
Incline2 = + 2.5
Incline3 = +5
Incline4 = +7.5
Incline5 = -2.5
Incline6 = -5
Incline7 = -7.5
```

```
SubjectName = vicon.GetSubjectNames;
[path, TrialName]=vicon.GetTrialName;
TrialName=strrep(TrialName,' ','');

newpath = 'H:\Raw Data\NexusOutput\BBraunMasterMotionCaptureDatabase\';

% load master output Struct
load([newpath 'AvData.mat'])
load([newpath 'Output.mat'])

splitStr = regexp(path,'\','split'); %splits up path where "\" is < to retrieve
participant group
```

```
%get affected side
[~, ~, affectedside] = xlsread('C:\Users\gwb13225\ShareFile\Personal Folders\Phd\3
Data\affected side.xlsx','Sheet1','A2:B88');


affectedside(cellfun(@(x) ~isempty(x) && isnumeric(x) && isnan(x),affectedside)) =
{''};

AffectedSide = strfind(affectedside,(SubjectName{1}));

ix=cellfun(@isempty,AffectedSide);
AffectedSide(ix)={0}; %convert blank cells to 0
clearvars ix

AffectedSide = cell2mat(AffectedSide);
if sum(sum(AffectedSide))== 0
 Sides1 = {'Left','Right'};
end
row = find(AffectedSide);

AffSide = affectedside(row,2);
AffSide = cell2mat(AffSide);
if isempty(AffSide) == 1
 disp('Healthy Participant')
elseif AffSide(1) == 'R'
 Sides1 = {'Unoperated', 'Operated'};
elseif AffSide(1) == 'L'
 Sides1 = {'Operated', 'Unoperated'};
end

clearvars affectedside AffectedSide row
```

```
%Set up data
[Start, End]= vicon.GetTrialRegionOfInterest;
Sides={'Left','Right'}; %i
TrialName=strrep(TrialName,' ','');
Joint={'Ankle','Knee','Hip','Pelvis'}; %j

%preallocate
AvAng = zeros(101,3);
stDev = zeros(101,3);

beep
choice = menu('Which inclination is this?',...
 '+7.5','+5','0','-5','-7.5');
% Handle response
switch choice
 case 5
  InclineDeg ={'Incline7'};
 case 4
  InclineDeg ={'Incline6'};
 case 3
  InclineDeg ={'Incline1'};
 case 2
  InclineDeg ={'Incline3'};
 case 1
  InclineDeg ={'Incline4'};
end
```

```
% Extract parameters Average kinematics for all cycles in trial
for i = 1:2
 HSF = vicon.GetEvents(SubjectName{1}, Sides{i},'Foot Strike'); %top = left, bottom =
right
 HSF = HSF(HSF > Start);
 for j=1:4 %to get all joint angles
  TempVar=vicon.GetModelOutput(SubjectName{1}, [Sides{i}(1) Joint{j} 'Angles'])';
  for l = 1:length(HSF)
   if l <= length(HSF)-1
    ToNorm{l} = TempVar(((HSF(l)):(HSF(l+1))),:);
    %ToNorm{l} = TempVar(((HSF(l)+Start):(HSF(l+1)+Start)),:);
   elseif l == length(HSF)
    continue
   end
  end

  ToAvNorm = [] ;
  for m = 1:length(ToNorm)
   Norm{m} = interpft(ToNorm{m},101);
   ToAvNorm = cat(3,ToAvNorm,(Norm{1,m}));
  end
  Output.(InclineDeg{1}).(splitStr{5}).(splitStr{6}).(Sides1{i}).([Sides{i}(1)
Joint{j} 'Angles'])=ToAvNorm;
  save([newpath 'Output.mat'],'Output')
  clearvars ToNorm l

  for n = 1:size(ToAvNorm,2) %each column
   for o = 1:size(ToAvNorm,1) %each row
    AvAng (o,n) = mean(ToAvNorm(o,n,:)); %gives mean of each cell across pages
    stDev(o,n) = std(ToAvNorm(o,n,:));

   end
  end
  AngDev = horzcat (AvAng,stDev);

  clearvars ToAvNorm
  AvData.(InclineDeg{1}).(splitStr{5}).(splitStr{6}).(Sides1{i}).([Sides{i}(1)
Joint{j} 'Angles'])=AngDev; %save in avdata
  save([newpath 'AvData.mat'],'AvData')
  clearvars AngDev
 end
end
save([newpath 'Output.mat'],'Output')
```

```
% Get TO's as percentage of cycle

PercentTO = NaN(1,5);
for i = 1:2
 HSF = vicon.GetEvents(SubjectName{1}, Sides{i},'Foot Strike'); %top = left, bottom =
right
 HSF = HSF(HSF > Start);
 HSF = HSF(HSF < End);


 TOF = vicon.GetEvents(SubjectName{1}, Sides{i},'Foot Off');
 TOF = TOF(TOF > Start);
 TOF = TOF(TOF < End);

 %check got 1 less TOF than HSF
 if HSF(1) < TOF(1) %check got less TOF than HSF
  for a = 1:(size(HSF,2)-1)
   PercentTO(a) =(((TOF(a))- (HSF(a)))*100) /(HSF(a+1)- HSF(a));
  end
 else
  for a = 1:(size(HSF,2)-1)
   PercentTO(a) =(((TOF(a+1))- (HSF(a)))*100) /(HSF(a+1)- HSF(a));
  end
 end
 Output.(InclineDeg{1}).(splitStr{5}).(splitStr{6}).([Sides1{i}(1:4) 'ToeOff']) =
PercentTO;

 AvTO = nanmean(PercentTO);
 StdDevTO = nanstd(double(PercentTO));
 TOStdDev = horzcat(AvTO, StdDevTO);
 AvData.(InclineDeg{1}).(splitStr{5}).(splitStr{6}).([Sides1{i}(1:4)
'ToeOff'])=TOStdDev;
 clearvars HSF TOF PercentTO AvTO StDevTO TOStdDev
end

save([newpath 'AvData.mat'],'AvData')
```

```
% work out treadmill speed
```

```
PercentTO = NaN(1,5);MarkerNames = vicon.GetMarkerNames(SubjectName{1});
MarkerNames2= strfind(MarkerNames, 'TOE');
MarkerNames3= find(not(cellfun('isempty',MarkerNames2)));
Toes = MarkerNames(MarkerNames3);
clearvars MarkerNames MarkerNames2 MarkerNames3

for i = 1:2
 HSF = vicon.GetEvents(SubjectName{1}, Sides{i},'Foot Strike'); %top = left, bottom =
right
 HSF = HSF(HSF > Start);
 HSF = HSF(HSF < End);

 TOF = vicon.GetEvents(SubjectName{1}, Sides{i},'Foot Off');
 TOF = TOF(TOF > Start);
 TOF = TOF(TOF < End);

 if TOF(1) < HSF(1)
  for m = 2:size(TOF,2)
   TOF (m-1) = TOF(m);
  end
  TOF(m) = [];
 end

 for j = 1:length(HSF) - 1
  FootFlatFrames(j,:) = ((HSF(j) + TOF(j))/2)-Start;
 end
 clearvars j

 [~, y, ~, ~] = vicon.GetTrajectory((SubjectName{1}),(Toes{i})); %Get AP trajectory
of toe
 y = y(Start:End); %get ToeY coordinates within region of interest
 VelY = diff(y)/0.01; % v = d (/dt)differenciate traj to find velocity
 VelY(1,1) = 0; %ignore first value

 for k = 1:length(FootFlatFrames)
  ToAvVelocity(k,:) = VelY(FootFlatFrames(k))/1000; %divide to put speed as m/s from
mm/s
 end
 clearvars k
 clearvars FootFlatFrames

 AvVelocity(i,:) = mean(ToAvVelocity);
 clearvars ToAvVelocity
end

AvDev = mean(AvVelocity);
AvData.(InclineDeg{1}).(splitStr{5}).(splitStr{6}).BeltSpeed = AvDev;

save([newpath 'AvData.mat'],'AvData')
```

```
%get belt speed from ascii
if strcmp((splitStr{5}),{'YearPostOp'}) || strcmp((splitStr{5}),{'Control'})
 cd ('H:\Raw Data\DFlowOutput\BBraunMasterMotionCaptureDatabase') % Uni
 [filename,PathName] = uigetfile('*','MultiSelect','on');
 TotalFileName = strcat(PathName,filename);
 delimiter = '\t';
 formatSpec =
'%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s
%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%
*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*
s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s
%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%
*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%*s%s%[^\n\r]';
 fileID = fopen(TotalFileName,'r');
 dataArray = textscan(fileID, formatSpec, 'Delimiter', delimiter, 'ReturnOnError',
false);
 fclose(fileID);
 InclineSpeedOutput = [dataArray{1:end-1}];

 if strcmp((InclineSpeedOutput{1,1}),'LBeltSpeed') == 0
  disp ('Old Belt Speed Kept')
 else
  InclineSpeedOutput = InclineSpeedOutput(2:end);
  InclineSpeedOutput = str2double(InclineSpeedOutput);
  AvVelo = mean(InclineSpeedOutput);
  AvData.(InclineDeg{1}).(splitStr{5}).(splitStr{6}).BeltSpeed = AvVelo;
 end

 clearvars filename delimiter startRow formatSpec fileID dataArray ans;
end

save([newpath 'AvData.mat'],'AvData')
%
```

```
clearvars AvDev AvVelocity FootFlatFrames ToAvVelocity VelY PathName TotalFileName
choice AvVelo InclineSpeedOutput
clearvars AffectedSide IsLeft Sides1 Toes a End Heels i j Joint l m Sides Start y y1
y2 TempVar ToNorm HSF TOF AvTO StdDevTO TOStdDev prompt dlg_title num_lines
defaultans AvAng InclineDeg n o stDev Norm AngDev
```

9.4. Data Analysis

9.4.1. Quality Assurance

```
% Quality control, for visually checking data for any mislabelled markers or gaps
% prior to averaging - use script below for sloped walking data assurance. C Govind.

clc
clear

newpath = 'H:\Raw Data\NexusOutput\BBraunMasterMotionCaptureDatabase\';


% load output Struct
load([newpath 'Output.mat']) %all outputs from all trials

Groups ={'Control', 'PreOp','YearPostOp'};

TrialType = {'LevelWalking',...  %1
 'StairsAscent','StairsDescent'}; % 2 and 3

Sides= {'Left','Right'};

%%%%%%%%%%%%%%% FILL IN %%%%%%%%%%%%%%%%
splitStr{5} = Groups{2}; % 1-CONTROL 2-PREOP 3-POSTOP
splitStr{6} = 'aj1947';

% 6 graphs with 2 subplots will be shown: as per:
% 1 - Level L + R Flexion Angle
% 2 - Level L + R Flexion Moment
% 3 - Stair Ascent L + R Flexion Angle
% 4 - Stair Ascent L + R Flexion Moment
% 5 - Stair Descent L + R Flexion Angle
% 6 - Stair Descent L + R Flexion Moment
ax1 = [0,0];
ax2 = [0,0];
%kinematics
for ll = 1:3
 TrialNames = fieldnames(Output.(TrialType{ll}).(splitStr{5}).(splitStr{6}))';
 if strcmp(TrialNames{ll},'LevelWalking')
  TrialNames = TrialNames(cellfun('isempty', strfind(TrialNames,'ToeOffs'))); %for
level walking ignore toe off field
  clearvars TrialName
 end
 figure
 for jj = 1:2
  AxesHandle(jj) = subplot(1,2,jj);
  for ii = 1:length(TrialNames)
   IsLR =
fieldnames(Output.(TrialType{ll}).(splitStr{5}).(splitStr{6}).(TrialNames{ii}));
   if ismember(Sides{jj},IsLR)

plot(Output.(TrialType{ll}).(splitStr{5}).(splitStr{6}).(TrialNames{ii}).(Sides{jj}).
([Sides{jj}(1) 'KneeAngles'])(:,1))
    hold on
   end
  end
  hold off
  legend show
 end
 allYLim = get(AxesHandle, {'YLim'});
 allYLim = cat(2, allYLim{:});
 set(AxesHandle, 'YLim', [min(allYLim), max(allYLim)]);
 clearvars AxesHandle allYLim
 figure
 %kinetics
 for jj = 1:2
  AxesHandle(jj) = subplot(1,2,jj);
  for ii = 1:length(TrialNames)
   IsLR =
fieldnames(Output.(TrialType{ll}).(splitStr{5}).(splitStr{6}).(TrialNames{ii}));
   if ismember(Sides{jj},IsLR)
    if
```

```
isfield(Output.(TrialType{ll}).(splitStr{5}).(splitStr{6}).(TrialNames{ii}).(Sides{jj
}),([Sides{jj}(1) 'KneeMoment']))

plot(Output.(TrialType{ll}).(splitStr{5}).(splitStr{6}).(TrialNames{ii}).(Sides{jj}).
([Sides{jj}(1) 'KneeMoment'])(:,1))
     hold on
    end
   end
  end
  hold off
  legend show

 end
 allYLim = get(AxesHandle, {'YLim'});
 allYLim = cat(2, allYLim{:});
 set(AxesHandle, 'YLim', [min(allYLim), max(allYLim)]);
 clearvars AxesHandle allYLim
end
```

```
% For incline

clc
clear

newpath = 'H:\Raw Data\NexusOutput\BBraunMasterMotionCaptureDatabase\';

% load output Struct
load([newpath 'Output.mat']) %all outputs from all trials
Groups ={'Control', 'PreOp','YearPostOp'};


%%%%%%%%%%%%%%% FILL IN %%%%%%%%%%%%%%%%
splitStr{5} = Groups{3}; % 1-CONTROL 2-PREOP 3-POSTOP
splitStr{6} = 'mw1944';

if strcmp(splitStr{5},'Control')
 Sides= {'Left','Right'};
else
 Sides ={'Operated','Unoperated'};
end

InclineDeg = {'Incline7','Incline6','Incline1','Incline3','Incline4'};

for jj = 1:5
 if isfield(Output.(InclineDeg{jj}).(splitStr{5}),(splitStr{6}))
  figure
  for ii = 1:2
   FN = fieldnames(Output.(InclineDeg{jj}).(splitStr{5}).(splitStr{6}).(Sides{ii}));
   if FN{1,1}(1) == 'R'
    ToPlot =
permute(Output.(InclineDeg{jj}).(splitStr{5}).(splitStr{6}).(Sides{ii}).RKneeAngles,[
1 3 2]);
    ToPlot = ToPlot(:,:,1);
   elseif FN{1,1}(1) == 'L'
    ToPlot =
permute(Output.(InclineDeg{jj}).(splitStr{5}).(splitStr{6}).(Sides{ii}).LKneeAngles,[
1 3 2]);
    ToPlot = ToPlot(:,:,1);
   end

   subplot (1,2,ii)
   plot(ToPlot)
   title(InclineDeg{jj})
%    legend show
  end
 else
  disp ('no data')
  continue
 end
end
```

9.4.2. Average Participant

```
% This script averages all kinetic/kinematic data from trials of a type of
% a participant from the Output structure into a structure called AvData.
% Just have to press run, all averages will be computed. C.Govind

clc
clear
```

```
%Average Data

newpath = 'H:\Raw Data\NexusOutput\BBraunMasterMotionCaptureDatabase\';

% load master output Struct
load([newpath 'Output.mat']) %all outputs from all trials
load([newpath 'AvData.mat']) %averages of all trials
```

```
%put all kinematic data in 3D matrix and average
TempAng = NaN(101:3:10);
AvAng = zeros(101:3);
stDev = zeros(101:3);
checkKinem = NaN(1,2);
TempKinet = NaN(101:3:4);
checkTO = NaN(1,5);
ToAvTO = NaN(1,5);

Groups ={'Control', 'PreOp','YearPostOp'};

TrialType = {'LevelWalking',...  %1
 'StairsAscent','StairsDescent'}; % 2 and 3

Sides= {'Left','Right'}; %l
Joint = {'Ankle','Knee','Hip','Pelvis'}; %m
Parameters = {'Moment','Power'}; %n


for t = 1:3
 for g = 1:3
  for l= 1:2 %for each side
   %    g = 3; t = 3; l = 2;
   % Get inputs
   Names = fieldnames(Output.(TrialType{t}).(Groups{g}));
   clearvars TrialNames
   for name = 1:length(Names)
```

```
    TrialNames = fieldnames(Output.(TrialType{t}).(Groups{g}).(Names{name}))';

    if t == 1;
     TrialNames = TrialNames(cellfun('isempty', strfind(TrialNames,'ToeOffs'))); %for
level walking
    end

    % get affected side

[~, ~, affectedside] = xlsread('C:\Users\gwb13225\ShareFile\Personal Folders\Phd\3
Data\affected side.xlsx','Sheet1','A2:B88');
     affectedside(cellfun(@(x) ~isempty(x) && isnumeric(x) && isnan(x),affectedside))
= {''};

     AffectedSide = strfind(affectedside,(char(Names{name})));

     ix=cellfun(@isempty,AffectedSide);
     AffectedSide(ix)={0}; %convert blank cells to 0
     clearvars ix

     AffectedSide = cell2mat(AffectedSide);
     if sum(sum(AffectedSide))== 0
      Sides1 = {'Left','Right'};
      disp ('Healthy Participant')
     elseif sum(sum(AffectedSide)) == 1
      row = find(AffectedSide);

      AffSide = affectedside(row,2);
      AffSide = cell2mat(AffSide);

      if AffSide(1) == 'R'
       Sides1 = {'Unoperated', 'Operated'};
      elseif AffSide(1) == 'L'
       Sides1 = {'Operated', 'Unoperated'};
      end
     end

     clearvars affectedside AffectedSide
```

```
% average kinematics
    clearvars NZ1 checkKinem checkKin NZ2
    checkKin = false(1,(length(TrialNames)));
    for k = 1:length(TrialNames) %for each trial
     checkKinem(k) =
isfield(Output.(TrialType{t}).(Groups{g}).(Names{name}).([TrialNames{k}]),(Sides{l}))
;
     checkKin(k) =
```

```
isfield(Output.(TrialType{t}).(Groups{g}).(Names{name}).([TrialNames{k}]),(Sides{l}))
&&...

isfield(Output.(TrialType{t}).(Groups{g}).(Names{name}).([TrialNames{k}]).(Sides{l}),
([Sides{l}(1) 'KneeMoment']));
    end
    NZ1 = find(checkKinem);
    NZ2 = find(checkKin);
    if ~isempty (NZ1)
     for m= 1:4   %for each joint
      a = 0;
      for k = NZ1(1:(size(NZ1,2)))
       a = a+1;

TempAng(:,:,a)=Output.(TrialType{t}).(Groups{g}).(Names{name}).([TrialNames{k}]).(Sid
es{l}).([Sides{l}(1) Joint{m} 'Angles']);
      end

      %average each point through multidimensional matrix
      for i = 1:size(TempAng,2) %each column
       for j = 1:size(TempAng,1) %each row
        AvAng (j,i) = mean(TempAng(j,i,:)); %gives mean of each cell across pages
        stDev(j,i) = std(TempAng(j,i,:));

       end
      end
      MeanDev = horzcat (AvAng,stDev);
      AvData.(TrialType{t}).(Groups{g}).(Names{name}).(Sides1{l}).([Sides1{l}(1)
Joint{m} 'Angles'])=MeanDev(:,:);
      save([newpath 'AvData.mat'],'AvData')
      clearvars TempAng AvAng stDev MeanDev
     end
    end
```

```
% average kinetics- if no data, dont include in average
    if ~isempty(NZ2)
     for m = 1:3 %for each joint
      for n= 1:2 %for each parameter
       a = 0;
       switch n
        case 1
         for k = NZ2(1:(size(NZ2,2)))
          a = a+1;
          TempKinet(:,:,a) =
Output.(TrialType{t}).(Groups{g}).(Names{name}).([TrialNames{k}]).(Sides{l}).([Sides{
l}(1) Joint{m} Parameters{n}]);
         end
        case 2 %multiply power by 1000 previously divided
         for k = NZ2(1:(size(NZ2,2)))
          a = a+1;
          TempKinet(:,:,a) =
Output.(TrialType{t}).(Groups{g}).(Names{name}).([TrialNames{k}]).(Sides{l}).([Sides{
l}(1) Joint{m} Parameters{n}])*1000;
         end
       end

       %average each point through multidimensional matrix
       for i = 1:size(TempKinet,2) %each column
        for j = 1:size(TempKinet,1) %each row
         AvAng (j,i) = nanmean(TempKinet(j,i,:)); %gives mean of each cell across
pages
         stDev(j,i) = nanstd(TempKinet(j,i,:));
        end
       end
       MeanDev = horzcat (AvAng,stDev);
       AvData.(TrialType{t}).(Groups{g}).(Names{name}).(Sides1{l}).([Sides1{l}(1)
Joint{m} Parameters{n}])=MeanDev(:,:);
       save([newpath 'AvData.mat'],'AvData')
       clearvars TempKinet AvAng stDev MeanDev
      end

      save([newpath 'AvData.mat'],'AvData')

     end

    end
```

```
% average toe offs
    if t ~= 1 && ~isempty (NZ1) %Stairs
     TrialNames = fieldnames(Output.(TrialType{t}).(Groups{g}).(Names{name}))';
     for k = 1:length(TrialNames)
      checkTO(k) =
isfield(Output.(TrialType{t}).(Groups{g}).(Names{name}).([TrialNames{k}]),(Sides{l}))
```

```
;
     end
     ind = find(isnan(checkTO));
     checkTO(ind) = [];
     NZ1 = find(checkTO);
     clearvars checkTO ind
     for k = NZ1(1:(size(NZ1,2)))
      ToAvTO(k) =
Output.(TrialType{t}).(Groups{g}).(Names{name}).([TrialNames{k}]).(Sides{l}).([Sides{
l}(1) 'ToeOff']);
     end
     AvTO = nanmean(ToAvTO);
     StDevTO = nanstd(ToAvTO);
     TOStdDev = horzcat(AvTO, StDevTO);
     AvData.(TrialType{t}).(Groups{g}).(Names{name}).(Sides1{l}).([Sides1{l}(1)
'ToeOff']) = TOStdDev;
     save([newpath 'AvData.mat'],'AvData')
     clearvars TrialNames AvTO StdDevTO TOStedDev

    elseif t == 1 && ~isempty (NZ1) %average level TO
     Names = fieldnames(Output.(TrialType{t}).(Groups{g}));
     TrialNames =
fieldnames(Output.(TrialType{t}).(Groups{g}).(Names{name}).('ToeOffs'));
     Lto = TrialNames(cellfun('isempty', strfind(TrialNames,'RToeOff'))) ;
     Rto = TrialNames(cellfun('isempty', strfind(TrialNames,'LToeOff'))) ;
     if l == 1
      % Average left toe off
      LTO = NaN(1,1);
      for i= 1:length(Lto)
       c = Output.(TrialType{t}).(Groups{g}).(Names{name}).ToeOffs.(Lto{i});
       LTO = horzcat(LTO,c);
      end

      AvTO = nanmean(LTO);
      StdDevTO = nanstd(double(LTO));
      TOStdDev = horzcat(AvTO, StdDevTO);

      AvData.(TrialType{t}).(Groups{g}).(Names{name}).(Sides1{l}).([Sides1{l}(1)
'ToeOff']) = TOStdDev;
      save([newpath 'AvData.mat'],'AvData')
      clearvars LTO Lto c AvTO StdDevTO TOStedDev i

     elseif l == 2 % Average right toe off
      RTO = [];
      for i= 1:length(Rto)
       c = Output.(TrialType{t}).(Groups{g}).(Names{name}).ToeOffs.(Rto{i});
       RTO = horzcat(RTO,c);
      end

      AvTO = nanmean(RTO);
      StdDevTO = nanstd(double(RTO));
      TOStdDev = horzcat(AvTO, StdDevTO);

      AvData.(TrialType{t}).(Groups{g}).(Names{name}).(Sides1{l}).([Sides1{l}(1)
'ToeOff']) = TOStdDev;
      clearvars RTO Lto c AvTO StdDevTO TOStedDev i TrialNames
      save([newpath 'AvData.mat'],'AvData')
     end
    end
    clearvars NZ2 NZ1 n
```

```
   end
  end
 end
 clearvars Names name
end

clearvars i j l k m n NZ1 stDev a Groups TrialType Sides Joint Parameters g t Names
name

disp ('Data saved')
```

9.4.3. Sort to Groups

```
% This script collates the average data from AvData (for level and stairs
% activities, concatenates them next to each other in O/U/L/R RawAvData
% structures. Toe off and spatiotemporal parameters will also be averaged and saved.
% AvData > RawAvData + RawImplantData. C Govind
```

```
clc
clear
```

```
newpath = 'H:\Raw Data\NexusOutput\BBraunMasterMotionCaptureDatabase\';

load([newpath 'AvData.mat']) %averages of all people
load([newpath 'RawAvData.mat']) %raw average data of all participants
load([newpath 'RawImplantData.mat'])

Groups ={'Control', 'PreOp', 'YearPostOp'}; %o

TrialType = {'LevelWalking',...  %1
 'StairsAscent','StairsDescent',...% 2 and 3
 'Incline7','Incline6','Incline1',...
 'Incline3','Incline4'};  %4

%%%%%%%%%%%%%%%%%% FILL IN %%%%%%%%%%%%%%%%%%

splitStr{7} = TrialType{1};

Sides = {'Left', 'Right', 'Operated', 'Unoperated'}; %l
Joint = {'Ankle','Knee','Hip','Pelvis'}; %m
Parameters = {'Angles','Moment','Power'}; %n
Direction = {'Flexion','AbAd','Rot'}; %q
ImplantGroup = {'Implant1','Implant2','Implant3'}; %s

TrialNames{1,1} = fieldnames(AvData.(splitStr{7}).Control);
TrialNames{1,2} = fieldnames(AvData.(splitStr{7}).PreOp);
TrialNames{1,3} = fieldnames(AvData.(splitStr{7}).YearPostOp);

TrialNames{2,1} =
{'ab1942','kh1945','km1951','dt1952','ml1952','jn1949','rc1952','hf1942','sl1945','aj
1947','ph1953','pw1953'}; %IMPLANT 1
TrialNames{2,2} = {'js1944','ms1947','lm1962','ob1938','hp1945'}; %IMPLANT 2
TrialNames{2,3} = {'ht1949','to1936','jm1940','cm1944','sc1956','mq1947','mw1944'};
%IMPLANT 3

%
```

```
% average whole group data
% control kinetics and kinematics

for l= 1:2 %for each side
 for n = 1:3
  if n == 1 %kinematics
   for m = 1:4 %for each joint
    for x = 1:3 %direction
     TempVar = [];
     a = 1;
     for k = 1:length(TrialNames{1,1})
      if isfield(AvData.(splitStr{7}).Control,TrialNames{1,1}{k,1})
       c =
AvData.(splitStr{7}).Control.([TrialNames{1,1}{k,1}]).(Sides{l}).([Sides{l}(1)
Joint{m} Parameters{n}])(:,x);
       TempVar = horzcat (TempVar,c);
       b{a,:} = TrialNames{1,1}{k,1};
       a = a + 1;
      end
     end
     TempVar = array2table(TempVar,'VariableNames',b);
     clearvars c k b

     %export average control raw data to RawAvData. make table?
     switch x
      case 1
       RawAvData.(splitStr{7}).Control.([Sides{l}(1) 'RawAngles']).([Joint{m}
Direction{x}]) = TempVar;
       clearvars TempVar
      case 2
       RawAvData.(splitStr{7}).Control.([Sides{l}(1) 'RawAngles']).([Joint{m}
Direction{x}]) = TempVar;
       clearvars TempVar
      case 3
       RawAvData.(splitStr{7}).Control.([Sides{l}(1) 'RawAngles']).([Joint{m}
Direction{x}]) = TempVar;
       clearvars TempVar
     end
     save([newpath 'RawAvData.mat'],'RawAvData')
    end
   end
  elseif strcmp(splitStr{7},'LevelWalking') || strcmp(splitStr{7},'StairsAscent') ||
strcmp(splitStr{7},'StairsDescent')
```

```
    for m = 1:3 %joint kinetics excluding pelvis
     if n == 2 %moments
      for x = 1:3 %direction
       TempVar = [];
       a = 1;
       for k = 1:length(TrialNames{1,1})
        if isfield
(AvData.(splitStr{7}).Control.([TrialNames{1,1}{k,1}]).(Sides{l}),[Sides{l}(1)
Joint{m} Parameters{n}])
         c =
AvData.(splitStr{7}).Control.([TrialNames{1,1}{k,1}]).(Sides{l}).([Sides{l}(1)
Joint{m} Parameters{n}])(:,x);
         TempVar = horzcat (TempVar,c);
         b{a,:} = TrialNames{1,1}{k,1};
         a = a + 1;
        end
       end

       TempVar = array2table(TempVar,'VariableNames',b);
       clearvars c k a b

       %export average control raw data to RawAvData. make table?
       switch x %direction
        case 1
         RawAvData.(splitStr{7}).Control.([Sides{l}(1) 'RawMoment']).([Joint{m}
Direction{x}]) = TempVar;
         clearvars TempVar
        case 2
         RawAvData.(splitStr{7}).Control.([Sides{l}(1) 'RawMoment']).([Joint{m}
Direction{x}]) = TempVar;
         clearvars TempVar
        case 3
         RawAvData.(splitStr{7}).Control.([Sides{l}(1) 'RawMoment']).([Joint{m}
Direction{x}]) = TempVar;
         clearvars TempVar
       end
       save([newpath 'RawAvData.mat'],'RawAvData')
      end
     elseif n == 3
      TempVar = [];
      a = 1;
      for k = 1:length(TrialNames{1,1})
       if isfield
(AvData.(splitStr{7}).Control.([TrialNames{1,1}{k,1}]).(Sides{l}),[Sides{l}(1)
Joint{m} Parameters{n}])
         c =
AvData.(splitStr{7}).Control.([TrialNames{1,1}{k,1}]).(Sides{l}).([Sides{l}(1)
Joint{m} Parameters{n}])(:,3);
        TempVar = horzcat (TempVar,c);
        b{a,:} = TrialNames{1,1}{k,1};
        a = a + 1;
       end
      end

      TempVar = array2table(TempVar,'VariableNames',b);
      clearvars c k a b

      %export average control raw data to RawAvData. make table?
      RawAvData.(splitStr{7}).Control.([Sides{l}(1) 'RawPower']).([Joint{m} 'Power'])
= TempVar;
      clearvars TempVar

      save([newpath 'RawAvData.mat'],'RawAvData')
     end
    end
   end
  end
 end
end

RawImplantData.(splitStr{7}).Control = RawAvData.(splitStr{7}).Control;
save([newpath 'RawImplantData.mat'],'RawImplantData')

clearvars a AvAng i j k l m MeanDev o stDev TempAng
```

```
% Average patient kinetics and kinematics
for o = 2:3
 for l= 3:4 %for each side
  for n = 3%1:3 %parameters
   if n == 1
    for m= 1:4 %for each joint
     for x = 1:3 %direction
      TempVar = [];
      for k = 1:length(TrialNames{1,o})
```

```
        FN2 = fieldnames
(AvData.(splitStr{7}).(Groups{o}).([TrialNames{1,o}{k,1}]).(Sides{l}));
        if FN2{1,1}(1) == 'R'
         c =
AvData.(splitStr{7}).(Groups{o}).([TrialNames{1,o}{k,1}]).(Sides{l}).(['R' Joint{m}
'Angles'])(:,x);
         TempVar = horzcat (TempVar,c);
        elseif FN2{1,1}(1) == 'L'
         c =
AvData.(splitStr{7}).(Groups{o}).([TrialNames{1,o}{k,1}]).(Sides{l}).(['L' Joint{m}
'Angles'])(:,x);
         TempVar = horzcat (TempVar,c);
        else
         c =
AvData.(splitStr{7}).(Groups{o}).([TrialNames{1,o}{k,1}]).(Sides{l}).([Sides{l}(1)
Joint{m} 'Angles'])(:,x);
         TempVar = horzcat (TempVar,c);
        end
       end
       clearvars c k
       TempVar = array2table(TempVar,'VariableNames',(TrialNames{1,o}));

       for ii = 1:3
        a = ismember(TrialNames{2,ii},TrialNames{1,o});
        b = TrialNames{2,ii}(a);
        t0 = TempVar(:,b);
        t0.Properties.VariableNames = b;
        tables{ii} = t0;
        clearvars a b TempVarVars t0
        %export average patient raw data to RawAvData
        RawImplantData.(splitStr{7}).(Groups{o}).(ImplantGroup{ii}).([Sides{l}(1)
'RawAngles']).([Joint{m} Direction{x}]) = tables{1,ii};
       end

       RawAvData.(splitStr{7}).(Groups{o}).([Sides{l}(1) 'RawAngles']).([Joint{m}
Direction{x}]) = TempVar;
       clearvars TempVar tables b
       save([newpath 'RawImplantData.mat'],'RawImplantData')
       save([newpath 'RawAvData.mat'],'RawAvData')
      end
     end
    elseif strcmp(splitStr{7},'LevelWalking') || strcmp(splitStr{7},'StairsAscent') ||
strcmp(splitStr{7},'StairsDescent')
     for m = 1:3 %kinetics excluding pelvis
      if n == 2 %moment
       for x = 1:3 %direction
        a = 1;
        TempVar = [];
        for k = 1:length(TrialNames{1,o}) %does kinetics exist for that patient and
that limb: operated/non operated
         if isfield
(AvData.(splitStr{7}).(Groups{o}).([TrialNames{1,o}{k,1}]).(Sides{l}),[Sides{l}(1)
Joint{m} Parameters{n}])
          c =
AvData.(splitStr{7}).(Groups{o}).([TrialNames{1,o}{k,1}]).(Sides{l}).([Sides{l}(1)
Joint{m} Parameters{n}])(:,x);
          TempVar = horzcat (TempVar,c);
          b{a,:} = TrialNames{1,o}{k,1};
          a = a + 1;
         end
        end
        TempVar = array2table(TempVar,'VariableNames',b); %table of all available
patient kinetics
        clearvars c k a
        % divide TempVar into implant groups-
        % tables = Implant 1, 2 ,3
        for ii = 1:3
         a = ismember(TrialNames{2,ii},b);
         c = TrialNames{2,ii}(a);
         t0 = TempVar(:,c);
         t0.Properties.VariableNames = c;
         tables{ii} = t0;
         clearvars a t0 c
         %export average patient raw data to RawAvData
         RawImplantData.(splitStr{7}).(Groups{o}).(ImplantGroup{ii}).([Sides{l}(1)
'RawMoment']).([Joint{m} Direction{x}]) = tables{1,ii};
        end

        RawAvData.(splitStr{7}).(Groups{o}).([Sides{l}(1) 'RawMoment']).([Joint{m}
Direction{x}]) = TempVar;
        clearvars TempVar tables b
        save([newpath 'RawImplantData.mat'],'RawImplantData')
        save([newpath 'RawAvData.mat'],'RawAvData')
```

```
      end
     elseif n == 3
      a = 1;
      TempVar = [];

      for k = 1:length(TrialNames{1,o}) %does kinetics exist for that patient and
that limb: operated/non operated
       if isfield
(AvData.(splitStr{7}).(Groups{o}).([TrialNames{1,o}{k,1}]).(Sides{l}),[Sides{l}(1)
Joint{m} Parameters{n}])
        c =
AvData.(splitStr{7}).(Groups{o}).([TrialNames{1,o}{k,1}]).(Sides{l}).([Sides{l}(1)
Joint{m} Parameters{n}])(:,3);
        TempVar = horzcat (TempVar,c);
        b{a,:} = TrialNames{1,o}{k,1};
        a = a + 1;
       end
      end
      TempVar = array2table(TempVar,'VariableNames',b); %table of all available
patient kinetics
      clearvars c k a
      for ii = 1:3
       a = ismember(TrialNames{2,ii},b);
       c = TrialNames{2,ii}(a);
       t0= TempVar(:,c);
       t0.Properties.VariableNames = c;
       tables{ii} = t0;
       clearvars a t0 c

       %export average patient raw data to RawAvData
       RawImplantData.(splitStr{7}).(Groups{o}).(ImplantGroup{ii}).([Sides{l}(1)
'RawPower']).([Joint{m} 'Power']) = tables{1,ii};
      end

      RawAvData.(splitStr{7}).(Groups{o}).([Sides{l}(1) 'RawPower']).([Joint{m}
'Power']) = TempVar;
      clearvars TempVar tables b
      save([newpath 'RawImplantData.mat'],'RawImplantData')
      save([newpath 'RawAvData.mat'],'RawAvData')

     end
    end
   end
  end
 end
end
toc
beep
clearvars ii l m n o x Joint Direction Parameters
```

```
% Average TO
%control
for l = 1:2
 TO = [];
 FN = fieldnames(AvData.(splitStr{7}).Control);
 if strcmp(splitStr{7},'LevelWalking') || strcmp(splitStr{7},'StairsAscent') ||
strcmp(splitStr{7},'StairsDescent')
  for k = 1:length(FN)
   TO(k,:) = AvData.(splitStr{7}).Control.(FN{k}).(Sides{l}).([Sides{l}(1)
'ToeOff'])(1);
  end
 else %incline
  for k = 1:length(FN)
   TO(k,:) = AvData.(splitStr{7}).Control.(FN{k}).([Sides{l}(1:4) 'ToeOff'])(1);
  end
 end
 TO = array2table(TO','VariableNames',FN);
 RawAvData.(splitStr{7}).Control.([Sides{l}(1) 'ToeOff'])= TO;
 RawImplantData.(splitStr{7}).Control.([Sides{l}(1) 'ToeOff']) = TO;
 save([newpath 'RawAvData.mat'],'RawAvData')
 save([newpath 'RawImplantData.mat'],'RawImplantData')
 clearvars TO k FN
end
clearvars l k TO

%patients - put into correct implant
for o = 2:3
 for l = 3:4
  FN = fieldnames(AvData.(splitStr{7}).(Groups{o}));
  if strcmp(splitStr{7},'LevelWalking') || strcmp(splitStr{7},'StairsAscent') ||
strcmp(splitStr{7},'StairsDescent')
   for k = 1:length(FN)
    TO(k,:) = AvData.(splitStr{7}).(Groups{o}).(FN{k}).(Sides{l}).([Sides{l}(1)
```

```
'ToeOff'])(1);
    end
   else %incline
    for k = 1:length(FN)
     TO(k,:) = AvData.(splitStr{7}).(Groups{o}).(FN{k}).([Sides{l}(1:4) 'ToeOff'])(1);
    end
   end
   TO = array2table(TO','VariableNames',FN);
   RawAvData.(splitStr{7}).(Groups{o}).([Sides{l}(1) 'ToeOff'])= TO;
   save([newpath 'RawAvData.mat'],'RawAvData')
   %divide by implant
   for ii = 1:3
    a = ismember(TrialNames{2,ii},FN);
    c = TrialNames{2,ii}(a);
    tO= TO(:,c);
    tables{ii} = tO;
    RawImplantData.(splitStr{7}).(Groups{o}).(ImplantGroup{ii}).([Sides{l}(1)
'ToeOff']) = tables{1,ii};
    save([newpath 'RawImplantData.mat'],'RawImplantData')
   end
   clearvars a c tO TO ii k FN
  end
end
clearvars l o tables
```

```
% Average SPT for Level and Stairs
if strcmp(splitStr{7},'LevelWalking') || strcmp(splitStr{7},'StairsAscent') ||
strcmp(splitStr{7},'StairsDescent')
 SideSPT(1:10,2) = {'Cadence';'Walking Speed';'Stride Time';'Step Time';...
  'Opposite Foot Off';'Opposite Foot Contact';'Foot Off';'Double Support';'Stride
Length';'Step Length'};
 SideSPT(11:20,2) = {'Cadence';'Walking Speed';'Stride Time';'Step Time';...
  'Opposite Foot Off';'Opposite Foot Contact';'Foot Off';'Double Support';'Stride
Length';'Step Length'};
else
 SideSPT(1:7,2) = {'Cadence';'Stride Time';'Step Time';...
  'Opposite Foot Off';'Opposite Foot Contact';'Single Support';'Double Support'};
 SideSPT(8:14,2) = {'Cadence';'Stride Time';'Step Time';...
  'Opposite Foot Off';'Opposite Foot Contact';'Single Support';'Double Support'};
end

for o = 1:3
 Spatiotemp = [];
 FN = fieldnames(AvData.(splitStr{7}).(Groups{o}));
 if o == 1 %control
  for k = 1:length(FN)
   a = AvData.(splitStr{7}).(Groups{o}).(FN{k}).Spatiotemporal(:,3);
   Spatiotemp = horzcat(Spatiotemp,a);
  end

  if strcmp(splitStr{7},'LevelWalking') || strcmp(splitStr{7},'StairsAscent') ||
strcmp(splitStr{7},'StairsDescent')
   SideSPT(1:10,1) = {Sides(1)};
   SideSPT(11:20,1) = {Sides(2)};
  else
   SideSPT(1:7,1) = {Sides(1)};
   SideSPT(8:14,1) = {Sides(2)};
  end
  clearvars a k
 else %put all operated sides on top & unoperated data below
  for k = 1:length(FN)

   b = AvData.(splitStr{7}).(Groups{o}).(FN{k}).Spatiotemporal(1);
   if strcmp(b{1},'Operated')
    b = AvData.(splitStr{7}).(Groups{o}).(FN{k}).Spatiotemporal(:,3);
   elseif strcmp(b{1},'Unoperated')
    c = AvData.(splitStr{7}).(Groups{o}).(FN{k}).Spatiotemporal(:,3);
    if strcmp(splitStr{7},'LevelWalking') || strcmp(splitStr{7},'StairsAscent') ||
strcmp(splitStr{7},'StairsDescent')
     b (1:10) = c(11:20);
     b (11:20) = c(1:10);
    else
     b (1:7) = c(8:14);
     b (8:14) = c(1:7);
    end
    b = b';
   end
   Spatiotemp = horzcat(Spatiotemp,b);
  end
  SideSPT(1:7,1) = {Sides(3)};
  SideSPT(8:14,1) = {Sides(4)};
 end
 clearvars a k b c
```

```
 TabHeadings = vertcat('Side','Parameter',FN)';
 Merge = horzcat(SideSPT,Spatiotemp);
 MergeTab = array2table(Merge,'VariableNames',TabHeadings);
 RawAvData.(splitStr{7}).(Groups{o}).RawSPT = MergeTab;
 save([newpath 'RawAvData.mat'],'RawAvData');

 if o ~=1 %divide patient SPT by implant
  for ii = 1:3
   a = ismember(TrialNames{2,ii},FN);
   c = TrialNames{2,ii}(a);
   t0= horzcat(MergeTab(:,1:2),MergeTab(:,c));
   tables{ii} = t0;
   RawImplantData.(splitStr{7}).(Groups{o}).(ImplantGroup{ii}).RawSPT = tables{1,ii};
   save([newpath 'RawImplantData.mat'],'RawImplantData')
   clearvars a c t0
  end
 end
 clearvars TabHeadings Merge MergeTab Spatiotemp

end
RawImplantData.(splitStr{7}).Control.RawSPT = RawAvData.(splitStr{7}).Control.RawSPT;
clearvars b c o ii
```

```
% Incline task belt speed
if strcmp(splitStr{7},'LevelWalking') == 0 && strcmp(splitStr{7},'StairsAscent') == 0
&& strcmp(splitStr{7},'StairsDescent') == 0

 for o = 1:3
  for i = 1:5
   TrialNames{1,o} = fieldnames(AvData.(splitStr{7}).(Groups{o}));
   TempVar = [];
   a = 1;
   clearvars k b c
   for k = 1:length(TrialNames{1,o})
    if isfield(AvData.(splitStr{7}).(Groups{o}),(TrialNames{1,o}{k,1}))
     c = AvData.(splitStr{7}).(Groups{o}).(TrialNames{1,o}{k,1}).BeltSpeed;
     TempVar = horzcat (TempVar,c);
     b{a,:} = TrialNames{1,o}{k,1};
     a = a + 1;
    end
   end

   TempVar = array2table(TempVar,'VariableNames',b);
   RawAvData.(splitStr{7}).(Groups{o}).BeltSpeed = TempVar;
   save([newpath 'RawAvData.mat'],'RawAvData')
   if o == 1
    RawImplantData.(splitStr{7}).(Groups{o}).BeltSpeed = TempVar;
   else %implant organising
    for ii = 1:3
     a = ismember(TrialNames{2,ii},TrialNames{1,o});
     b = TrialNames{2,ii}(a);
     t0 = TempVar(:,b);
     t0.Properties.VariableNames = b;
     tables{ii} = t0;
     clearvars a b TempVarVars t0
     %export average patient raw data to RawAvData
     RawImplantData.(splitStr{7}).(Groups{o}).(ImplantGroup{ii}).BeltSpeed =
tables{1,ii};
     save([newpath 'RawImplantData.mat'],'RawImplantData')
    end
   end
  end
 end
end

% toc
```

9.4.4. Export Discrete Parameters

```
% Script to export paramaters for statistical analysis. Values are saved in Stats
% struct of RawImplantData structure - below are edits to the script to allow data
% exported from stance only. C Govind

clc
clear
```

```
%set up
newpath = 'H:\Raw Data\NexusOutput\BBraunMasterMotionCaptureDatabase\';
load([newpath 'RawImplantData.mat'])

Groups ={'Control', 'PreOp','YearPostOp'}; %o
Sides = {'Left', 'Right', 'Operated', 'Unoperated'}; %l
ImplantGroup = {'Implant1','Implant2','Implant3'}; %s %for o = 2%:3
Parameters = {'Angles','Moment','Power'}; %n

TrialType = {'LevelWalking',...  %1
 'StairsAscent','StairsDescent',...% 2 and 3
 'Incline7','Incline6','Incline1',...
 'Incline3','Incline4'}; %t
```

```
%FILL IN

% make pop up?
MeasureParam = Parameters{1};

if strcmp(MeasureParam, 'Angles')
 FN =
{'PelvisFlexion';'PelvisAbAd';'PelvisRot';'HipFlexion';'HipAbAd';'HipRot';'KneeFlexio
n';'KneeAbAd';'KneeRot';'AnkleFlexion';'AnkleAbAd';'AnkleRot'};
elseif strcmp(MeasureParam, 'Moment')
 FN =
{'HipFlexion';'HipAbAd';'HipRot';'KneeFlexion';'KneeAbAd';'KneeRot';'AnkleFlexion';'A
nkleAbAd';'AnkleRot'};
else
 FN = {'HipPower';'KneePower';'AnklePower'};
end
```

```
%Output Control data to do stats on and save in RawImplantData.Activity.Control.stats
tic
for t = 1:8
 if isfield(RawImplantData.(TrialType{t}).Control,['BothRaw' MeasureParam])
  for jj = 1:length(FN)
   RawData = table2array(RawImplantData.(TrialType{t}).Control.(['BothRaw'
MeasureParam]).(FN{jj}));
   RawTable = RawImplantData.(TrialType{t}).Control.(['BothRaw'
MeasureParam]).(FN{jj});
   Colheadings = RawImplantData.(TrialType{t}).Control.(['BothRaw'
MeasureParam]).(FN{jj}).Properties.VariableNames;

   %max min each row = person, each column = joint + direction
   maxVal(:,jj) = max(RawData);
   minVal(:,jj) = min(RawData);

   %kinematics
   if strcmp(MeasureParam,'Angles')
    % knee stance peak/trough during level walking/incline only
    if strcmp(TrialType{t},'LevelWalking') && jj == 7 ||
sum(ismember(TrialType{t},'Incline')) == 7 && jj == 7
     for kk = 1:length(Colheadings)
      StancePeak = max(RawData(1:40,:))';
      StanceTrough = min(RawData(15:60,:))';
     end
     RawImplantData.(TrialType{t}).Control.Stats.(MeasureParam).MaxStanceKneeFlex =
array2table(StancePeak,'RowNames',Colheadings);
     RawImplantData.(TrialType{t}).Control.Stats.(MeasureParam).MinStanceKneeFlex =
array2table(StanceTrough,'RowNames',Colheadings);
    end

    %sagittal data at HS and TO
    if jj == 1 || jj == 4 || jj == 7 || jj == 10
     DataAtHS(:,jj) = RawData(1,:); %data at 1st gait cycle % = heel strike
     for kk = 1:length(Colheadings)
      TOP =
round(RawImplantData.(TrialType{t}).Control.BothToeOff.(Colheadings{kk}));
      DataAtTO(kk,jj) = table2array(RawTable(TOP,Colheadings{kk}));
      clearvars TOP
     end
     DataAtTO( :, all(~DataAtTO,1) ) = [];
     DataAtHS( :, all(~DataAtHS,1) ) = [];
     RawImplantData.(TrialType{t}).Control.Stats.(MeasureParam).DataAtHS =
array2table(DataAtHS,'RowNames',Colheadings);
     RawImplantData.(TrialType{t}).Control.Stats.(MeasureParam).DataAtTO =
array2table(DataAtTO,'RowNames',Colheadings);
    end

    %ROM
    ROM = maxVal - minVal;
    RawImplantData.(TrialType{t}).Control.Stats.(MeasureParam).ROM =
array2table(ROM,'RowNames',Colheadings);
```

```
    save([newpath 'RawImplantData.mat'],'RawImplantData')

   elseif sum(ismember(TrialType{t},'Incline')) <= 7 %KINETICS
    %max min DURING STANCE export more peaks in KFM,KAM?
    for kk = 1:length(Colheadings)
     TOP = round(RawImplantData.(TrialType{t}).Control.BothToeOff.(Colheadings{kk}));
     StanceMom = table2array(RawTable(1:TOP,Colheadings{kk}));
     maxVal(kk,jj) = max(StanceMom);
     minVal(kk,jj) = min(StanceMom);
     clearvars StanceMom TOP
    end

    %flexion angle at max momemtnt
    if strcmp(MeasureParam,'Moment') && jj == 4 %no pelvis moment so jj = 4 instead
of 7
     for kk = 1:length(Colheadings)
      [~,MaxKFMF] =
max(RawImplantData.(TrialType{t}).Control.BothRawMoment.KneeFlexion.(Colheadings{kk})
); %Max knee flexion moment frame
       FlexAtMaxMom(kk,:) =
RawImplantData.(TrialType{t}).Control.BothRawAngles.KneeFlexion.(Colheadings{kk})(Max
KFMF);
      clearvars MaxKFMF
     end
     RawImplantData.(TrialType{t}).Control.Stats.(MeasureParam).FlexAtMaxMom =
array2table(FlexAtMaxMom,'RowNames',Colheadings);
     clearvars FlexAtMaxMom kk
    end

    %KAM Peaks
    if strcmp(MeasureParam,'Moment') && jj == 5
     KAM(:,1) = max(RawData(1:30,:)); %early stance
     KAM(:,2) = min(RawData(20:50,:)); %midstance
     KAM(:,3) = max(RawData(40:80,:)); %late stance
     RawImplantData.(TrialType{t}).Control.Stats.(MeasureParam).KAMPeaks =
array2table(KAM,'RowNames',Colheadings);
     clearvars KAM
    end

    %KFM Peaks
    if strcmp(MeasureParam,'Moment') && jj == 4
     KFM(:,1) = max(RawData(1:30,:));
     KFM(:,2) = min(RawData(20:50,:));
     KFM(:,3) = max(RawData(40:80,:));

     RawImplantData.(TrialType{t}).Control.Stats.(MeasureParam).KFMPeaks =
array2table(KFM,'RowNames',Colheadings);
     clearvars KFM
    end

    %ankle power peaks
    if strcmp(MeasureParam,'Power') && jj == 3
     AnklePowerPeaks(:,1) = max(RawData(40:80,:));
     AnklePowerPeaks(:,2) = min(RawData(30:60,:));

     if strcmp(TrialType{t},'StairsAscent') %EXTRA peak/trough for stairs only
      AnklePowerPeaks(:,3) = max(RawData(1:40,:));
     elseif strcmp(TrialType{t},'StairsDescent')
      AnklePowerPeaks(:,3) = min(RawData(1:40,:));
     end
     RawImplantData.(TrialType{t}).Control.Stats.(MeasureParam).APPeaks =
array2table(AnklePowerPeaks,'RowNames',Colheadings);
     clearvars AnklePowerPeaks
    end

    %knee power peaks
    if strcmp(MeasureParam,'Power') && jj == 2
     KneePowerPeaks(:,1) = max(RawData(10:30,:));
     if strcmp(TrialType{t},'StairsAscent') %peak in Stair Ascent
      KneePowerPeaks(:,2) = max(RawData(50:70,:));
     else %trough in level and stair descent
      KneePowerPeaks(:,2) = min(RawData(50:70,:));
     end
     RawImplantData.(TrialType{t}).Control.Stats.(MeasureParam).KPPeaks =
array2table(KneePowerPeaks,'RowNames',Colheadings);
     clearvars KneePowerPeaks
    end
    save([newpath 'RawImplantData.mat'],'RawImplantData')
   end
  end
  RawImplantData.(TrialType{t}).Control.Stats.(MeasureParam).maxVal =
array2table(maxVal,'RowNames',Colheadings);
```

```
  RawImplantData.(TrialType{t}).Control.Stats.(MeasureParam).minVal =
array2table(minVal,'RowNames',Colheadings);
  clearvars RawData RawTable

  save([newpath 'RawImplantData.mat'],'RawImplantData')
  clearvars Colheadings kk ii jj maxVal minVal RawData ROM StancePeak StanceTrough
DataAtHS DataAtTO ToeOffs
 else
  disp('no data')
 end
end

disp('Control done')
```

```
%patients
for t = 1:8
 for o = 2:3
  for s = 1:3
   for ii = 3:4
    if
isfield(RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}),([Sides{ii}(1)
'Raw' MeasureParam]))
     for jj = 1:length(FN)
      RawData =
table2array(RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).([Sides{ii}(1
) 'Raw' MeasureParam]).(FN{jj}));
      RawTable =
(RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).([Sides{ii}(1) 'Raw'
MeasureParam]).(FN{jj}));
      Colheadings =
RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).([Sides{ii}(1) 'Raw'
MeasureParam]).(FN{jj}).Properties.VariableNames;

      maxVal(:,jj) = max(RawData);
      minVal(:,jj) = min(RawData);

      %KINEMATICS
      if strcmp(MeasureParam,'Angles')
       % knee stance peak/trough during walking
       if strcmp(TrialType{t},'LevelWalking') && jj == 7 ||
sum(ismember(TrialType{t},'Incline')) == 7 && jj == 7
        StancePeak = max(RawData(1:40,:))';
        StanceTrough = min(RawData(15:70,:))';

RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).Stats.(Sides{ii}).(Measur
eParam).MaxStanceKneeFlex = array2table(StancePeak,'RowNames',Colheadings);

RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).Stats.(Sides{ii}).(Measur
eParam).MinStanceKneeFlex = array2table(StanceTrough,'RowNames',Colheadings);
       end

       % Sagittal sata at HS & TO
       if jj == 1 || jj == 4 || jj == 7 || jj == 10
        DataAtHS(:,jj) = RawData(1,:);
        for kk = 1:length(Colheadings)
         TOP =
round(RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).([Sides{ii}(1)
'ToeOff']).(Colheadings{kk}));
         DataAtTO(kk,jj) = table2array(RawTable(TOP,Colheadings{kk}));
         clearvars TOP
        end
        DataAtTO( :, all(~DataAtTO,1) ) = [];
        DataAtHS( :, all(~DataAtHS,1) ) = [];

RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).Stats.(Sides{ii}).(Measur
eParam).DataAtHS = array2table(DataAtHS,'RowNames',Colheadings);

RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).Stats.(Sides{ii}).(Measur
eParam).DataAtTO = array2table(DataAtTO,'RowNames',Colheadings);
       end

       %ROM
       ROM = maxVal - minVal;

RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).Stats.(MeasureParam).ROM
= array2table(ROM,'RowNames',Colheadings);

       save([newpath 'RawImplantData.mat'],'RawImplantData')

      elseif sum(ismember(TrialType{t},'Incline')) <= 7 %KINETICS
       %max min DURING STANCE
       for kk = 1:length(Colheadings)
        TOP =
```

```
round(RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).([Sides{ii}(1)
'ToeOff']).(Colheadings{kk}));
        StanceMom = table2array(RawTable(1:TOP,Colheadings{kk}));
        maxVal(kk,jj) = max(StanceMom);
        minVal(kk,jj) = min(StanceMom);
        clearvars StanceMom TOP
       end

       %flexion angle at max momemtn
       if strcmp(MeasureParam,'Moment') && jj == 4 %no pelvis moment so jj = 4
instead of 7
        for kk = 1:length(Colheadings)
         [~,MaxKFMF] =
max(RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).([Sides{ii}(1)
'RawMoment']).KneeFlexion.(Colheadings{kk})); %Max knee flexion moment frame
         FlexAtMaxMom(kk,:) =
RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).([Sides{ii}(1)
'RawAngles']).KneeFlexion.(Colheadings{kk})(MaxKFMF);
        clearvars MaxKFMF
       end

RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).Stats.(Sides{ii}).(Measur
eParam).FlexAtMaxMom = array2table(FlexAtMaxMom,'RowNames',Colheadings);
        save([newpath 'RawImplantData.mat'],'RawImplantData')
        clearvars FlexAtMaxMom kk
       end

       %KAM Peaks
       if strcmp(MeasureParam,'Moment') && jj == 5
        KAM(:,1) = max(RawData(1:30,:)); %early stance
        KAM(:,2) = min(RawData(20:50,:)); %midstance
        KAM(:,3) = max(RawData(40:80,:)); %late stance

RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).Stats.(Sides{ii}).(Measur
eParam).KAMPeaks = array2table(KAM,'RowNames',Colheadings);
        clearvars KAM
       end

       %KFM Peaks
       if strcmp(MeasureParam,'Moment') && jj == 4
        KFM(:,1) = max(RawData(1:30,:));
        KFM(:,2) = min(RawData(20:50,:));
        KFM(:,3) = max(RawData(40:80,:));


RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).Stats.(Sides{ii}).(Measur
eParam).KFMPeaks = array2table(KFM,'RowNames',Colheadings);
        clearvars KFM
       end

       %ankle power peaks
       if strcmp(MeasureParam,'Power') && jj == 3
        AnklePowerPeaks(:,1) = max(RawData(40:80,:));
        AnklePowerPeaks(:,2) = min(RawData(30:60,:));

        if strcmp(TrialType{t},'StairsAscent') %EXTRA peak/trough for stairs only
         AnklePowerPeaks(:,3) = max(RawData(1:40,:));
        elseif strcmp(TrialType{t},'StairsDescent')
         AnklePowerPeaks(:,3) = min(RawData(1:40,:));
        end

RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).Stats.(Sides{ii}).(Measur
eParam).APPeaks = array2table(AnklePowerPeaks,'RowNames',Colheadings);
        clearvars AnklePowerPeaks
       end

       %knee power peaks
       if strcmp(MeasureParam,'Power') && jj == 2
        KneePowerPeaks(:,1) = max(RawData(10:30,:));
        if strcmp(TrialType{t},'StairsAscent') %peak in Stair Ascent
         KneePowerPeaks(:,2) = max(RawData(50:70,:));
        else %trough in level and stair descent
         KneePowerPeaks(:,2) = min(RawData(50:70,:));
        end

RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).Stats.(Sides{ii}).(Measur
eParam).KPPeaks = array2table(KneePowerPeaks,'RowNames',Colheadings);
        clearvars KneePowerPeaks
       end
      end
     end

RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).Stats.(Sides{ii}).(Measur
```

```
eParam).maxVal = array2table(maxVal,'RowNames',Colheadings);

RawImplantData.(TrialType{t}).(Groups{o}).(ImplantGroup{s}).Stats.(Sides{ii}).(Measur
eParam).minVal = array2table(minVal,'RowNames',Colheadings);

     save([newpath 'RawImplantData.mat'],'RawImplantData')
     clearvars ToeOffs Colheadings kk jj maxVal minVal RawData ROM StancePeak
StanceTrough DataAtHS DataAtTO RawData
    end
    clearvars ROM maxVal minVal RawData ROM StancePeak StanceTrough DataAtHS DataAtTO
Colheadings kk
   end
  end
 end
 clearvars ii jj
end
clearvars FN
disp('data done')
beep
```

```
%script to extract peak knee ab/ad agle DURING STANCE. Can paste into excel
%to remove NaN then paste into SPSS to do stats.CG 11/01/19

clc
clear
load('D:\LaptopData\NexusOutput\BBraunMasterMotionCaptureDatabase\RawImplantData.mat'
)
```

```
ImplantGroup = {'Implant1','Implant2','Implant3'}; %s %for o = 2%:3
OpState = {'PreOp','YearPostOp'};
dir =
{'LevelWalking','StairsAscent','StairsDescent','Incline1','Incline3','Incline4','Incl
ine6','Incline7'};

for l = 1:8
 Contdata = RawImplantData.(dir{l}).Control.BothRawAngles.KneeAbAd;
 ContTO = RawImplantData.(dir{l}).Control.BothToeOff;
 for m = 1:width(ContTO)
  name = ContTO.Properties.VariableNames{m};
  TOnum = round(ContTO.(name),0);
  data(m,:) = max(Contdata.(name)(1:TOnum,1)); %change to min if want to calculate
ROM = Max - min
 end
 dump{:,l} = data;
 clearvars Contdata ContTO name TOnum data
end
clearvars a l m

for l = 1:8
 a = 2;
 for o = 1:2
  for i = 1:3
   Patientdata =
RawImplantData.(dir{l}).(OpState{o}).(ImplantGroup{i}).ORawAngles.KneeAbAd;
   PaitTO = RawImplantData.(dir{l}).(OpState{o}).(ImplantGroup{i}).OToeOff;
   for m = 1:width(PaitTO)
    name = PaitTO.Properties.VariableNames{m};
    TOnum = round(PaitTO.(name),0);
    data(m,:) = max(Patientdata.(name)(1:TOnum,1)); %change to min if want to
calculate ROM = Max - min
   end
   dump{a,l} = data;
   a=a+1;
   clearvars Patientdata PaitTO name TOnum data
  end
 end
end
clearvars a i l m n o
%dump set up: each column is an activity in line with dir
%each row is a participant group order as so:
%Control
%preCRDD
%postCRDD
%preUC
%postUC
%preUCR
%postUCR

a = dump;
b = cellfun('size',a,1);
c = max(b');

d = 1;
```

```
f = [];
for i = 1:(length(c))
 e = ones(c(i),1)*d;
 f = vertcat(f,e);
 d = d+1;
end
b1 =
vertcat(ones(c(1),1)*7,ones(c(2),1)*1,ones(c(3),1)*2,ones(c(4),1)*3,ones(c(5),1)*4,on
es(c(6),1)*5,ones(c(7),1)*6); %each group cont = 7
b2 =
vertcat(ones(c(1),1)*7,ones(c(2),1)*0,ones(c(3),1)*1,ones(c(4),1)*0,ones(c(5),1)*1,on
es(c(6),1)*0,ones(c(7),1)*1); %pre/post pre = 0, post = 1
b3 =
vertcat(ones(c(1),1)*7,ones(c(2),1)*1,ones(c(3),1)*1,ones(c(4),1)*2,ones(c(5),1)*2,on
es(c(6),1)*3,ones(c(7),1)*3); % implant, crdd = 1, uc =2, ucr = 3

%fill gaps to make full matrix
for i = 1:size(a,2) %per activity
 for j = 1:7 %per participant group
  a{j,i}(end+1:c(j))=nan;
 end
end

merged = vertcat(a{:});
A = reshape(merged,[],8); %change A to A2 if wanting to calculate ROM
A = horzcat(b1,b2,b3,A2);
clearvars b d e a c f i j merged b1 b2 b3
disp ('!!! Paste into Excel and replace NaN for .!!')

ROM = A - A2
```

9.4.5. Export to SPSS

```
% Script to extract all parameters per activity to paste into SPSS later. Group data
outputted is of both sides and control data is of 1 randomly selected limb. Below
scipt handles exporting STP data. C Govind.


clc
clear
```

```
% set up
newpath = 'H:\Raw Data\NexusOutput\BBraunMasterMotionCaptureDatabase\';
load([newpath 'RawImplantData.mat'])

Sides = {'Left', 'Right', 'Operated', 'Unoperated'}; %1
ImplantGroup = {'Implant1','Implant2','Implant3'}; %s %for o = 2%:3

dir =
{'LevelWalking','StairsAscent','StairsDescent','Incline1','Incline3','Incline4','Incl
ine6','Incline7'};
Parameters = {'Angles','Moment','Power','SPT'};

[s,~] = listdlg('PromptString','Trial type?',...
 'SelectionMode','single',...
 'ListString',dir);

[t,~] = listdlg('PromptString','Parameter?',...
 'SelectionMode','single',...
 'ListString',Parameters);

MeasureParam = Parameters{t};
splitStr{7} = dir{s};


choice = menu('What type of data is this?',...
 'Max Val','ROM','Data at HS and TO','Min val', 'Max/Min Stance', 'Flex at max
moment','SPT');

ToStats = {'maxVal'; 'ROM'; 'DataAtHS'; 'DataAtTO'; 'minVal'; 'MaxStanceKneeFlex';...
 'MinStanceKneeFlex'; 'FlexAtMaxMom';};

switch choice
 case 1 %max
  ToKernel = ToStats{1};
 case 2 %ROM
  ToKernel = ToStats{2};
 case 3 %for horzcat later on of TO and HS data. Order: Pelvis, hip, knee, ankle
```

```
  ToKernel{1} = ToStats{3};
  ToKernel{2} = ToStats{4};
 case 4 % Min data- for moments
  ToKernel = ToStats{5};
 case 5 %for horzcat later on of max and min stance data
  ToKernel{1} = ToStats{6};
  ToKernel{2} = ToStats{7};
 case 6 %knee flexion at max moment
  ToKernel = ToStats{8};
 case 7 %knee flexion at max moment
  ToKernel = ToStats{8};

end

clearvars s t dir ToStats
```

```
% get data
%preallocate
Contdata = {0; 0};
Predata = {0 0 0; 0 0 0};
Postdata = {0 0 0; 0 0 0};

ContStats = RawImplantData.(splitStr{7}).Control.Stats.(MeasureParam);


try
 ContZeros = array2table(zeros(height(ContStats.(ToKernel)),3));
catch
 ContZeros = array2table(zeros(height(ContStats.(ToKernel{1})),3));
end

if choice ~= 3 && choice ~= 5 && choice ~= 7 %if choice is ROM, Max, Min or flexion
at max moment
 if strcmp(MeasureParam, 'Angles')
  Contdata = table2array(ContStats.(ToKernel));
  for l = 1:2
   for ii = 1:3 %operated on top, unop below
    Predata{l,ii} =
table2array(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).
(MeasureParam).(ToKernel));
    Postdata{l,ii} =
table2array(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).Stats.(Sides{l
+2}).(MeasureParam).(ToKernel));
   end
  end
 elseif strcmp(MeasureParam, 'Moment')
  if strcmp(ToKernel,'maxVal')  %3x hip moment     %2x max KFM peaks         %3x KAM
peaks             %knee rotation and ankle moment peaks
   Contdata =
table2array(horzcat(ContStats.(ToKernel)(:,1:3),ContStats.KFMPeaks(:,1),ContStats.KFM
Peaks(:,3),ContStats.KAMPeaks(:,1),ContStats.KAMPeaks(:,2),ContStats.KAMPeaks(:,3),Co
ntStats.(ToKernel)(:,6:9)));
   for l = 1:2
    for ii = 1:3
     PreStats =
RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(MeasurePara
m);
     PostStats =
RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(Measur
eParam);
     Predata{l,ii} =
table2array(horzcat(PreStats.(ToKernel)(:,1:3),PreStats.KFMPeaks(:,1),PreStats.KFMPea
ks(:,3),PreStats.KAMPeaks(:,1),PreStats.KAMPeaks(:,2),PreStats.KAMPeaks(:,3),PreStats
.(ToKernel)(:,6:9)));
     Postdata{l,ii} =
table2array(horzcat(PostStats.(ToKernel)(:,1:3),PostStats.KFMPeaks(:,1),PostStats.KFM
Peaks(:,3),PostStats.KAMPeaks(:,1),PostStats.KAMPeaks(:,2),PostStats.KAMPeaks(:,3),Po
stStats.(ToKernel)(:,6:9)));
    end
   end
  elseif strcmp(ToKernel,'minVal')
   %horzcat with 0's for spss
   Contdata =
table2array(horzcat(ContStats.(ToKernel)(:,1:3),ContStats.KFMPeaks(:,2),ContZeros(:,1
),ContStats.(ToKernel)(:,5),ContZeros(:,2),ContZeros(:,3),ContStats.(ToKernel)(:,6:9)
));
   for l = 1:2
    for ii = 1:3
     PreStats =
RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(MeasurePara
m);
     PostStats =
RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(Measur
```

```
eParam);
     PreZeros = array2table(zeros(height(PreStats.KFMPeaks(:,2)),3));
     PostZeros = array2table(zeros(height(PostStats.KFMPeaks(:,2)),3));
     Predata{l,ii} =
table2array(horzcat(PreStats.(ToKernel)(:,1:3),PreStats.KFMPeaks(:,2),PreZeros(:,1),P
reStats.(ToKernel)(:,5),PreZeros(:,2),PreZeros(:,3),PreStats.(ToKernel)(:,6:9)));
     Postdata{l,ii} =
table2array(horzcat(PostStats.(ToKernel)(:,1:3),PostStats.KFMPeaks(:,2),PostZeros(:,1
),PostStats.(ToKernel)(:,5),PostZeros(:,2),PostZeros(:,3),PostStats.(ToKernel)(:,6:9)
));
    end
   end
  elseif strcmp(ToKernel,'FlexAtMaxMom')
   Contdata = table2array(ContStats.(ToKernel));
   for l = 1:2
    for ii = 1:3 %operated on top, unop below
     Predata{l,ii} =
table2array(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).
(MeasureParam).(ToKernel));
     Postdata{l,ii} =
table2array(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).Stats.(Sides{l
+2}).(MeasureParam).(ToKernel));
    end
   end
  end
 elseif strcmp(MeasureParam, 'Power')
  if strcmp(ToKernel,'maxVal')
   if strcmp(splitStr{7},'LevelWalking')
    Contdata =
table2array(horzcat(ContStats.(ToKernel)(:,1),ContStats.KPPeaks(:,1),ContZeros(:,1),C
ontStats.APPeaks(:,1),ContZeros(:,2)));
    for l = 1:2
     for ii = 1:3
      PreStats =
RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(MeasurePara
m);
      PostStats =
RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(Measur
eParam);
      PreZeros = array2table(zeros(height(PreStats.KPPeaks(:,2)),2));
      PostZeros = array2table(zeros(height(PostStats.KPPeaks(:,2)),2));
      Predata{l,ii} =
table2array(horzcat(PreStats.(ToKernel)(:,1),PreStats.KPPeaks(:,1),PreZeros(:,1),PreS
tats.APPeaks(:,1),PreZeros(:,2)));
      Postdata{l,ii} =
table2array(horzcat(PostStats.(ToKernel)(:,1),PostStats.KPPeaks(:,1),PostZeros(:,1),P
ostStats.APPeaks(:,1),PostZeros(:,2)));
     end
    end
   elseif strcmp(splitStr{7},'StairsAscent')
    Contdata =
table2array(horzcat(ContStats.(ToKernel)(:,1),ContStats.KPPeaks(:,1),ContStats.KPPeak
s(:,2),ContStats.APPeaks(:,1),ContStats.APPeaks(:,3)));
    for l = 1:2
     for ii = 1:3
      PreStats =
RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(MeasurePara
m);
      PostStats =
RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(Measur
eParam);
      PreZeros = array2table(zeros(height(PreStats.KPPeaks(:,2)),2));
      PostZeros = array2table(zeros(height(PostStats.KPPeaks(:,2)),2));
      Predata{l,ii} =
table2array(horzcat(PreStats.(ToKernel)(:,1),PreStats.KPPeaks(:,1),PreStats.KPPeaks(:
,2),PreStats.APPeaks(:,1),PreStats.APPeaks(:,3)));
      Postdata{l,ii} =
table2array(horzcat(PostStats.(ToKernel)(:,1),PostStats.KPPeaks(:,1),PostStats.KPPeak
s(:,2),PostStats.APPeaks(:,1),PostStats.APPeaks(:,3)));
     end
    end
   elseif strcmp(splitStr{7},'StairsDescent')
    Contdata =
table2array(horzcat(ContStats.(ToKernel)(:,1),ContStats.KPPeaks(:,1),ContZeros(:,1),C
ontStats.APPeaks(:,1),ContZeros(:,2)));
    for l = 1:2
     for ii = 1:3
      PreStats =
RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(MeasurePara
m);
      PostStats =
RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(Measur
eParam);
```

```
        PreZeros = array2table(zeros(height(PreStats.KPPeaks(:,2)),2));
        PostZeros = array2table(zeros(height(PostStats.KPPeaks(:,2)),2));
        Predata{l,ii} =
table2array(horzcat(PreStats.(ToKernel)(:,1),PreStats.KPPeaks(:,1),PreZeros(:,1),PreS
tats.APPeaks(:,1),PreZeros(:,2)));
        Postdata{l,ii} =
table2array(horzcat(PostStats.(ToKernel)(:,1),PostStats.KPPeaks(:,1),PostZeros(:,1),P
ostStats.APPeaks(:,1),PostZeros(:,2)));
       end
      end
     end
    elseif strcmp(ToKernel,'minVal') %extra 0 column for level walking ankle power
     if strcmp(splitStr{7},'LevelWalking')
      Contdata =
table2array(horzcat(ContStats.(ToKernel)(:,1),ContStats.KPPeaks(:,2),ContZeros(:,1),C
ontStats.APPeaks(:,2),ContZeros(:,2)));
      for l = 1:2
       for ii = 1:3
        PreStats =
RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(MeasurePara
m);
        PostStats =
RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(Measur
eParam);
        PreZeros = array2table(zeros(height(PreStats.KPPeaks(:,2)),2));
        PostZeros = array2table(zeros(height(PostStats.KPPeaks(:,2)),2));
        Predata{l,ii} =
table2array(horzcat(PreStats.(ToKernel)(:,1),PreStats.KPPeaks(:,2),PreZeros(:,1),PreS
tats.APPeaks(:,2),PreZeros(:,2)));
        Postdata{l,ii} =
table2array(horzcat(PostStats.(ToKernel)(:,1),PostStats.KPPeaks(:,2),PostZeros(:,1),P
ostStats.APPeaks(:,2),PostZeros(:,2)));
       end
      end
     elseif strcmp(splitStr{7},'StairsAscent')
      Contdata =
table2array(horzcat(ContStats.(ToKernel)(:,1),ContStats.(ToKernel)(:,2),ContZeros(:,1
),ContStats.APPeaks(:,2),ContZeros(:,2)));
      for l = 1:2
       for ii = 1:3
        PreStats =
RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(MeasurePara
m);
        PostStats =
RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(Measur
eParam);
        PreZeros = array2table(zeros(height(PreStats.KPPeaks(:,2)),2));
        PostZeros = array2table(zeros(height(PostStats.KPPeaks(:,2)),2));
        Predata{l,ii} =
table2array(horzcat(PreStats.(ToKernel)(:,1),PreStats.(ToKernel)(:,2),PreZeros(:,1),P
reStats.APPeaks(:,2),PreZeros(:,2)));
        Postdata{l,ii} =
table2array(horzcat(PostStats.(ToKernel)(:,1),PostStats.(ToKernel)(:,2),PostZeros(:,1
),PostStats.APPeaks(:,2),PostZeros(:,2)));
       end
      end
     elseif strcmp(splitStr{7},'StairsDescent')
      Contdata =
table2array(horzcat(ContStats.(ToKernel)(:,1),ContStats.KPPeaks(:,2),ContZeros(:,1),C
ontStats.APPeaks(:,2),ContStats.APPeaks(:,3)));
      for l = 1:2
       for ii = 1:3
        PreStats =
RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(MeasurePara
m);
        PostStats =
RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(Measur
eParam);
        PreZeros = array2table(zeros(height(PreStats.KPPeaks(:,2)),2));
        PostZeros = array2table(zeros(height(PostStats.KPPeaks(:,2)),2));
        Predata{l,ii} =
table2array(horzcat(PreStats.(ToKernel)(:,1),PreStats.KPPeaks(:,2),PreZeros(:,1),PreS
tats.APPeaks(:,2),PreStats.APPeaks(:,3)));
        Postdata{l,ii} =
table2array(horzcat(PostStats.(ToKernel)(:,1),PostStats.KPPeaks(:,2),PostZeros(:,1),P
ostStats.APPeaks(:,2),PostStats.APPeaks(:,3)));
       end
      end
     end
    end
   end

elseif choice == 3 %merge HS/TO data
```

```
 Contdata =
horzcat(table2array(ContStats.(ToKernel{1})),table2array(ContStats.(ToKernel{2})));
 for l = 1:2
  for ii = 1:3
   Predata{l,ii,:} =
horzcat(table2array(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).Stats.(Side
s{l+2}).(MeasureParam).(ToKernel{1})),table2array(RawImplantData.(splitStr{7}).PreOp.
(ImplantGroup{ii}).Stats.(Sides{l+2}).(MeasureParam).(ToKernel{2})));
   Postdata{l,ii,:} =
horzcat(table2array(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).Stats.
(Sides{l+2}).(MeasureParam).(ToKernel{1})),table2array(RawImplantData.(splitStr{7}).Y
earPostOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(MeasureParam).(ToKernel{2})));
  end
 end
elseif choice == 5 %merge stance max/min
 if strcmp(splitStr{7},'LevelWalking') || sum(ismember(char(splitStr{7}),'Incline'))
== 7
  Contdata =
horzcat(table2array(ContStats.(ToKernel{1})),table2array(ContStats.(ToKernel{2})));
  for l = 1:2
   for ii = 1:3
    Predata{l,ii,:} =
horzcat(table2array(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).Stats.(Side
s{l+2}).(MeasureParam).(ToKernel{1})),table2array(RawImplantData.(splitStr{7}).PreOp.
(ImplantGroup{ii}).Stats.(Sides{l+2}).(MeasureParam).(ToKernel{2})));
    Postdata{l,ii,:} =
horzcat(table2array(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).Stats.
(Sides{l+2}).(MeasureParam).(ToKernel{1})),table2array(RawImplantData.(splitStr{7}).Y
earPostOp.(ImplantGroup{ii}).Stats.(Sides{l+2}).(MeasureParam).(ToKernel{2})));
   end
  end
 end
elseif choice == 7 %SPT, L and operated on top
 if strcmp(splitStr{7},'LevelWalking')
  Contdata{1,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).Control.RawSPT(1:10,3:end)))';
  Contdata{2,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).Control.RawSPT(11:20,3:end)))';
  for ii = 1:3
   Predata{1,ii,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).RawSPT(1:1
0,3:end)))';
   Predata{2,ii,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).RawSPT(11:
20,3:end)))';
   Postdata{1,ii,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).RawSP
T(1:10,3:end)))';
   Postdata{2,ii,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).RawSP
T(11:20,3:end)))';
  end
 elseif strcmp(splitStr{7},'StairsAscent') || strcmp(splitStr{7},'StairsDescent')
  Contdata{1,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).Control.RawSPT(1:8,3:end)))';
  Contdata{2,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).Control.RawSPT(11:18,3:end)))';
  for ii = 1:3
   Predata{1,ii,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).RawSPT(1:8
,3:end)))';
   Predata{2,ii,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).RawSPT(1:8
,3:end)))';
   Postdata{1,ii,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).RawSP
T(1:8,3:end)))';
   Postdata{2,ii,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).RawSP
T(1:8,3:end)))';
  end
 elseif sum(ismember(char(splitStr{7}),'Incline')) == 7
  Contdata{1,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).Control.RawSPT(1:7,3:end)))';
  Contdata{2,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).Control.RawSPT(1:7,3:end)))';
  for ii = 1:3
   Predata{1,ii,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).RawSPT(1:7
,3:end)))';
   Predata{2,ii,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).RawSPT(1:7
,3:end)))';
```

```
    Postdata{1,ii,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).RawSP
T(1:7,3:end)))';
    Postdata{2,ii,:} =
cell2mat(table2array(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).RawSP
T(1:7,3:end)))';
   end
  end
end

PlotGroups = {Contdata,Predata,Postdata}; %operated on top, unop below, control is
randomly selected limb

clearvars choice l ii Parameters ContStats ContZeros PreStats PreZeros PostStats
PostZeros
```

```
% Paste into SPSS
ForSPSS =
vertcat(Contdata,Predata{1,1},Predata{1,2},Predata{1,3},Postdata{1,1},Postdata{1,2},
Postdata{1,3});
NoContPrePost =
horzcat(size(Contdata,1),size(Predata{1,1},1),size(Predata{1,2},1),size(Predata{1,3},
1),...
 size(Postdata{1,1},1),size(Postdata{1,2},1),size(Postdata{1,3},1));

b1 = [];
b2 = [];
b3 = [];
for ii = 1:length(NoContPrePost)
 if ii == 1
  a(1:NoContPrePost(ii)) = 7;
  b1 = horzcat(b1,a);
  b2 = horzcat(b2,a);
  b3 = horzcat(b3,a);
  clearvars a
 else
  a(1:NoContPrePost(ii)) = ii - 1;
  b1 = horzcat(b1,a);

  if ii == 2 || ii == 3 || ii == 4 %create pre/post
   a(1:NoContPrePost(ii)) = 0; %pre op
   b2 = horzcat(b2,a);
   clearvars a
  else
   a(1:NoContPrePost(ii)) = 1;% post op
   b2 = horzcat(b2,a);
   clearvars a
  end

  if ii == 2 || ii == 5 %implant groups
   a(1:NoContPrePost(ii)) = 1; %implant 1
   b3 = horzcat(b3,a);
   clearvars a
  elseif ii == 3 || ii == 6
   a(1:NoContPrePost(ii)) = 2; %implant 2
   b3 = horzcat(b3,a);
   clearvars a
  elseif ii == 4 || ii == 7
   a(1:NoContPrePost(ii)) = 3; %implant 3
   b3 = horzcat(b3,a);
   clearvars a
  end
  clearvars a
 end
end
b1 = b1';
b2 = b2';
b3 = b3';
ForSPSS = horzcat(b1,b2,b3,ForSPSS);
clearvars k o b1 ii NoContPrePost b2 b3
```

```
% set up
newpath = 'H:\Raw Data\NexusOutput\BBraunMasterMotionCaptureDatabase\';
load([newpath 'RawImplantData.mat'])

Sides = {'Left', 'Right', 'Operated', 'Unoperated'}; %l
ImplantGroup = {'Implant1','Implant2','Implant3'}; %s %for o = 2%:3

dir =
{'LevelWalking','StairsAscent','StairsDescent','Incline1','Incline3','Incline4','Incl
ine6','Incline7'};

[s,~] = listdlg('PromptString','Trial type?',...
```

```
 'SelectionMode','single',...
 'ListString',dir);

splitStr{7} = dir{s};

clearvars s t dir
```

```
% extract
ContStats = {0};
Predata = {0 0 0};
Postdata = {0 0 0};

if strcmp(splitStr{7},'LevelWalking') || strcmp(splitStr{7},'StairsAscent') ||
strcmp(splitStr{7},'StairsDescent')
 ContStats = num2cell(table2array(RawImplantData.(splitStr{7}).Control.RawSPT)');
 for ii = 1:3
  Predata{1,ii} =
table2array(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).RawSPT(1:10,3:end))
';
  Postdata{1,ii} =
table2array(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).RawSPT(1:10,3:
end))';
 end
else
 ContStats1 = table2array(RawImplantData.(splitStr{7}).Control.RawSPT);
 ContStats =
num2cell(vertcat(ContStats1(1,:),table2array(RawImplantData.(splitStr{7}).Control.Bel
tSpeed),ContStats1(2:7,:)))';
 for ii = 1:3
  Predata1 =
table2array(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}).RawSPT(1:7,3:end));
  Postdata1 =
table2array(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGroup{ii}).RawSPT(1:7,3:e
nd));
  Predata{1,ii} =
vertcat(Predata1(1,:),table2cell(RawImplantData.(splitStr{7}).PreOp.(ImplantGroup{ii}
).BeltSpeed),Predata1(2:7,:))';
  Postdata{1,ii} =
vertcat(Postdata1(1,:),table2cell(RawImplantData.(splitStr{7}).YearPostOp.(ImplantGro
up{ii}).BeltSpeed),Postdata1(2:7,:))';
 end
end
clearvars ContStats1 Predata1 Postdata1
```

```
% for SPSS
ForSPSS =
vertcat(ContStats,Predata{1,1},Predata{1,2},Predata{1,3},Postdata{1,1},Postdata{1,2},
Postdata{1,3});
NoContPrePost =
horzcat(size(ContStats,1),size(Predata{1,1},1),size(Predata{1,2},1),size(Predata{1,3}
,1),...
 size(Postdata{1,1},1),size(Postdata{1,2},1),size(Postdata{1,3},1));

b1 = [];
b2 = [];
b3 = [];
for ii = 1:length(NoContPrePost)
 if ii == 1
  a(1:NoContPrePost(ii)) = 7;
  b1 = horzcat(b1,a);
  b2 = horzcat(b2,a);
  b3 = horzcat(b3,a);
  clearvars a
 else
  a(1:NoContPrePost(ii)) = ii - 1;
  b1 = horzcat(b1,a);

  if ii == 2 || ii == 3 || ii == 4 %create pre/post
   a(1:NoContPrePost(ii)) = 0; %pre op
   b2 = horzcat(b2,a);
   clearvars a
  else
   a(1:NoContPrePost(ii)) = 1;% post op
   b2 = horzcat(b2,a);
   clearvars a
  end

  if ii == 2 || ii == 5 %implant groups
   a(1:NoContPrePost(ii)) = 1; %implant 1
   b3 orzcat(b3,a);
   clearvars a
  elseif ii == 3 || ii == 6
   a(1:NoContPrePost(ii)) = 2; %implant 2
```

```
    b3 = horzcat(b3,a);
    clearvars a
   elseif ii == 4 || ii == 7
    a(1:NoContPrePost(ii)) = 3; %implant 3
    b3 = horzcat(b3,a);
    clearvars a
   end
   clearvars a
  end
end
b1 = num2cell(b1');
b2 = num2cell(b2');
b3 = num2cell(b3');
ForSPSS = horzcat(b1,b2,b3,ForSPSS);

clearvars b1 b2 b3 ContStats ii ImplantGroup NoContPrePost Postdata Predata
```

9.4.6. Plot 3D Graphs

```
% Script to plot 3d kinematics or kinetics of all knee implant groups and
% controls. Boxes allow for selections to be made - may have to manually update graph
% axes labels so these are correct for the data being plotted. C. Govind.

clc
clear
```

```
% set up
newpath = 'H:\Raw Data\NexusOutput\BBraunMasterMotionCaptureDatabase\';
load([newpath 'AvImplant.mat'])
```

```
% Select plot
Sides = {'Left', 'Right', 'Operated', 'Unoperated'};
ImplantGroup = {'Implant1','Implant2','Implant3'};
Joint = {'Pelvis','Hip', 'Knee','Ankle'};
Parameters = {'Angles','Moment','Power'};
dir =
{'LevelWalking','StairsAscent','StairsDescent','Incline1','Incline3','Incline4','Incl
ine6','Incline7'};

[s,~] = listdlg('PromptString','Trial type?',...
 'SelectionMode','single',...
 'ListString',dir);

[t,~] = listdlg('PromptString','Parameter?',...
 'SelectionMode','single',...
 'ListString',Parameters);

if t == 1
 ToPlot = {'AnkleFlexion'}; %enter kinematics
elseif t == 2
 ToPlot = {'KneeAbAd','HipFlexion'}; %enter moment
elseif t == 3
 ToPlot = {'AnklePower','AnklePower'}; %enter power
end

MeasureParam = Parameters{t};
splitStr{7} = dir{s};
```

```
% plot single
gait = (AvImplant.LevelWalking.Control.Angles.KneeFlexion(:,1))'; %FILL IN - TO PLOT
stdevno = (AvImplant.LevelWalking.Control.Angles.KneeFlexion(:,2)*2)';%FILL IN - %
Standard deviation * 2

x = linspace (0,100,101);
plot (x,gait)

hold on
fill ([x fliplr(x)],[gait-stdevno fliplr(gait+stdevno)],'k');
xlabel('% Gait Cycle','FontWeight','bold','Color',[0.5 0.5 0.5]);
ylabel('Ext  Deg  Flex','FontSize',10,'Color',[0.5 0.5 0.5]);
alpha(0.15)
axis square
hold off
```

```
% set up graph
scrsz = get(groot,'ScreenSize');
figure('Position',[100 scrsz(4)/6 scrsz(3)/1.7 scrsz(4)/1.1])

XTicks = [1 10 20 30 40 50 60 70 80 90 100];
XTicksLabel = {'' '10' '' '30' '' '50' '' '70' '' '90' ''};

x = linspace (0,100,101);
xlim = [0 101];
if strcmp(MeasureParam,'Angles')
 plotTitles = {'CR DD','UC','UCR'};
 spfh = gobjects(1,3); %preallocate
 for i = 1:3
  spfh(i) = subplot(1,3,i);
  set(spfh(i), 'XTick', XTicks, 'XTickLabel', XTicksLabel, 'NextPlot', 'add',
'xlim',xlim);
  title(plotTitles(i), 'FontWeight', 'bold', 'Color', [0.1 0.1 0.1]);
  plot(spfh(i),[0 100],[0 0],'Color',[0 0 0]);
  xlabel('% Gait Cycle','FontWeight','bold','Color',[0.5 0.5 0.5]);
  axis square
 end
 ylabel(spfh(1),'Ext  Deg  Flex','FontSize',10,'Color',[0.5 0.5 0.5]);

else %kinetics
 plotTitles = {'CR DD','UC','UCR','CR DD','UC','UCR'};
 spfh = gobjects(2,3); %preallocate
 for i = 1:6
  spfh(i) = subplot(2,3,i);
  set(spfh(i), 'XTick', XTicks, 'XTickLabel', XTicksLabel, 'NextPlot', 'add',
'xlim',xlim);
  title(plotTitles(i), 'FontWeight', 'bold', 'Color', [0.1 0.1 0.1]);
  plot(spfh(i),[0 100],[0 0],'Color',[0 0 0]);
  xlabel('% Gait Cycle','FontWeight','bold','Color',[0.5 0.5 0.5]);
  axis square
 end
 if strcmp(MeasureParam,'Moment')
%   ylabel(spfh(1),'Ext  N.m/kg  Flex','FontSize',10,'Color',[0.5 0.5 0.5]);
  ylabel(spfh(1),'Abd  N.m/kg  Add','FontSize',10,'Color',[0.5 0.5 0.5]);
 elseif strcmp(MeasureParam,'Power')
  ylabel(spfh(1),'Ecc  W/kg  Con','FontSize',10,'Color',[0.5 0.5 0.5]);
  ylabel(spfh(4),'Ecc  W/kg  Con','FontSize',10,'Color',[0.5 0.5 0.5]);
 end
end

clearvars plotTitles XTicks XTicksLabel
```

```
% plot each implant + control as shaded stdev
% toe off
PreTO =
horzcat(AvImplant.(splitStr{7}).PreOp.Implant1.OToeOff(1),AvImplant.(splitStr{7}).Pre
Op.Implant2.OToeOff(1),AvImplant.(splitStr{7}).PreOp.Implant3.OToeOff(1));
PostTO = horzcat(AvImplant.(splitStr{7}).YearPostOp.Implant1.OToeOff(1),
AvImplant.(splitStr{7}).YearPostOp.Implant2.OToeOff(1),
AvImplant.(splitStr{7}).YearPostOp.Implant3.OToeOff(1));
TO3 = AvImplant.(splitStr{7}).Control.ToeOff;

if strcmp(MeasureParam,'Angles')
 for i = 1:3 %implant
  spfh(i)= subplot(1,3,i);
  yPre = AvImplant.(splitStr{7}).PreOp.(ImplantGroup{i}).(MeasureParam).(['O'
ToPlot{1}]); %preop
  yPost = AvImplant.(splitStr{7}).YearPostOp.(ImplantGroup{i}).(MeasureParam).(['O'
ToPlot{1}]); %postop
  yCont = AvImplant.(splitStr{7}).Control.(MeasureParam).(ToPlot{1}); % control
  yPreMean = yPre(:,1)'; %mean angles
  yPreStDev = (yPre(:,2))'*2; %STDEV of means
  yPostMean = yPost(:,1)'; %mean angles
  yPostStDev = (yPost(:,2))'*2; %STDEV of means
  yContMean = yCont(:,1)'; %mean angles
  yContStDev = (yCont(:,2))'*2; %STDEV of means

  plot(spfh(i),yPreMean,'r')
  plot(spfh(i),yPostMean,'b')
  subplot (1,3,i)
  fill ([x fliplr(x)],[yPreMean-yPreStDev fliplr(yPreMean+yPreStDev)],'r:');
  fill ([x fliplr(x)],[yPostMean-yPostStDev fliplr(yPostMean+yPostStDev)],'b:');
  fill ([x fliplr(x)],[yContMean-yContStDev fliplr(yContMean+yContStDev)],'k');
  alpha(0.15)
  allYLim = get(spfh, {'YLim'});
  allYLim = cat(2, allYLim{:});
  set(spfh, 'YLim', [min(allYLim), max(allYLim)]);
  clearvars AxesHandle allYLim
 end
```

```
 %plot toe off as vertical line
 for i = 1:3
  subplot (1,3,i)
  line([PreTO(i) PreTO(i)],get(spfh(i),'YLim'),'Color',[1 0 0]); %plot pre toe off as
vertical red line
  line([PostTO(i) PostTO(i)],get(spfh(i),'YLim'),'Color',[0 0 1]); %plot post toe off
as vertical blue line
  line([TO3(1) TO3(1)],get(spfh(i),'YLim'),'Color',[0 0 0]); %plot control toe off as
vertical black line
 end

else %moment or power
 for l = 1:2
  for i = 1:3 %implant
   yPre = AvImplant.(splitStr{7}).PreOp.(ImplantGroup{i}).(MeasureParam).(['O'
ToPlot{l}]); %preop
   yPost = AvImplant.(splitStr{7}).YearPostOp.(ImplantGroup{i}).(MeasureParam).(['O'
ToPlot{l}]); %postop
   yCont = AvImplant.(splitStr{7}).Control.(MeasureParam).(ToPlot{l}); % control
   if size(yPre,2) == 1
    tomerge = zeros (101,1);
    yPre = horzcat(yPre,tomerge);
   end
   yPreMean = yPre(:,1)'; %mean angles
   yPreStDev = (yPre(:,2))'*2; %STDEV of means
   yPostMean = yPost(:,1)'; %mean angles
   yPostStDev = (yPost(:,2))'*2; %STDEV of means
   yContMean = yCont(:,1)'; %mean angles
   yContStDev = (yCont(:,2))'*2; %STDEV of means

   switch l
    case 1
     spfh(i) = subplot(2,3,i);
     plot(spfh(i),yPreMean,'r')
     plot(spfh(i),yPostMean,'b')
     subplot (2,3,i)
     %      if i == 1 || i == 3 %for stair descent
     fill ([x fliplr(x)],[yPreMean-yPreStDev fliplr(yPreMean+yPreStDev)],'r:');
     fill ([x fliplr(x)],[yPostMean-yPostStDev fliplr(yPostMean+yPostStDev)],'b:');
     fill ([x fliplr(x)],[yContMean-yContStDev fliplr(yContMean+yContStDev)],'k');
     alpha(0.15)
     %      else
     %      fill ([x fliplr(x)],[yPostMean-yPostStDev
fliplr(yPostMean+yPostStDev)],'b:');
     %      fill ([x fliplr(x)],[yContMean-yContStDev
fliplr(yContMean+yContStDev)],'k');
     %      alpha(0.15)
     %      axis square
     %      end
%      allYLim = [get(spfh, {'YLim'});
%      allYLim = cat(2, allYLim{:});
     set(spfh, 'YLim', [-0.5, 1])%[min(allYLim), max(allYLim)]);
     clearvars AxesHandle allYLim
    case 2
     spfh2(i) = subplot(2,3,i+3);
     plot(spfh(i+3),yPreMean,'r')
     plot(spfh(i+3),yPostMean,'b')
     subplot (2,3,i+3)
     fill ([x fliplr(x)],[yPreMean-yPreStDev fliplr(yPreMean+yPreStDev)],'r:');
     fill ([x fliplr(x)],[yPostMean-yPostStDev fliplr(yPostMean+yPostStDev)],'b:');
     fill ([x fliplr(x)],[yContMean-yContStDev fliplr(yContMean+yContStDev)],'k');
     alpha(0.15)
%      allYLim = get(spfh, {'YLim'});
%      allYLim = cat(2, allYLim{:});
%      set(spfh, 'YLim', [min(allYLim), max(allYLim)]);
     set(spfh, 'YLim', [-2, 4.5])
     clearvars AxesHandle allYLim
   end

  end
 end
 %plot toe off as vertical line
 for i = 1:6
  subplot (2,3,i)
  if i == 1 || i == 2 || i == 3
   line([PreTO(i) PreTO(i)],get(spfh(i),'YLim'),'Color',[1 0 0]); %plot pre toe off
as vertical red line
   line([PostTO(i) PostTO(i)],get(spfh(i),'YLim'),'Color',[0 0 1]); %plot post toe
off as vertical blue line
   line([TO3(1) TO3(1)],get(spfh(i),'YLim'),'Color',[0 0 0]); %plot control toe off
as vertical black line.
  else
```

```
    line([PreTO(i-3) PreTO(i-3)],get(spfh(i),'YLim'),'Color',[1 0 0]);
    line([PostTO(i-3) PostTO(i-3)],get(spfh(i),'YLim'),'Color',[0 0 1]);
    line([TO3(1) TO3(1)],get(spfh(i),'YLim'),'Color',[0 0 0]);
  end
 end
end

clearvars dir i l s PostTO PreTO t TO3 scrsz Sides x xlim yCont yContMean yContStDev
yPost yPostMean yPostStDev yPre yPreMean yPreStDev
```

Appendix 10: Stair Navigation Kinetic Data

10.1. Stair Ascent

Table 0-4: Peak joint moments during stair ascent of all participants.

Peak Joint Moment (N.m/kg)	CR DD		UC		UCR		Control (N = 8)
	Pre-Op (N = 8)	Post-Op (N = 9)	Pre-Op (N = 2)	Post-Op (N = 4)	Pre-Op (N = 3)	Post-Op (N = 3)	
Hip Flexion	0.60±0.29	**0.92±0.19††**	0.58±0.29	0.91±0.11	0.53±0.19	**1.08±0.16††**	0.69±0.32
Hip Extension	0.31±0.44	**0.14±0.13****	0.23±0.13	0.37±0.13	0.44±0.17	**0.07±0.06***	0.43±0.2
Early Stance KFM (1)	0.38±0.19	**0.28±0.16***	0.20±0.01	**0.1±0.08***	0.34±0.16	0.24±0.27	0.63±0.33
Mid-stance KEM (2)	0.11±0.28	**0.39±0.19**,†**	0.11±0.04	0.30±0.14	-0.07±0.09	0.28±0.24	0.09±0.07
Late Stance KFM (3)	0.14±0.23	**0.07±0.01****	0.02±0	**0.07±0.04***	0.11±0.08	0.10±0.07	0.16±0.06
Early Stance KAM (1)	0.39±0.12	0.39±0.19	0.29±0.1	0.30±0.12	0.43±0.09	0.31±0.06	0.59±0.17
Mid-stance KAM (2)	0.19±0.07	0.09±0.17	0.19±0.14	0.14±0.16	0.24±0.19	**0.03±0.18†**	0.12±0.09
Late stance KAM (3)	0.31±0.11	0.18±0.15	0.30±0.27	0.34±0.14	0.29±0.19	0.09±0.08	0.27±0.21
Knee Abduction	0.04±0.02	0.08±0.04	0.04±0.01	0.08±0.09	0.05±0.03	**0.18±0.09††**	0.10±0.07
Ankle Dorsiflexion	**1.22±0.18***	1.40±0.19	**1.07±0.31***	**1.49±0.3†**	1.09±0.2*	1.35±0.13	1.59±0.24
Ankle Plantarflexion	0.0±0.03	0.02±0.04	0.01±0.04	0.04±0.11	0.02±0.02	0.0±0.03	0.04±0.04

Table 0-5: Peak joint powers during stair ascent of all participants.

Peak Joint Power (W/kg)	CR DD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Concentric							
Hip	1.13±0.67	**1.67±0.38†**	1.12±0.51	1.72±0.4	0.68±0.22	**1.70±0.56†**	1.42±0.66
Knee (Heel Strike)	**1.04±0.3****	0.90±0.45	**0.58±0.02***	0.43±0.29	**0.82±0.17***	0.82±0.27	2.01±0.72
Knee (Toe Off)	**0.46±0.13*****	**0.72±0.2†**	**0.22±0.01*****	**0.78±0.47†**	**0.43±0.12*****	0.51±0.25	1.18±0.23
Ankle	**2.26±0.71****	2.62±0.48	**1.85±0.74***	**3.35±1.26†**	**2.16±0.45***	2.56±0.82	3.74±0.89
Eccentric							
Hip	0.22±0.24	0.08±0.13	0.07±0.09	0.24±0.25	0.10±0.08	0.09±0.08	0.17±0.18
Knee	**0.14±0.1***	**0.33±0.19†**	0.27±0.09	0.45±0.26	0.14±0.10	0.26±0.15	0.34±0.16
Ankle	0.25±0.11	0.13±0.14	0.28±0.06	0.42±0.39	0.08±0.20	0.22±0.12	0.42±0.42

10.2. Stair Descent

Table 0-6: Peak joint moments during stair descent of all participants.

Peak Joint Moment (N.m/kg)	CR DD		UC		UCR		Control
	Pre-Op (N = 8)	Post-Op (N = 3)	Pre-Op (N = 1)	Post-Op (N = 3)	Pre-Op (N = 3)	Post-Op (N = 2)	(N = 7)
Hip Flexion	0.17±0.14	0.41±0.12	0.40±0.0	0.31±0.08	0.08±0.20	0.28±0.20	0.11±0.15
Hip Extension	0.29±0.30	0.14±0.03	0.25±0.0	0.21±0.06	0.42±0.28	0.32±0.28	0.39±0.19
Early Stance KFM (1)	0.32±0.22	0.06±0.11	0.07±0.0	0.02±0.11	0.41±0.13	0.42±0.12	0.56±0.32
Mid-stance KEM (2)	-0.14±0.22	**0.12±0.26****	0.0±0.0	**0.09±0.11***	-0.2±0.19	-0.28±0.02	-0.23±0.23
Late Stance KFM (3)	0.47±0.19	0.36±0.12	0.42±0.0	0.24±0.03	0.55±0.09	0.58±0.13	0.66±0.30
Early Stance KAM (1)	0.46±0.26	0.34±0.06	0.34±0.0	0.38±0.07	0.57±0.09	0.42±0.09	0.60±0.29
Mid-stance KAM (2)	0.18±0.13	0.20±0.10	0.13±0.0	0.14±0.07	0.31±0.07	0.20±0.05	0.23±0.12
Late stance KAM (3)	0.32±0.13	0.31±0.11	0.30±0.0	0.34±0.11	0.44±0.04	0.36±0.12	0.47±0.23
Knee Abduction	0.05±0.06	0.06±0.10	0.02±0.0	0.06±0.07	0.03±0.03	0.10±0.12	0.03±0.03
Ankle Dorsiflexion	**1.07±0.18***	1.39±0.19	0.85±0.0	**1.36±0.13†**	1.01±0.18	1.00±0.22	1.35±0.16
Ankle Plantarflexion	0.01±0.03	0.0±0.05	0.02±0.0	0.01±0.03	0.0±0.01	-0.06±0.08	-0.04±0.03

Table 0-7: Peak joint powers during stair descent of all participants.

Peak Joint Power (W/kg)	CR DD		UC		UCR		Control
	Pre-Op	Post-Op	Pre-Op	Post-Op	Pre-Op	Post-Op	
Concentric							
Hip	**0.25±0.13***	0.18±0.05	0.21±0.0	0.27±0.38	0.29±0.04	0.23±0.07	0.73±0.40
Knee	0.12±0.16	0.15±0.12	0.06±0.0	0.27±0.21	0.12±0.03	0.19±0.05	0.27±0.50
Ankle	**0.83±0.55***	0.85±0.42	0.83±0.0	0.93±0.45	0.80±0.25	0.56±0.05	1.61±0.44
Eccentric							
Hip	0.45±0.32	0.45±0.28	0.48±0.0	0.73±0.47	0.38±0.22	0.53±0.21	0.05±0.18
Knee	**1.66±0.78***	**1.04±0.27***	0.93±0.0	**1.27±0.82***	1.94±0.25	1.75±0.16	3.10±0.89
Ankle	**1.65±0.71****	2.76±1.45	1.31±0.0	3.24±0.22	2.24±0.72	2.14±0.47	3.89±1.39