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Kinematics in Total Knee Arthroplasty

Allan Roy Sekeitto, *The University of Western Ontario*

Supervisor: Lanting, Brent, *The University of Western Ontario*
Co-Supervisor: Willing, Ryan, *The University of Western Ontario*

A thesis submitted in partial fulfillment of the requirements for the Master of Science degree in Surgery

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Abstract

Total knee arthroplasty (TKA) continues to be a successful procedure for the past 50 years. We investigated the kinematics after under- and overstuffing the joint space with the polyethylene (PE) insert in both cruciate-retaining (CR) TKA and posterior-stabilized (PS) TKA. We compared CR and PS designs to evaluate their respective kinematic differences. This study employed a hybrid computational-experimental joint motion simulation on a 6 degrees of freedom joint motion simulator. Physical prototypes of a virtually performed TKA based on cadaveric CT scans and a virtual ligament model were utilized. We demonstrated understuffing decreases stability and the joint compressive forces, with the inverse occurring with overstuffing. In isolation, understuffing did not result in instability during activities of daily living in CR-TKA. Notably, a 2 mm increase or decrease in PE thickness altered the post-cam mechanism engagement. The PS design demonstrated greater stability, but overall, the kinematics between CR and PS designs were similar. This study demonstrates the ability of a hybrid model to further the understanding of kinematics in TKA.

Keywords

Total knee arthroplasty, kinematics, Cruciate-retaining, Posterior-stabilized, post-cam mechanism, Instability, Activities of daily living, Joint motion simulator, Virtual ligament model, Hybrid computational-experimental model.

Summary for Lay Audience

Total knee arthroplasty (TKA) continues to be a successful procedure for the past 50 years in treating patients with osteoarthritis of the knee. It entails resurfacing the joint surfaces of the knee with implants and a polyethylene (PE) insert which lies between them. We investigated the kinematics after under- and overstuffing the tibiofemoral joint space of the knee with the PE insert in both cruciate-retaining (CR) and posterior-stabilized (PS) designs of TKA. The under- and overstuffed joint spaces were simulated by 2mm with respective to a reference joint space. The CR design retains the posterior cruciate ligament (PCL), and the PS design substitutes the PCL with a post cam mechanism. We compared CR and PS designs to evaluate their respective kinematic differences. This study employed a hybrid computational-experimental joint motion simulation on a 6 degrees of freedom joint motion simulator machine. Physical prototypes of a virtually performed TKA based on cadaveric CT scans were mounted on to the joint motion simulator. A virtual ligament model developed from the literature was utilized and virtually installed around the TKA on the joint motion simulator. We demonstrated understuffing decreases stability by reducing the soft tissue tension and the joint compressive forces, with the inverse occurring with overstuffing the joint space with the PE insert. In isolation, understuffing did not result in instability during activities of daily living in CR-TKA. Notably, a 2 mm increase or decrease in PE thickness altered the post-cam mechanism engagement in PS-TKA. The PS design demonstrated greater stability, but overall, the kinematics between CR and PS designs were similar. This study demonstrates the ability of *in vitro* hybrid models to further the understanding of kinematics in TKA.

Co-Authorship Statement

This study was performed in collaboration with the Departments of Orthopaedic Surgery and Mechanical & Materials Engineering members. The study was conducted under the supervision of Dr Brent Lanting MD, MSc, FRCSC and Dr Ryan Willing PhD, B.Sc.E as the co-supervisor. This study was conducted based on the preliminary work performed by Dr Jance McGale MD, MSc, FRCSC and Mr. Liam Montgomery BSc in their respective masters in developing the hybrid computational-experimental model and methodologies. I formulated the study question and developed the study design, which included the study topics and testing parameters. I was responsible for performing the various simulations, data collection, subsequent data analysis, manuscript writing and the final preparation of the thesis. I authored this original thesis under the guidance and supervision of Dr Brent Lanting and Dr Ryan Willing.

Acknowledgments

I would like to acknowledge my master's supervisors Dr Brent Lanting, for his guidance on this study and for guiding me on balance between the clinical and basic science spheres in research. Dr Ryan Willing for his patience and guidance in enlightening my understanding of the hybrid computational-experimental model, the simulations and sharing his analytical thought process. Collectively they were both pivotal in my growth as a student, academic and surgeon.

Secondly, Dr Jance McGale and Mr. Liam Montgomery who were always available to answer my clarifying questions on their preliminary work. All the educators who endeavoured to empower us throughout the master's program this year. This was no mean feat given the ongoing COVID pandemic and its impact on our learning.

Lastly, I would like to thank my wife Boipelo, family and friends who supported my decision to pursue this master's degree.

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Abbreviations

6 DoF	6 degrees of freedom
ADL	Activities of daily living
AP	Anteroposterior
AS	Anterior - stabilized
CR	Cruciate - retaining
CS	Condylar - stabilized
IE	Internal/external rotation
IPE	Isolated polyethylene insert exchange
JCFs	Joint compressive forces
KA	Kinematic alignment
LCL	Lateral collateral ligament
MA	Mechanical alignment
PCL	Posterior cruciate ligament
PE	Polyethylene
PS	Posterior-stabilized
RCTs	Randomised controlled trials
ROM	Range of motion
SD	Standard deviation
sMCL	Superficial medical collateral ligament
TKA	Total knee arthroplasty
VV	Varus/valgus

Chapter 1

1.1 Introduction

Total knee arthroplasty (TKA) is a surgical procedure with a success story spanning almost 50 years.(1) The primary goal of TKA is to provide a painless stable knee with an appropriate range of motion (ROM) and good function. TKA has provided pain relief and functional restoration in patients with advanced osteoarthritis of the knee (*Figure 1.1&1.2*).(2, 3) It has been reported that up to 43% of patients forget they underwent TKA.(3) In the past decade, the survivorship, kinematics and patient-reported outcomes in TKA have improved.(4)

Despite all the advances in TKA in the last five decades, there remains a significant amount of people dissatisfied with their results.(5, 6) The literature reports that up to 20-25% of patients who underwent surgery for a TKA remain dissatisfied with their results.(7) Patients with residual pain are dissatisfied, have a lower quality of life and require higher health care resources.(8) The demand for TKA is increasing.(9) If the incidence of revision surgeries stays the same, this will lead to a greater number of revision surgeries being performed. However, research efforts have attempted to eliminate or decrease the main reasons for revision surgery. This would potentially lead to better patient outcomes, quality of life, and decreased revision surgery burden. Instability is currently one of the top three most common reasons for TKA revision.(10, 11) Approximately one-third of early TKA failures are due to soft tissue imbalances.(12, 13) Soft tissue balancing is subjective to the surgeons feel in most cases, and it is unclear what is optimum for the soft tissue balance, tension, bone cuts, polyethylene (PE) sizing, alignment or how these factors interact with each other to obtain favourable outcomes.(14)

Recent advances in joint motion simulator technology have enabled these machines to analyze TKA mechanics while simulating different alignments and soft tissue balancing by changing the properties of parametric virtual ligaments.(15) The work of this thesis will be to further understand TKA kinematics using *in vitro* analysis with a joint motion simulator linked to a virtual ligament model. This hybrid computational-experimental study explored specific contemporary topics in TKA. The following topics were investigated from a kinematic perspective; prosthesis controversy between the cruciate-retaining (CR) and posterior-stabilized (PS) designs, TKA alignment philosophies, PE insert sizing and instability. We introduced the posterior tibial slope adjustments as a possible area of research focus for future follow-on studies.

Figure 1.1: A preoperative anteroposterior x-ray demonstrating bilateral knee osteoarthritis.

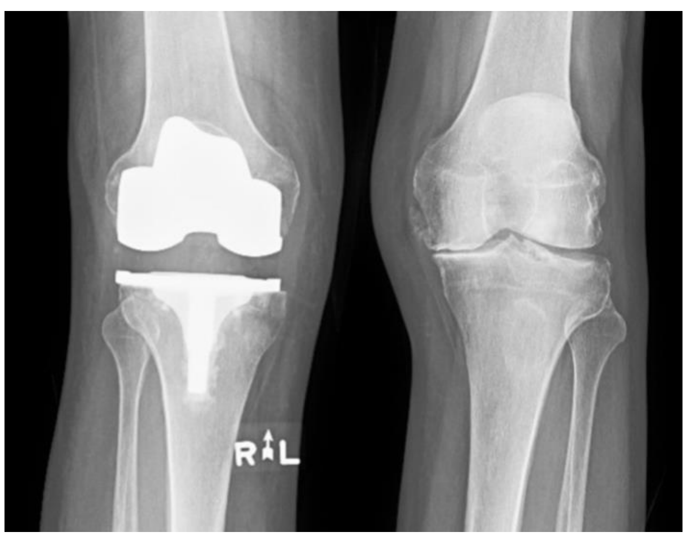

Figure1.2: A postoperative anteroposterior x-ray after a right total knee arthroplasty.

1.2 Total Knee Arthroplasty Design

The knee prosthesis used in TKA should provide maximal ROM and stability.(4) TKA prostheses designs have come close to mimicking the kinematics of the native knee(9) with newer alignment strategies and implant designs. However, knee kinematics after TKA is still not the physiological kinematics of the native knee.(16)

The 1970s was the founding decade with the development of modern TKA, and this condylar knee design entailed resurfacing the entire tibiofemoral joint. In 1974 the Total Condylar Prosthesis (Zimmer, Warsaw, IN) (*Figure 1.3)* was introduced. This prosthesis was a posterior cruciate ligament (PCL) sacrificing and not substituting design. This was the first true globally utilized TKA design. Early anatomic TKA designs preserved both the cruciate ligaments. However, their failures were linked to; inadequate tibial fixation, thin PE inserts and instability.(17) In the early 1970s, the first uncemented cruciate-sparing TKA was implanted by Dr Yamamoto and Professor T. Kodma. Dr Charles Townley led designing the first cemented condylar CR TKA during the same period, which was later manufactured by Depuy (Warsaw, IN).(17) By 1978, the introduction of the Insall Burstein PS prosthesis with the first post-cam mechanism was driven by the need to improve ROM and prevent posterior tibial translation during flexion.(18)

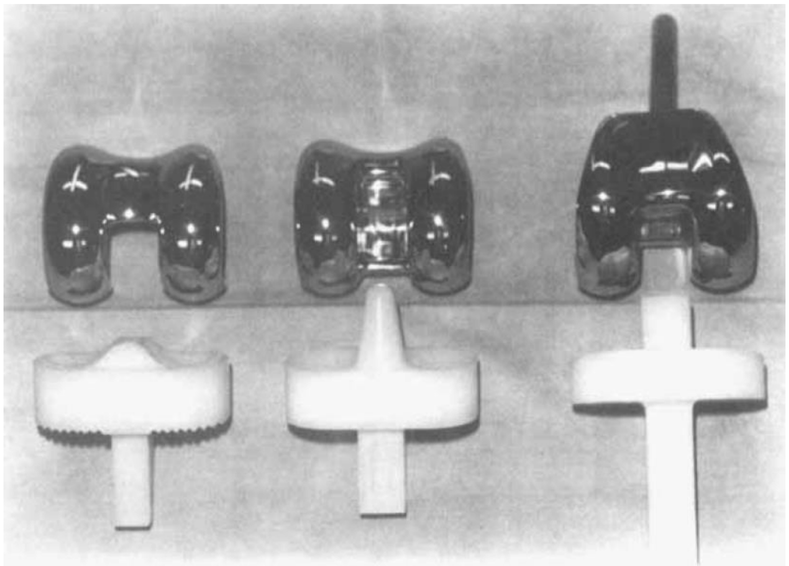

Figure 1.3: From left to right, the prostheses are the total condylar, total condylar II and total condylar III. Reproduced with permission, Insall et al.(19)

Primary TKA prostheses designs vary depending on their respective philosophies. Bicruciate-retaining designed TKA retains both the anterior and posterior cruciate ligaments. CR designed TKA retains the PCL and sacrifices the anterior collateral ligament (ACL); PE insert geometry is shown in *Figure 1.4*. PS designed TKA sacrifices both cruciate ligaments and replaces the PCL functionality with a post-cam mechanism; the PE insert geometry is shown in *Figure 1.5*. The medial pivot design sacrifices both cruciate ligaments. It has a deep constrained but congruent medial design and non-congruent lateral design to allow for; lateral translation and a medial pivot motion to mimic the native knee.(16) Lastly, a conforming ultra-congruent medial and lateral surface design sacrifices both the cruciate ligaments. This articular philosophy comprises a range of implants and includes several names in the market and includes a range of implant designs specific to each manufacturer: highly congruent, deep-dish bearing, bicruciate-stabilized, condylar-stabilized (CS) or anterior-stabilized (AS). (20-22). The CS PE insert is shown in *Figure 1.6*.

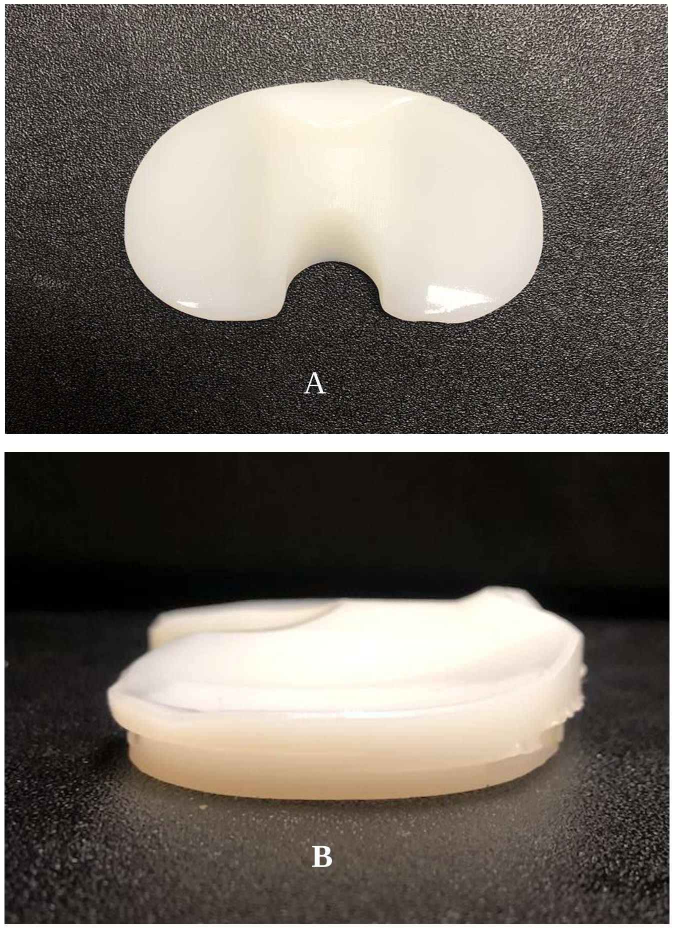

Figure 1.4: Cruciate-retaining polyethylene insert. A- side profile, B- superior view.

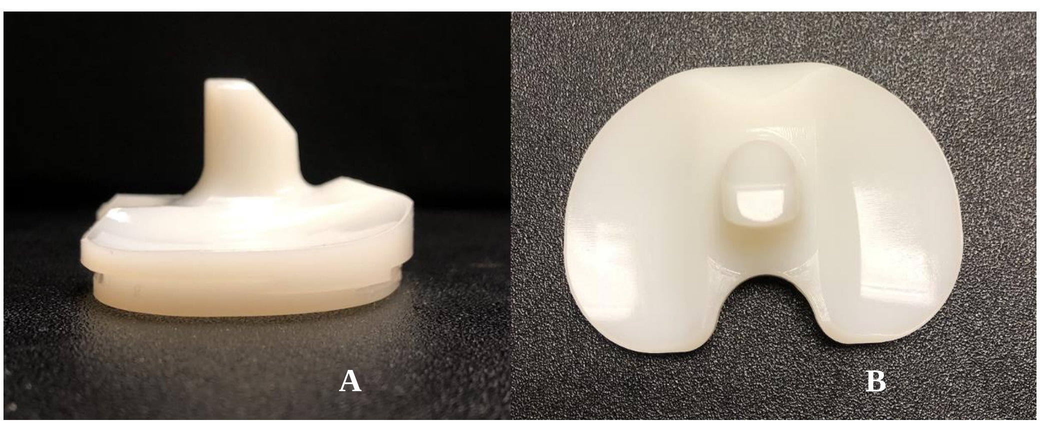

Figure 1.5: Posterior-stabilized polyethylene insert. The post projects perpendicular from the articulating surface. A- side profile, B- superior view

In CR-TKA the function of the PCL is to stabilize the knee and in flexion to keep the femur in a posterior position to aid femoral rollback and therefore deepen flexion of the knee.(23) It also prevents paradoxical anterior translation of the femur relative to the tibia during flexion.(24) The PS-TKA works through a post-cam mechanism. As the knee flexes, the transverse cam on the femoral component engages with the posterior aspect of the post on the PE insert to direct the femur posteriorly and initiate femoral rollback. The CS design prevents anterior translation of the femoral component due to the prominent anterior lip. It functions as a posterior-stabilizing insert while sparing the femoral bone lost with PS-TKA required for the box that houses the post-cam mechanism.

Figure 1.6: Condylar-stabilized polyethylene insert. Note the prominent anterior lip.

The CR and PS are both popular implant design choices amongst orthopaedic surgeons who perform TKA. Historically, the decision making in utilizing either a CR or PS design has been debated.(25) CR-TKA represents the minimum constraint in TKA. Although early literature discouraged using a CR prosthesis in the setting of rheumatoid arthritis, significant deformities and a previous patellectomy, this thinking has challenged in modern literature. (26-30) Archibeck et al.(27), in a prospective study, followed up 72 CR-TKAs in patients with rheumatoid arthritis and found satisfactory clinical and radiological results at follow up (mean 10.5 years). Ünkar et al.(30) retrospectively reviewed 112 TKAs with a mean preoperative mechanical tibiofemoral angle of 20.1^{o} varus. The CR and PS groups consisted of 58 and 54 TKAs, respectively. At a mean follow up of 56.6 months, the two groups had comparable outcomes. Abdel et al.(31) performed a retrospective comparative study between CR- and PS-TKA. Their subgroup analysis combined knees with preoperative flexion and angular deformities of >15° into a deformity group. The 20-year survival in 308 CR-TKAs was 89.8% (95% CI, 85.7% to 94.1%) compared to 56 PS-TKAs of 70.5% (95% CI, 55.3% to 89.8%). Reinhardt et al.(29) evaluated the survivorship and functional outcomes in 23 CR-TKAs with a previous patellectomy. The reported aseptic loosening-free survival was 100% at 5 and 10 years. Survival with revision for any reason as the outcome was 96% at five years (95% confidence interval [CI], 87.7%-100%) and 84% at ten years (95% CI, 69.5%-100%) and the Knee Society scores improved from 36+/-13 preoperatively to 92+/-9.6 at follow up. This represents the ever-evolving landscape in TKA and the application of the various implant designs. Contraindications for CR-TKA are PCL insufficiency, posterolateral instability, bone loss requiring augments and major revision procedures.(32)

In selecting to perform CR-TKA, the status of the PCL should be assessed preoperatively and intraoperatively. Preoperatively during the physical examination, a posterior sag sign is observed in PCL insufficiency. Intraoperatively excessive femoral roll back is observed in the setting of a PCL contracture, resulting in a limitation of full knee flexion. A PCL recession or “PCL balancing” should be performed in the setting of a PCL contracture. However, this may result in a posterior sag or anteroposterior laxity. The CR design itself provides no additional coronal stability. A manufacturer indicated their CR design was limited to 15° internal and external rotation.(33) Intraoperatively, if the quality of the PCL is questionable on inspection or with tension, the decision can be made to switch to a PS-TKA.(32) In a PCL-substituting TKA resection of the PCL in cadaveric studies has shown to increase the flexion gap by 1.8 mm to 4.8 mm.(34-37) Increasing the distal femur cut by 2 mm has been recommended to match the extension gap to the flexion gap.(37) This is to achieve adequate coronal balance in flexion and extension in PS-TKA and is common practice for many orthopaedic surgeons.

In PS-TKA, the PCL is sacrificed and therefore, there is no need to evaluate its status. This makes the PS design a favourably alternative to the CR design in clinical scenarios where the PCL is absent or deficient. The post-cam mechanism substitutes the PCL. The post projects from the PE insert *Figure 1.5*, and the cam is added to the femoral component. An additional bone resection creating an intercondylar box on the femoral side is required to house this mechanism. Historically the incidence regarding intraoperative fractures related to the preparation of the intercondylar box has been debated.(38, 39) The post-cam mechanism has been reported to engage at around 75^o of flexion. This engagement prevents anterior femoral translation and initiates the femoral rollback.(40, 41) PS-TKA has been shown to increase the sagittal constraint of the implanted knee, preventing posterior knee sag.(42) Although the design offers increased sagittal constraint, there is no increase in the coronal constraint compared to the CR design.(43) The PS design varies with the manufacturer permitting from 7.5 – 12^o of internal and external rotation.(33, 44) The post-cam mechanism does not remove the need to balance the knee. Balancing the flexion and extension gap remains vital to prevent the cam on the femoral side from "jumping" the post, i.e. post dislocation.(45) The PS design is often the minimum constraint required when dealing with bone defects while making use of augments and stems in difficult primary or revision knee arthroplasties.

The CS design is an alternative to the PS design. The prominent anterior lip provides additional stability but preventing paradoxical anterior femoral translation and posterior tibial translation during flexion. (20) When compared to the PS design, the CS design been shown not to produce the same anteroposterior stability.(46) In addition, the CS design has been found not to compensate for the lack of the PCL in an *in vitro* cadaveric study.(22) A five year follow up comparison has found the two designs comparable in clinical and functional outcomes.(47) A recent level 1 therapeutic study reported the CS design is inferior to the PS design in terms of posterior stability, flexion, and clinical scores at the two years follow up.(48)

Additional TKA designs are present in the market; mobile-bearing articulations, varus/valgus constrained and hinged designs. These are beyond the scope of this thesis and therefore have not been discussed.

1.3 Total Knee Arthroplasty Kinematics

The implanted knee, like the native knee, permits movement with 6 degrees of freedom (6 DoF): 3 rotations (flexion and extension, internal-external (IE) rotation, varus and valgus) and three translations (anteroposterior (AP), medial and lateral, compression and distraction).(49) Knee kinematics is defined as the analysis of motion patterns in native and prosthetic knees.(50) TKA kinematics has been studied *in vivo, in vitro* and *in silico*. *In vivo* studies are performed with patients implanted with a TKA, in *vitro* are performed in the laboratory setting and *in silico* with computational modeling.

1.3.1 Anteroposterior translation

The AP translation is of importance in the knee as the femur translates posteriorly on the tibia during flexion. This motion is referred to as femoral rollback. In the native knee, it starts as knee flexion is initiated and mostly occurs in the first 30^o of flexion.(51) In full extension, the tibia is externally rotated relative to the femur. The femur is located anterior to the midpoint of the tibia. As flexion is initiated, the lateral femoral condyle translates posteriorly, so too does the medial femoral condyle to a reduced magnitude compared to the lateral. Literature reports an average of 19 mm and 1.5 mm posterior translation of lateral and medial femoral condyles.(52) This difference in medial and lateral translation results in the femur externally rotating or the tibia being in an internally rotated position relative to the femur.

Femoral roll back prevents impingement of the femur on the tibia in deep flexion. Greater femoral roll back results in increased knee flexion.(53) Banks et al.(54) studied 121 knees assessed with fluoroscopy and shape matching to review knee kinematics in a weight-bearing deep flexion activity. They found a correlation between the posterior translation of the femur on the tibia and increased knee flexion. They found increased AP translation in the PS knees -13.5 mm ±2.5, compared to -6.0 mm ±4.9 in the CR knees ($p<0.001$). The overall linear regression demonstrated 1.4^o more flexion per 1 mm of femoral posterior translation ($R = 0.64$, $p < 0.001$). Cates et al.(55) evaluated the *in vivo* kinematics using fluoroscopy in 30 CR- and PS-TKAs. The mean AP translation was -4.9mm and -6.4mm in CR and PS knee on the lateral femoral condyle, respectively (P=.195). The mean AP translation on the medial femoral condyle was -1.0 mm and -4.2mm in the CR and PS knees, respectively (P=.028). Paradoxical anterior translation of the femur with flexion has been reported in CR-TKA.(56, 57) This movement occurs during flexion with the femur translating anterior on the tibia instead of posteriorly. Prostheses designs that control the position of the tibia and femur have demonstrated more femoral rollback. The soft tissue may influence rollback, translation of the femur, with increased laxity in the flexion space.(58)

Watanabe et al.(59), in their prospective study, reviewed 56 asymmetrical AS-TKAs with dynamic radiography at one-year post-surgery. There were 27 TKAs that retained the PCL, and in the remaining 29 TKAs, the PCL was sacrificed. The PCL-sacrificing TKAs demonstrated greater knee flexion during kneeling (122° versus 115°). They proposed this result was linked to contracted PCL in severely deformed knees likely were the cause of limited flexion in the PCL-retaining group. The cumulative data comparing CR-TKA to PS-TKA shows less anterior translation in PS-TKA.(50) AS-TKA has been performed in settings with or without a PCL. A recent study demonstrated that there is no influence on the post-operative clinical results in AS-TKA, whether the PCL is sacrificed or retained.(21) However, an *in vitro* cadaveric study has shown that the AS design without the PCL has reduced AP stability, therefore, does not fully compensate for the PCL. (22) The literature highlights the importance of AP translation kinematics in relation to the clinical performance of TKA.

1.3.2 Internal-external rotation

The AP translation is comparatively higher with the lateral femoral condyle compared to the medial femoral condyle in TKA. This results in axial rotation of the knee as the tibia internally rotates relative to the femur as the knee flexes. The external rotation of the tibia with extension is referred to as the screw home mechanism. Cates et al.(55) evaluated the *in vivo* kinematics using fluoroscopy, reported in CR and PS-TKA that from full extension to 90°, both designs underwent internal rotation of the tibia. The IE or axial rotation mean angles at full extension were; 0.1° and -1.0°, respectively. The mean axial rotation angle was 4.9° and 1.9° for the CR and PS knees at maximum flexion, respectively (P=.011). The internal rotation of the tibia with flexion has been demonstrated in other studies.(60, 61) However, reversal of this normal axial rotation has also been reported in the literature.(62) AS-TKA has been reported to demonstrate a more normal axial rotation pattern when compared to the native knee than other designs.(59) Overall, the magnitudes of axial rotation in the literature appear to be similar between the different TKA designs.(50)

1.3.3 Varus/Valgus

The varus-valgus (VV) kinematics are evaluated in the coronal plane. There is limited literature specifically reviewing the VV kinematics in TKA. TKA has been reported not to restore the normal VV kinematics of the native knee.(63) The VV kinematics are primarily influenced by the alignment philosophy and the soft tissue tension in TKA. The medial and lateral collateral ligaments are the primary constraints in the coronal plane. The medial collateral ligament (MCL) has a superficial and deep component. The deep MCL is routinely released during the approach and exposure of the knee during TKA and has limited clinical relevance in the implanted knee. The superficial MCL (sMCL) has anterior and posterior fibres. The anterior fibres of superficial MCL are tight in flexion. The posterior fibres of the superficial MCL are tight at full extension. The sMCL primarily resists a valgus force with maximum resistance at 30^{o} flexion. The sMCL secondarily resists internal rotation and anterior translation of the tibia. The lateral collateral ligament (LCL) is the primary restraint to a varus force, with maximum resistance at 30^{o} flexion. The laxity and tensions of these ligaments influence the VV kinematics in TKA.

Siston et al.(63) reported on the intraoperative passive kinematics with a surgical navigation system in PS-TKA. The alignment of the mechanical axis at 10^{o} of flexion ($0.9^{o} \pm 2.2^{o}$ valgus) was different ($p<0.02$) from the arthritic knees ($1.3^{o} \pm 2.2^{o}$ varus) ($p=0.11$). Between 10^{o} and 60^{o} of flexion, there was no specific pattern to the VV kinematics. However, in deeper flexion, the implanted knees were in varus. Hino et al.(64) in their comparative study of 20 CR-TKA and 20 PS-TKA, described five VV kinematic patterns through unloaded flexion using navigation. Pattern A with increasing varus through the flexion arc. Pattern B with increasing varus till roughly 80^{o} and tapering back towards 0^{o} with deeper flexion. Pattern C was neutral throughout the range of motion. Pattern D with increasing valgus till roughly 80^{o} and tapering back towards 0^{o} with deeper flexion. Pattern E with increasing valgus through the flexion arc. All these post-operative patterns were observed in CR knees. The type C pattern was predominant in the PS knees ($p=0.028$). The VV kinematics are not uniform in the implanted knee.

1.3.4 Activities of daily living

Activities of daily living (ADL) are activities performed by individuals daily. They represent walking, lunging, kneeling, squatting, sitting, stair ascent and stair descent etc. *(Figure 1.7)*. During gait on a level surface, knee flexion of 45- 55^{o} is required.(65) In performing stair activities, stair ascent and descent both require at least 85^{o} of knee flexion.(65) In comparing axial rotation, slight variation has been reported, with an average of 4 – 7^{o} during the stance phase and was found to increase during stair activity.(66) Banks et al.(67) compared kinematics in different TKA designs in stair-stepping, which consisted of single-leg step-up and down activity with the foot on a 25-cm step. They used fluoroscopy to study 213 knees in 173 patients with 25 implant designs within three groups; fixed- bearing PS, fixed-bearing CR and mobile bearing. They found similar IE rotation kinematics within all the groups. They found differences in AP translation with the majority (75%) of the PS-TKA group demonstrating a medial centre of rotation consistent with posterior femoral back with flexion. For the CR-TKAs, 63% of the group demonstrated a lateral centre of rotation consistent with anterior femoral translation with flexion. However they concluded that the motion in TKA is a function of the constraint of the prosthesis. Komnik et al.(68) investigated the transverse plane kinematics of TKA during ADL: walking, stair ascent and stair descent. They found no statistical significance in their TKA group between the CR and PS knees ($p<0.05$).

Figure 1.7: Patient performing activities of daily living during an *in vivo* study. Reproduced with permission, Kutzner et al.(69)

1.4 TKA Alignment Philosophies

The alignment (coronal, axial and sagittal) of the knee is vital for prothesis stability, prevents accelerated PE wear, and ensures patient satisfaction.(3) The intraoperative alignment techniques in TKA are broadly two groups, mechanical or kinematic. In mechanical alignment (MA), the components are implanted perpendicularly to the tibial and femoral mechanical axis, respectively.(70) In the kinematic alignment (KA) philosophy, purported improved functional outcomes are based on implanting the components according to the patient's pre-arthritic alignment.(70) The goal of the MA philosophy is to evenly distribute the forces going through the implant and implant-bone interface to facilitate implant longevity.

The "neutral" MA is non-anatomic, as the native knee has a mean alignment of 1.3^{o} varus.(71) Constitutional varus alignment has been described as a varus alignment of $\geq 3^{o}$.(71) Bellemans et al.(71) and Vandekerckhove et al.(72) reported 32% of males and 17% of females versus 46% of males and 23% of females in their study cohorts had constitutional varus respectively. The femoral rotation in MA is set parallel to the trans epicondylar axis (TEA). A human database study demonstrated the femoral rotation differed up to 11.3^{o} from the TEA in some individuals.(73) Therefore, MA will not restore the native knee rotation for all patients. KA maintains the native femoral rotation with symmetrical posterior femoral cuts of the pre-arthritic knee.(74) This results in the femoral rotation being neutral in KA and externally rotated in MA. The majority of published literature on KA is based on the CR designs, which requires a posterior tibial slope of 5-7^{o} to optimise the PCL function.(75) Overall, the KA is meant to result in bone preservation and require less soft tissue release and improved patient satisfaction.(76)

A recent systematic review investigated the clinical outcomes between the two alignment philosophies.(75) A total of 15 articles were reviewed. They found that the improvement in patient-reported outcomes and objective knee outcomes reported with KA by 2 cases series and two case-control trials were not reproduced in the six randomised controlled trials (RCTs). They concluded that further evaluation was required to determine if there was clinical superiority of either design. A meta-analysis of the literature included four studies and reported on 229 kinematic and 229 conventional TKA patients. The kinematic group demonstrated a higher combined postoperative

Knee Society Score than the conventional TKA group (mean difference, 9.1 points; 95% confidence interval, 5.2-13.0 points; $P < .001$).(77) An *in vivo* gait analysis study of 15 MA patients and 14 KA patients evaluated level walking was reviewed at a minimum 2-year follow up post-surgery. A 9-camera motion analysis system was utilised for the analysis. They concluded the overall differences in gait parameters were unable to demonstrate the superiority of either alignment philosophy.(78)

Previous literature advocated for TKA to restore a neutral to valgus (mechanical alignment) for the best implant survivorship when compared with the varus (kinematic) alignment, which historically resulted in failures.(79, 80) Recent evidence has indicated that tibia component alignment beyond the historical $\pm 3^{o}$ "safe zone" does not impact implant survivorship.(81) Proponents of KA in TKA have demonstrated that KA restores function without the risk of catastrophic failure.(82) A retrospective review of KA in 222 knees in 216 patients performed using patient-specific instrumentation reported implant survivorship of 97.5% for revision of any cause and 98.4% for aseptic loosening.(83) This study cohort was based on a single surgeon. A meta-analysis of the literature included nine studies reported on an aggregated 877 kinematic TKAs, and the cumulative survivorship was 97.4% at a weighted mean follow-up of 37.9 months.(77) They reported no difference in complication rates in their subgroup analysis. Notably, there is a paucity of high-quality KA long term follow up studies.

The optimal alignment for TKA is still debated. It remains unclear what the optimal soft tissue balancing and alignment is for implant survival and the most remarkable patient satisfaction.(84)

1.5 Polyethylene thickness

The PE insert engages on to the tibial component and articulates with the femoral component. Historically the literature refers to "thin", and "thick" PE inserts with varying definitions.(85, 86) Early literature was concerned with the minimum thickness of the PE inserts to preserve the subchondral bone and prevent PE deformation.(87) The concern with thick PE inserts >16 mm was the association with failure and the required larger tibial bone resection and were associated with ligament imbalances ($P < .0001$).(85) In the era of modularity and cross-linked PE, we have finer increments of PE insert sizing with modern systems offering between 1-4 mm thickness increments.

Changing the thickness of the PE insert will influence the soft tissue tension in both flexion and extension. Oversizing of the tibiofemoral joint space can result in limitation in knee ROM.(88) This can happen in the clinical setting by “over stuffing” the joint space with the PE insert. Intraoperatively the appropriate PE thickness is determined by overall knee balance and soft tissue tension.(89) Laskin et al.(88), in their paper on stiffness in TKA, coined the term “stuffing” of the joint space, which results in decreased ROM due to the inappropriate oversizing of implants (femur and tibia). Overstuffing of the joint with increased PE inserts thickness results in increased soft tissue tension, decreasing ROM and increasing joint stability. In addition to influencing the ROM and the overall functioning in TKA can be affected.(90, 91) The use of thicker PE inserts has been utilised to obtain ligamentous stability in the setting of instability.(85)

Understuffing the joint space by decreasing the PE insert thickness resulted in decreased soft tissue tension and increased ROM, but joint stability was decreased.(90) The under-sizing of the PE insert is a recognised cause of instability due to the resultant soft tissue laxity.(92) In TKA, soft tissue laxity can result in instability and cause pain.(85) Stiffness is defined as a ROM of less than 10^{o} - 90^{o} at six weeks post operatively.(93) In the treatment of stiffness in TKA, upsizing the PE insert for instability or downsizing in settings where the ligaments are excessively tight has been called an isolated PE insert exchange (IPE).(94) In the study by Green et al.(95) on IPE for flexion instability, all the PE inserts were upsized. They reported the PE inserts were increased by a mean of 3.5 mm, resulting in 7.6^{o} mean loss of ROM ($P = .0659$), and a mean increase of 3.6 mm resulted in an 11.9^{o} mean loss of ROM ($P=.0068$) in their CR and PS group respectively. Their case series presented a case of flexion instability and pain after TKA, which was successfully treated with an IPE from 9 mm to 11 mm PE insert.(95)

Kishimura et al.(91) in 127 knees looked at the relationship between increased PE insert thickness and developing a flexion contracture due to over stuffing of the tibiofemoral joint. They found that this worsened up to 3 months post-operative then improved, but at two years, the flexion contracture persisted due to the PE thickness. Okamato et al.(96), in their prospective comparative study, found that the clinical significance of overstuffing the tibiofemoral joint results in a flexion contracture due to increased soft tissue tension in their study of 75 knees. They concluded an extension laxity of >1 mm intraoperatively is required to prevent a flexion contracture. Lanting et al.(90), in their clinical trial of 35 TKAs, reported that PE insert thickness was statistically significant in influencing ROM and coronal plane joint stability ($P<.001$). They reported that a base size 0 PE and an increase of 2 mm thickness resulted in a loss of 3° of extension while a decrease of 2mm resulted in 5° of hyperextension(90). Asano et al.(97) measured the soft tissue tension intraoperatively and evaluated the knees at one year post operatively. They found the extension deficit was larger in a setting of increased soft tissue tension ($P<0.05$). The varus laxity in full extension and 30° flexion and valgus laxity at 30° was reduced with an increase of soft tissue tension ($P<0.05$). Other studies have demonstrated similar results in the effect of intraoperative soft tissue tension on ROM.(98) Bandi et al.(99) assessed the effect of increments of PE insert thickness on laxity and kinematics using *in vitro* testing and numerical simulation. They used mounted cadaveric knees and demonstrated that by increasing PE insert thickness, the soft tissue tension increased, influencing the VV, IE and AP kinematics ($p < 0.001$). The change doubled from 1 mm to 2 mm. They concluded the smaller increments in PE insert thickness will help achieve the optimal balance of the knee. Achieving the optimal soft tissue tension is linked to adequate kinematics in TKA.(85) Additional studies are required to understand how changes in the PE insert size affect the resultant TKA kinematics.

1.6 Instability in flexion

Approximately one-third of early TKA failures are due to soft tissue imbalance resulting in instability being a common indication for revision surgery.(12, 13, 100-102) TKA stability is intraoperatively assessed at full extension and at 90° of flexion but not routinely in the mid-flexion ranges. Flexion and mid-flexion instability are controversially used interchangeably to describe instability in a flexed implanted knee.(103-106) This instability in flexion is multifactorial and is currently a poorly understood phenomenon.(105) During activities of daily living (ADL), the knee is maintained in a flexed position.(107) Patients with instability present with vague complaints with walking and stair activity.(104)

The definitions of flexion and mid-flexion instability are often used interchangeably in the literature.(103-106, 108) The classical "mid-flexion" instability in TKA literature refers to coronal and AP instability.(109) This has been reproduced in both cadaveric and computer model studies with an elevation of the joint line.(103, 110) Martin and Whiteside performed their landmark study using ten cadaveric knees.(109) The knees were mounted onto a machine, a 45N vertical load and 10 Nm moment were applied with testing at 15^{o} intervals of flexion up to 90^{o}. In five knees, they shifted the femoral component proximal and anterior by 5 mm. In the remaining five knees, they shifted the femoral component distal and posterior by 5 mm. They assessed the VV, AP displacement in mm at the respective flexion angles. They reported increased VV and AP displacements in the proximal and anterior group compared to control group. They reported the opposite in the distal and posterior group with decreased VV and AP displacements. Mid flexion instability is illustrated with joint line elevation in *Figure 1.8*. Flexion instability was first described as an extension and flexion gap mismatch. The knee is stable at full extension but lax at 90^{o} of flexion.(111) Patients report instability, particularly when bearing weight on a flexed knee. As the knee flexes, the stability decreased compared to full extension due to the soft tissue laxity. The laxity allows the femur to glide on the tibia, which is perceived as instability by the patient and results in recurrent effusions and pain.

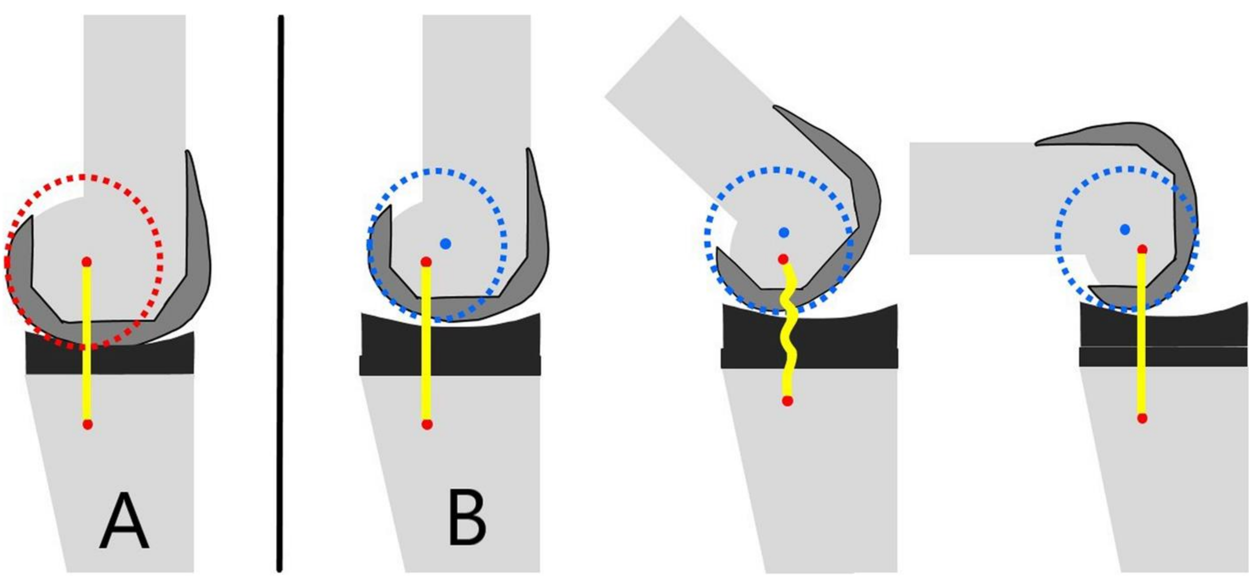

Figure 1.8: Mid-flexion instability after joint line elevation. Situation "A" represents a non-elevated joint line TKA, the center of rotation (red/blue dot) is restored by complete

restoration of the posterior condylar offset and joint line height. The medial collateral ligament (marked yellow) will keep its isometry throughout the entire range of motion. Situation “B” represents an elevated joint line TKA with ticker insert to compensate. The axis of flexion (blue dot) no longer coincides with the MCL insertion (red dot). Therefore, the knee is stable in extension and 90° of flexion but laxity occurs in the mid-flexion range. The medial collateral ligament loses its isometric function in mid-flexion. Reproduced from van Lieshout et al.(112)

A recent systematic review and meta-analysis reviewed literature investigating the possible causes of mid-flexion instability.(113) The identified investigated risk factors in the literature were joint line elevation, conformity of the TKA prosthesis, PS vs CR design, the radius of curvature of the femoral condyle, gap balancing at 0º and 90º, mechanical vs anatomical alignment, bicruciate retaining design, anterior placement of the femur, reverse vs classical gap balancing, posterior condyle offset, bearing type in PS-TKA, preoperative joint gap/laxity, implant size and activity type. Most investigations regarding instability in flexion are *in vitro* studies.(104, 113) This highlights the role of *in vitro* studies in understanding findings from the clinical setting.

1.7 Posterior tibial Slope

The posterior tibial slope (PTS) is the bony cut performed on the proximal tibia during preparation for the tibial component. The various techniques in determining and performing the PTS cut intraoperatively are debated.(114) The PTS influences femoral rollback and stability in TKA. The PTS is measured on a lateral x-ray of the knee, and this can be performed on both the implanted and native knee *Figure1.9.*

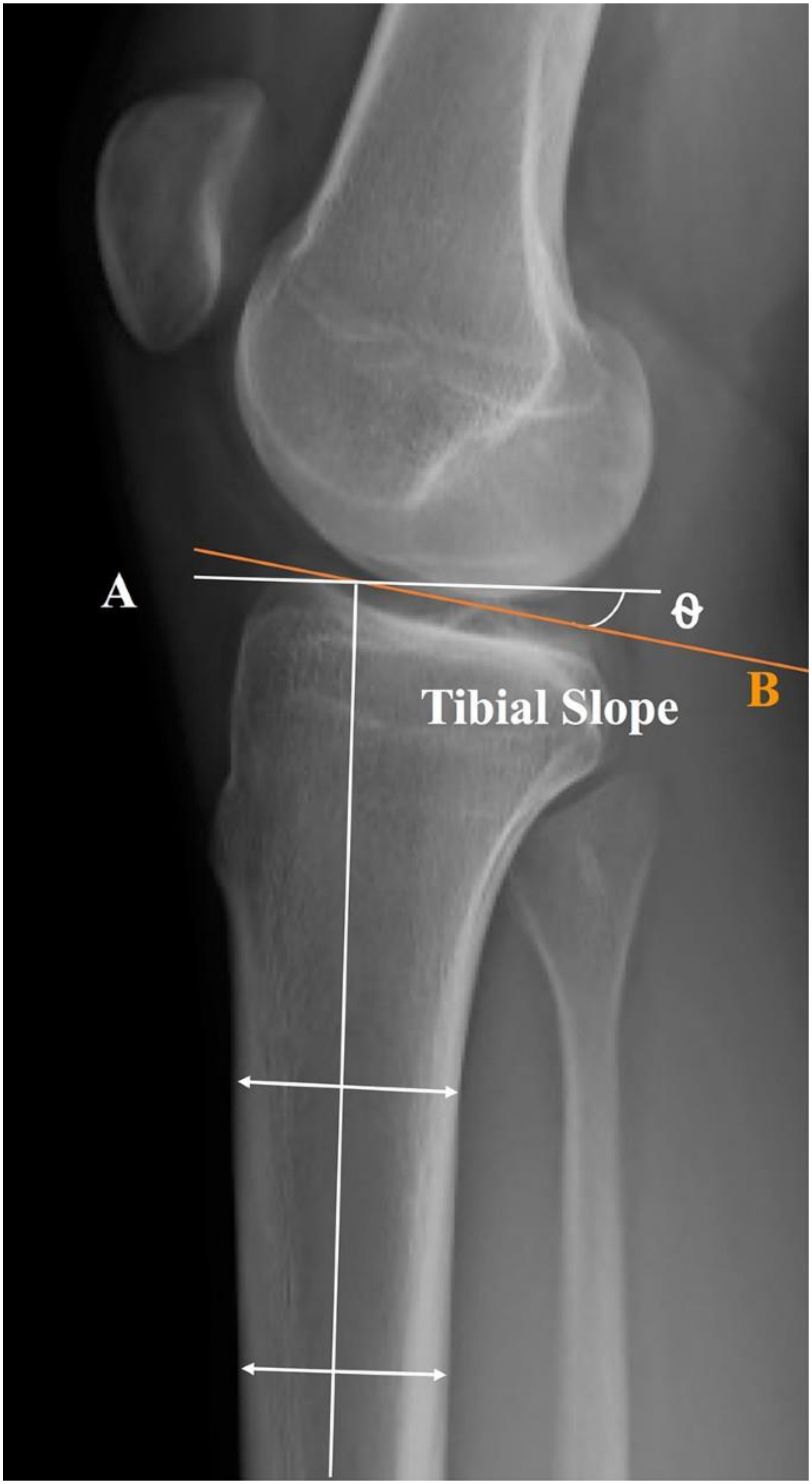
A
θ
B
Tibial Slope

Figure 1.9: 'True lateral' radiograph of a knee. Posterior tibial slope (*θ*) is the angle between the perpendicular to the tibial diaphysis (a), and the tangent to the anterior and posterior edges of the medial tibial plateau (b). Reproduced with permission, Dejour et al.(115)

Increasing the tibial slope results in greater femoral rollback and therefore increases maximum flexion of the knee. In TKA, the desired slope is influenced by the prosthesis design, and in PS-TKA, the aim is 0^o - 3^o and in CR-TKA, 5-7^o or to match the anatomical slope of the patient. Khasian et al.(116) performed a computational sensitivity analysis in CR-TKA, increasing the tibial slope from 0–8^o in 2^o increments. Comparing 0^o and 8^o of PTS in full extension, the lateral femoral condyle contact point was posterior to the midline at 5.63 mm and 11.38 mm, respectively. In the medial femoral condyle, the contact point moved from 7.29 mm to 13.07 mm at full extension due to changing the slope. The overall femoral rollback in the lateral condyle is reduced by 2.12 mm to 0.22 mm at 0^o and $8^{o,}$ respectively. They concluded that although the femoral rollback reduced, the more posterior femur position with increased slope prevented femoral impingement on the tibia, therefore allowing deeper knee flexion. Fujito et al.(117) utilized fluoroscopic surveillance to analyse kinematics in CR-TKA with $\leq 7^o$ or less tibial slope compared to $\geq 8^o$. They concluded that increasing the PTS did not affect the AP translation or the external rotation but increased the ROM (P=.0053). The effects of PTS are influenced by gravity and weight-bearing. (118) Chamber et al. (119) performed a cadaveric fluoroscopic study with CR-TKA increasing the PTS from 1^o to 4^o resulted in; an average increase of 2.3^o of flexion, mean femoral back increase of 1 mm per 1^o of PTS (P <.05). In AS-TKA, PTS of $\geq 10^o$ resulted in anterior impingement, and PTS $\geq 5^o$ resulted in instability both during stair ascent.(114) Bellemans et al.(120), in their cadaveric simulation of 21 CR-TKAs, demonstrated increasing ROM with a mean maximal flexion angle of 104^o, 112^o and 120^o with a PTS of 0^o, 4^o and 7^o, respectively. Overall, 1.7^o of flexion is gained with each additional 1^o of PTS in their study. The literature broadly suggests increased PTS in CR-TKA results in increased ROM.

The literature demonstrates decreasing PCL tension with an increasing PTS.(121, 122). Singerman et al.(121) measured the PCL strain with a differential variable reluctance transducer (Microstrain, Burlington, VT) to measure changes in ligament length. The PTS were 5^o, 8^o and $10^{o,}$ and they found as the PTS increased, the PCL strain decreased. In addition, they found the PCL strain increased up to 100^o then began to decrease with deeper flexion irrespective of the PTS. Kuriyama et al.(123) demonstrated using a musculoskeletal computer model that 2^o increments in the PTS decreased the force at the PCL by up to 41%.

The relevance of the PTS in PS-TKA has been investigated. Kang et al.(124) analysed PS-TKA with a PTS from -3^o to 15^o in 3^o increments using a finite element model. They demonstrated increased posterior AP translation with a net increase of 4.9 mm and increased contact post pressure with increasing the PTS. They also reported decreased collateral tension with increasing PTS, raising concerns of clinical instability in PS-TKA. Shi et al.(125) analysed 65 PS-TKAs divided into three groups of PTS; Group 1: ≤ 4^o, Group 2: 4^o - 7 ^{o} and Group: > 7^o. The mean maximal flexions were 101^o (SD 5), 106^o (SD 5) and 113 ^{o} (SD 9) in Groups 1, 2 and 3, respectively (P<.001). In their cohort, an increase of 1^o in the tibial slope resulted in a 1.8^o flexion increase (r = 1.8, R2 = 0.463, P<.001). The AP translation at 30^o flexion was; 7.4 mm (SD 1.3), 7.6 mm (SD 1.2) and 7.9mm (SD 1.1) in groups 1, 2 and 3, respectively. The AP translation at 90^o flexion were; 5.2 mm (SD 0.5), 5.3 mm (SD 0.5) and 5.5 mm (SD 0.5) in groups 1, 2 and 3 respectively. Overall the differences in AP translation were found not to be statistically significant. In PS-TKA, increasing the PTS still remains controversial.

The PTS can influence soft tissue laxity and knee stability. Inadequate PTS in CR -TKA can result in a tight flexion space with subsequent decreased ROM.(126) Various studies have shown that by increasing the PTS, the soft tissue laxity increases. This compromises the collateral ligaments causing flexion instability.(122, 127)

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Chapter 2
Thesis Objectives

The current literature on kinematics in total knee arthroplasty (TKA) highlights numerous unclear and debated areas within the field of study. Furthermore, the current limitations in understanding the kinematics in TKA have been suggested as part of the multifactorial cause of dissatisfaction within TKA patients. Therefore, this study aimed to further the understanding of kinematics in TKA.

Recent advances in joint motion simulator technology facilitate further *in vitro* investigations. This study employed a hybrid computational-experimental joint motion simulation on a 6 degrees of freedom joint motion simulator. In addition, physical prototypes of a virtually performed TKA based on cadaveric CT scans and a virtual ligament model were utilized.

We investigated specific contemporary topics in TKA from a kinematic perspective. In Chapter 3, we studied the kinematic and laxity differences between cruciate-retaining (CR), and posterior-stabilized (PS) designed TKA during activities of daily living (ADL) and laxity testing. Our hypothesis in this chapter was that there would be no difference between the designs during the simulations.

Chapter 4 investigated the influence of TKA alignment philosophies and polyethylene (PE) insert thickness in PS- TKA during neutral flexion and laxity testing. Our hypothesis in this chapter was that understuffing the joint space with a downsized PE insert would increase laxity and alter the post-cam mechanism. Conversely, the inverse would occur with overstuffing the joint space.

Chapter 5 investigated the influence of the PE insert thickness on instability in flexion in CR-TKA during ADL. Our hypothesis in this chapter was that understuffing the joint space would cause instability with flexion during ADL.

The outcome measures were anteroposterior translation, axial rotation, coronal displacement, joint compressive forces, and ligament tensions. These were acquired through simulations of loads and motions of neutral flexion, laxity testing and ADL in the respective chapters.

Chapter 3

Cruciate-retaining and posterior-stabilized total knee arthroplasty: a comparative study

Abstract

Modern total knee arthroplasty (TKA) has been a success for the past 50 years. However, there is controversy regarding the selection of prosthesis design in TKA, particularly in cruciate-retaining (CR) versus posterior-stabilised (PS) and which design results in the optimal knee kinematics. The study aimed to identify differences between CR- and PS-TKA.

This study employed a hybrid computational-experimental joint motion simulation on a VIVO 6 degrees of freedom joint motion simulator (AMTI, Watertown, MA, USA). Physical prototypes of a virtually performed TKA based on cadaveric CT scans and a virtual ligament model were utilised. Activities of daily living (ADL) and laxity testing, loads and motions were simulated for both the CR and PS configurations.

During the gait cycle (40-85%) there is a difference in axial rotations were -11.7^{o} ±3.9 (-16.8 to -6.6) and -9.4^{o} ±3.1 (-13.2 to -5.8) in the CR and PS configurations, respectively. In stair descent, the stance phase (20-60%), -2.9mm ±3.8 and -1.8mm ±3.3 in CR and PS, respectively. In the stair stance phase (20-60%), -2.9mm ±3.8 and -1.8mm ±3.3 in CR and PS, respectively.

Surgeons who choose to perform either CR or PS-TKA should understand the design indications and the associated kinematic differences. Overall, during ADL, we found the kinematics between the two implant designs to be similar. However, there is increased stability in PS-TKA during ADL.

Introduction

The primary goal of total knee arthroplasty (TKA) is to provide a painless stable knee and restoration of joint function in patients with advanced osteoarthritis. The knee prosthesis should be performed to provide maximal range of motion (ROM) and stability.(4, 128) The kinematics in TKA can be influenced by the prosthesis design. We investigated the differences between CR- and PS-TKA.

There is controversy regarding prosthesis design in TKA, particularly in cruciate-retaining (CR) versus posterior-stabilised (PS). This design results in optimal knee kinematics and function.(129, 130) The CR design has been reported to reproduce more physiological kinematics and with retention of posterior cruciate ligament (PCL), resulting in improved proprioception of the implanted knee.(129, 131, 132) However, a Cochrane systematic review found that preserving the PCL compared to PCL sacrifice may not result in improved ROM, pain, function or patient satisfaction.(133) The clinical outcomes between the two TKA prosthesis designs have been an ongoing area of interest in arthroplasty related research.(31, 134-137)

Recent advances in joint motion simulator technology have enabled these machines to analyze TKA mechanics while simulating different alignments and soft tissue balancing by changing the properties of parametric virtual ligaments.(15) The objective of this study was to determine the differences between CR- and PS-TKA in simulated activities of daily living (ADL) and laxity testing using a joint motion simulator machine linked to a virtual ligament model. We hypothesised that there would be no difference between the designs during the simulations.

Methods

This study employed a hybrid computational-experimental joint motion simulation on a VIVO 6 degrees of freedom (6-DoF) joint motion simulator (AMTI, Watertown, MA, USA) shown in *Figure 3.10*. These simulations measure the kinematics of physical implant components in response to applied loads, but with forces imposed due to simulated stretching of surrounding “virtual ligaments”. These simulated one-dimensional point-to-point springs are virtually installed around the implant components, will become tensioned or slack in response to insertion kinematics (insertion coordinates defined relative to the implant components), and the forces they generate will contribute to joints kinematics and stability. The virtual ligaments employed in the current study were designed to replicate the relative insertion coordinates, tensioning, and stiffness of real ligaments around a mechanically aligned TKA, using the following procedure.

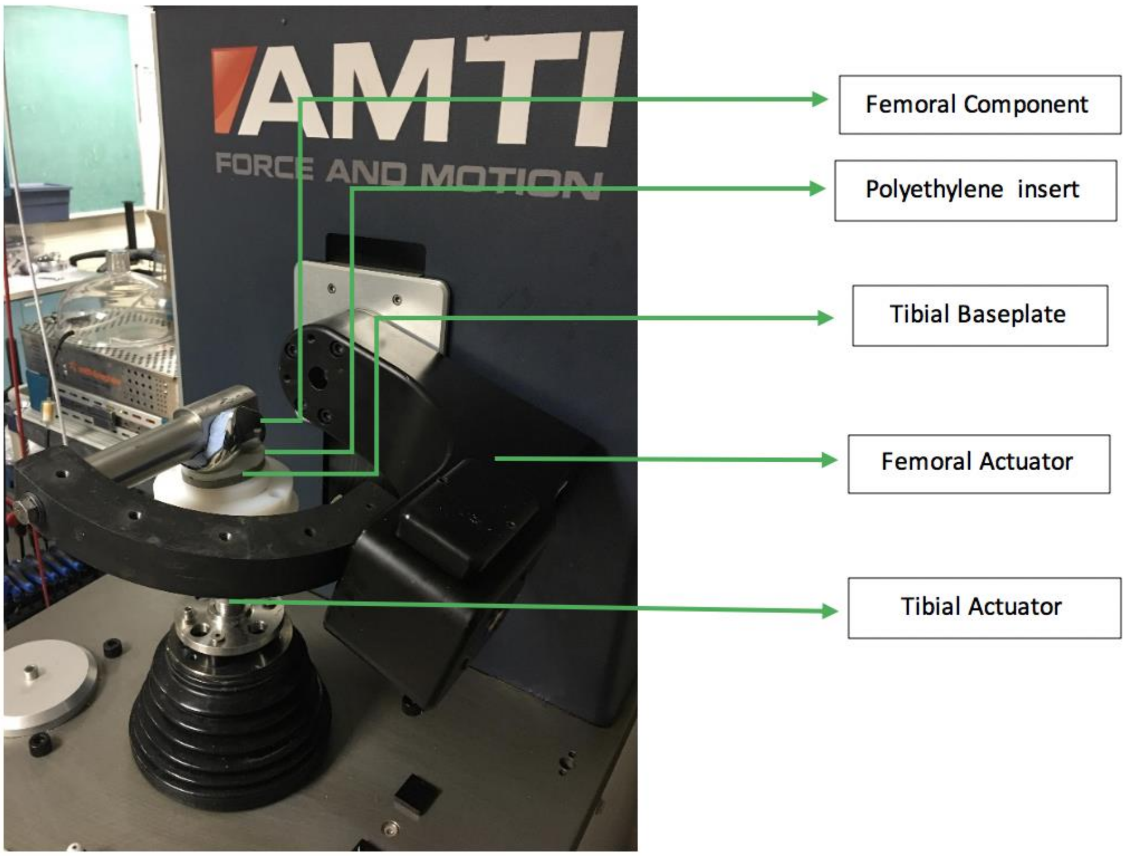


Figure 3.10: VIVO joint motion simulator with components labelled.

Virtual ligament design

The distal femur and proximal tibia were reconstructed from a CT scan of a single cadaver knee (female 53-year-old) in neutral extension using 3D Slicer(138) and exported as stereolithography (.stl) files. These 3D surface models were imported into the commercially available CAD software, SolidWorks 2020® (Dassault Systèmes SolidWorks Corporation, Waltham, MA). In SolidWorks, this native knee geometry was used to identify relevant ligament insertions based on established bony landmarks and previous literature.(139-146) The femoral and tibial insertions of the posterior cruciate ligament (PCL), superficial medial collateral ligament (sMCL), and lateral collateral ligament (LCL) were each identified. They would serve as connection points for the single-bundle virtual ligaments. The anterior cruciate ligament and deep MCL were not represented as they are routinely released in TKA procedures as part of the soft tissue dissection required for exposure or bony resection. The PCL was “turned off” in the PS simulations. The insertion-to-insertion distance of each ligament was noted. Acknowledging that ligaments may be slack or stretched when in neutral extension, these distances could not be used to define the actual resting length (unstretched or “slack” length) of each ligament. The slack length, however, was estimated based on literature-derived intact knee ligament reference strains and the insertion-to-insertion distance in extension.(147-149)

Surface models (.stl files) of a tibial and femoral Triathlon® (Stryker Corp., Mahwah, NJ) TKA components were imported into the CAD software. A size five tibial tray with a 9 mm thick CR and PS polyethylene (PE) insert component was used, in combination with a size five femoral component. Virtual TKA was performed with mechanical alignment (MA), which required the distal femur bone model to be cut perpendicular to the femur’s mechanical axis, and for the proximal tibia to be cut perpendicular to the tibia’s mechanical axis. The femoral component was aligned with the approximated trans-epicondylar axis; this was determined by externally rotating 3° from the posterior condylar axis. The posterior sloped tibial resection was performed at 5º for CR TKA and 3° for PS TKA. The implant components were aligned to the cut surfaces of the femur and tibia, and then the entire tibia (plus tibial component) was translated and rotated such that the femoral and tibial components were neutrally positioned and aligned (in extension, with the femoral condyles dwelling at the deepest point in the PE dishes). The coordinates of the ligament femoral and tibial insertions were measured with respect to the femoral component, as were their new lengths, which changed after TKA. These data were sufficient to define virtual ligaments around the TKA (CR and PS) along with the ligament stiffness. Please see Appendix B for a further in-depth explanation.

Joint motion simulator

Real physical prototypes of the same implant components were mounted onto the VIVO. The femoral component was mounted to the mounting axle with poly methyl methacrylate cement (Bosworth Fastray; Keystone Industries GmbH, Singen, Germany), and the tibial baseplate component was anchored into the tibial fixture using dental model stone (Modern Materials Golden Denstone Labstone; Modern Materials, Kulzer GmbH, Hanau, Germany). We used polydimethylsiloxane (silicone)-based lubricant (HAS0001-OS, Horizon Fitness, Cottage Grove, WI) as an articulation lubricant and applied it consistently throughout the experiment. The VIVO was used to apply loads and motions representative of laxity testing and ADL. The resulting kinematics (measured outcome) were sensitive to the implant component geometries, alignments, and virtual ligament properties. This *in vitro* technique of measuring motions and kinematics with simulated virtual ligaments has been previously described.(15)

Input loads and motions

Motions were simulated as follows:

(i) Posterior Laxity: With a 10 N compressive force applied parallel to the long axis of the tibia and passing through the centre of the joint, the joint was flexed from 0° to 90° in 15° increments. A 100 N posterior-directed was applied at each fixed flexion angle, causing a relative posterior displacement of the tibia. All other degrees of freedom were unconstrained. After testing at each flexion angle, the joint returned to 0°, and the entire process was repeated for a total of four iterations. Data were recorded during the 3rd and 4th iteration. Recorded data included the posterior displacement of the tibia (relative to the corresponding neutral flexion kinematics at the same flexion angle), net ligament forces and individual ligament tensions.

(ii) Varus/Valgus (VV) Laxity: VV laxity testing was accomplished using a similar technique described for (i), but by applying a 10 Nm varus or valgus joint torque instead of a 100 N posterior force. Recorded data included the varus/valgus angulation (relative to the corresponding neutral flexion kinematics at the same flexion angle), net ligament forces and individual ligament tensions.

(iii) Activities of Daily Living: Simulated ADL were performed as in previous studies(22) by applying loads and motions, which were based on data measured using load-sensing TKA implants.(150) Gait, stair ascent and descent were simulated. For the gait files, the load cycle begins at the flat foot and goes through the gait cycle (flat foot, heel off, toe-off, swing phase, heel strike, flat foot etc.). This splits the gait cycle into the first 60% stance phase and the last 40% swing phase. The stair ascent and descent load cycles both begin and end with the middle of swing phase. This splits the stair ascent and descent into the first and last 20% swing phase and the middle 60% as stance phase. The gait cycle, stair ascent and descent loads, and motions based on the AVG75 dataset from the Orthoload website database were used (https://orthoload.com/).(150) All degrees of freedom were operated in force control, with the exception of flexion angle, which was prescribed. Each activity was simulated for four cycles, with data recorded during the third and fourth cycle which included 6-DoF kinematics.

Data analysis and statistics

Recorded data were smoothed using a low-pass Butterworth filter followed by a spline interpolation function in Matlab (The MathWorks, Natick, MA), and then down-sampled to only include data at 15° intervals of flexion and only during the flexion phase of the complete flexion/extension motion in the laxity testing. During the ADL testing, the joint motion was sampled throughout the cycle. We extracted the anteroposterior (AP), internal/external rotation (IE) and VV kinematic data in each of the 6-DoFs. We also collected posterior, varus and valgus motion limits in each of the 6-DoFs at these limits. The smoothed and processed data were used for statistical analysis. Descriptive statistics were used to compare the results of each implant design. All statistical analyses were completed in Microsoft® Excel (v.16.45).

RESULTS

Laxity testing

The results of the posterior laxity testing are shown in *Figure 3.11*. The mean translation 0-90° for posterior laxity *(Figure 3.11-A)* was 7.6 mm ±3.5 (3.0 to 11.6) and 4.9 mm ±3.7 (0.5 to 9.1) for the CR and PS configurations, respectively. A 57% increase in the mean posterior laxity in the CR configuration was compared with the PS configuration. The coronal laxity results are shown in *Figure 3.11-B*. The coronal laxity results were similar from 0-45° of knee flexion. Within 45-90° of knee flexion, the laxities were 4.2° ±0.6 (3.8 to 5.0) and 4.9° ±0.4 (4.6 to 5.5) for the CR and PS configurations, respectively. The CR configuration yielded a 14% decrease in mean coronal laxity compared to the PS configuration during this range (45-90°) of knee flexion.

Gait cycle

The kinematics during the gait cycle are shown in *Figure 3.12*. Regarding the AP kinematics, both models had the same relative AP position at the start of the gait cycle *(Figure 3.12-A)*. Across the stance phase, the mean values were -3.5 mm ±3.4 and -2.9 mm ±2.7 for the CR and PS configurations, respectively. During the stance phase at 10-35% of the cycle, the CR model translated into a more posterior position of -6.6mm ±1.1 and -5.2mm ±0.2 for CR and PS configurations, respectively. In the swing phase, the mean values were 1.4mm ±1.7 and 1.4mm ±1.3 for CR and PS, respectively. The overall wave patterns were similar. The axial rotation kinematics are shown in *Figure 3.12-B*. The CR configuration maintains a more internally rotated position throughout the gait cycle, and the mean values were -7.5° ±6.2 and -5.9° ±5.5 in the CR and PS configurations, respectively. During the gait cycle (40-85%), there is a difference in the waveforms of -11.7° ±3.9 (-16.8 to -6.6) and -9.4° ±3.1 (-13.2 to -5.8) in the CR and PS configurations, respectively. During this period, the CR configuration demonstrates greater rotation. The coronal kinematics are shown in *Figure 3.12-C*. The overall mean coronal kinematics were similar between designs and were -0.6° ±0.8 and -0.5° ±0.7 in the CR and PS configurations, respectively.

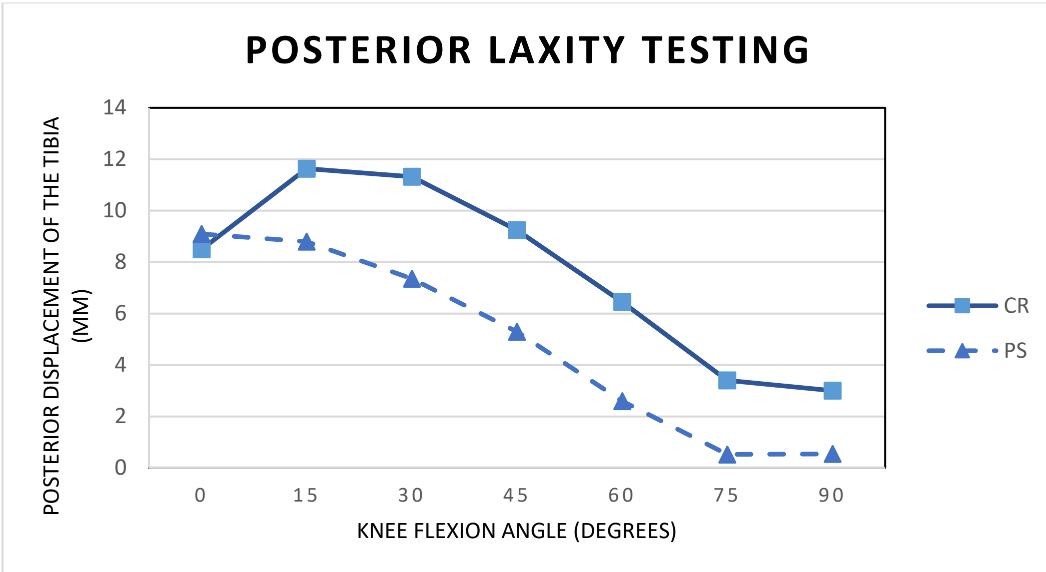
Stair descent

The stair descent kinematics are shown in *Figure 3.13*. The AP kinematics *(Figure 3.13-A)* were similar between the design throughout the stair descent cycle 0.5 mm ±4.6 and 0.6 mm ±4.4 in the

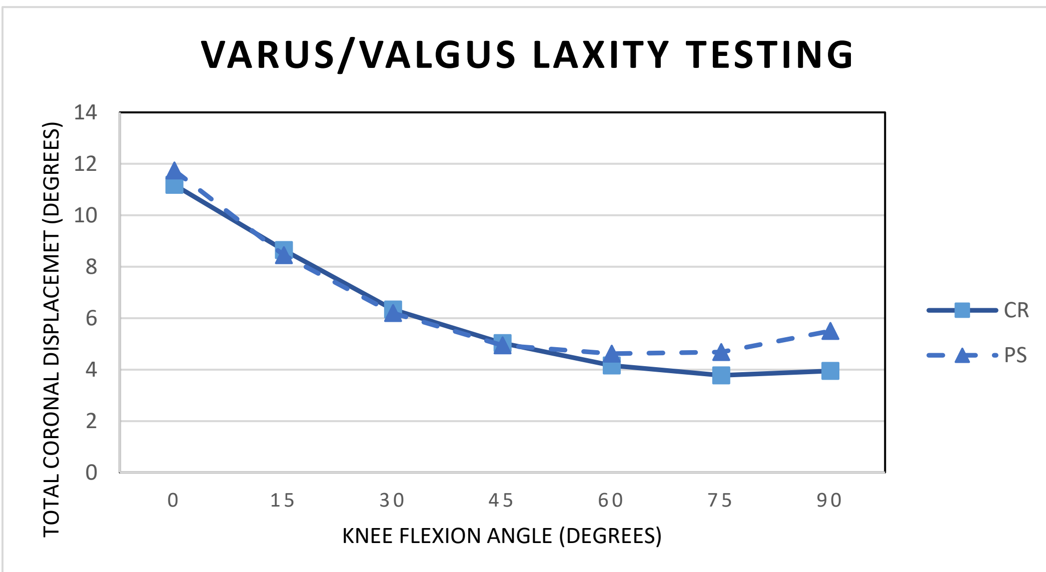
CR and PS configurations, respectively. The axial rotation kinematics are shown in *Figure 3.13-B*. The overall mean axial rotation kinematics were -7.4^o ±6.0 and -6.5^o ±4.6 in the CR and PS configurations, respectively. Differences were observed in the stance phase (40-80%) with -12.2^o ±3.7 and -9.3^o ±2.5 in CR and PS, respectively. The PS configuration demonstrated greater stability. The coronal kinematics *(Figure 3.13-C)* were similar between the designs throughout the stair descent cycle, with the mean value of -0.5^o ± 0.5 in both models.

Stair ascent

The stair ascent kinematics are shown in *Figure 3.14*. The overall AP translation kinematics *(Figure 3.14-A)* were -0.1 mm ±4.5 and 0.3 mm ±4.1 in the CR and PS configurations. Differences were observed during the stance phase (20-60%), -2.9 mm ±3.8 and -1.8 mm ±3.3 in CR and PS, respectively. The PS configuration demonstrated greater stability, with the CR tibia translating more posteriorly. The axial rotation kinematics are shown in *Figure 3.14-B*. The overall axial rotation kinematics were -6.8^o ±4.4 and -5.9^o ±3.8 in the CR and PS configurations, respectively. The starting point was the same for both designs. However, from 20-35% and 45-100% of the cycle, the CR configuration was in more internal rotation than the PS configuration. During 20-35%, the values were -8.2^o ±4.0 and -7.3^o ±2.7 for CR and PS, respectively. During 45-100%, the values were -7.9^o ±4.3 and -6.7^o ±4.0 for the CR and PS configurations, respectively. The coronal kinematics *(Figure 3.14-C)* were similar between the designs throughout the stair ascent cycle. The mean was -0.4^o ±0.5 in both models.



A



B

Figure 3.11: Graphs representing the laxity testing results. A-Posterior laxity testing. B-Varus/valgus laxity testing.

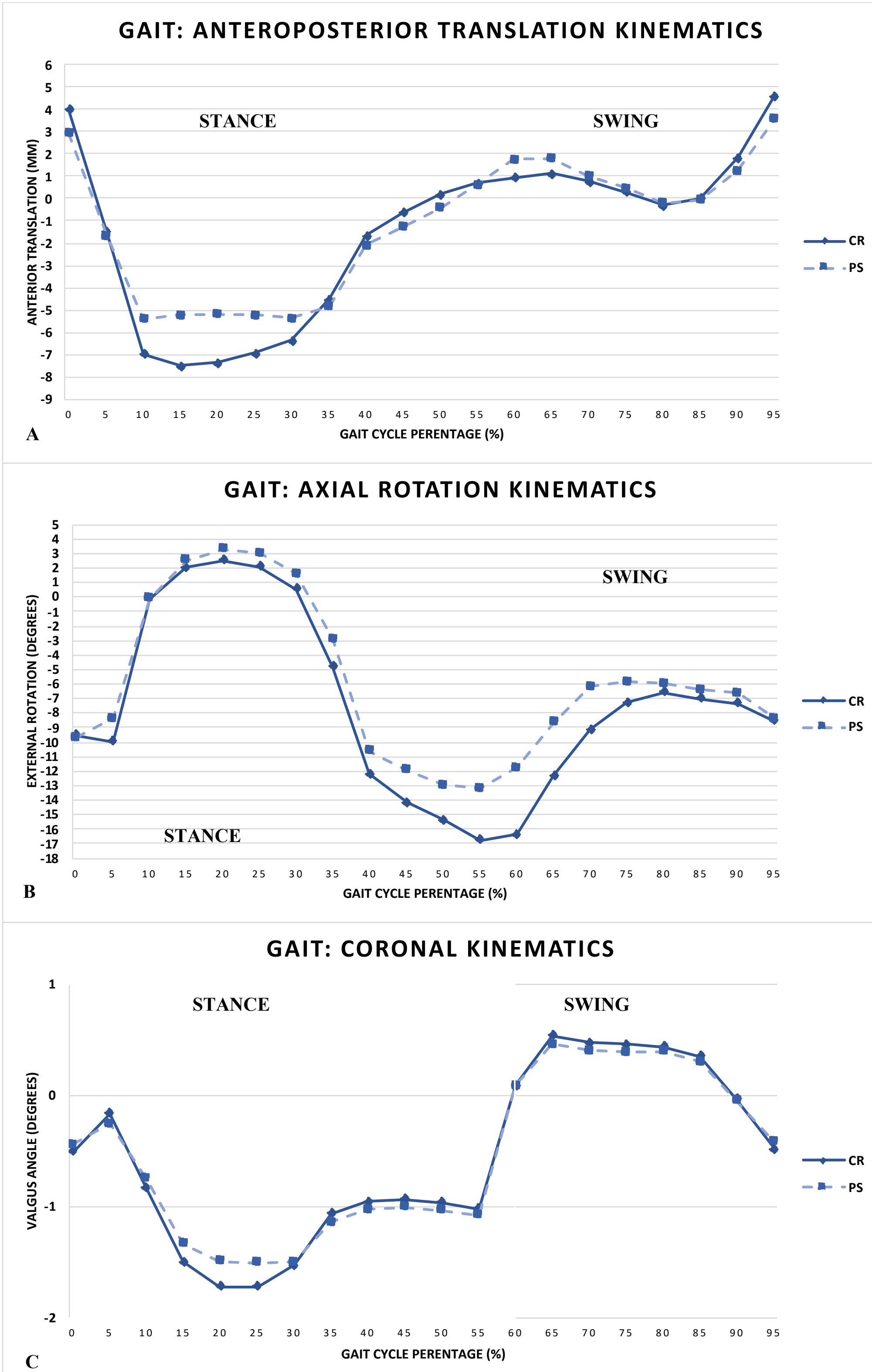


Figure 3.12: Graphs representing the knee kinematics during gait. A- Anteroposterior kinematics. B- Axial rotation kinematics. C- Coronal kinematics.

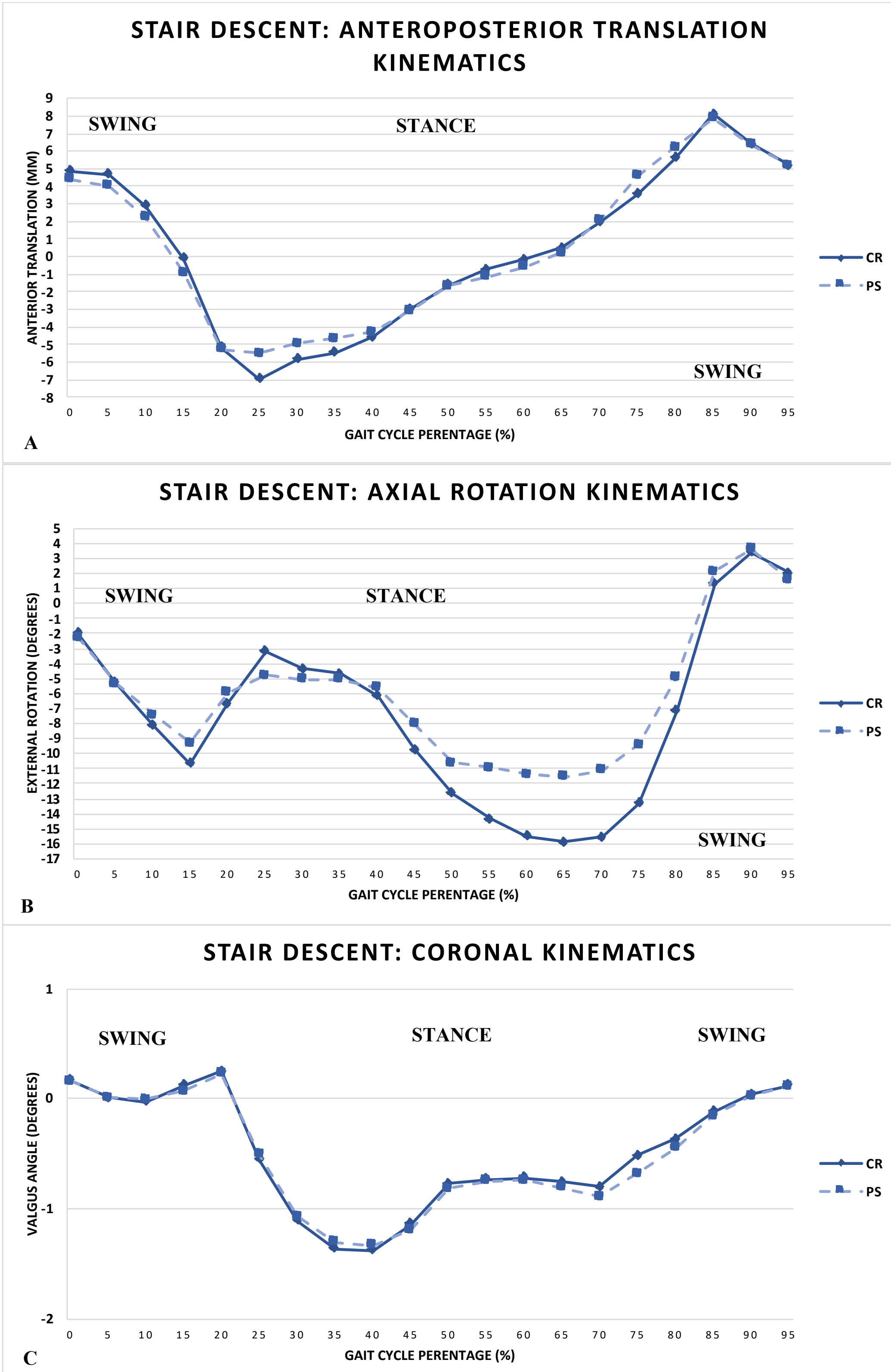


Figure 3.13: Graphs representing the knee kinematics during stair descent. A- Anterior posterior kinematics. B- Axial rotation kinematics. C- Coronal kinematics.

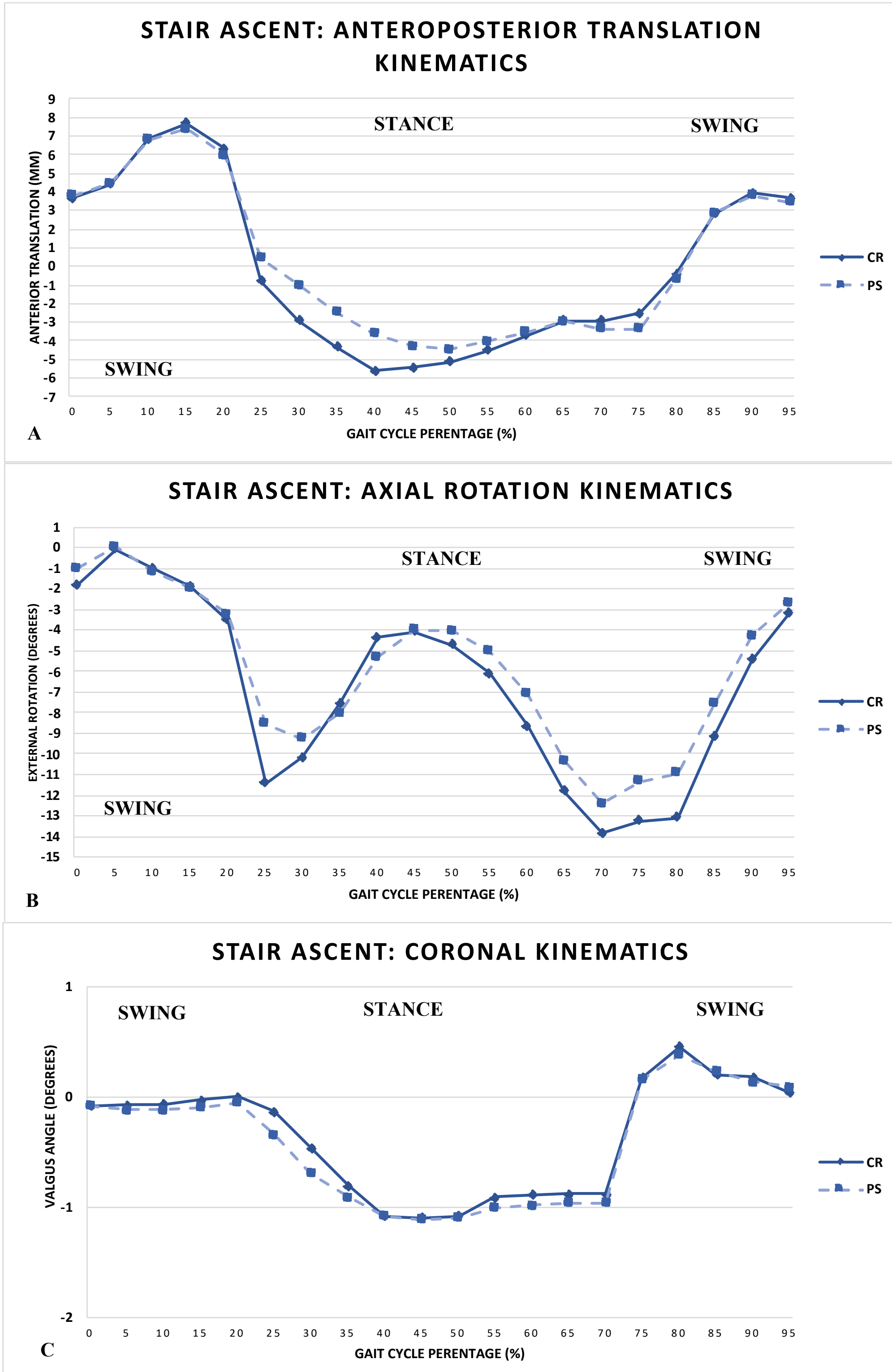


Figure 3.14: Graphs representing the knee kinematics during stair ascent. A-Anterior, posterior kinematics. B-Axial rotation kinematics. C-Coronal kinematics.

Discussion

CR and PS-TKA are both popular implant designs. Our study evaluated laxity testing and kinematics during ADL between the two designs. The kinematic studies in comparing CR- and PS-TKA are focused mainly on the AP translation kinematics, femoral roll back, and ROM.(40, 41, 151, 152) In CR-TKA, the function of the PCL is to act as a restraint to anterior femoral translation while facilitating femoral rollback.(151) In PS-TKA, the function of the PCL is substituted by the post-cam mechanism. The post-cam mechanism engages at around 75^{o} of flexion. This engagement prevents anterior femoral translation of the femur and facilitates the femoral rollback in PS-TKA.(40, 41) PS-TKA increases the sagittal constraint of the knee.(42) The 57% increase in the mean posterior laxity in the CR configuration compared with the PS configuration observed in our study was expected for the implant design. During the gait cycle, the initial anterior translation of the femur on the tibia demonstrated in our results (*Figure 3.12-A*) has been reported in CR and PS-TKA using *in vivo* fluoroscopic analysis.(153) Hamai et al.(154) evaluated CR- and PS-TKA with respective to their *in vivo* kinematics using radiographic-based image-matching techniques. They found the post-cam mechanism did not engage during stair climbing, but this was due to the dynamic flexion angle in their study being less than 75^{o}. They found that CR-TKA demonstrated more sagittal stability. In our study, the PS configuration demonstrated greater stability, with the CR tibia translating more posteriorly during the stair ascent stance phase. The post-cam mechanism created a hard stop after engagement in the PS configuration. While a gradual translation was seen in CR configuration is related to the elasticity of the PCL. Overall, we found increased AP stability in the PS configuration, particularly during the stance phase in all the ADL.

In reviewing the coronal kinematics between the two implant designs, Perkins et al.(155) performed a retrieval analysis study with 47 cadaveric knees measuring laxity patterns and wearing a custom knee testing machine. They reported the PS knees had increased varus laxity at 60^{o} (PS =10.1^{o} vs CR =6.9^{o}; P=.022) and 90^{o} of flexion (PS=12.4^{o} vs CR=7.5^{o}; P=.017). Additionally, the PS implants had a statistically significant increased total coronal laxity at 60^{o} when compared to the CR knees (PS=16.9^{o} vs CR=13.3^{o}; P=.05). In the PS cohort, the laxity correlated with wear, but this was not the case in the CR cohort. In a computer navigation study of 34 knees, Hino et al.(156) demonstrated increased coronal stability in CR- TKA compared to PS-TKA. The differences were statistically significant at 10° (p = 0.0093), 20° (p = 0.0098) and 30° of flexion (p = 0.0252). Hamai et al.(154) reported increased coronal stability in CR-TKA compared to PS-TKA in stair ascent. During stair ascent, we found coronal kinematics were similar between the designs throughout the stair ascent cycle. The mean was -0.4^{o} ±0.5 in both models. In our study, the CR configuration yielded a 14% decrease in mean coronal laxity compared to the PS configuration during this range (45-90^{o}) of knee flexion. These findings correlate with the stability in flexion with PCL retention. However, during all the ADL in our study, the coronal stability was similar between the two implant designs.

The CR and PS designs allow ± 9° and ± 12° of IE rotation, respectively.(33, 44) In our study, all designs, irrespective of alignment, demonstrated the internal rotation of the tibia with flexion. Cates et al.(55) evaluated the in vivo kinematics using fluoroscopy, reported in CR and PS-TKA that from full extension to 90^{o,} both designs underwent internal rotation of the tibia. The axial rotation means angles at full extension were 0.1^{o} and -1.0^{o}. The mean axial rotation angle was 4.9° and 1.9° for the CR and PS knees at maximum flexion, respectively (P = .011). They showed significantly increased axial rotational arc in the CR knees. Dennis et al.(153), in their *in vivo* fluoroscopic study of TKA, reported mean axial rotation of 0.1^{o} (-11.2 to 7.5) and -0.1^{o} (-5.4 to 10.0) during gait in CR and PS-TKA, respectively. Another study reported during step-up/down activities a net tibial internal rotation of 5-7^{o} with no statistical significance between the two implant designs.(157) In our study, the mean axial rotations were -7.5^{o} ±6.2 and -5.9^{o} ±5.5 in the CR and PS configurations, respectively, during gait. During stair descent -7.4^{o} ±6.0 and -6.5^{o} ±4.6 in the CR and PS configurations, respectively. During stair descent -6.8^{o} ±4.4 and -5.9^{o} ±3.8 in the CR and PS configurations, respectively. This is in keeping with clinical findings. The internal rotation of the tibia with flexion we observed in TKA has been demonstrated in other studies.(60, 61)

A recent study compared the contact kinematics in CR- and PS-TKA at a mean follow up of nine years. They suggested that the post-cam mechanism better maintains kinematics and functional in PS-TKA due to the PCL insufficiency observed in CR-TKA.(158) A single-institution review of 8117 TKAs found at a 15-year review the survival was 90% and 77% in CR- and PS-TKA, respectively.(31) Kolisek et al.(134) suggested that between CR and PS TKA neither showed superiority and suggest that the surgeon preference should dictate implant choice. Vertullo et al.(159) reported increased revision risk for the surgeons who preferred PS-TKA was significantly higher for all causes when compared to CR-TKA (HR 1.45 [95% CI, 1.30 to 1.63]; $p < 0.001$), for loosening or lysis (HR 1.93 [95% CI, 1.58 to 2.37]; $p < 0.001$), and for infection (HR 1.51 [95% CI, 1.25 to 1.82]; $p < 0.001$). Overall, there was a 45% higher risk; however, it was limited to male patients within the study. A meta-analysis of randomized controlled trials comparing CR to PS-TKA concluded the clinical outcomes were similar with no difference in the short- and medium-term follow up.(135) This was followed by another systematic review and meta-analysis that found no differences in function or outcomes and concluded not to make a recommendation as to which implant design demonstrated superiority.(160) Another meta-analysis found no difference in knee scores, radiological outcomes, complications and found improved ROM in PS-TKA, but this did not impact the clinical outcomes.(161) However, a recent meta-analysis of clinical trials reported improved ROM in the PS-TKA (2.18^{o}) but with prolonged surgery time (6.87 mins) but overall no statistically significant relevant differences in complications.(136) This shared the findings of an earlier meta-analysis whose statistical significance favoured PS-TKA regarding flexion and ROM.(162) The plethora of meta-analyses in the clinical literature fuel the ongoing debate. Our study elucidates the kinematics between the two designs and offers a further understanding of TKA biomechanics.

The limitations in this study were the use of point-to-point ligaments rather than bundles of ligaments which does not entirely represent the native ligament properties. There is still a significant variation in the literature regarding the representation of ligaments.(149) The computational models are based on approximations and assumptions made to simplify the complexity of the human knee. Another limitation is the loading parameters for the ADL, as these are based on TKA parameters using PS PE.(22) Our model lacked the patellofemoral joint, and therefore its effect on TKA kinematics was excluded in our study. Our study only had a single difference which was the post-cam mechanism and PCL between configurations.

Conclusion

The debate between superiority between CR- and PS-TKA continues. Therefore, surgeons who choose to perform either CR or PS TKA should understand the design indications and limitations and the associated kinematic differences. Overall, during ADL, we found the kinematics between the two implant designs to be similar. However, there is increased stability in PS-TKA during ADL.

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Chapter 4
Posterior-stabilized Total Knee Arthroplasty Kinematics and Joint Laxity

Abstract

Posterior-stabilized (PS)-TKA arose as an alternative to cruciate-retaining (CR)-TKA in the 1970s. Since then, it has become a popular utilized TKA design with comparable outcomes to CR-TKA. The post-cam mechanism is unique to PS-TKA as it substitutes the function of the posterior cruciate ligament (PCL). The study aimed to understand the kinematic and laxity changes in PS-TKA with under- and overstuffing the tibiofemoral joint space with the polyethylene insert.

This study employed a hybrid computational-experimental joint motion simulation on a VIVO 6 degrees of freedom (6-DoF) joint motion simulator (AMTI, Watertown, MA, USA). Physical prototypes of a virtually performed TKA in mechanical alignment (MA) and kinematic alignment (KA) based of cadaveric CT scans and a virtual ligament model were utilised. The reference, understuffed (down 2 mm) and overstuffed (up 2 mm) joint spaces were simulated, and neutral flexion and laxity were testing loads and motions were performed for each configuration.

The PE insert thickness influenced post-cam engagement, which occurred after 60^{o} in the overstuffed configurations, after 60-75^{o} in the reference configurations and after 75^{o} in the understuffed configurations. The understuffed configurations compared to the reference configurations resulted in a mean 2.0^{o} (28%) and 2.0^{o} (31%) increase in the coronal laxity in MA and KA, respectively. The overstuffed compared to the reference configuration resulted in mean joint compressive forces (JCFs) increase by 73N (61%) and 77N (62%) in MA and KA models, respectively.

The under- and overstuffing in PS-TKA alter the kinematics with variable effects. Understuffing decreases stability JCFs and inverse with overstuffing. Subtle changes in the PE insert thickness alters the post-cam mechanics.

Introduction

The primary goal of total knee arthroplasty (TKA) is to provide a painless stable knee and restoration of joint function in patients with advanced osteoarthritis. The knee prosthesis should be performed to provide a maximal range of motion (ROM) while providing stability.(4, 100) There have been numerous advances in the designs utilised in TKA.

Posterior-stabilized (PS)-TKA arose as an alternative to cruciate-retaining (CR)-TKA in the 1970s. Since then, it has become a popular utilized TKA design with comparable outcomes to CR-TKA.(163) The post-cam mechanism *(Figure 4.15)* is unique to PS-TKA as it substitutes the degenerative posterior cruciate ligament (PCL), preventing posterior tibial subluxation, facilitating femoral back and improving ROM.(164-166) Whilst there has been a great success in TKA, there is a subset of patients who remain dissatisfied with their results.(5, 6) In TKA, pain and instability are common indications for revision.(100-102) Review of the literature has shown that post-operative stiffness is a relatively common outcome 4-16%.(167) This highlights the challenge of providing a stable and painless TKA.

Recent advances in TKA include increasing technology in the operating room.(168) However, the judgement as to what polyethylene (PE) insert size to use – what is too tight, too loose, symmetric tightness and soft tissue balancing remain unclear. Manual techniques utilized over and above traditional balancing include spacer blocks, lamina spreaders, tensiometers, and stress tests, are common practice but are subjective.(97) Technology such as intra-compartment force-sensing devices were developed to aid intraoperative surgeon decision making.(169) The early data indicates this technology may improve outcomes in TKA.(170, 171) However, it remains unclear what is the correct answer, as there is no consensus as to what defines the balanced knee.

The advances in joint motion simulator technology have enabled these machines to analyze TKA mechanics while simulating different alignments and soft tissue balancing by changing the properties of parametric virtual ligaments.(15) We hypothesized that either under- or overstuffing with the PE insert would affect knee kinematics. The objective of this study was to determine the differences in kinematics with PE thickness in PS-TKA using a joint motion simulator machine linked to a virtual ligament model.

Figure 4.15: Sagittal cut of Scorpio NRG prosthesis, demonstrating the cam on the femoral component, engaging with the post on the polyethylene insert. Reproduced with permission, Akasaki et al. (172)

Methods

This study employed a hybrid computational-experimental joint motion simulation on a VIVO 6 degrees of freedom (6-DoF) joint motion simulator (AMTI, Watertown, MA, USA). These simulations measure the kinematics of physical implant components in response to applied loads, but with forces imposed due to simulated stretching of surrounding “virtual ligaments”. These simulated one-dimensional point-to-point springs are virtually installed around the implant components, will become tensioned or slack in response to insertion kinematics (insertion coordinates defined relative to the implant components), and the forces they generate will contribute to joints kinematics and stability. The virtual ligaments employed in the current study were designed to replicate the relative insertion coordinates, tension and stiffness of natural ligaments around a mechanically aligned TKA, using the following procedure.

Virtual ligament design

The distal femur and proximal tibia were reconstructed from a CT scan of a single cadaver knee as described in Chapter 3 in neutral extension using 3D Slicer(138) and exported as stereolithography (.stl) files. These 3D surface models were imported into the commercially available CAD software, SolidWorks 2020® (Dassault Systèmes SolidWorks Corporation, Waltham, MA). In SolidWorks, this native knee geometry was used to identify relevant ligament insertions based on established bony landmarks and previous literature.(139-146) The femoral and tibial insertions superficial medial collateral ligament (sMCL) and lateral collateral ligament (LCL) were each identified and would serve as connection points for the single-bundle virtual ligaments. The anterior and posterior cruciate ligaments and deep MCL were not represented as they are routinely released in PS-TKA procedures as part of the soft tissue dissection required for exposure or bony resection. The insertion-to-insertion distance of each ligament was noted. Acknowledging that ligaments may be slack or stretched when in neutral extension, these distances could not be used to define the true resting length (unstretched or "slack" length) of each ligament. The slack length, however, was estimated based on literature-derived intact knee ligament reference strains and the insertion-to-insertion distance in extension.(147-149)

Surface models (.stl files) of a tibial and femoral Triathlon® (Stryker Corp., Mahwah, NJ) TKA components were imported into the CAD software. A size five tibial tray with a 9 mm thick PS PE insert component was used, in combination with a size five femoral component. Virtual TKA was performed with mechanical alignment (MA) and kinematic alignment. In MA, the distal femur bone model to be cut perpendicular to the femur's mechanical axis and for the proximal tibia to be cut perpendicular to the tibia's mechanical axis. The femoral component was aligned with the approximated trans-epicondylar axis; this was determined by externally rotating 3° from the posterior condylar axis. In kinematic alignment (KA), the distal femur resection was made 3°

valgus to the femur's mechanical axis, and the proximal tibial resection was made 3° varus to the tibia's mechanical axis. In addition, the femoral rotation was set to align parallel with the posterior condylar axis. In both alignments, the posterior tibial slope resection was 3°.

The implant components were aligned to the cut surfaces of the femur and tibia. Then the entire tibia (plus tibial component) was translated and rotated such that the femoral and tibial components were neutrally positioned and aligned (in extension, with the femoral condyles dwelling at the deepest point in the PE dishes). Finally, the coordinates of the ligament femoral and tibial insertions were measured with respect to the femoral component, as were their new lengths, which changed after TKA. Along with the ligament stiffness, these data were sufficient to define virtual ligaments around the TKA.

Joint motion simulator

Real physical prototypes of the same implant components were mounted onto the VIVO. The femoral component was mounted to the mounting axle with poly methyl methacrylate cement (Bosworth Fastray; Keystone Industries GmbH, Singen, Germany), and the tibial baseplate component was anchored into the tibial fixture using dental model stone (Modern Materials Golden Denstone Labstone; Modern Materials, Kulzer GmbH, Hanau, Germany). We used polydimethylsiloxane (silicone)-based lubricant (HAS0001-OS, Horizon Fitness, Cottage Grove, WI) as an articulation lubricant and applied it consistently throughout the experiment. The VIVO was used to apply loads and motions representative of ADL, and the resulting kinematics (measured outcome) were sensitive to the implant component geometries, alignments, and virtual ligament properties. This *in vitro* technique of measuring motions and kinematics with simulated virtual ligaments has been previously described.(15)

Joint space

Three different joint spaces were simulated (See Appendix C):

1. Reference joint space: The 9 mm PE insert was utilised. This created our reference joint space relative to our virtual ligament model.
2. Understuffed joint space: The 9 mm PE insert was undersized by 2 mm. This was simulated by the VIVO and resulted in an understuffed joint space.
3. Overstuffed joint space: The 9 mm PE insert was oversized by 2 mm. This was simulated by the VIVO and resulted in an overstuffed joint space.

Input loads and motions

Motions were simulated as follows:

(i) Neutral Flexion Kinematics: With a 10 N compressive force applied parallel to the long axis of the tibia and passing through the centre of the joint, the femur was flexed to 90° and extended back to 0° at a rate of 25 s/cycle. All other degrees of freedom were unconstrained (set to maintain 0 N or 0 Nm of load). Four flexion/extension cycles were simulated; resulting 6-DoFs joint kinematics, net ligament forces and individual ligament tensions were recorded during the third and fourth iteration.

(ii) Posterior Laxity: With a 10 N compressive force applied parallel to the long axis of the tibia and passing through the centre of the joint, the joint was flexed from 0° to 90° in 15° increments. A 100 N posterior-directed force was applied to the tibia at each fixed flexion angle, causing its relative posterior displacement. This posterior displacement was limited by the combined contributions of the concave congruency of the condyles and tensioning of the virtual ligaments. All other degrees of freedom were unconstrained. After testing at each flexion angle, the joint returned to 0°, and the entire process was repeated for a total of four iterations. Data were recorded during the third and fourth iteration. Recorded data included the posterior displacement of the tibia (relative to the corresponding neutral flexion kinematics at the same flexion angle), net ligament forces and individual ligament tensions.

(iii) Varus/Valgus (VV) Laxity: VV laxity testing was accomplished using a similar technique described for (ii), but by applying a 10 Nm varus or valgus joint torque instead of a 100 N posterior force. Recorded data included the varus/valgus angulation (relative to the corresponding neutral flexion kinematics at the same flexion angle), net ligament forces and individual ligament tensions.

Data analysis and statistics

Recorded data were smoothed using a low-pass Butterworth filter followed by a spline interpolation function in Matlab (The MathWorks, Natick, MA), and then down-sampled to only include data at 15° intervals of flexion and only during the flexion phase of the complete flexion/extension motion in the laxity testing. During the ADL testing, the joint motion was sampled throughout the cycle. We extracted the anteroposterior (AP), internal/external rotation (IE) and varus/valgus (VV) kinematic data in each of the 6-DoFs. We also collected posterior, varus and valgus motion limits in each of the 6-DoFs at these limits. The smoothed and processed data were used for statistical analysis. Descriptive statistics were used to compare the results between each joint space configuration and alignment. All statistical analyses were completed in Microsoft® Excel (v.16.45).

RESULTS

Neutral flexion

The PE insert thickness influenced the ligament tension during neutral flexion *(Figure 4.16).* The means values across the entire neutral flexion arc (0-90^{o}) for the 2mm increase (overstuffed) or decrease (understuffed) in PE insert thickness the sMCL tension increased or decreased by a mean 32N (105%) and 36N (85%) in MA and KA respectively. For the 2mm increase (overstuffed) or decrease (understuffed) PE insert thickness, the LCL tension increased or decreased by a mean of 42N (47%) and 41N (50%) in MA and KA, respectively. The adduction moment observed in the configurations resulted in the LCL tension being higher than sMCL tension.

The neutral flexion AP translation kinematics are shown in *Figure 4.17-A.* The PE insert thickness influenced when the post engaged with the cam, which occurred after 60^o in the overstuffed configurations, after 60-75^o in the reference configurations and after 75^o in the understuffed configurations. In the overstuffed configurations, there were no differences between KA and MA. MA caused contact locations that were initially anterior to the KA configurations in the reference and understuffed configurations, but at the 90^o flexion, they were all at the same position.

The axial rotation kinematics are shown in *Figure 4.17-B.* At full extension, all configurations were internally rotated relative to their position at 90^o. Increasing the PE insert thickness increased the internal rotation for each alignment. However, in flexion of 75-90^o, all those in MA had decreased internal rotation compared to those in KA regardless of the PE insert thickness. During 15-75^o of neutral flexion, the two alignments yielded different waveforms.

The coronal kinematics are shown in *Figure 4.17-C.* In the first 30^o of knee flexion, all the configurations were in a varus alignment, but by 30^o of knee flexion, all demonstrate a neutral coronal alignment. This degree of initial varus was influenced by the PE insert thickness and associated ligament tension. Increasing PE insert thickness decreased initial varus as the sMCL tension increased more than the LCL.

Laxity testing

The posterior laxity is shown in *Figure 4.18-A.* The posterior laxity is most significant at full extension and decreases as the knee flexion increases. After the post-cam mechanism engages, the observed posterior laxity becomes negligible due to the mechanical block of the post on the cam. In addition, the PE insert thickness influences the laxity; a 2 mm change of PE insert in the configurations resulted in a mean laxity difference of 0.7 mm (14%) and 0.6 mm (12%) in MA and KA respectively. This difference in laxity increased with understuffing and decreased with overstuffing of the joint space. This corresponds to previously observed ligament tension changes with PE insert thickness during neutral flexion *(Figure 4.16).*

The VV laxity testing results are shown in *Figure 4.18-B*. The coronal constraint is positively influenced by the PE thickness, with increased thickness demonstrating less coronal laxity. The understuffed configurations compared to the reference configurations resulted in a mean 2.0° (28%) and 2.0° (31%) reduction in the coronal laxity in MA and KA, respectively. The least coronal laxity is observed at 60° of knee flexion, which coincides with peak sMCL tensions, and the LCL tension is at its peak at 75° of knee flexion.

Joint compressive forces

The joint compressive forces (JCFs) results are shown in *Figure 4.19*. The under- and overstuffed configurations influenced the JCFs. The overstuffed compared to the reference configuration resulted in a mean JCFs increase by 73N (61%) and 77N (62%) in MA and KA, respectively. The inverse was yielded with the understuffed configuration. The peak forces are recorded from 60-75° of knee flexion peaking at 75°. The peak forces in the overstuffed configurations were 329N and 340N flexion in MA and KA, respectively. The peak forces in the understuffed configurations were 152N and 155N in MA and KA, respectively. The peak forces in the reference configurations were 235N and 245N in MA and KA, respectively. This coincidence with the peak collateral ligament tension observed *(Figure 4.16)*, the engagement of the post-cam mechanism *(Figure 4.17-A)* and the least coronal laxity *(Figure 4.18-B)*. This illustrates the positive relationship between the ligament tension, stability and JCFs.

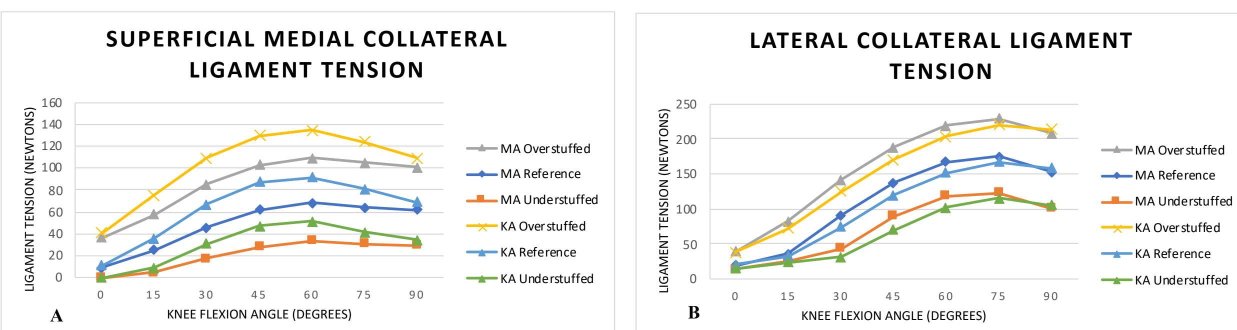

Figure 4.16: Graphs demonstrating the relationship between ligament tension and polyethylene insert thickness. A- Superficial medial collateral ligament, B- Lateral collateral ligament

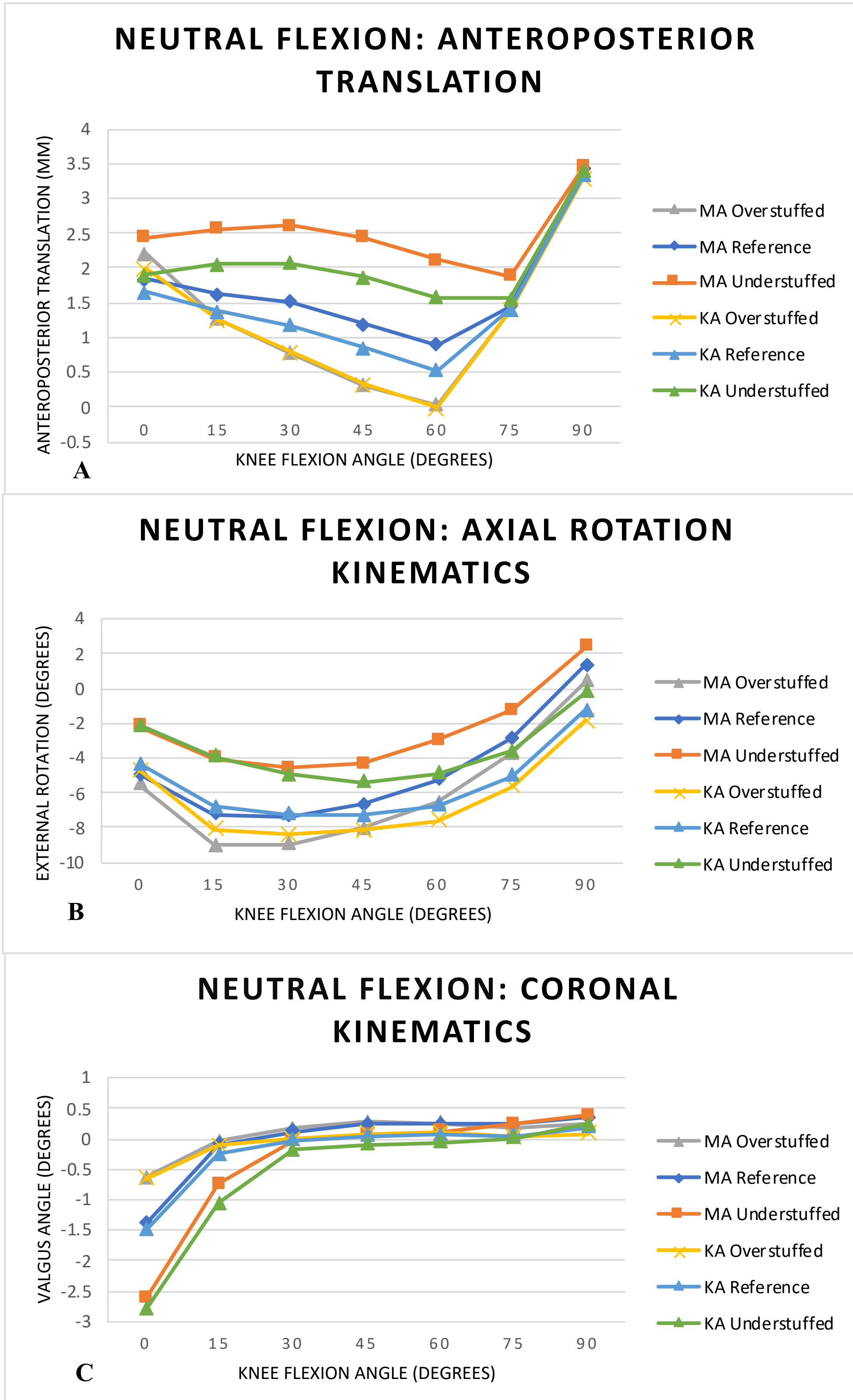

Figure 4.17: Graphs demonstrating the neutral flexion kinematics. A- Anteroposterior, B- Axial Rotation, C- Varus/valgus.

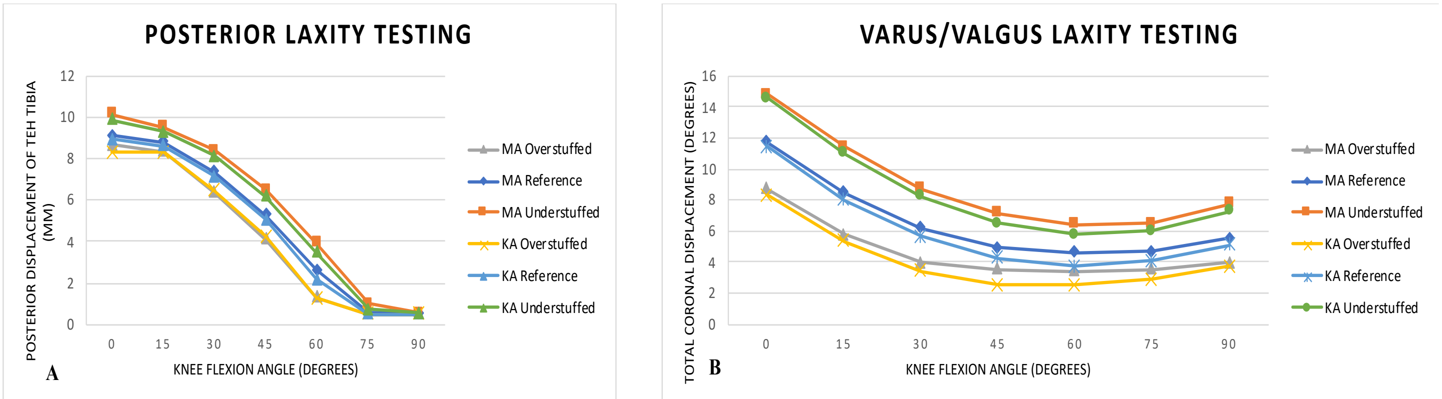

Figure 4.18: Graphs illustrating the joint laxity test results. A- Posterior, B- Varus/valgus

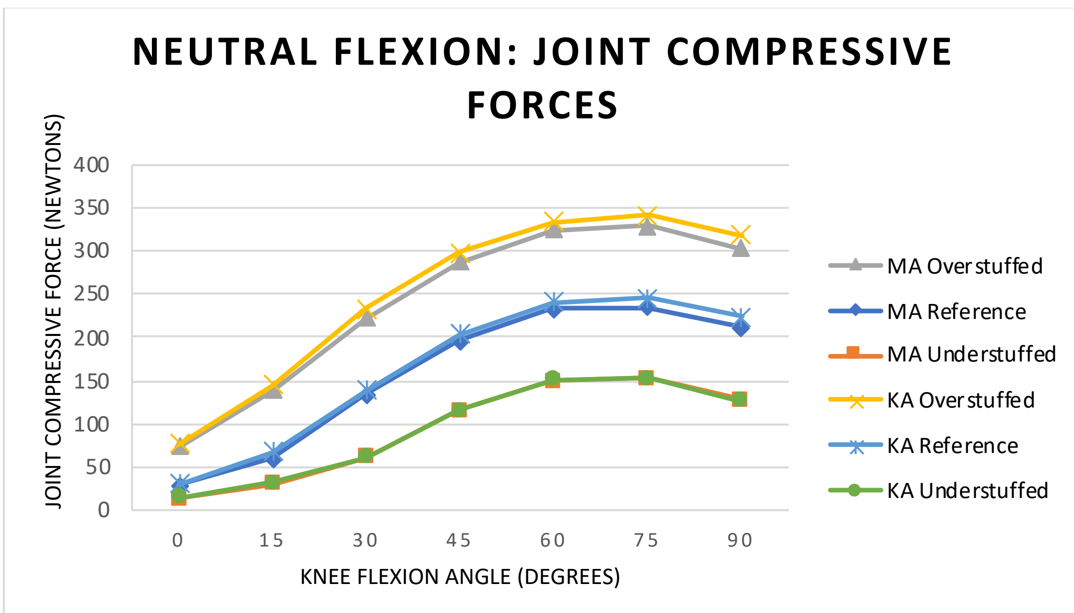

Figure 4.19: Graph illustrating the joint compressive forces during neutral flexion.

Discussion

The goal of our study was to describe the effect of under- and overstuffing the joint space in PS-TKA on knee kinematics, joint forces, and laxity. Using MA and KA, we measured and recorded data during neutral flexion and laxity testing. To our knowledge, this is the first study to evaluate the effects of under- and overstuffing the tibiofemoral joint in both KA and MA in relation to neutral flexion, laxity, JCFs, and post-cam engagement in PS-TKA.

In the PS-TKA, the function of the PCL is substituted by the post-cam mechanism. The post-cam mechanism engages at around 75° of flexion. This engagement prevents anterior femoral translation of the femur and facilitates the femoral rollback in PS-TKA.(40, 41) PS-TKA has been shown to increase the sagittal constraint of the implanted knee.(42) Hamai et al.(154) evaluated cruciate-retaining (CR) and PS-TKA in vivo kinematics using radiographic-based image-matching techniques. They found the post-cam mechanism did not engage during stair climbing, but this was due to the dynamic flexion angle in their study being less than 75°. They found the CR-TKA demonstrated more sagittal stability. Several studies have evaluated the post-cam mechanism under the following areas of interest; geometry, wear, contact forces, kinematics, and design.(172-177) Our study suggests that increasing the PE insert thickness may infer greater sagittal stability during activities of daily living, such a stair climbing. This is based on our observation that increased PE insert thickness and overstuff the joint space resulted in the post-cam mechanism engaging earlier after 60° of flexion demonstrated in our neutral flexion AP translation kinematics in the overstuffed configurations *4.17-A*. The ligament tension increases likely to drive the observations in our study yielded in the overstuffed configurations.

We observed a positive relationship between increasing PE insert thickness (overstuffing the joint space) and ligament tension *Figure 4.16*. Whilst observing increasing PE insert thickness occurs with decreased joint laxity *Figure 4.18*. Shimizu et al.(178) investigated the effects of weight-bearing on PS-TKA kinematics. They reported post-cam engagement at 93.4° ± 3.3° and 70.5° ± 7.2° in weight-bearing and non-weight-bearing, respectively. In loading the joint, the ligaments become lax resulting in later engagement of the post-cam. This phenomenon of later post-cam engagement with laxity was observed in our understuffed configurations *Figure 4.17-A* and *Figure*

4.18. It has been suggested that delayed post-cam engagement can facilitate increased maximum knee flexion.(179, 180) Suggs et al.(180) showed a correlation between the initial post-cam contact angle and the maximum flexion angle r = 0.505 (p = 0.019). Arnout et al.(181) demonstrated in their *in vitro* study using a dynamic knee kinematic simulator that earlier post-cam engagement facilitated more physiological motion. The post-cam mechanism determines the posterior femoral translation and facilities movement in deeper flexion shown in a computation model study.(182) We observed that understuffing the joint space results in increased joint laxity with delayed post-cam engagement. Overall, this offers a biomechanical explanation of observations in the clinical literature that reports increased ROM with increased joint laxity.(90)

There are studies that have shown that early post-cam engagement can lead to increased contact stress and accelerate post wear.(183, 184) However, these effects are further highlighted in our posterior laxity results testing shown in *Figure 4.18-A*. This is driven by the related JCFs. As the post-cam mechanism engages, the observed posterior laxity becomes negligible due to the mechanical block of the post on the cam. The posterior laxity yielded in our study at full extension when the native is most stable is likely due to the lack of secondary knee stabilisers in our virtual model. A radio stereometric analysis study demonstrated that kinematics in PS-TKA could impact tibial component migration through alterations in force transmission.(185) The kinematic differences we were observed were coupled with changes in the JCFs. These subtle variations in JCFs may impact long-term implant survivorship. Further studies are required to evaluate how the JCFs change we observed translate into contact stress patterns.

A manufacturer indicated their PS-TKA designs allow up to ± 12° of axial rotation.(33) Cates et al.(55) evaluated the *in vivo* kinematics using fluoroscopy and reported in PS-TKA that from full extension to 90^{o} flexion, internal rotation of the tibia was observed. The axial rotation means angles at full extension were -1.0^{o} in PS-TKA, and at maximum flexion, the mean axial rotation angle was 1.9° for PS-TKA. Tamaki et al.(186) evaluated the *in vivo* kinematics of 20 TKA reported the femoral component demonstrated a mean of 13.5^{o} (5.2^{o} to 21^{o}) external rotation of increasing external rotation with flexion. We observed similar axial rotation, i.e., internal tibial rotation in the first 30^{o} and 45^{o} in MA and KA configurations, respectively. For native knee kinematics, in the initial 30^{o} of flexion, the tibial internally rotate after that it remains within 1^{o} of that position while

PS-TKA typically continues to internally rotation.(187) Our configurations demonstrated reduced internal rotation with continued flexion. This observation isn't abnormal for the implanted knee, as Suggs et al.(180) reported that there is a reduction in internal tibial rotation with post-cam engagement. The coronal laxity decreased in our study with overstuffing the joint. This can be explained by the increased ligament tension, which increases the stability of the configurations.

Our study findings suggest that TKA alignment and not PE insert thickness many play a greater role in axial rotation in PS-TKA with greater internal rotation in KA compared to MA irrespective of the PE insert thickness *Figure 4.17-B*. The internal rotation of the tibia with flexion is related to the screw-home mechanism observed in the knee. The greater internal rotation observed in KA is in keeping with literature that supports KA as being more physiological.(188, 189) Another *in silico* study demonstrated more physiological knee kinematics with KA.(190) Amongst the two alignments, KA allows a greater contribution of the soft tissues to balance and stabilize the knee. The increased contribution of the soft tissues is evidenced by the increased JCF and soft tissue tension proportionately in KA compared to MA. Our study yielded higher JCFs with KA (*Figure 4.19*). High contact pressures have been reported in KA, resulting in increased PE wear, but they have not been shown to cause PE failure.(190) Our study demonstrated minor differences between the alignments with regards to the joint kinematics during neutral flexion, laxity testing in conjunction with under- and overstuffing the tibiofemoral joint. This is representative of the current literature regarding alignment, which remains controversial.(84, 188, 189, 191)

The limitations in this study were the use of point-to-point ligaments rather than bundles of ligaments which does not entirely represent the native ligament properties. There is still a significant variation in the literature regarding the representation of ligaments.(149) The computational models are based on approximations and assumptions made to simplify the complexity of the human knee.(22) Our model lacked the patellofemoral joint, and therefore its effect on TKA kinematics was excluded in our study.

Conclusion

Surgeons who utilise PS-TKA need to be aware that PE size changes have a variable effect on the kinematics of the implanted knee. These effects are in conjunction with other features such as alignment and the post-cam mechanism. This body of work contributes to the understanding of TKA kinematics in PS-TKA. This knowledge can aid the surgeon in intraoperative decision making to individualize patient treatment. Improvement in judgement can improve patient satisfaction and mitigate against pain, instability and subsequent revisions while optimizing kinematics.

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Chapter 5

Instability in Cruciate Retaining Total Knee Arthroplasty

Abstract

Approximately one-third of early TKA failures are due to soft tissue imbalances resulting in instability. Flexion and mid-flexion instability are used to describe instability in a flexed implanted knee. This instability in flexion is multifactorial and currently a poorly understood clinical phenomenon. During activities of daily living (ADL), the knee is maintained in a flexed position potentiating the possible instability. Therefore, we investigated the contribution towards the instability of under- and or overstuffing the tibiofemoral joint space with the polyethylene (PE) insert.

This study employed a hybrid computational-experimental joint motion simulation on a VIVO 6 degrees of freedom joint motion simulator (AMTI, Watertown, MA, USA). Physical prototypes of a virtually performed TKA in mechanical alignment based on cadaveric CT scans and a virtual ligament model were utilised. The reference, understuffed (downsized 2 mm) and overstuffed (upsized 2 mm) joint spaces were simulated, and ADL loads and motions were performed for each configuration.

Our results demonstrated differences in the TKA kinematics within the configurations. However, no overt instability was demonstrated. During the gait cycle (10-35%), the mean AP translations were -4.3 mm ±1.1, -5.2 mm ±0.8, -5.8 mm ±0.6. During stair descent (85-100%), the mean AP translations were 11.8 mm ±1.7, 10.8 mm ±2.0 and 9.8 mm ±2.3. During stair ascent (40-60%), the mean axial rotations were -3.4° ±1.8), -5.6° ±1.9 and -7.3° ±1.9. This was in the understuffed, reference and overstuffed configurations, respectively.

In implant modularity, changes in the PE insert sizing affect the TKA kinematics, soft tissue laxity, and ligament tension during ADL. This hybrid biomechanical study highlights that understuffing

the joint with a 2 mm smaller PE insert in isolation is unlikely to create instability with flexion during gait, stair ascent and descent.

Introduction

The aim of total knee arthroplasty (TKA) is in patients with advanced knee osteoarthritis is to provide a painless stable knee and restore joint functionality.(2, 3) The implanted knee should provide a maximal range of motion (ROM) and stability for the patient to perform their required activities of daily living (ADL).(4) Whilst there has been a great success in TKA, there is a subset of patients who remain dissatisfied with their results.(5, 6) Proper ligament balancing and stability are requirements for achieving good functional outcomes and long-term implant survival.(101) Patients with residual pain are dissatisfied, have a lower quality of life and higher health care resource burdens.(8)

Approximately one-third of early TKA failures are due to soft tissue imbalance resulting in instability requiring subsequent revision.(12, 13, 100-102) Intraoperatively TKA stability is manually assessed at full extension and 90° of flexion but not routinely in the mid-flexion ranges. Flexion and mid-flexion instability are controversially used interchangeably to describe instability in a flexed knee.(103-106) This instability in flexion is multifactorial and is currently a poorly understood phenomenon.(105) During ADL, the knee is maintained in a flexed position.(107) Patients with instability present with vague complaints with walking and stair activity.(104)

The advances in joint motion simulator technology have enabled these machines to analyze TKA mechanics while simulating different joint spaces and soft tissue balancing by changing the properties of parametric virtual ligaments.(15) A recent systematic review and meta-analysis found literature investigating 14 possible causes of mid-flexion instability, none of which evaluated if under- or overstuffing the tibiofemoral contributed to this phenomenon.(113) It remains unclear the effects of under- or overstuffing the tibiofemoral joint and its contribution towards instability in TKA during ADL.

The objective of this study was to evaluate and identify instability in cruciate-retaining (CR) TKA during ADL using; a reference, understuffed and overstuffed joint spaces. We hypothesized that understuffing the joint space would affect TKA kinematics and cause instability.

Methods

This study employed a hybrid computational-experimental joint motion simulation on a VIVO 6 degrees of freedom (6-DoF) joint motion simulator (AMTI, Watertown, MA, USA). These simulations measure the kinematics of physical implant components in response to applied loads, but with forces imposed as a result of simulated stretching of surrounding "virtual ligaments". These simulated one-dimensional point-to-point springs are virtually installed around the implant components, will become tensioned or slack in response to insertion kinematics (insertion coordinates defined relative to the implant components), and the forces they generate will contribute to joints kinematics and stability. The virtual ligaments employed in the current study were designed to replicate the relative insertion coordinates, tensioning, and stiffness of actual ligaments around a mechanically aligned TKA, using the following procedure.

Virtual ligament design

The distal femur and proximal tibia were reconstructed from a CT scan of a single cadaver knee as described in Chapter 3 in neutral extension using 3D Slicer (138) and exported as stereolithography (.stl) files. These 3D surface models were imported into the commercially available CAD software, SolidWorks 2020® (Dassault Systèmes SolidWorks Corporation, Waltham, MA). In SolidWorks, this native knee geometry was used to identify relevant ligament insertions based on established bony landmarks and previous literature.(139-146) The femoral and tibial insertions of the posterior cruciate ligament (PCL), superficial medial collateral ligament (sMCL), and lateral collateral ligament (LCL) were each identified and would serve as connection points for the single-bundle virtual ligaments. The anterior cruciate ligament and deep MCL were not represented as they are routinely released in TKA procedures as part of the soft tissue dissection required for exposure of bony resection. The insertion-to-insertion distance of each ligament was noted. Acknowledging that ligaments may be slack or stretched when in neutral extension, these distances could not be used to define the correct resting length (unstretched or "slack" length) of each ligament. However, the slack length was estimated based on literature-derived intact knee ligament reference strains and the insertion-to-insertion distance in extension.(147-149)

Surface models (.stl files) of a tibial and femoral Triathlon® (Stryker Corp., Mahwah, NJ) TKA components were imported into the CAD software. A size five tibial tray with a nine mm thick CR polyethylene (PE) insert component was used, in combination with a size five femoral component. Virtual TKA was performed with mechanical alignment (MA), which required the distal femur bone model to be cut perpendicular to the femur's mechanical axis, and for the proximal tibia to be cut perpendicular to the tibia's mechanical axis. The femoral component was aligned with the approximated trans-epicondylar axis; this was determined by externally rotating 3° from the posterior condylar axis. The posterior sloped tibial resection was performed at 5°. The implant components were aligned to the cut surfaces of the femur and tibia, and then the entire tibia (plus tibial component) was translated and rotated such that the femoral and tibial components were neutrally positioned and aligned (in extension, with the femoral condyles dwelling at the deepest point in the PE dishes). The coordinates of the ligament femoral and tibial insertions were measured with respect to the femoral component, as were their new lengths, which changed after TKA. Along with the ligament stiffness, these data were sufficient to define virtual ligaments around the TKA.

Joint motion simulator

Real physical prototypes of the same implant components were mounted onto the VIVO. The femoral component was mounted to the mounting axle with polymethylmethacrylate cement (Bosworth Fastray; Keystone Industries GmbH, Singen, Germany), and the tibial baseplate component was anchored into the tibial fixture using dental model stone (Modern Materials Golden Denstone Labstone; Modern Materials, Kulzer GmbH, Hanau, Germany). We used polydimethylsiloxane (silicone)-based lubricant (HAS0001-OS, Horizon Fitness, Cottage Grove, WI) as an articulation lubricant and applied it consistently throughout the experiment. The VIVO was used to apply loads and motions representative of ADL, and the resulting kinematics (measured outcome) were sensitive to the implant component geometries, alignments, and virtual ligament properties. This *in vitro* technique of measuring motions and kinematics with simulated virtual ligaments has been previously described.(15)

Joint space

Three different joint spaces were simulated (See Appendix C):

(iv) Reference joint space: The 9 mm PE insert was utilised. This created our reference joint space relative to our virtual ligament model.

(v) Understuffed joint space: The 9 mm PE insert was undersized by 2 mm. This was simulated by the VIVO and resulted in an understuffed joint space.

(vi) Overstuffed joint space: The 9 mm PE insert was oversized by 2 mm. This was simulated by the VIVO and resulted in an overstuffed joint space.

Input loads and motions

Motions were simulated as follows for each joint space:

(i) Activities of Daily Living: Simulated ADL was performed as in previous studies(22) by applying loads and motions based on data measured using load-sensing TKA implants.(150) walking, stair ascent and descent were simulated. The load cycle begins at the flat foot for the walking files and goes through the gait cycle (flat foot, heel off, toe-off, swing phase, heel strike, flat foot, etc.). This splits the gait cycle into the first 60% stance and the last 40% swing phase. The stair ascent and descent load cycles both begin and end with the middle of swing phase. This splits the stair ascent and decent into the first and last 20% swing phase and the middle 60% as stance phase. The walking, stair ascent and descent loads, and motions based on the AVG75 dataset from the Orthoload website database were used (https://orthoload.com/).(150) All degrees of freedom were operated in force control, except flexion angle, which was prescribed. The ADL was simulated for each joint space configuration. Each activity was simulated for four cycles, with data recorded during the third and fourth cycles, including 6-DoF kinematics.

Data analysis and statistics

Recorded data were smoothed using a low-pass Butterworth filter followed by a spline interpolation function in Matlab (The MathWorks, Natick, MA), and then down-sampled to only include data at 15° intervals of flexion and only during the flexion phase of the complete flexion/extension motion in the laxity testing. During the ADL testing, the joint motion was sampled throughout the cycle. We extracted the anteroposterior (AP), internal/external rotation (IE) and varus/valgus (VV) kinematic data in each of the 6-DoFs. We also collected posterior, varus and valgus motion limits in each of the 6-DoFs at these limits. The smoothed and processed data were used for statistical analysis. Descriptive statistics were used to compare the results between each joint space configuration. All statistical analyses were completed in Microsoft® Excel (v.16.45).

Results

We identified regions in the resultant kinematic waveforms results where the under- and overstuffed configurations deviated from the reference configuration. These areas of differences were quantified, and the knee flexion ranges were reported. The net ligament forces were increased in the overstuffed configuration and decreased in the understuffed configuration when compared to the reference configuration.

Gait

The gait AP, axial and coronal kinematics are shown in *(Figure 5.20).* In the AP kinematics (*Figure 5.20-A*) during the stance phase (10-35% gait cycle), there was an increased mean translation in the understuffed configuration -5.8 mm ±0.6 and decreased mean translation in the overstuffed configuration -4.3 mm ±1.1 compared to the reference configuration -5.2 mm ±0.8. During this period, the knee was flexed between 13-22^o. During the stance phase (10-30% gait cycle), the axial kinematics (*Figure 5.20-B)* demonstrated a mean rotation of 2.9^o ±0.9 in the understuffed configuration -0.2^o ±1.6 in the overstuffed configuration compared to the reference configuration 1.4^o ±1.1. During this period, the knee flexion was 15-22^o. During the swing phase, there was rotation instability observed in the understuffed configuration. This is demonstrated by the peak in the understuffed waveform (*Figure 5.20-B*), as the knee was extending from 61^o to 10^o of flexion. In the coronal kinematics (*Figure 5.20-C*), in the stance phase, there was more varus in the understuffed configuration and less varus in the overstuffed configuration compared to the reference configuration. In addition, there was less valgus in the understuffed configuration in the swing phase and more valgus in the overstuffed configuration compared to the reference configuration. However, these observed differences were within less than 0.5^o of each other, respectively.

Stair descent

The stair descent AP, axial and coronal kinematics are shown in (*Figure 5.21*). In the AP kinematics (*Figure 5.21-A*), the three configurations behaved similarly during the stance phase. During the swing phase (85-100%), the mean AP translations were 11.8 mm ±1.7, 10.8 mm ±2.0 and 9.8 mm ±2.3 in the understuffed, reference and overstuffed configurations, respectively. The knee flexion angle during this period was 48-98^o. The axial kinematics (*Figure 5.21-B*) during 0-50% yielded a spread of the waveforms of the configurations. The knee flexion range was 13-33^o. The mean axial rotations 0-50% were -5.1^o ±3.1, -7.3^o ±3.1 and -9.0^o ±2.7 in the understuffed, reference and overstuffed configurations, respectively. In the coronal kinematics (*Figure 5.21-C*), there was more varus in the understuffed configuration and less varus in the overstuffed configuration compared to the reference configuration. However, these observed differences were within less than 0.5^o of each other, respectively.

Stair ascent

The stair ascent AP, axial and coronal kinematics are shown in (*Figure 5.22*). The AP kinematics (*Figure 5.22-A*), during swing phase 90-15% yielded mean AP translations were 11.4 mm ±1.7, 9.9 mm ±1.8 and 9.7 mm ±1.4 in the understuffed, reference and overstuffed configurations respectively, the knee was flexed between 63-93^o. The IE kinematics (*Figure 5.22-B*), during 40-60% of the cycle, the mean axial rotations were -3.4^o ±1.8, -5.6^o ±1.9 and -7.3^o ±1.9 in the understuffed, reference and overstuffed configurations respectively, the knee flexion was 21-40^o. In the VV kinematics (*Figure 5.22-C*), there was more varus in the understuffed configuration and less varus in the overstuffed configuration than the reference configuration. However, these observed differences were within less than 0.5^o of each other, respectively.

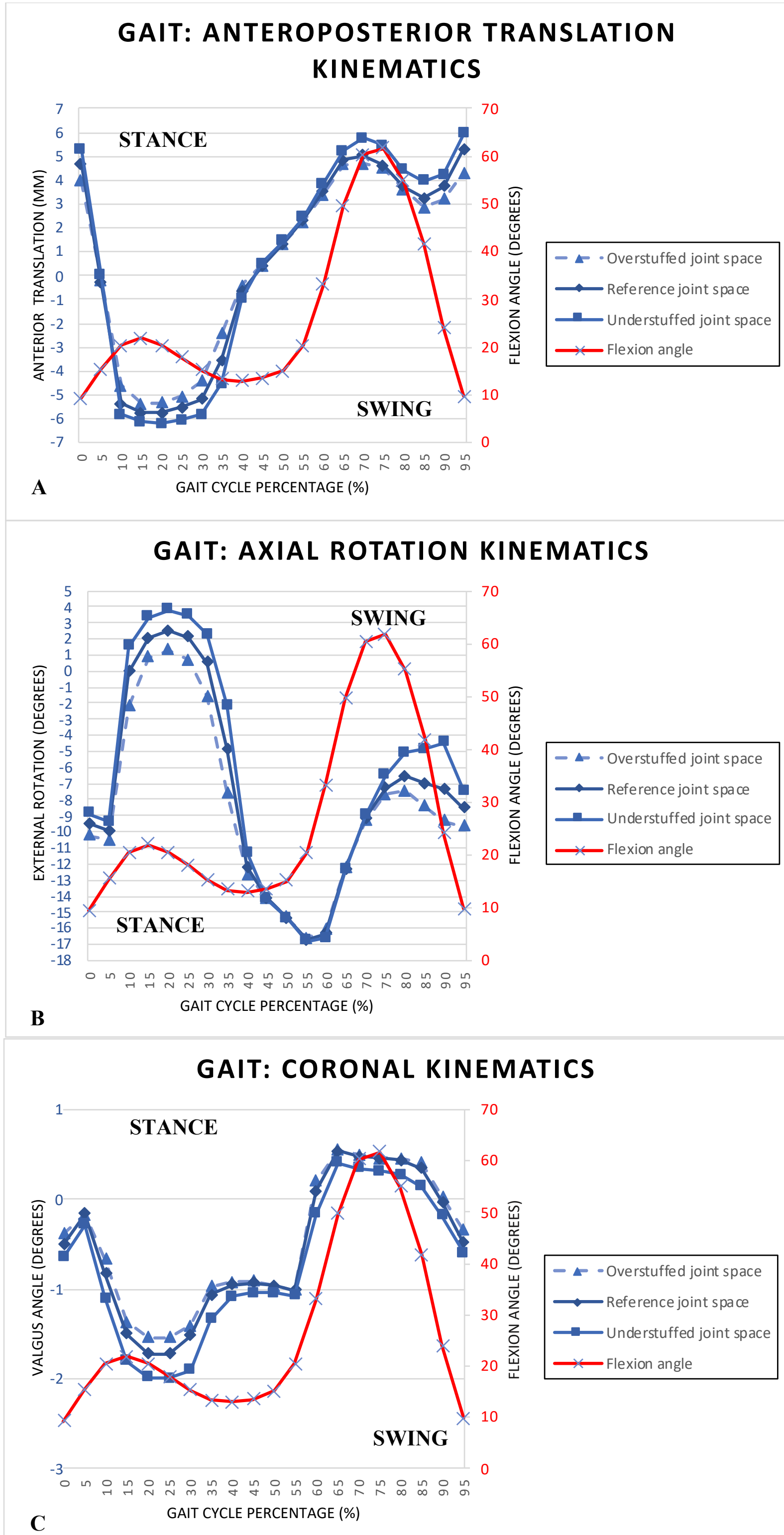

Figure 5.20: Graphs representing the knee kinematics during walking. A- Anteroposterior kinematics. B- Axial rotation kinematics. C- Coronal kinematics.

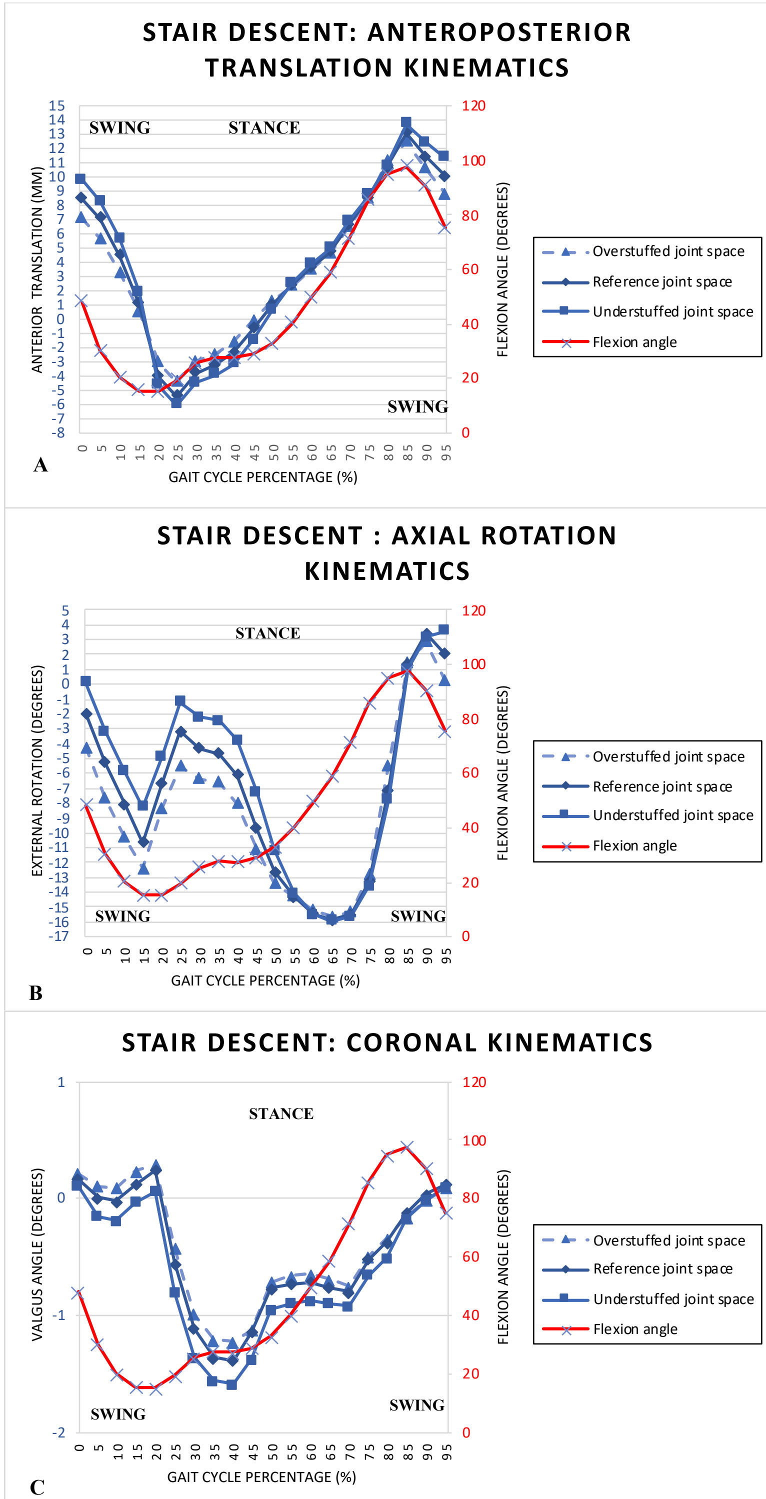

Figure 5.21: Graphs representing the knee kinematics during stair descent. A- Anteroposterior kinematics. B- Axial rotation kinematics. C- Coronal kinematics.

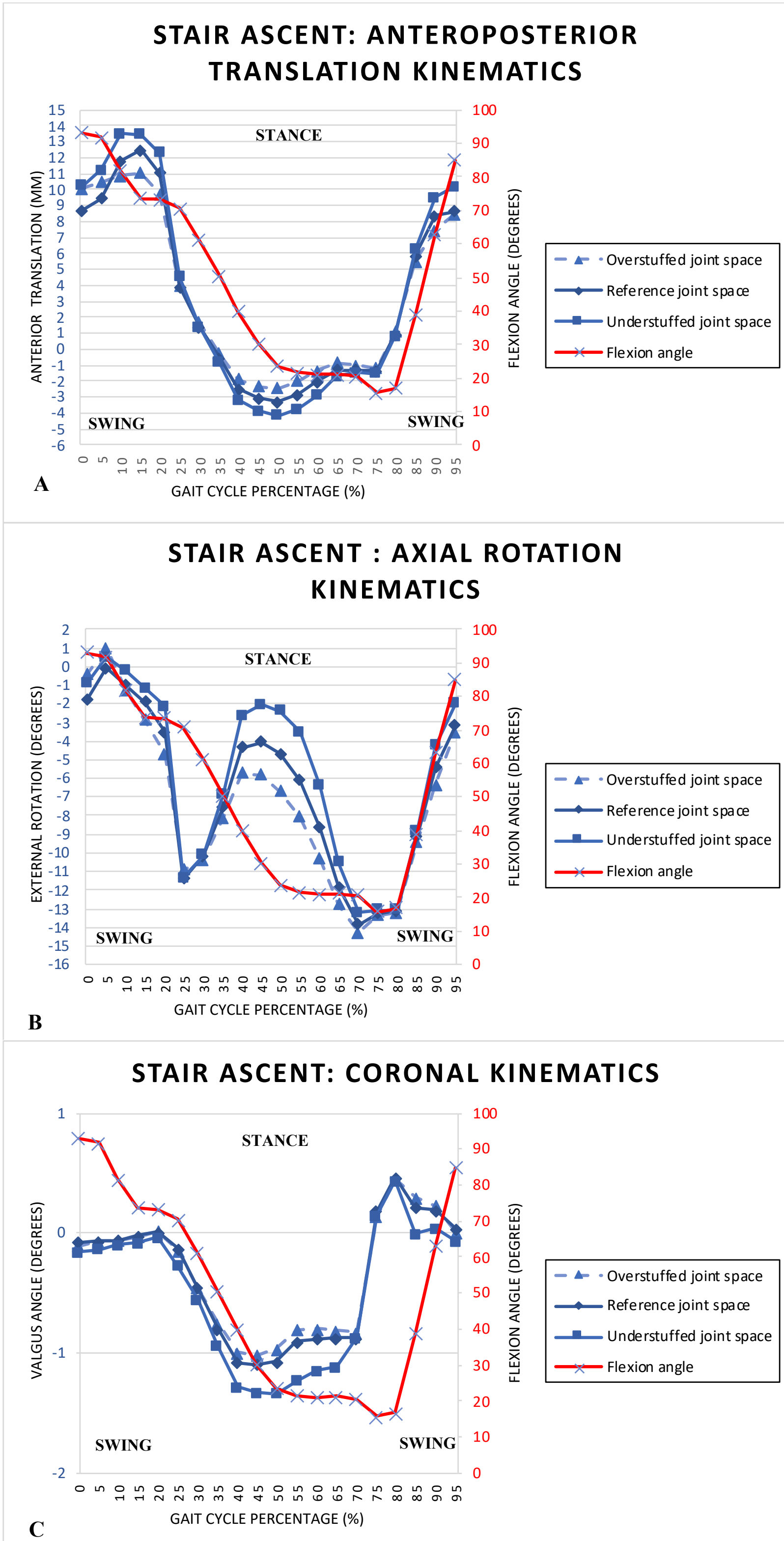

Figure 5.22: Graphs representing the knee kinematics during stair ascent. A- Anteroposterior kinematics. B- Axial rotation kinematics. C- Coronal kinematics.

Discussion

This study aimed to evaluate CR TKA for instability during ADL using; a reference, understuffed and overstuffed joint spaces. To our knowledge, this is the first study to evaluate and report on the effects of under- and overstuffing to elucidate instability during ADL.

The definitions of flexion and mid flexion instability are often used interchangeably in the literature.(103-106, 108) The classical "mid-flexion" instability in TKA literature refers to coronal and AP instability.(109) This has been reproduced in both cadaveric and computer model studies with the joint line elevation.(103, 110) Martin and Whiteside (109) performed their landmark study using ten cadaveric knees. The knees were mounted onto a machine, a 45N vertical load and 10 Nm moment were applied with testing at 15^o intervals of flexion up to 90^o. In five, they shifted the femoral component proximal and anterior by 5 mm. In the remaining five, they shifted the femoral component distal and posterior by 5 mm. They assessed the coronal and AP displacement in mm at the respective flexion angles. They reported increased coronal and AP displacements in the proximal and anterior group compared to control group. They reported the opposite in the distal and posterior group with decreased coronal and AP displacements.

Instead of manipulating the femoral component position, our study performed adjusts of the "joint space" simulating PE insert sizes, which is a routine intraoperative decision made by orthopaedic surgeons who perform TKA. Our study yielded similar results. During the simulated ADL, the understuffed configuration yielded increased AP, axial and coronal displacement compared to the reference configuration. The overstuffed configuration yielded decreased AP, axial and coronal displacement compared to the reference configuration. This stiffing seen in our overstuffed configuration was also reported by Martin and Whiteside.(109) In our study, we did not observe any coronal instability. The coronal kinematics in the understuffed configuration were within less than 0.5^o of the reference configuration. The magnitude of which is unlikely to have any clinical significance. This is likely a result of the virtual ligament in our study.

A systematic review indicated that mid-flexion instability had been reported not only coronally but AP and rotationally.(113) During the gait cycle, our understuffed configuration demonstrated rotatory instability in the swing phase as the knee decelerated while extending from 61^o to 10^o of flexion *(Figure 5.20-B*). This was found during the second half of the swing phase as the flexion angle rapidly decreased. Rotatory instability has been reported in sacrificing both cruciate ligaments.(64) We observed the rotatory instability during the swing phase in gait and not reproduced in stair ascent and descent configurations. In the clinical setting, patients can compensate for instability while walking on a level surface by stiffing the knee. Pes anserine bursitis causing medial side tenderness after TKA with ADL, particularly stair ascent, has been reported.(192) Our results may suggest this pain is a result of hamstring contraction in an attempt to stabilise rotatory instability. However, the rotation instability is likely linked to our virtual ligament and the lack of secondary stabilisers.

Flexion instability was first described as an extension and flexion gap mismatch. The knee is stable at full extension but lax at 90^o of flexion.(111) Patients report instability, particularly when bearing weight on a flexed knee. The laxity allows the femur to glide on the tibia, which is perceived as instability by the patient and results in recurrent effusions and pain. A computational analysis assessing instability was performed by taking a balanced TKA and simulating joint line elevation, implant sizing and translating the femoral component anteriorly.(103) They assessed the ligament load between 15-75^o of knee flexion. They found in CR TKA that the overall tibiofemoral ligament load was reduced significantly with femoral proximalization and anterior femur translation ($P < .001$). This was not observed with implant size changing ($P > .6$). The simulation yielded no changes in the PCL load ($P > .9$). Another computational model study assessed joint line elevation at 5 and 10 mm while measuring the ligament tension with an ultra-congruent PE. They concluded that elevation of the joint line was insufficient to be the main contribution of coronal flexion instability.(193) We found similar results in our study. We did not observe any coronal instability. The axial kinematics in the under- and overstuffed configuration were within less than 0.5^o of the reference configuration. The magnitude of which is unlikely to have any clinical significance. However, our simulation affected the extension and flexion gap by understuffing the joint space by reducing the "joint space" with the PE insert size. We observed a decrease of our net ligament tension in our understuffed configuration during ADL causing relative laxity. Minoda et al.(194) evaluated 30 varus osteoarthritic knee and performed posterior-stabilized TKA, and intraoperatively measured the joint gap using a tensor device with two different trial femoral components. The first component and had a 9 mm distal and posterior thickness, and the second had a 7 mm distal and posterior thickness. The differences in the joint gap between 0-90^{o} of flexion were not statistically significant. They concluded this elevation of the joint line was not associated with mid-flexion instability. This study however didn't consider the TKA kinematics during physiological weight bearing with ADL where instability is commonly reported.

Laskin et al.(88) coined the term "stuffing" of a joint which results in decreased ROM related due to the inappropriate sizing of implants (femur and tibia). In the clinical literature, under stuffing the joint with decreased PE thickness, decreased soft tissue tension increases ROM, but the joint stability decreases.(90) Achieving the optimal soft tissue tension is linked to adequate kinematics in TKA.(85) The under-sizing of the PE Insert is a recognised cause of instability due to the soft tissue laxity.(92) In TKA, soft tissue laxity can result in not only instability but also pain.(85) Overstuffing of the tibiofemoral joint space can result in limitation in knee ROM. (88) In addition to influencing the ROM and the overall functioning in TKA can be affected.(90, 91) However, our simulation showed increased AP, axial and coronal stability in the overstuffed configurations. Our study is unable to report on ROM as the simulations prescribed this.

Intraoperatively the surgeon determines the appropriate PE thickness based on the overall knee balance and soft tissue tension. (89) This method does not factor in the mid-flexion kinematics of the PE insert sizing choice as the second constraints, such as the posterior capsule, are removed with knee flexion. Our results demonstrated that under- and overstuffing did not globally affect the kinematics during ADL. However, deviations from the reference configuration happened within specific knee flexion ranges during the respective activities. In our study, overstuffing resulted in increased stability, while the inverse was observed in understuffing the joint space *(Figure 5.20-5.22)*. The positive findings were demonstrated during the gait cycle between a flexion range of 10-61^{o} as the knee extended *(Figure 5.20).* In addition, the positive findings were demonstrated during the stair descent between a flexion range of 13-33^{o} and 48-98^{o} *(Figure 5.20).* Similarly, positive findings were demonstrated during the stair ascent between a flexion range of 21-40^{o} and 63-93^{o} *(Figure 5.22).*

The patient could perceive the increased translations and displacements seen in the understuffed configuration as the knee giving way and result in instability complaints. The sense of shifting or the slide of the knee has been reported in clinical literature as instability.(111) This phenomenon has been reported in CR TKA with loss of anterior restraints and associated quadriceps weakness resulting in instability and anterior knee pain.(195) However, the magnitude of kinematic changes we observed during ADL are unlikely to be clinically significant. Our findings add to the current body of literature in the understanding of instability during flexion. Our study confirmed that instability in flexion could not result from isolated understuffing of the tibiofemoral joint space with the PE insert during ADL.

The limitations in this study were the use of point-to-point ligaments rather than bundles of ligaments which does not wholly represent the native ligament properties. In addition, there is still a significant variation in the literature regarding the representation of ligaments.(149) The computational models are based on approximations and assumptions made to simplify the complexity of the human knee. Another limitation is the loading parameters for the ADLs, as these are based on TKA parameters using PS PE.(22) Our model lacked the patella-femoral joint and other soft tissue stabilisers of the knee, and therefore their effects on TKA kinematics were excluded in our study.

Conclusion

In implant modularity, changes in the PE insert sizing affect the TKA kinematics, soft tissue laxity, and ligament tension during ADL. This hybrid biomechanical study highlights that understuffing the joint with a 2 mm smaller PE insert in isolation is unlikely to create instability with flexion during gait, stair ascent and descent. However, further studies are required to understand the reported phenomenon of instability during ADL fully.

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Appendices

Appendix A: An in-depth description of the virtual model creation as explained in the following excerpt from the methods section of a previous work by Montgomery et al.

Virtual Model Development

Creating a virtual knee model started with the segmentation of CT scans taken for a previous cadaveric study using the program Slicer Version 4.11.0. A threshold segmentation technique was used to extract individual bone segments as separate files. Three separates segmentations were extracted: proximal femur including the femoral head, distal femur, and proximal tibia. The proximal tibia and the distal femur – the two parts of the leg tested in the previous cadaveric study – were then assigned unique individual coordinate systems based on the coordinate system developed by Grood and Suntay. The femoral coordinate system originated at the middle of the line connecting the centre of the two spheres made by the condyles (epicondylar axis). The z-axis was defined as a line that passed from this origin to the center of a sphere-fit of the femoral head and was positive proximally[21]. The y-axis was the anteriorly positive cross-product of the epicondylar axis and z-axis. The x-axis was parallel to the sagittal plane of the femur and was the result of cross-multiplying the z and y axes. It was positive to the right. The tibial coordinate system originated at the center of the intercondylar notch. Its z-axis extended proximally from the center of the ankle joint – calculated as the midpoint of the lateral and medial malleoli – to the center of the intercondylar eminences and is positive in the proximal direction. The y-axis was calculated by cross multiplying the z-axis with a line connecting the centers of the two tibial plateaus and was positive in the anterior direction. Finally, the x-axis was the right-facing cross product of y and z axes. Note that the z-axes for both bones are coincident with the respective bone's mechanical axis. The finalized models were then saved as stereolithographic files so that they could be used in CAD software. The stereolithographic files for the TKR prosthesis were obtained directly from the manufacturer: Stryker Corporation. Both the femoral component and tibial components were given coordinate systems as well. The x-axis of the femoral component was taken as the line connecting the centre of the sphere-fits made to each of the condyles. The z-axis was the result of a cross multiplication between the x-axis and a horizontal line taken from the bottom surface of the component. The anterior facing y-axis was the cross product of the z and x axes. This coordinate system was situated at the midpoint between the two condylar sphere-fit centres. The tibial component's coordinate system was centered at the front lip of the central hole. The z-axis was defined as a line parallel to the back edge of the component. The cross product of this and a line connecting the two posterior protuberances gave the y-axis, and the x-axis was a cross product of the z and y axis. In all cases, the z-axis was positive superiorly, the x-axis was positive to the right and the y-axis was positive anteriorly.

Appendix B: An in-depth description of the virtual model creation as explained in the following excerpt from the methods section of a previous work by McGale et al.

Virtual Ligament Model

We simulated soft tissue balancing that is performed during TKA by applying virtual ligaments that recreate the soft tissues used to balance TKAs. The PCL, the sMCL and the LCL were included in the virtual soft tissue envelope. The ACL and dMCL were not used, as these are routinely released in most TKAs as part of the soft tissue dissection required for exposure or boney resection. The ligament insertion points were determined from previous anatomic studies and defined on our anatomic models[13–20]. Once defined on the model, it was possible to determine the relative position of the ligament insertions with respect to the local coordinate systems of the implant components. The same real implant components were mounted onto a joint motion simulator, and it was then possible to determine the coordinates of the insertion points with respect to the simulator axes based on knowing the position of the implant components on the machine. The insertion points were referenced in relation to femoral component, which allowed poly changes without affecting insertion points. However, switching between MA and KA, the ligament insertion points were adjusted accordingly to simulate the change in implant component alignment.

Ligament properties including stiffness, reference strain, reference length and zero load length were adapted from the literature and calculated to fit with our virtual model. A combination of computational TKA models and native knee properties were used to create the ideal ligamentous properties in our work[16,21–25]. Ligament properties had to be defined with respect to a distinct pose, or starting position, on the VIVO. This reference pose was defined at 0° of extension with the application of a 100N compressive load across the joint. This position was used to record the resulting equilibrium pose. With the knee in full extension, the ligament's length can be defined from our models and ligament insertion points. Using this pose, we can calculate the reference strain of each ligament. We used the native ligament length at the same pose and strain on each ligament, as reported in the literature, to calculate the values for zero-load length or slack length. The following calculation for the reference strain of each ligament was used: (current length – original length) / original length x 100%. Qualitatively, this defines the amount of deformation in the ligament at full extension due to the anatomical force placed on the ligaments. For example, the PCL in mechanical alignment has a reference strain of -3.42%, which equates to approximately

1.3mm of slack in the ligament (zero-load length of 37.9mm vs mechanical load length of 36.6mm). Another example is the sMCL in mechanical alignment that has a reference strain of 2.73%, which equates to the ligament being on stretch by approximately 2.4mm from its slack length. The ligament properties of stiffness, reference strain, and ligament length used for each alignment are shown in Table 4-1.

Table 4-1. Ligament properties adapted from literature and calculated to fit our virtual ligament model. All lengths are in millimeters, reference (ref.) strains are given as a percentage of zero-load length, stiffness is in units of Newtons per unit strain. (A) CR femoral component MA and KA, (B) PS femoral component MA and KA. *MA=Mechanically Aligned, KA = Kinematically Aligned*

Ligament CR Component	**Zero-Load Length (mm)**	**MA Length (mm)**	**MA Ref. Strain (%)**	**KA Length (mm)**	**KA Ref. Strain (%)**	**Stiffness (N/ε)**
PCL	37.93	36.63	-3.42	37.29	-1.67	9000
sMCL	89.75	92.21	2.73	92.45	3.00	2200
LCL	63.25	64.61	2.14	64.26	1.60	1800

A

Ligament PS Component	**Zero-Load Length (mm)**	**MA Length (mm)**	**MA Ref. Strain (%)**	**KA Length (mm)**	**KA Ref. Strain (%)**	**Stiffness (N/ε)**
sMCL	89.75	92.38	2.93	92.58	3.15	2200
LCL	63.25	64.91	2.62	64.59	2.12	1800

B

Appendix C: Simulation of under- and overstuffed joint spaces

Understuffed Joint Space

Prior to running the simulation, the VIVO was put in displacement mode. The VIVO was positioned in full extension with the femoral component in contact with the 9 mm polyethylene insert. Using specimen mode menu, we increased the joint space by 2mm creating a gap between the femoral component and 9 mm polyethylene insert. This was then saved as the reference pose for the virtual ligament model. Once in program mode the simulation was loaded and initiated the femoral component returned to a contact position with the 9 mm polyethene insert creating laxity in the virtual ligament model which simulated a 2 mm understuffed joint space.

Overstuffed Joint Space

Prior to running the simulation, the VIVO was put in displacement mode. The VIVO was then positioned in full extension. Using specimen mode menu, the joint space was distracted, and the 9 mm polyethylene insert was removed. The joint space was then set to 7mm “reducing gap for 9 mm polyethylene insert by 2mm”. This was then saved as the reference pose for the virtual ligament model. The joint space was distracted, and the 9 mm polyethylene insert was then re-inserted, and the components placed in a position of contact. The virtual ligament model responded with tensioning with 9 mm polyethylene insert occupying a 7 mm potential space. Once in program mode the simulation was loaded and initiated the virtual ligament model behaved in a manner representative of 2 mm overstuffing of the joint.

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Curriculum Vitae

BIOGRAPHICAL INFORMATION

Full Name: Allan Roy Sekeitto

POST-SECONDARY EDUCATION AND DEGREES:

University of Witwatersrand, Health Sciences Faculty, South Africa

Master of Medicine: Orthopaedic Surgery

"Costing Total Hip Arthroplasty in a South African State Tertiary Hospital"

Research report passed with distinction (2019)

College of Medicine South Africa

Fellowship College of Orthopaedic Surgeons (FC ORTH)

FC Orth(SA) Obtained (May 2019)

University of Witwatersrand, Wits Business School, South Africa

Post Graduate Diploma in Management (PDM) (2012-2013)

Achievements: Dean's List for two semesters. Distinctions: Principles of Strategic Management; Analytical Decision Making and Technology and Information Management.

University of Limpopo, Medunsa Campus, South Africa

Bachelor of Medicine, Bachelor of Surgery (MBCHB) (2005-2010) ***Achievements***: Distinction in Orthopaedic Surgery. PPS Bursary Award (2008).

WORK EXPERIENCE

London Health Sciences Centre, University Hospital

Adult Hip & Knee Reconstructive Surgery Fellow, (August 2020 – Present)

University of the Witwatersrand, Johannesburg South Africa.
Chief Registrarship, Charlotte Maxeke Johannesburg Academic Hospital (July 2019 – June 2020)

University of the Witwatersrand, Johannesburg South Africa.
Registrar, Associate Lecturer (Department of Orthopaedics) (July 2015 – June 2019)

Helen Joseph/Rahima Moosa Mother and Child Hospital Complex, Johannesburg South Africa.
Medical Officer (Department of Orthopaedics)(July 2014 – June 2015)

Charlotte Maxeke Academic Hospital, Johannesburg South Africa.
Medical Officer (Department of Surgery) (January 2014 – June 2014)

Bertha Gxowa Hospital, Germiston, South Africa.
Medical Officer, January 2013 – December 2013

Charlotte Maxeke Academic Hospital, Johannesburg South Africa.
Medical Officer (Internship)(January 2011 – December 2012)

Presentations (Relevant to this thesis)

- Sekeitto AR, McGale JG, Montegomery L, Vasarhelyi E, Willing R, Lanting R. Biomechanical analysis of Polyethylene Insert Thickness in Posterior- stabilized Total Knee Arthroplasty (TKA) using TKA Implants Linked to a Virtual Ligament Model. Canadian Orthopaedic Association Annual Meeting 2021.
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- Sekeitto AR, Van der Jagt K, Sikhauli N, Mokete L, Van der Jagt DR. Intraprosthetic dislocation after a revision hip replacement: a case report. SA Orthop J 2021;20(1):49-52.
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- Almeida RP, Sekeitto AR, Sikhauli N, Mokete L, Pietrzak, JRT. The draining surgical wound post total hip and knee arthroplasty. What are my options? A narrative review. EFORT Open Rev. (Accepted)