

BIOMECHANICAL EFFECTS OF ASYMMETRIC CRANK ARM LENGTHS FOR
ERGOMETER CYCLING IN PATIENTS WITH TOTAL KNEE ARTHROPLASTY

A Dissertation Presented for the

Doctor of Philosophy

Degree

The University of Tennessee, Knoxville

Walter Menke

August 2025

Copyright 2025 © by Walter Gabriel Menke

All rights reserved.

DEDICATION

To my father, Henry, and sister, Paige, for inspiring me to reach towards more “hard science” goals, and especially to my fiancée, Grace, who has stuck with me through arduous writing and coding sessions.

And most importantly to my mother, Sharon, who is not able to be here to celebrate this accomplishment, but without whom this never would have happened. You were my first and strongest cheerleader.

ACKNOWLEDGEMENTS

I would like to first extend my gratitude and thanks to the International Society of Biomechanics and the Kinesiology, Recreation, and Sport Studies Department at UTK for providing financial support for this project. Recruitment was tough with a financial incentive for participants, so without this support, it would have been a rougher road. An additional thanks to the Tennessee Orthopaedic Alliance East and specifically to Dr. Harold Cates and Dr. Katie Stevens for their assistance with the logistics of patient recruitment and study design.

I'm highly grateful to my advisor, Dr. Songning Zhang, for helping to develop my writing and research skillset. I've come out of this process as an improved researcher in many ways thanks to your help. Additionally, this wouldn't be possible without my committee members. Dr. Josh Weinhandl has been an incredible teacher and source of help in practical application of biomechanics as well as helpful during the OpenSim and Matlab courses I took with him. Dr. Hamparsum Bozdogan has been a wealth of statistics knowledge and without whom, I wouldn't have expanded the number of things I still don't know. I'm so glad I took many statistics courses under you- I've become much more critical of papers' conclusions now as a result! And to Dr. Jeffrey Reinbolt who was invaluable in helping understand some of the intricacies of OpenSim and interpretation of simulation results. It has been a team effort, and I'm thankful to have such a competent, intelligent team in my corner.

An enormous thanks to fellow lab members Alex Ovispo-Martinez and Tsung-Lin Lu for their time and efforts in getting data collections set up and running smoothly. I have appreciated your help immensely! And a final thanks to my fiancée, Grace, for putting up with the sound of my mechanical keyboard in the guest room for, what seems like, an uncountable number of hours.

ABSTRACT

Total knee arthroplasty (TKA) is one of the most common treatments of end stage knee osteoarthritis. Not all individuals receiving these surgeries biomechanically recover to a desired level of physical function or improve subjective satisfaction during and after rehabilitation. Investigating accommodating alternatives to traditional rehabilitation exercises for individuals recovering from their TKA surgery is important for promoting improved outcomes, physically and mentally for these individuals. Cycling is often recommended for this population, but there is room to adapt cycling for the recovering knee specifically. This research investigates patients recovering from a TKA soon after their surgery while cycling with asymmetrically adjusted crank arms for the recovering limb through the use of traditional biomechanics and musculoskeletal simulation.

The first study, Chapter 4, evaluated the peak biomechanical responses of the lower limbs of patients 1-2 months post-TKA surgery when cycling on an ergometer using varying lengths of asymmetric and symmetric crank arms. The evidence from Chapter 4 suggests that asymmetric crank arms reduce knee range of motion while maintaining peak knee extension moments. These findings indicate that asymmetric crank arm usage is beneficial in accommodating difficulties with knee stiffness and range of motion early in recovery from TKA without introducing unwanted loading at the knee.

The second study, Chapter 5, calculated the differences between limbs of peak biomechanical variables between groups of participants who had unilateral or bilateral TKAs. Chapter 5 findings report almost no changes in the relationship between limbs of participants with unilateral or bilateral TKA. These findings suggest that the acute asymmetric crank arm

intervention implemented will equivalently affect both groups and clinicians can prescribe this modality for unilateral and bilateral TKA.

The final study, Chapter 6, implemented musculoskeletal modeling simulations to investigate the effects of asymmetric crank arm usage in this population on the replaced limb's estimated vertical knee contact forces as well as the peak muscle forces for musculature surrounding the knee. Chapter 6 results suggest that total vertical knee contact forces are lowered with decreased crank arm length. Future work should evaluate the longitudinal effects of this crank arm methodology to understand chronic adaptations that may occur in this population.

TABLE OF CONTENTS

Chapter 1: Introduction	1
Background	2
Problem Statement and Research Hypotheses	8
Significance.....	10
Delimitations.....	10
Limitations	11
Chapter 2: Literature Review.....	12
Total Knee Arthroplasty	13
Pathology and Epidemiology.....	13
Implant Type and Alignment.....	18
TKA Rehabilitation.....	22
Recommendations	22
Satisfaction and Outcomes	25
Cycling in Healthy Populations	27
Benefits.....	27
Power and Force Measurement	28
Bike Fit and Modification.....	30
Electromyography	37
Cadence	39

Workload and Workrate	43
Crank Arms.....	45
Cycling in OA and TKA Populations	49
Knee OA Populations	49
TKA Populations	51
Limb Alignment Effects	54
Musculoskeletal Modeling.....	57
Muscle Force Estimation	58
Fiber Dynamics.....	59
Model Parameters	68
Muscle Path and Moment Arm.....	72
Geometry	75
OpenSim Modeling.....	78
Scaling	78
Inverse Kinematics	80
Joint Contact Forces	82
Computational Methods.....	84
Static Optimization	84
Forward Dynamics	86
Electromyography in Modeling.....	87

Computed Muscle Control.....	89
Optimization Method Comparison	91
Knee Models	93
Usage and Validation.....	93
Subject Specific Models	95
Cycling Models.....	98
Chapter 3: Materials and Methods.....	101
Participants.....	102
Instrumentation	102
Experimental Protocol	104
Statistical Analysis.....	108
Chapter 4: Effects of Asymmetrical Crank Arm Length Changes on Knee Biomechanics of TKA Patients During Ergometer Cycling.....	109
Abstract.....	110
Introduction.....	110
Methods.....	112
Participants	112
Instrumentation	113
Experimental Protocol	115
Data Analysis.....	116

Statistical Analysis	116
Results.....	117
Discussion	118
Conclusion	124
Chapter 5: Preliminary Investigation of Asymmetric Cycling Biomechanics in Early Unilateral and Bilateral Patients of TKA.....	126
Abstract	127
Methods.....	130
Participants	130
Instrumentation	131
Experimental Protocol	133
Data Analysis.....	133
Statistical Analysis	134
Results.....	135
Discussion	135
Conclusion	140
Chapter 6: Knee Contact Forces in TKA Patients During Ergometer Cycling with Asymmetrically Adjusted Crank Arm Lengths	141
Abstract	142
Methods.....	145

Participants	145
Instrumentation	146
Musculoskeletal Simulation	147
Data Analysis	148
Statistical Analysis	149
Results	149
Discussion	150
Conclusion	154
Chapter 7: Conclusions	156
References	159
Appendices	200
Vita	245

LIST OF TABLES

Table 1: Participant Demographics Information (mean $\pm$ std)	200
Table 2: Mean Results for Peak Pedal Reaction Force and Peak Knee Joint Angle, ROM and Kinetic Variables (mean $\pm$ std)	201
Table 3: Mean Results for ROM and Peak Moment Variables for the Ankle and Hip (mean $\pm$ std)	202
Table 4: Mean Results for Pain and RPE of the Replaced Limb During Cycling (mean $\pm$ std). 203	
Table 5: Mean Results for Differences between Limbs of Peak Pedal Reaction Force, Knee Kinematic, and Knee Kinetic Variables (mean $\pm$ std)	204
Table 6: Mean Results for Differences between Limbs of ROM and Peak Moment Variable Differences for the Ankle and Hip (mean $\pm$ std).....	205
Table 7: Mean Peak Knee Contact Forces and Peak Vertical Pedal Reaction Force (mean $\pm$ std)	206
Table 8: Mean Peak Muscle Group Forces (mean $\pm$ std).....	206
Table 9: Mean Peak Individual Muscle Forces (mean $\pm$ std)	206
Table 10: Mean Peak Residuals and Peak Moments (mean $\pm$ std)	206
Table 11: Participant Qualitative Demographics (mean $\pm$ std).....	206
Table 12: Participant Quantitative Demographics (mean $\pm$ std).....	207
Table 13: Study Inclusion and Exclusion Criteria	208
Table 14: Description of Cycling Crank Arm Length Conditions.....	208
Table 15: Survey Scores (mean $\pm$ std)	208
Table 16: Clinical Metrics and Measurements (mean $\pm$ std)	209
Table 17: Pain in the Replaced Knee during Cycling (mean $\pm$ std).....	209

Table 18: Rating of Perceived Exertion (RPE, Borg Scale) during Cycling (mean $\pm$ std).....	209
Table 19: Peak Vertical Pedal Reaction Force (N, mean $\pm$ std)	210
Table 20: Sagittal Knee ROM ($^{\circ}$, mean $\pm$ std).....	210
Table 21: Peak Knee Extension Angle ($^{\circ}$, mean $\pm$ std).....	211
Table 22: Peak Knee Extension Moment (Nm, mean $\pm$ std)	211
Table 23: Peak Knee Abduction Moment (Nm, mean $\pm$ std)	212
Table 24: Peak Knee Power (Nm/s, mean $\pm$ std)	212
Table 25: Sagittal Ankle ROM ($^{\circ}$, mean $\pm$ std)	213
Table 26: Peak Ankle Plantarflexion Moment (Nm, mean $\pm$ std)	213
Table 27: Sagittal Hip ROM ($^{\circ}$, mean $\pm$ std)	214
Table 28: Peak Hip Extension Moment (Nm, mean $\pm$ std).....	214
Table 29: Vertical Pedal Reaction Force (N, mean $\pm$ std) - Difference.....	215
Table 30: Sagittal Knee ROM ($^{\circ}$, mean $\pm$ std) - Difference	215
Table 31: Knee Extension Angle ($^{\circ}$, mean $\pm$ std) - Difference	215
Table 32: Knee Extension Moment (Nm, mean $\pm$ std) - Difference.....	216
Table 33: Knee Abduction Moment (Nm, mean $\pm$ std) - Difference.....	216
Table 34: Knee Power (Nm/s, mean $\pm$ std) - Difference	216
Table 35: Sagittal Ankle ROM ($^{\circ}$, mean $\pm$ std) - Difference	217
Table 36: Ankle Plantarflexion Moment (Nm, mean $\pm$ std) - Difference.....	217
Table 37: Sagittal Hip ROM ($^{\circ}$, mean $\pm$ std) - Difference.....	217
Table 38: Hip Extension Moment (Nm, mean $\pm$ std) - Difference	218
Table 39: Replaced Knee Total Contact Force (N, mean $\pm$ std).....	218
Table 40: Replaced Knee Medial Compartment Contact Force (N, mean $\pm$ std).....	218

Table 41: Replaced Knee Lateral Compartment Contact Force (N, mean $\pm$ std)	219
Table 42: Peak Quadriceps Muscle Force (N, mean $\pm$ std)	219
Table 43: Peak Hamstrings Muscle Force (N, mean $\pm$ std)	219
Table 44: Peak Gastrocnemius Muscle Force (N, mean $\pm$ std).....	219
Table 45: Peak Rectus Femoris Muscle Force (N, mean $\pm$ std).....	220
Table 46: Peak Semitendinosus Muscle Force (N, mean $\pm$ std)	220
Table 47: Peak Medial Gastrocnemius Muscle Force (N, mean $\pm$ std)	220
Table 48: Peak Tibialis Anterior Muscle Force (N, mean $\pm$ std).....	220

LIST OF FIGURES

Figure 1: Data collection setup of ergometer, motion capture, and participant.....	221
Figure 2: Ensemble curves for sagittal knee joint angle across all conditions. C1: 140/170, C2: 155/170, C3: 170/170, C4: 155/155, C5: 140/140.....	221
Figure 3: Ensemble curves for sagittal hip joint angle across all conditions. C1: 140/170, C2: 155/170, C3: 170/170, C4: 155/155, C5: 140/140.....	222
Figure 8: Double burst electromyography pattern of selected musculature, Suzuki [150]	225
Figure 9: Bicycle crank diagram reference system, adapted from Coyle [334]	225
Figure 10: Fiber length relationship at varying joint positions, Chen and Franklin [223]	226

LIST OF EQUATIONS

Equation 1: Calculation of tendon force.	60
Equation 2: Calculation of muscle force.....	60
Equation 3: Musculotendon length.	61
Equation 4: Musculotendon length and pennation angle interaction.....	61
Equation 5: Muscle force accounting for pennation angle.	61
Equation 6: Tendon slack length.....	65
Equation 7: Calcium dynamics.	70
Equation 8: Minimized least squares for kinematics.	80
Equation 9: Objective function including squared muscle activations.	82
Equation 10: Reaction force at the knee, Steele [277].....	82

NOMENCLATURE

<i>ACL</i>	Anterior Cruciate Ligament
<i>ADL</i>	Activities of Daily Living
<i>ACSA</i>	Anatomical Cross Sectional Area
<i>ANOVA</i>	Analysis of Variance
<i>API</i>	Application Programming Interface
<i>AP</i>	Anteroposterior
<i>BCR</i>	Bi-Cruciate-Retaining
<i>BDC</i>	Bottom Dead Center
<i>BF</i>	Biceps Femoris (muscle)
<i>BMI</i>	Body Mass Index
<i>BW</i>	Bodyweight
<i>CAMS</i>	Comprehensive Assessment of the Musculoskeletal System
<i>CMC</i>	Computed Muscle Control
<i>CR</i>	Cruciate-Retaining
<i>CS</i>	Cruciate-Substituting
<i>DoF</i>	Degree of Freedom
<i>EMG</i>	Electromyography
<i>FEA</i>	Finite Element Analysis
<i>IK</i>	Inverse Kinematics
<i>KA</i>	Kinematic Alignment
<i>KAbM</i>	Knee Abduction Moment
<i>KEM</i>	Knee Extension Moment

<i>kg</i>	Kilogram
<i>KOOS</i>	Knee Injury and Osteoarthritis Outcome Score
<i>KSS</i>	Knee Society Scoring System
<i>LC</i>	Lateral (Compartment) Contact Force
<i>MA</i>	Mechanical Alignment
<i>MCF</i>	Medial (Compartment) Contact Force
<i>MG</i>	Medial Gastrocnemius (muscle)
<i>MHC</i>	Myosin Heavy Chain
<i>MRI</i>	Magnetic Resonance Imaging
<i>MVIC</i>	Maximal Voluntary Isometric Contraction
<i>N</i>	Newtons
<i>Nm</i>	Newton-Meters
<i>NRS</i>	Numerical Rating Scale
<i>OA</i>	Osteoarthritis
<i>PARQ+</i>	Physical Activity and Readiness Questionnaire
<i>PCS</i>	Pain Catastrophizing Scale
<i>PCSA</i>	Physiological Cross-Sectional Area
<i>PCL</i>	Posterior Cruciate Ligament
<i>PRF</i>	Pedal Reaction Force
<i>PROs</i>	Patient Reported Outcomes
<i>PS</i>	Posterior-Stabilized
<i>rKA</i>	Restricted Kinematic Alignment
<i>RMS</i>	Root Mean Square

<i>RMSE</i>	Root Mean Square Error
<i>ROM</i>	Range of Motion
<i>RPE</i>	Rating of Perceived Exertion
<i>RPM</i>	Revolutions Per Minute
<i>RRA</i>	Residual Reduction Algorithm
<i>RSA</i>	Radiostereometric Analysis
<i>RF</i>	Rectus Femoris (muscle)
<i>SENIAM</i>	Surface EMG for a Non-Invasive Assessment of Muscles
<i>SO</i>	Static Optimization
<i>ST</i>	Semitendinosus (muscle)
<i>TA</i>	Tibialis Anterior (muscle)
<i>TCF</i>	Total Contact Force
<i>TDC</i>	Top Dead Center
<i>TKA</i>	Total Knee Arthroplasty
<i>TSK</i>	Tampa Scale of Kinesiophobia
<i>TUG</i>	Timed Up and Go
<i>VAS</i>	Visual Analog Scale
<i>VM</i>	Vastus Medialis (muscle)
<i>W</i>	Watts
<i>WOMAC</i>	Western Ontario and McMaster Universities Arthritis Index

Chapter 1: Introduction

Background

Total knee arthroplasty (TKA) is a surgical procedure that involves reshaping and resection of bone and meniscus at the patellofemoral and tibiofemoral joints {Varacallo, 2023 #251;Parsons, 2021 #259}. TKA procedures have had an increasing growth rate since 2000 and have a projected annual growth rate of 4.44% while surpassing 480,000 surgeries in 2019 {Shichman, 2023 #224}. The purpose of the surgery in many cases is to remedy primary knee osteoarthritis (OA) by restoring an individual's quality of life and physical function, decreasing pain, and attempting to halt structural decay of the involved bones {Varacallo, 2023 #251;Guo, 2018 #260}. Women undergo more TKA procedures than men, which is likely a result of articular cartilage loss at four times the yearly rate as men {Hanna, 2009 #307;O'Connor, 2011 #310}. Women also tend to have more unfavorable outcomes such as implant failure, pain and depressions levels, obesity, and additional symptomatic joints after TKA than men, but some authors have concluded that sex specific TKA implants result in similar outcomes {Vina, 2020 #311;Mehta, 2015 #312;Johnson, 2011 #308}. Further factors for knee OA progression consideration include high body mass index (BMI), greater radiographic disease burden, slower chair-stand time, and larger bone marrow lesions {Driban, 2020 #402}. In conjunction with higher BMI, physical activity levels for TKA recipients after surgery decrease below their pre-surgery levels, which often fail to meet activity guidelines {Arnold, 2016 #207}. In the interest of curbing BMI increases after surgery, there is an interest in investigating strategies for increasing exercise adherence after surgery.

Patient satisfaction with their TKA surgery has been reported in the range of 75-89%, and there is a potential that rehabilitation protocols and activities contribute to a portion of those who are dissatisfied {Vissers, 2010 #188}. If rehabilitation for patients receiving TKA were

standardized, it may be possible to determine which factors contribute to those dissatisfied. However, TKA rehabilitation guidelines vary by provider and as such, have caused issues with determining the efficacy of the protocols {Mistry, 2016 #313}. Current literature suggests that low-impact activities such as aquatic therapy, ergometer cycling, level-surface walking, group exercise therapy, and continuous passive motion therapy are all acceptable and beneficial activities for increasing clinical outcome scores, muscle strength, and gait speed {Papalia, 2020 #238;Zech, 2015 #220}. Certain clinics begin rehabilitation activities before performing a TKA procedure, but a meta-analysis determined that this only shortened the hospital stay as opposed to more common metrics such as pain and knee extension capacity {Su, 2022 #316}. In fact, weight loss programs started after surgery have been shown to be more beneficial than programs started before surgery at 26 weeks post-operation {Pellegrini, 2018 #208}. As previously mentioned, physical activity levels after surgery are lower postoperatively and BMI tends to increase for TKA recipients which highlights the importance of weight loss and cardiovascular components in rehabilitation programs {Arnold, 2016 #207}. High and low intensity postoperative rehabilitation programs have been compared and showed no significant changes at 2 or 3 months in variables of interest such as knee range of motion (ROM), quadriceps, and hamstrings strength {Bade, 2017 #317}. Bade et al. [18] had all patients perform high or low intensity exercises in an outpatient facility, then sent patients home with exercises to do in an unsupervised setting. They reported that home exercise adherence for the high intensity group was 7.7% lower which may indicate a decreased desire to perform activities at this level of intensity in an unsupervised setting {Bade, 2017 #317}. Many surgeons tend to advise against high or medium impact activities postoperatively, approximately 25% and 47%, respectively, of TKA patients perform these activities at nearly seven years postoperative follow-up {Papalia,

2020 #238;Mayr, 2015 #326}. High impact activities included activities such as rock climbing, dancing, and tennis while medium impact activities included mountain hiking and Nordic walking.

Cycling is well-studied modality for increasing cardiovascular fitness and a wealth of evidence attests to its benefits for reducing all-cause mortality, morbidity, and mortality in patients with cancer, coronary heart disease morbidity and mortality, type II diabetes, and obesity {Oja, 2011 #217;Logan, 2023 #216}. Stationary cycling provides the benefit of safety from motor vehicles and offers the same benefits as commuter or road cycling {Chavarrias, 2019 #278;Johan de Hartog, 2010 #279}. Specifically, indoor cycling has a potential benefit to decrease systolic and diastolic pressure, increase high density lipoproteins, decrease low density lipoproteins, and aerobic capacity, but should be combined with diet interventions to maximize outcomes {Chavarrias, 2019 #278}. The reductions in blood pressure, specifically, are accomplished with increasing effectiveness of cardiovascular musculature, while lipid profiles are more likely to be improved in conjunction with exercise and diet interventions. In a study using a self-reported physical activity survey, cycling has been associated with weight gain attenuation in middle-aged populations, while other common activities such as slow walking and swimming were not {Littman, 2005 #342}. Cycling is an activity that reduces required weight-bearing loads of the knee which decreases tibiofemoral forces compared to gait while maintaining cardiovascular benefits such as lowered all-cause mortality, coronary heart disease risk, and obesity {Kutzner, 2012 #4;Logan, 2023 #216;Oja, 2011 #217}. For TKA populations, cycling provides cardiovascular benefits while decreasing peak resultant knee joint loads compared to walking or other daily activities {Kutzner, 2012 #4}. Moreover, at 40 revolutions per minute (RPM) and a power setting of 90 Watts (W) peak resultant knee contact force was

approximately 160% bodyweight (BW), on average, compared to the 200-350% BW of other daily activities such as stair descent, stair ascent, standing up, and sitting down {Kutzner, 2010 #408}. As a result, cycling can be recommended to TKA recipients as an excellent way to engage in physical activity while minimizing harmful impact and wear on their implant.

Early in the TKA rehabilitation process, stationary cycling has been implemented and frequently recommended to facilitate ROM increases, pain improvement, and patient reported outcome scores via surveys {Zech, 2015 #220;Lester, 2022 #412}. Zech et al. {, 2015 #220} reported that knee ROM improved, pain decreased, and Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) scores increased with an exercise regimen started approximately 2 weeks after TKA. Continuous x-ray images of ergometer cycling in patients at least six months after TKA provided evidence that stationary cycling is encouraged over activities such as golf in order to maintain acceptable loading of the implant's polyethylene insert {Hamai, 2008 #319}. Specifically, Hamai et al. {, 2008 #319} demonstrated that TKA implant flexion angle excursion in the sagittal plane was favorable and that axial rotation was acceptable during cycling. Maintaining the mechanical health of the polyethylene insert is important for minimizing the likelihood of implant loosening or failure {Zhao, 2007 #249}.

Despite the overall decreased kinetic requirements of cycling due to partial bodyweight support, TKA patients have shown significantly reduced peak knee extension moments (KEM) of the replaced limb compared to the native limb on a cycle ergometer and may find it functionally difficult to perform shortly after, e.g., one to two months, after surgery {Hummer, 2021 #200}. Hummer et al. {, 2021 #200} recruited TKA patients 6 to 18 months post unilateral TKA and found that peak KEM was 18.7 ± 5.2 and 24.7 ± 6.9 Nm for the replaced and non-replaced limb, respectively, for a work rate of 80 W at 80 RPM. A lower peak KEM in the

replaced limb is also present in this population with other tasks, such as stair climbing {Yocum, 2022 #222}. Decreased KEM during physical activities is likely a result of decreased quadriceps strength post-TKA which is further exacerbated by muscle attenuation after surgery that does not recover at six months post operation {Paravlic, 2022 #301}. In order to accommodate kinematic and kinetic deficits of the replaced limb post-TKA, one aspect of bike modifications that may be beneficial is adjustment of the crank arm lengths. When shortening crank arms symmetrically, a significantly reduced knee range of motion is demonstrated, which is theorized to be beneficial for patients managing pathologies such as knee osteoarthritis while maintaining joint-specific powers {Barratt, 2016 #508, Barratt, 2011 #286, Martin, 2001 #30}. For individuals recovering from a TKA, however, symmetrical reductions in crank length may not be necessary and minimal work has been conducted on asymmetrically shortening the crank arms specifically for the affected limb of individuals with unilateral lower limb pathologies. Childers et al. {Childers, 2014 #225} and Millour et al. {Millour, 2020 #226} have presented asymmetrical crank arm usage in transtibial amputees and limb length difference populations. Research on asymmetrical cycling interventions, acute or longitudinal, have not been reported in the TKA population. The gap in the literature for rehabilitation recommendations for those recovering from TKA is evident and acute investigation is warranted into asymmetric crank arm usage and its effects on kinematics and kinetics.

Beyond traditional biomechanics, utilizing musculoskeletal modeling in the TKA population is of interest for estimating internal knee contact forces. During cycling, a 23.5% decrease has been reported in peak medial tibiofemoral compressive force for the replaced limb compared to the non-replaced limb in TKA patients {Hummer, 2022 #169}. Short cycle sprint testing has been implemented in TKA patients at 3-, 6- and 12-months post-operation and

showed significantly decreased lower limb power output at 3 months and significantly greater power output at 12 months {Hurworth, 2021 #229}. Hurworth et al. {, 2021 #229} additionally provided Oxford Knee Scores and KOOS scores that improved for each follow-up. These findings show support for tracking rehabilitation progress by measuring muscle power in TKA populations as a dynamic value instead of maximum muscle force output or more simple metrics such as knee range of motion. Musculoskeletal modeling in cycling populations has been performed using the standard gait2392 model as well as the Lerner model developed in 2015 {Lerner, 2015 #171;Hummer, 2022 #169;Hummer, 2021 #200}. Both models consist of one degree of freedom (DoF) (flexion/extension) for the tibiofemoral joint. The Lerner model is able to dissolve the knee contact forces between the medial and lateral compartments, while the gait2392 model is only able to calculate overall tibiofemoral contact force. In a study consisting of 6–18-month post-operative TKA patients, the Lerner model was used to assess medial and lateral compartmental loading during stationary cycling at 80 RPM with power settings of 80 and 100 W. Hummer et al. {, 2022 #169} reported that the TKA patients cycled with balanced medial and lateral compartment loadings at 80 W, but the medial loading increased with power more rapidly in the replaced limb. Specifically, the replaced limb at 80 W had 456.9 and 489.4 N of contact force for the medial and lateral components, respectively, while the 100 W condition increased the loadings to 597.2 and 411.9 N. An additional study investigated the effect of q-factor (pedal width) on medial compartment loading in stationary cycling using the Lerner model {Thorsen, 2021 #173}. Important findings from this study included the finding that regression analyses revealed that knee abduction moment and KEM accounted for 87.4% of the variance in medial compartment contact force when q-factor was considered. The findings from Hummer et al. {, 2022 #169} show an approximately 21% increase in peak KEM in the nonreplaced limb of

TKA patients with no accompanying change in knee abduction moment between limbs. This finding is in agreement with reports by Thorsen et al. {, 2021 #173} associating medial compartment loading with KEM.

Problem Statement and Research Hypotheses

Given the known deficits present in TKA populations in replaced limb kinetics during stationary cycling, there is still no literature available that proposes asymmetric cycle ergometer adjustments designed to accommodate for these deficits. Asymmetric crank arms have only been implemented in two demographics: transtibial amputees and individuals with limb length discrepancies {Millour, 2020 #226;Childers, 2014 #225}. In transtibial amputees, kinematic symmetry in hip and knee range of motion were obtained, but not kinetic symmetry {Childers, 2014 #225}. In individuals with limb length discrepancies, knee range of motion significantly decreased while perceived exertion was lower; this was accompanied by an increase in maximal power output {Millour, 2020 #226}. Cycling intervention studies in TKA populations have indeed been conducted, but the use of asymmetric crank arm lengths to ease the patient into standard cycling crank lengths has not been seen {Liebs, 2010 #411}. This intervention study, however, only found significant beneficial clinical scores for hip arthroplasty patients and not for TKA patients. Asymmetrically shortening the crank arm for the replaced limb could be iteratively lengthened as rehabilitation progresses to a full-length crank arm, thus gradually increasing the required knee range of motion for these patients. Given the existing evidence of potential clinical benefits of asymmetric crank lengths in cycling, the frequent recommendations to engage in cycling, and the lack of cycling interventions in the TKA population, there is a gap in the literature that this research can fill. Therefore, the purposes and hypotheses of this work were as follows:

Study One

The primary purpose of this study was to compare the biomechanical profiles of patients with unilateral or bilateral TKAs during asymmetrical crank arm length ergometer cycling.

It is hypothesized that:

- a. Knee ROM, and peak KEM and knee extension power will be lower in the most recently replaced knee during asymmetrical crank length conditions when compared to the contralateral knee of patients with at least one TKA.
- b. Replaced knee pain and rating of perceived exertion (RPE) will be lower for asymmetrical crank length conditions compared to the symmetrical conditions.

Study Two

The primary purpose of this study was to assess the quantitative differences between limbs of patients with unilateral or bilateral TKAs during asymmetrical crank arm length stationary cycling.

It is hypothesized that:

- a. The difference of peak KEM between limbs will be significantly lower and peak hip extension or peak ankle plantarflexion moment differences will be significantly greater in bilateral TKA participants compared to unilateral TKA participants for all cycling conditions.

Study Three

The primary purpose of this study was to assess knee joint loading with OpenSim during asymmetrically shortened crank arm conditions for the replaced limb shortly after a TKA procedure.

It is hypothesized that:

- a. The peak TCF of the replaced limb will be unchanged during the asymmetric crank length conditions.
- b. The peak quadriceps muscle force will be significantly lower at shorter crank arm lengths compared to the symmetrical, control lengths, which is theorized to occur due to the expected decrease in knee ROM.

Significance

Cycling is a commonly agreed upon acceptable exercise modality for rehabilitation of patients with TKA and further encouraged for long term physical health. Cycling is an exercise that provides significant cardiovascular benefits while reducing potentially harmful, chronic knee joints loads that other, higher impact activities impart. Reduction of these loads is not only important for implant survival, but also for ensuring the implant recipients can continue to engage in exercise unhindered. However, it is a known concern in the TKA population that physical activity after surgery declines from a sub-standard pre-operation level. Finding a way to engage this population in physical activity as early as possible after rehabilitation could be important to their continued adherence to activity post-rehabilitation as well as enhancement of their rehabilitation outcomes. TKA participants exhibit significant kinetic deficits in their replaced limbs in stationary cycling after a range of 6-18 months after receiving an implant. To remedy these asymmetries, this research may provide an alternative method for TKA populations to begin cycling earlier in their rehabilitation process.

Delimitations

The inclusion criteria for TKA participants are found in **Table 3** (all tables are in the appendix).

Limitations

The greatest limitation of this work is the sample size of 7 participants, which is exacerbated when splitting this pool into unilateral and bilateral participants. Future work should significantly bolster recruitment efforts by recruiting from more than one clinic to obtain participants. One of the larger sources of error for this work relates to the placement of the retroreflective marker placement on each subject. Markers positions were placed by the same researcher, but placement on the same subject for both sessions may have been different. This would affect inverse dynamics as well as musculoskeletal simulation results. Additionally, kinematic data has an inherent limitation with skin motion creating unwanted marker movement during dynamic movements which may be increased in participants with a greater BMI. A final, and significant, limitation of this work relates to the small sample size. Although a power of 0.80 was reached after a power analysis, this work can be considered exploratory with the primary goal of an expanded subject pool in the future.

Chapter 2: Literature Review

Total Knee Arthroplasty

Pathology and Epidemiology

Total knee arthroplasty (TKA) is a surgical procedure that can involve reshaping and resection of bone and meniscus at the patellofemoral and tibiofemoral joints {Varacallo, 2023 #251;Parsons, 2021 #259}. TKA procedures have had a positive growth rate since 2000 and have a projected annual growth rate of 4.44% while surpassing 480,000 surgeries in 2019 {Shichman, 2023 #224}. The purpose of the surgery in many cases is to remedy primary knee osteoarthritis (OA) by restoring an individual's quality of life, decreasing pain, and attempting to halt structural decay of the involved bones {Varacallo, 2023 #251;Guo, 2018 #260}. Women undergo more TKA procedures than men, which is likely a result of articular cartilage loss at four times the yearly rate as men {Hanna, 2009 #307;O'Connor, 2011 #310}. Additional factors proposed to affect women's likelihood for OA development include increased q-angle, lower limb muscle strength, increased knee laxity, and hormonal differences {Black, 2022 #410}. Women also tend to have more unfavorable outcomes such as implant failure, pain and depressions levels, obesity, and additional symptomatic joints after TKA than men, but some authors have concluded that sex specific TKA implants result in similar outcomes {Vina, 2020 #311;Mehta, 2015 #312;Johnson, 2011 #308}. Further factors for knee OA progression consideration include high body mass index (BMI), slower chair-stand time, and larger bone marrow lesions {Drihan, 2020 #402}. Additionally, individuals who developed knee OA within a four-year follow-up period exhibited more radiographic symptoms at the initial time point which identified it as a risk factor compared to peers who did not develop knee OA.

Patients who undergo one TKA have a 46% chance of having a contralateral TKA procedure done within ten years {Shao, 2013 #205}. Maximizing TKA outcomes such as

implant survival requires a wide scope of factors to be considered to prevent revision TKA surgeries that are higher risk of complications such as increased hospital stay, operation time, and economic cost than the initial implant procedure {Nham, 2023 #184}. The most common comorbidities for a revision TKA include obesity, hypertension, and diabetes {Nham, 2023 #184;Mak, 2020 #203}. Obesity and BMI are also significant indicators of needing an initial TKA procedure {Kulkarni, 2016 #204;Nicholls, 2012 #202}. Having a BMI score in the obese range introduces more risk factors into post-operative joint arthroplasty recovery, as well, such as delayed functional recovery and health complications such as infection or thromboembolic events that have a high mortality rate {Lui, 2015 #209;Vaqr, 2023 #210}. BMI increases after TKA surgery are common and there is a large discrepancy in physical activity levels of post-surgery TKA patients and healthy controls {Barahona, 2023 #206;Arnold, 2016 #207}. Arnold et al. {, 2016 #207} demonstrated that TKA patients tend to perform less physical activity a year from surgery compared to their healthy control peers. However, exercise interventions designed to promote weight loss started after surgery have been shown to be more effective relative to the same program started prior to surgery {Pellegrini, 2018 #208}. A high BMI is not only a predictor for TKA likelihood but also is associated with increased inpatient surgery costs and episode of care costs {Dowsey, 2011 #415}. An episode of care consists of the entire instance of medical care for a patient at a hospital, excluding surgery, which consists of hospital inpatient costs, readmissions, and emergency presentations. Dowsey et al. {, 2011 #415} reported a BMI-associated increase of \$1,226.89 for inpatient surgery and an increase of \$1,821.36 for episode of care costs. Further, between obese and non-obese patients, there was a significant increase in diabetes mellitus occurrence, complications, and unplanned readmissions.

Knee OA is often quantitatively assessed radiographically with the Kellgren-Lawrence (KL) scale proposed in 1957 {Kellgren, 1957 #136}. While other systems for OA severity classification exist such as the Fairbank and Ahlbäck scales, the KL system is the most widely used. The KL system has been shown to have good inter- and intra-observer correlation coefficients of 0.83 {Kohn, 2016 #33}. Higher scores on the KL scale are commonly a precursor to TKA approval by insurance companies and is graded at five levels from 0 to 4 and, which ranges from no joint space narrowing visible or deformities to obvious joint space narrowing, large osteophyte formation, and bone end deformities {Kohn, 2016 #33}. The medial and lateral compartments of the knee are involved with a TKA, as opposed to unicompartmental knee arthroplasty, and a tibial tray with a polyethylene insert are used to replace the proximal tibia. Alignment of the femur and tibia in the frontal plane along their respective longitudinal axes are of interest in knee OA and, subsequently, TKA populations.

Individual lower limb alignment can be classified into neutral, valgus, and varus alignments with a radiographically measured knee mechanical axis angle, or commonly referred to as the hip knee angle, of $180^{\circ} \pm 2^{\circ}$, $>182^{\circ}$, and $<178^{\circ}$, respectively {Sharma, 2010 #474}. Statically, varus alignment results in greater contact between the femur and tibia at the medial compartment, while valgus alignment results in greater contact at the lateral compartment {Ginesin, 2018 #416}. Ginesin et al. {, 2018 #416} reported that children frequently have more severe degrees of valgus knee alignment, but this valgus alignment often self-corrects to approximately 7° or 8° . Additionally, due to pelvic structure and consequent q-angle changes, women tend to have a greater degree of valgus {Mitani, 2017 #510}. This is most likely a contributing factor to the increase in OA prevalence among women and their subsequent increase in likelihood for TKA. Generally, varus alignment relates to the risk of medial knee OA, while

valgus alignment relates to the risk of lateral knee OA. The association of valgus alignment with OA risk is further strengthened with lateral compartment OA progression specifically, with varus alignment increasing the odds of medial compartment OA progression {Sharma, 2010 #474}. In a longitudinal study conducted with cohorts from the Multicenter Osteoarthritis Study and the Osteoarthritis Initiative, approximately 73% of the knees that developed OA were in the medial compartment, with the remaining 27% in the lateral compartment {Wei, 2019 #511}. Results from this work reported that obesity and varus knee alignment were significantly more associated with medial knee OA than lateral knee OA by 95% and 466%, respectively. These findings suggest that varus malalignment and medial knee OA are the most commonly occurring, with even greater incidence in obese populations. Increased amounts of valgus combined with the frequent obesity of knee OA populations are important factors for TKA prediction.

According to a longitudinal study of 1,030 participants to assess lateral knee OA risk factors, the prevalence of normal, valgus, and varus alignments were 29.9%, 56.5%, and 13.7%, respectively {Modenese, 2018 #163}. This study reported that medial knee OA was present in 16% of the participants, and lateral knee OA was present in 4.3% of the participants. In an observational study that assessed 2958 knees, it has been reported that the percentage of neutral, valgus, and varus knees was approximately 39.2%, 19.4%, and 41.4%, respectively {Sharma, 2010 #474}. Sharma et al. {, 2010 #474} reported that medial knee OA was significantly more likely to develop in individuals with varus alignment, but not valgus, when accounting for variables such as knee strength, laxity, injury, BMI, gender, and age. Malalignment can lead to increased grades of knee OA and in some cases, the cartilage damage can lead to further malalignment and joint space narrowing {Ginesin, 2018 #416;Duffaut, 2023 #409}. Injury to knee ligaments may also lead to compromised alignment and knee OA progression regardless of BMI or behavioral

factors such as physical activity levels {Duffaut, 2023 #409}. A main concern of accelerated cartilage damage due to malalignment after unilateral TKA is the likelihood of the patient requiring a contralateral TKA in the future. Interestingly, biomechanical factors such as peak knee (external) adduction moment, knee ROM, and peak knee flexion and extension angles were not significantly different in unilateral TKA patients that did and did not receive a contralateral implant within approximately nine years of their original surgery {Aljehani, 2022 #306}. As a result, contralateral TKA likelihood may be based on additional factors that were not measured by Aljehani et al. {, 2022 #306}.

Prevalence of radiographic knee OA in men, for either knee, has been reported to be approximately 27.7% and 34.6% in women from a large cohort (n=4151, age range 22-93) taken from the Copenhagen Osteoarthritis Substudy database {Laxafoss, 2013 #418}. This database provides a unique opportunity to assess the behavior of knee alignment through a wide age range. Laxafoss et al. {, 2013 #418} concluded that diseased knees tended to produce varus alignment while knees without radiographic OA signs tended to shift to valgus alignment with increasing age. This conclusion is unique and rare because of the longitudinal nature of the work and helpful in the interpretation of other, commonly cross-sectional, studies of knee OA populations in the field of biomechanics. Progression of compartmental OA tends to be accelerated for the medial compartment according to data obtained from the Osteoarthritis Initiative {Everhart, 2019 #419}. Participants underwent magnetic resonance imaging (MRI) at baseline and at a two year follow up to assess cartilage defect progression characteristics. It was reported that medial defects increased significantly in size and independent predictors of defect size progression were varus alignment and increased weight of the participant. These data suggest that body mass is a significant variable for knee OA progression and that medial

compartment, as well as total, knee loading should be minimized whenever possible in OA and TKA populations.

Implant Type and Alignment

Multiple TKA implant design types exist and vary based on original anatomy preservation and implant DoF or motion restriction {Varacallo, 2023 #251}. Conflicting evidence has been presented to address implant design superiority for native knee characteristics, functional outcomes, survivorship, contact positions, kinematics, and kinetics between popular designs such as cruciate-retaining (CR), bi-cruciate-retaining (BCR), cruciate-substituting (CS) and posterior-stabilized (PS) designs {Song, 2019 #261;LeDuc, 2023 #254;Li, 2014 #255;Zelle, 2009 #257;Gray, 2020 #256;Broberg, 2020 #258;Tsai, 2019 #80;Boese, 2021 #268;Arnout, 2020 #267}.

CR designs retain the original posterior cruciate ligament of the patient and a recent review indicated that clinical scores between CR and PS designs are undetectable, have improved survivability, and decreased operating times {Yang, 2023 #403}. Debate continues amongst surgeons and researchers about the level of importance in TKA procedures of the posterior cruciate ligament (PCL), which acts to constrain posterior tibial translation while allowing for femoral rollback during activities of deep flexion {Movassaghi, 2023 #404}. The removal of the PCL for PS designs involves a cam-post structure on the tibial tray that is designed to perform similar roles to the native PCL that allows for a certain degree of femoral rollback. Complete sacrifice of the cruciate ligaments for PS designs has been associated with problematic outcomes such as accelerated implant wear and patellar clunk {Movassaghi, 2023 #404}. Patellar clunk is a symptom that results from scar tissue development superior to the patella and manifests as a painful, sometimes audible clunk, during active knee extension in the

range of approximately 30-45° of flexion {Putman, 2019 #405}. CS designs are not new in the implant space, but with the improvement in modern polyethylene compounds has reduced the concerns regarding wear that were common in the past {Volkmar, 2022 #406}. CS implants, theoretically, can combine benefits of PS and CR designs by not requiring the PCL to be adjusted to balance flexion and extension. Additionally, CS implants do not need a femoral box, the use of which can be linked to crepitus that may present in PS implants . One of the primary differences in CS designs compared to CR designs is an increased degree of congruency in the polyethylene inserts between the tibial and femoral components for CS designs {Volkmar, 2022 #406}. BCR designs maintain the original anterior cruciate ligament (ACL) and PCL and are the most modern and as such, have the least amount of literature regarding long-term outcomes {Sabatini, 2021 #269}. A meta-analysis by LeDuc et al. {, 2023 #254} concluded that no significant differences were present between CR and CS implant groups in survivorship, clinical scores, or ROM. When comparing PS to CR implants, Li et al. {, 2014 #255} found an approximately 11° increase in knee flexion ROM in the PS group, but no differences in knee society scores (KSS) or survivorship. Broberg et al. {, 2020 #258} performed radiostereometric analysis (RSA) from 0° to 100° of knee flexion for PS and CR implant designs and found no significant differences in condylar separation frequency, medial excursion, or clinical outcome scores. The purpose of RSA is to examine implant component migration and assess implant survival based on actual or predicted implant loosening. RSA is performed by embedding tantalum, or another similar material, markers into the patient's bone and examining them at regular 6-month intervals to quantify any motion occurring between the bone and the implant. However, the PS group exhibited greater lateral excursion as well as decreased timed-up-and-go (TUG) test timings {Broberg, 2020 #258}.

Once an implant design has been chosen, aligning the implant kinematically or mechanically becomes a surgical choice that has no clear consensus within contemporary literature {Blakeney, 2019 #190;Risitano, 2022 #192;Blakeney, 2019 #190;Williams, 2020 #46}. Inconclusive evidence has been presented as to whether kinematically or mechanically aligned implants are more favorable to achieve desired patient outcomes and gait characteristics {Blakeney, 2019 #190;Rivière, 2017 #315}. Kinematic alignment (KA) approaches aim to reinstitute native knee kinematics in the replaced knee by restoring tibiofemoral axial and frontal orientation as well as surrounding soft tissue laxity {Blakeney, 2019 #190}. Further, restricted kinematic alignment (rKA) has been proposed with a five-goal approach that attempts to maintain as much native bone and soft tissue structure as possible {Vendittoli, 2021 #272}. Evidence has been presented to support the superiority of rKA over mechanical alignment (MA) based on a decrease in excessive varus and valgus alignment while providing similar outcome metrics and implant failure rates {Risitano, 2022 #192}. It is important to note that the systematic review done by Risitano et al. {, 2022 #192} did not compare rKA to KA and that long term follow-ups were not available at the time of publication.

MA approaches require frontal plane x-ray views of the patient to determine coronal alignment implant orientation {Moreland, 1987 #177}. MA methodology has been considered the dominant strategy for alignment historically and attempts to equally distribute medial and lateral compartment loads to reduce implant wear {Hohman, 2015 #271}. Determining coronal alignment should be done in a standing, weight-bearing position with a full-length view to determine the hip-knee-angle (HKA) as radiograph short views are inadequate for this purpose {Diamond, 2016 #127;Dargel, 2016 #244}. HKA is defined as the angle between the mechanical axes of the femur and tibia. Coronal alignment determined on the short anteroposterior (AP)

view of the knee can result in an approximately 14% or 33% alignment misclassification for preoperative and postoperative images, respectively {Diamond, 2016 #127}. The standard, close-up anteroposterior (AP) view of the knee can be beneficial for implant alignment assessment post-surgery unless a weight-bearing position to assess the hip to ankle mechanical axis is desired {Skyttä, 2009 #246}. Both long and short radiographs are beneficial for assessing implant alignment and quality after surgery, which is critical for early identification of problems such as aseptic loosening, ligament laxity, and polyethylene wear {Kumar, 2014 #186}. Additionally, a skyline view of the knee is beneficial for assessing patellar morphology which has been linked to pain changes during stair navigation tasks {Aït-Si-Selmi, 2020 #270}. Specifically, Aït-Si-Selmi et al. {, 2020 #270} described TKA patients without patellar resurfacing as having more pain during stair ascent and descent based on characteristics such as larger or smaller medial patellar facets. Mechanical alignment can be achieved through dependent or independent bone cuts and no difference has been found between the methodologies when gait parameters, patient reported outcomes, and patient satisfaction are compared {Miozzari, 2021 #305}.

Longitudinal studies comparing alignment strategies in TKA biomechanically are rare, however, work presenting patient reported outcomes (PROs) between the alignment types has been published {Dossett, 2023 #502; Miura, 2023 #503; Gao, 2020 #504}. Dossett et al. {, 2023 #502} followed 48 patients with a mixture of kinematically and mechanically alignment knee implants over a 13-year period. Upon completing five commonly utilized surveys, it was determined that there were no statistical differences in PROs, major and minor reoperations, or patient satisfaction. Miura et al. {, 2023 #503} performed a meta-analysis on implants aligned via MA, KA, or rKA for ten large scale randomized control trials and 1,008 knees with 1.5 years

as the median follow-up time. Overall, the findings were inconclusive as all presented results were with very low confidence. Confidence, or certainty of evidence, in healthcare decision making arise from comparing the sample confidence interval to some defined clinical decision making boundary. Miura et al. {, 2023 #503} do not explicitly state their boundary for their meta-analysis. The only reported differences occurred between the tibial and femoral components in their frontal plane alignments between the methodologies. Results from this analysis should be considered carefully and with caution. Gao et al. {, 2020 #504} also performed a meta-analysis on randomized control trials in TKA patients receiving MA or KA implants, but only in the follow-up range of 6 to 24 months after surgery. The KA group did demonstrate improved knee flexion ROM at the follow-up, but no other differences were found from a wide range of variables such as knee extension ROM, tibiofemoral alignment, tibial component slope angle, or survey scores.

TKA Rehabilitation

Recommendations

Unlike knee OA populations, TKA rehabilitation and management recommendations are not consolidated or commonly disseminated amongst clinicians. Organizations such as the Osteoarthritis Research Society International and the American College of Rheumatology/Arthritis have statements and guidelines about OA, while analogs for TKA do not exist {Bannuru, 2019 #8; Miozzari, 2021 #305; Kolasinski, 2020 #9}. As a result, TKA rehabilitation procedures vary among providers, which, has caused issues with determining the efficacy of the protocols {Mistry, 2016 #313}. In fact, a review of recommendations for patients having received a hip or knee prosthesis reported that no study quantified recommendations for limiting sedentary behavior or increasing physical activity to a certain threshold {Mooiweer,

2022 #407}. If patients are being directed to rehabilitation sessions with minimal to no guidance on their behavior after the rehabilitation protocol is complete, ensuring optimal or even positive outcomes is likely to be difficult. The amount of rehabilitation sessions and length of sessions is variable, but some studies see patients three times a week in their first six weeks post-operation and then decrease the visits to twice a week afterwards for the next five weeks {Bade, 2017 #317}. Current literature suggests that low-impact activities such as aquatic therapy, ergometer cycling, level-surface walking, group exercise therapy, and continuous passive motion therapy are all acceptable and beneficial activities for increasing clinical outcome scores, muscle strength, and gait speed {Papalia, 2020 #238;Zech, 2015 #220}. Certain clinics begin rehabilitation activities before performing a TKA procedure, but a meta-analysis determined that these pre-surgery efforts only shortened the initial hospital stay as opposed to improving more common outcomes such as pain and knee extension capacity {Su, 2022 #316}. Weight loss programs started after surgery have been shown to be more beneficial than programs started before surgery when compared at 26 weeks post-operation {Pellegrini, 2018 #208}. As previously mentioned, physical activity levels after surgery are lower postoperatively and BMI tends to increase for TKA recipients which highlights the importance of weight loss and cardiovascular components in rehabilitation programs {Arnold, 2016 #207}.

High and low intensity postoperative rehabilitation programs have been compared with no significant changes at 2 or 3 months in variables of interest such as knee ROM, and quadriceps and hamstrings strength {Bade, 2017 #317}. The high intensity group received more balance, agility, weight-bearing, and walking prescriptions than the low intensity group. The main differences for the low intensity group were that the initial rehabilitation focus was on isometric and ROM exercises, a slower transition to weight-bearing activities, no added

resistance beyond body weight or elastic bands, and a slower progression into more difficult weight-bearing activities. Interestingly, the high intensity group was given a goal of 30 minutes of uninterrupted walking before progressing to commonly recommended, lower impact activities, such as swimming and cycling. Bade et al. {, 2017 #317} reported that home exercise adherence for the high intensity group was 7.7% lower which may indicate a decreased desire to perform activities at this level of intensity. While many surgeons tend to advise against high or medium impact activities postoperatively, approximately 25% and 47%, respectively, of TKA patients perform these activities at nearly seven years postoperative follow-up {Papalia, 2020 #238;Mayr, 2015 #326}. High impact activities included activities such as rock climbing, dancing, and tennis while medium impact activities included mountain hiking and Nordic walking. Mayr et al. {, 2015 #326} did not find evidence of component wear or loosening with radiographic follow-up scans for patients participating in activities at any level. Comprehensive review of recommendations for amount physical activity and interventions for reducing sedentary activity has resulted in the conclusion that little or no attention is given to the patient's post-operative physical activity levels {Mooiweer, 2022 #407}. Despite a consensus on specific activities that are encouraged or discouraged, it is important to prescribe amounts and frequencies of activity for this population that tend to have difficulty adhering to activity schedules.

Surgeon recommendations for return to sport time and allowed activities are mostly guided by experience and moderated by higher patient expectations {Vu-Han, 2020 #421}. Vu-Han et al. {, 2020 #421} designed a questionnaire for German surgeons concerning arthroplasty and their management of patients with high expectations to return to sport . Surgeon responses showed that 54.5% considered physical activity 'important' after TKA, 29.7% considered it 'very important,' and approximately 11% stated it was 'unimportant' or 'irrelevant'. 36.6% of

responses did not recommend a return to high impact sports at all, while 53.5% recommended high impact sports after ‘adequate training.’ No further elaboration was provided for what constitutes ‘adequate training,’ which contributes to the lack of unity in surgeon and clinician recommendations for these patients. The results from this survey demonstrate that measuring outcomes and determining rehabilitation efficacy is difficult not only due to implant types and patient factors, but also inconsistent surgeon recommendations and guidelines. Study results including arthroplasty patients from a wide variety of surgeons, physical therapists, and rehabilitation protocols make interpretation and comparisons difficult on a large scale. Future work in epidemiology may consider consolidating some of these variables in order to isolate the effects of specific components of the surgical process, such as rehabilitation.

Satisfaction and Outcomes

Despite recent surgical approach innovations and implant design improvements, significant long-term pain is reported in approximately 20% of TKA patients {Beswick, 2012 #253}. Pain measurement scales commonly used in clinical scoring are the Numerical Rating Scale (NRS) and the Visual Analog Scale (VAS) which have previously been validated to measure pain intensity and measure pain from 0 (no pain) to 10 (severe pain) {Thong, 2018 #262}. These scales measure pain as a single ordinal value. surveys such as the Pain Catastrophizing Scale (PCS) have been developed using multiple ordinally scored questions to identify the psychological and psychosocial aspect of pain in clinical populations. The PCS score ranges from 0 to 52 and consists of 13 questions on a 5-point Likert scale of 0 (never) to 4 (always) {Franchignoni, 2022 #327}. The PCS has been deployed in knee OA populations before and after surgery as a predictor of chronic pain more than three months after surgery {Burns, 2015 #263; Ong, 2021 #264; Birch, 2019 #252}. Birch et al. {, 2019 #252} reported that

patients with a PCS score over 21 possessed poorer physical function and general health before and after TKA than those with a PCS score below 11.

Kinesiophobia is defined as pain-related fear of movement that can alter physical activity levels. Kinesiophobia, as scored by the Tampa Scale of Kinesiophobia (TSK), is a documented barrier to TKA outcomes and it is important to develop strategies for encouraging exercise modalities that are more comfortable and perceived to be less injury prone for these patients {Kazarian, 2021 #215}. When comparing high and low kinesiophobia groups, 60% of pain level and 47% of knee flexion ROM deficits could be explained by the group affiliation {Güney-Deniz, 2017 #218}. Using the TSK, patients can be placed into groups based on improvement trajectory that are predictable based on BMI, heart disease and pain level {Yan, 2023 #265}. Activities for rehabilitation need to be objectively less demanding and subjectively less painful or anxiety-inducing to elicit rehabilitation program adherence which already has multidimensional barriers {Bakaa, 2022 #219}. The American Association for Hip and Knee Surgeons has previously provided a consensus that no activity restrictions should be placed on low-impact tasks such as swimming, level walking, and level bicycling {Swanson, 2009 #239}.

Many additional surveys have been developed to encompass pain, quality of life, and daily functional capacity in OA or TKA populations such as the Western Ontario and McMaster Universities Arthritis Index (WOMAC), the Knee Injury and Osteoarthritis Outcome Score (KOOS), and the Knee Society Knee Scoring System (KSS). The WOMAC encompasses three domains: pain, stiffness, and physical function with ratings on an ordinal 0 (never) to 4 (always) Likert scale for a total of 96 possible points; the majority of which relate to physical function. Pain and physical function subsection reliability, validity, and responsiveness in orthopedic groups have been investigated and deemed acceptable, however, the stiffness subscale has not

shown sufficient test-retest reliability {McConnell, 2001 #328}. The KOOS survey is an extension of the WOMAC that evaluates short- and long-term outcomes in a variety of clinical populations and is a valid, reliable, and responsive survey for TKA populations {Nilsson, 2003 #142}. Additionally, the KOOS has been compared to the WOMAC in TKA populations and evidence has shown that the KOOS has improved validity. The KOOS is a 42-item survey with five subscales: pain, other symptoms, function in daily living, function in sport and recreation, and quality of life. The largest effect size for the KOOS most commonly comes from the quality of life subscale, followed by the pain subscale {Nilsson, 2003 #142}. The KOOS score ranges from 0 (extreme problems) to 100 (no problems). The KSS survey consists of four subscales: knee score, patient satisfaction score, patient expectation score, and functional activity score. The KSS survey was updated in 2012 after its inception in 1989 and has shown moderate-sized Spearman correlation coefficients with the KOOS but did not demonstrate advantages over the KOOS in validity or internal consistency {Culliton, 2018 #331}.

Cycling in Healthy Populations

Benefits

Cycling is a well-studied modality for increasing cardiovascular fitness and a wealth of evidence attests to its benefits for reducing all-cause mortality, cancer morbidity and mortality, coronary heart disease morbidity and mortality, type II diabetes, and obesity {Oja, 2011 #217; Logan, 2023 #216}. Stationary, or indoor, cycling provides the benefit of safety from motor vehicles and offers the same benefits as commuter or road cycling {Chavarrias, 2019 #278; Johan de Hartog, 2010 #279}. Specifically, research indicates indoor cycling has a potential benefit to decrease blood pressure, increased serum HDL, decreased serum LDL, decreased cholesterol, and increased aerobic capacity, but should be combined with diet

interventions to maximize outcomes. In self-reported physical activity surveys, cycling has been associated with weight gain attenuation in middle-aged populations, while other common activities such as slow walking and swimming were not {Littman, 2005 #342}. For TKA populations, cycling provides cardiovascular benefits while decreasing peak resultant knee joint loads compared to walking or other daily activities {Kutzner, 2012 #4}. Moreover, at 40 RPM and a power setting of 90 Watts (W) peak resultant force was approximately 160% bodyweight, on average, compared to the 200-350% BW of other daily activities such as stair descent, stair ascent, standing up, and sitting down {Kutzner, 2010 #408}. As a result, cycling can be recommended to TKA recipients as an excellent way to engage in physical activity while decreasing harmful impact and wear on their implant.

Power and Force Measurement

Rider power output on a bicycle is most frequently measured in Watts as a measure of work per unit time. Early laboratory measurements of cycling power in the 1970's relied on instrumentation such as strain gauges applied to the bicycle cranks {Daly, 1976 #339}. Strain gauges operate on the principle of electrical resistance changes as a result of force application on an object or surface attached to the strain gauge. Force plates using strain gauges in each of the four corners operate on this idea. Once the resistance change has been quantified, it can be translated into force or a variety of other mechanical variables such as tension. Measuring applied force at the pedal requires force sensing sensor(s), proper constructions of a pedal that can accommodate force measuring sensor(s), and proper determination of sensor locations within the pedal. Additional math and calculations are needed in order to determine the crank torque and power, as there is a distance from the measurement point at the crank to the point of force application- the pedal itself. In 1979, a publication demonstrated perhaps the first creation of a

set of pedals that were instrumented with strain gauges to allow for more direct measurement of forces at the rider-pedal interface {Soden, 1979 #340}. Pedal-based strain gauge power measurement systems are still being developed contemporarily and are popular commercially for their ease of transfer between bicycles, wireless capabilities, and affordability {Montalvo-Pérez, 2021 #34;Rodríguez-Rielves, 2021 #35}. Piezoelectric transducers have also been introduced to measure pedal reaction forces (PRF) measured in Newtons (N) triaxially in laboratory settings which has allowed for high resolution analyses of individual pedal cycles {Fang, 2016 #52;Hummer, 2021 #200;Hummer, 2021 #200}. Piezoelectric sensors, however, tend to have an output drift under load for extended periods of time, which is not as prevalent an issue for strain gauge measurements.

Monitoring force applied to the pedal, torque applied to the crank, and power output of the rider is of interest in areas such performance and rehabilitation to track progress or monitor recovery. Not only can force application be quantified, but joint moment estimations can be gained in combination with experimentally measured kinematics during cycling. The assessment of bilateral force application is important not only to assist with performance, but to evaluate risk for pathologies such as knee OA which may affect one limb more than the other. Joint moments are a critical component of investigating human movement and especially for evaluation of interventions that have a clinical relevance. For example, without the measurement of pedal force it would be impossible to determine the effects of cycling interventions on the magnitude of sagittal plane knee moments. Determining the power output of the rider is typically done by taking the product of force tangential to the crank, crank length, and crank angular velocity {Bini, 2014 #324}. Formulas to calculate power output vary depending on the modality of measurement device as some sensors provide triaxial force, normal force, or shear forces. Some

examples of sensor types are strain gauges, capacitive load cells, chain tension sensors, and piezoelectric sensors {Bini, 2014 #324}. Sensor type when combined with the use of cleats is important to understand because cleats restrict the user to a specific position on the pedal while toe clips allow for more variance in foot position on the pedal {Bini, 2014 #324}.

Bike Fit and Modification

Fitting the bicycle to the individual is a critical component of acquiring accurate and reproducible kinematic and kinetic data because adjustments in bike components elicit different biomechanical responses. Bike fit component adjustment includes, but is not limited to, modifications in body position, saddle (seat) position, handlebar position, and crank arm lengths. Additional factors of interest that alter biomechanical responses in cycling are cadence, commonly reported as crank RPM, and resistance such as workload and workrate, commonly reported as kilogram (kg) or N of braking force or Watts {Fang, 2016 #52;Bini, 2010 #108}. While no explicit bike fit consensus has been published, an overview of a variety of methods has been presented below.

Overall upper body position during cycling has an important effect on cycling in terms of power output and knee and hip joint work {Bini, 2014 #324;Hubenig, 2011 #281}. Hubenig et al. {, 2011 #281} recruited females with aerobic training and reported that a decreased power output resulted from an aerodynamic position (152.7 ± 28 W) compared to an upright position (159.7 ± 33.1 W). An aerodynamic position is commonly adopted in competitive cycling and means the rider is in a more hunched position with greater trunk and hip flexion. Bini et al. {, 2014 #280} performed a more comprehensive biomechanical analysis on body position by comparing preferred, most forward, and most backward positions. The positions were defined by the rider moving their hips forward or backwards on the saddle and did not include movement of

the saddle itself in the AP direction. Two-dimensional instrumented pedals were used to record applied pedal force and electromyography (EMG) was implemented to determine muscle activity changes while kinematics were recorded and synchronized with an external trigger. A forward position resulted in significant differences in knee joint angle, knee joint work, rectus femoris activation, and hip joint work while no significant changes in ankle kinetics or kinematics were discovered {Bini, 2014 #324}. Knee range of motion was reduced by 8% and knee work increase by 6% at the most forward position on the saddle compared to the backward and preferred position. Hip work was reduced by 28% at the forward position compared to the preferred position. These results show a general trend that the knee contributes more to each revolution of the crank at a more forward position and, therefore, that the body's position on the saddle must be considered for populations with hip or knee joint concerns.

Saddle height adjustments are critical considerations for certain patient populations as decreased saddle height has been shown to increase patellofemoral compressive forces, likely as a result of increased anteroposterior distance of the knee from the pedal {Ericson, 1987 #283}. A wide range of methods for determining saddle height have been proposed in the past, with many of them falling into categories based on limb length, trochanteric height, inseam length, or knee joint angle measurements {Bini, 2011 #105}. Named methods that appear frequently in the literature are the LeMond method, the Hamley and Thomas method, and the Holmes et al. method {Hamley, 1967 #332; Holmes, 1994 #333}. The LeMond method requires the distance from the top of the saddle to the center of the bottom bracket to be 88.3% of the inseam leg length of the rider {Bini, 2011 #105}. While not scientifically developed, the method was used by multiple Tour de France champion Greg LeMond, lending some credence to the methodology. The Hamley and Thomas method simply requires the saddle height to be 109% of

the inseam leg length. The Holmes et al. method utilizes a knee flexion angle or range of angles at bottom dead center of the pedal stroke to determine saddle height for a variety of clinical populations. Specifically, Holmes et al. {, 1994 #333} recommended knee flexion angle ranges of 25°-30° for quadriceps tendinitis and patellofemoral ligament injuries while 30°-35° is recommended for iliotibial band syndrome and biceps tendinitis.

Whole body kinematics on a bicycle are inherently tied to the saddle, or seat, height and even minor saddle height changes have shown changes in cardiovascular function and lower body kinematic and kinetic behavior {Ferrer-Roca, 2014 #282;Bini, 2010 #108;Bini, 2011 #105}. Ferrer-Roca et al. {, 2014 #282} adjusted saddle heights $\pm 2\%$ from a preferred position in trained cyclists and found that VO₂ max was negatively impacted by higher saddle height and that ROM of the hip, knee, and ankle increase by 1°, 3° and 4°, respectively with increased saddle height. Concurrent with the ROM changes, only the knee and ankle angular velocity were significantly affected by the higher compared to preferred and lower seat heights. Adjustments in muscle work, joint ROM, and energetics are unsurprising, as saddle height changes muscle lengths, joint angles, and muscle moment arms, which in turn alter length-tension and force-velocity relationships of the involved musculature {Delp, 1990 #465}. Bini et al. {, 2010 #108} used the trochanteric height method to determine seat height and altered in by $\pm 3\text{cm}$ for a higher and lower condition in male participants without competitive cycling experience. With increasing saddle height, the ankle joint work contribution to total work increases significantly, the knee joint work contribution decreases, and the hip joint work contribution was not affected {Bini, 2010 #108}. From low to high saddle height positions, the ankle work contribution increased from approximately 3.5% to 4.25% while the knee joint contribution changed from approximately 42% to 39%. The authors proposed that the knee work reduction was beneficial

for soft tissue off-loading and that consideration should be given to ensure saddle height is not lowered 3 cm below trochanteric height in populations with knee injury or local anterior pain {Ericson, 1987 #283;Bini, 2011 #105;Hamley, 1967 #332;Bini, 2011 #105}.

Following the presentation of the various saddle height methodologies, it has been debated amongst the community which is most optimal, and a review done by Bini et al. {, 2011 #105} summarizes key findings and missing information in cycling literature relating saddle height to injury risk and performance. It should be noted that ‘most optimal’ bike fit depends on the population in question as professional cyclists do not have the same priorities or considerations that clinical populations do. Limb length measurements to determine saddle height can be problematic due to the variability in lower limb morphologies of the femur, tibia, and feet {Peveler, 2011 #423}. Even a 4 or 5% increase or decrease in saddle height can result in up to 25% difference in knee ROM {Sanderson, 2009 #424}. Muscle activity changes due to saddle height do not provide conclusive evidence for saddle height recommendations alone, but increased quadricep and hamstring activity has been shown when changing saddle height from 100 to 95% trochanteric height {Jorge, 1986 #356}. The review by Bini et al. {, 2011 #105} concluded that cycling saddle height should be set to elicit 25-30° of knee flexion to minimize injury and oxygen uptake.

Bini et al. {, 2013 #323} defined backward and forward positions abstractly as a hill-climb or time-trial position, respectively, while maintaining a consistent upper body angle. The displacement at the backward to forward positions were approximately 0.28 to 0.39m, respectively. Saddle fore-aft position, or saddle setback, is an additional saddle modification that involves the anteroposterior movement of the rider or saddle. The predominant kinematic outcome of the saddle or rider moving anteriorly without moving the saddle itself is a 5-6°

increase in knee flexion angle, while kinetically, tibiofemoral shear force increases in the range of 19-26% as the rider moves from backward to forward positions {Bini, 2013 #323}. Some literature calculates and reports force and work effectiveness indices to allow for bike fit modifications to be compared {Menard, 2016 #325}. Menard et al. {, 2016 #325} defined force effectiveness as the ratio of force perpendicular to the crank arm to total force applied to the pedal and work effectiveness simply as the minimization of crank work. The authors concluded that sitting further back on the saddle decreased total work by 5% but increased the work and force effectiveness compared to sitting forward. The application to performance of this conclusion is that the forward position lends to maximal power output, but the more backwards position leads to greater effectiveness during steady state pedaling. Further work on saddle position utilizing musculoskeletal modeling approaches demonstrate 14 and 15% increases in mean and peak tibiofemoral compression forces, respectively, in backward saddle positions compared to forward saddle positions {Menard, 2020 #338}. It can be concluded that a saddle position too far forward or too far backward can create high, unwanted patellofemoral or tibiofemoral contact forces. Therefore, proper adjustment of the rider on the saddle in the anteroposterior direction is of vital importance to populations where tibiofemoral force magnitudes are a concern.

Handlebar position on a bicycle is a secondary component that affects body position and methods have been implemented using handlebar position to alter hip flexion angles during sub-maximal cycling to determine bike fit {Bini, 2020 #285}. Bini et al. {, 2020 #285} reported that sub-maximally determined bike fit using handlebar position to alter hip joint angles was not maintained during sprinting trials and therefore this method may not be suitable when fitting for varying workloads. Common handlebar positions for cyclists include the tops, drops, and hoods

of the handles. The tops position consists of holding on to the handlebar portion that is perpendicular to the main bike's tube. The drops position consists of holding on to the lower curved portion of the handlebars behind the brakes of a racing bike. The hoods position consists of holding on to the handlebars directly above the brake levers on a racing bike. However, these positions are not available on all cycle ergometer models and literature results should be interpreted carefully and caution should be taken when extrapolating to clinical ergometers without these handlebar configurations. Across external work rates of 100 W, 200 W, and a varying Wattage matching participant lactate threshold, no main effect of handgrip position on lower limb joint powers were found when changing hand positions between tops, drops, and hoods positions {Skovereng, 2020 #284}. However, the drops compared to tops position did result in a significant decrease in knee joint work. For recreational riders, the trunk flexion angle changed for the tops, hoods, and drops positions from approximately 47° to 43° to 35°, respectively. The change in hip angle for recreational cyclists going from tops to drops resulted in a decrease from approximately 104° to 97°. Work done to investigate bicycle configurations and flexibility has demonstrated that lowered handlebar positions were associated with commensurate increases in peak power output in trained cyclists {Holliday, 2021 #427}. However, it should be noted that the R^2 value of the correlation was 0.19 for this correlation relationship. The authors conjectured that the lower handlebar setting for riders relates to hamstring flexibility which could produce a more powerful position. Lowered handlebars likely require the rider to tilt the pelvis anteriorly which in turn increases the length of the hamstrings and shifts the force capabilities on the length-tension curve. Quantification of this shift would be considerably complex due to the biarticular nature of the hamstrings.

Three of the most common pedal types are clipless, platform, and toe cage pedals. Clipless pedals are used by road cyclists and the majority of triathletes and require special shoes that physically integrate with the surface of the pedal {Cruz, 2001 #343}. This pedal design allows for force application to the pedal through both the power phase and recovery phase of a single cycle. The ability to pull the pedal up using the hamstrings during the recovery phase implies greater muscle activity during the recovery phase, however, research on triathletes showed decreased semitendinosus and semimembranosus activity for clipless compared to toe clip pedals {Cruz, 2001 #343}. This research should be interpreted with caution on account of the small sample size and unreported power setting for the 100 RPM trials. Platform pedals are commonly used by mountain bikers and allow for complete freedom of foot placement on the pedal. Toe cages, toe clips, or metatarsal strap pedals provide varying levels of restraint of the foot inside the pedal and are seen in recreational and clinical pedals. Monark ergometer pedals tend to have a strap over the metatarsals to provide some restriction, while other pedals employ a full toe cage that provides more consistent foot position between participants. It has been demonstrated that platform and toe-clip pedals do not significantly affect VO₂ at 60 RPM with power settings of 60, 120, 180, and 240 W {Ostler, 2008 #344}. The previously discussed pedals are designed to allow specific movements of the rider's foot. For example, the platform pedals allow the rider to place their foot at mostly any position along the pedal's plane. Toe cage pedals somewhat restrict movement to a primarily AP plane. A pedal's allowable DoF should be considered, especially for clinical populations, as a two DoF pedal has been shown to reduce valgus knee moment when compared to a one DoF pedal during stationary cycling at 90 RPM and 250 W {Boyd, 1997 #345; Wootten, 1992 #347}.

Electromyography

Surface EMG is commonly used in laboratory and clinical settings to record muscular activity for the purpose of assessing muscle coordination patterns, activation magnitudes, and fatigue assessment in cycling. Standardization of EMG electrode placement is a critical component of methodological rigor, and most researchers abide by the Surface EMG for a Non-Invasive Assessment of Muscles (SENIAM) guidelines presented after development in 2000 {Hermens, 2000 #349}. The SENIAM guidelines provide instructions for electrode placement using percentage distances between specific bone landmarks as well as muscle fiber direction to place them. Modern wireless EMG sensors have minimized the non-physiological signal effect of wire and sensor movement, but crosstalk is still an important phenomenon to be aware of. Crosstalk is when EMG sensors pick up electrical signals from nearby muscular activity that does not originate from the muscle of interest. Crosstalk is an important consideration when interpreting EMG signals as crosstalk and co-contraction may sometimes be indistinguishable {Talib, 2019 #426}. A secondary concern when using surface EMG electrodes is the amount of adipose tissue present around the muscle of interest. Characteristics such as BMI, waist circumference, and sex influence adipose tissue density and distribution. It is known that increased adipose tissue beneath an EMG sensor decreases the recorded signal which can lead to higher normalized EMG values in participants with more adipose tissue, making comparison validity questionable {Arauz, 2019 #84;De Luca, 1997 #357;Ptaszkowski, 2019 #359}. Raw EMG signals are essentially uninterpretable and producing a linear envelope to further obtain the root mean square (RMS) values is recommended and frequently used over other proposed processing methods such as integrated EMG {Basmajian, 1985 #350;Dorel, 2008 #351}. EMG

processing generally includes a Butterworth bandpass filter, rectification of the signal, and the application of a moving RMS.

Normalization of EMG signals for comparison between dynamic tasks is difficult and many publications have recommended maximal voluntary isometric contraction (MVIC) trials to determine the theoretical highest level of muscle activity {Marsh, 1995 #353;Ericson, 1986 #352}. The MVIC attempts to elicit a maximal contraction from the participant, which theoretically includes the greatest number of motor units available for recruitment. Despite this popular method, it may not be advisable for certain tasks that place joints at different positions than they were during the MVIC trials, such as cycling. For example, dynamically recorded activity levels have been reported over 100% of the MVIC during maximal cycling which indicates that the MVIC trial was insufficient to produce enough muscular activity {Hunter, 2002 #354}. However, this finding was derived from a study on cycling fatigue protocols and therefore is likely not generalizable to clinical cycling protocols which are not often designed to induce maximal fatigue. Other methods involving maximal isokinetic or all-out trials for EMG normalization during cycling have been proposed and shown good within-subject reliability, but validity of these methods have not been assessed rigorously {Fernández-Peña, 2009 #366}. It is vital to not conflate increased muscle activity as a direct signal of increased muscle force output because muscle activity increases with fatigue given a constant external load or task {Tabasi, 2023 #355}. Some researchers have employed EMG to force ratios in an attempt to quantify the relationship between signal amplitude and external force output in cycling. During 50 RPM sub-maximal cycling tests performed after sprints, the EMG to force ratio of agonist musculature, such as the vastus lateralis, increased while antagonist muscles decreased their activation during fatigued states, indicating a reduction in co-contraction {Hautier, 2000 #364}.

A selection of common muscles to assess during cycling are the vastus medialis (VM), rectus femoris (RF), biceps femoris (BF), semitendinosus (ST), medial gastrocnemius (MG), and the tibialis anterior (TA). Previous cycling research at 120 W demonstrated that EMG activity of biarticular muscles such as the RF were 22% lower than its MVIC values whereas monoarticular muscles such as the VM were 45% lower than the respective MVIC {Ericson, 1986 #352}. Biarticular musculature such as the RF and the BF display double burst activations during each cycling phase (**Figure 8**, all figures are in the appendix) {Suzuki, 1982 #365}.

The VM activation onset is just before top dead center (TDC) while the RF onset is earlier in the pedal cycle at approximately 270° and offset occurs around 90° (Appendix B, **Figure 9**) {Dorel, 2008 #351}. ST muscle activity behavior is more variable between studies and has been shown to have ranges of activity from TDC to BDC or from TDC to 270° {Dorel, 2008 #351; Jorge, 1986 #356}. In a review of pedaling EMG, the authors reported that this discrepancy in ST activation may be a result of differences in bike fit components such as body position or shoe-pedal interface {Hug, 2009 #348}. GM activity has an onset to offset range of approximately 30°-270° {Dorel, 2008 #351}. The TA activity primarily occurs between bottom dead center (BDC) and TDC, specifically starting at 270° to just after TDC, as the ankle actively dorsiflexes during the recovery phase {Dorel, 2008 #351}. Unsurprisingly, increasing workloads during cycling increase lower extremity EMG amplitudes significantly at workloads of 60, 80, and 90% of participant's maximum heart rates {Holliday, 2023 #360}.

Cadence

Cadence is the RPM of the crank arm and studies have investigated ranges of 40 to 120 RPM for rehabilitation, recreational, and competitive purposes {Wilkinson, 2020 #297; Aasvold, 2019 #214; Bini, 2010 #108}. Lower extremity joint work is one of many sets of variables related

to cadence that are often reported to assess the concentric and eccentric activity of muscle groups during movement. Joint work is most often calculated by first multiplying a joint angular velocity by its corresponding joint moment to obtain a measure of mechanical power for each frame of data throughout the crank cycle. Then, the integral of this curve is taken through a variety of methods such as the trapezoidal method in order to compute joint work in Joules, or Joules per kilogram. It should be noted that joint work is usually not normalized by body mass because the saddle alters the effective weight of the participant. In cycling trials performed by healthy individuals with no competitive cycling experience, ankle joint contributions to the total lower limb work increase with cadence when comparing 40 to 70 RPM {Bini, 2010 #108}. Bini et al. {, 2010 #108} did not report any significant changes in knee or hip contributions to total lower limb work at these cadence ranges. The approximate work contributions to the total lower limb work across both cadences were 5, 40, and 55% for the ankle, knee, and hip, respectively. While differences in the relative contributions were found, no effect of pedaling cadence was observed for the hip, knee, or ankle mechanical work. Other literature reports that absolute hip and knee joint power (W) differences begin to occur at 60 RPM in a study that assessed cadences from 40 to 100 RPM {Aasvold, 2019 #214}. However, it should be noted that Aasvold et al. {, 2019 #214} differentiated cyclists by skill level (elite and recreational) which demonstrated a differing joint power contribution pattern, suggesting that experience level may affect this relationship. When assessing all participants, this work still found an increase in knee joint contribution and decrease in hip joint contribution to joint work as cadence increased above 60 RPM. Elite cyclists tend to prefer higher cadences around 90 RPM when allowed to freely choose cadence during level ground time trials of 50km and flat, long group stages of 190km

from actual professional cyclists' race data. Therefore, this may not be an adequate population to use when interpreting results for other populations that cycle recreationally {Lucía, 2001 #298}.

Cadence increases from 70 to 120 RPM can result in an approximately 17% increase in hip joint power contribution to net joint power, while decreasing knee power contribution by approximately 9% in agreement with Aasvold et al. {, 2019 #214} {Wilkinson, 2020 #297}. *In vivo* hip joint loads during ergometer cycling revealed that a cadence setting at 40 RPM elicited a greater joint reaction force compared to 60 and 90 RPM, with 60 being the most ideal for decreasing hip joint loads {Damm, 2017 #300}. Cadence increases from 60 to 100 RPM have been shown to increase relative knee moment contribution by only 4% ($P < 0.05$), with increasing power, while the knee decreases its relative moment contribution when increasing cadence {Mornieux, 2007 #299}. When investigating cadences of 70, 80 and 90 RPM, the increasing cadence has been reported to have no effect on peak knee abduction moment, but did elicit an increased peak knee flexion moment when cycling on a pedal with a toe cage {Fang, 2016 #52}. No significant kinematic differences were reported from knee extension ROM or peak adduction angle across cadences. The knee flexion moment significantly increased from -16.69 Nm at 60 RPM to -21.51 Nm at 90 RPM. The knee extension moment did not significantly increase in this range but was 20.23 Nm and 24.50 Nm at 60 RPM and 90 RPM, respectively. Additionally, greater anterior PRFs were found to occur with increasing cadence from 34.25 to 52.04 N at 60 and 90 RPM, respectively. These findings from Fang et al. {, 2016 #52} suggest the possibility of different frontal plane knee moment patterns being exhibited by participants. Moreover, approximately 39% of the study's participants displayed an abduction moment while the remaining 61% displayed an adduction moment during the power phase. The breakdown of participants by sex was not listed in this study, but the age range of 40 to 79 was given which

may provide some explanation for these patterns due to cartilage wear from age or sex-specific differences that were unmeasured. While participants were deemed healthy for the inclusion criteria, without radiographic verification of cartilage health, it is difficult to rule out the effects of potential age-related joint health changes.

Quantification of limb function asymmetry has been of interest since the mid 1970's and has since been studied extensively for clinical gait and cycling as well as performance applications in cycling {Bini, 2015 #322;Daly, 1976 #339;Gale, 2019 #335;Hunt, 2004 #228;Smak, 1999 #98}. Competitive cyclists performing a 4-km time trial demonstrated that larger limb effective force asymmetries between limbs were related to improved performance while also exhibiting low intra-limb variability {Bini, 2015 #322}. Force asymmetry was quantified using an index of effectiveness defined as the ratio between angular impulse of the tangential force applied to the crank and linear impulse of total applied force to the pedal. Normal and anteroposterior forces were measured via strain gauge instrumented bicycle pedals, and angular potentiometers allowed for measurement of the angle of the pedal to crank angle. 11-21% more total force was applied by the dominant limb during these time trials. During cycling in 11 male competitive cyclists at a constant 250 W and varying cadences of 60, 75, 90, 105, and 120 RPM it was reported that the dominant leg produced between 0.5 and 2.0% more total average crank power ($p = 0.04$) than the non-dominant leg {Smak, 1999 #98}. Additionally, it was found that the dominant leg produced significantly less positive hip power because of an increase in non-dominant leg hip extensor moments during the power phase. The general trend of asymmetry increases from approximately 6 to 16% were with an accompanying increase in cadence of 60 and 120 RPM, respectively.

Workload and Workrate

The effects of workload and workrate on the metabolic and biomechanical variables at a given cadence have been widely studied, but often in skilled populations or at considerably higher power settings than most clinical populations would perform. Observational studies of cycling kinematics in healthy populations have provided useful ROM data and comparisons amongst age and sex groups {Yum, 2021 #318}. Yum et al. {, 2021 #318} provided a comprehensive kinematic overview demonstrating sagittal ROM of 43.9°, 75.2°, and 26.9° at the hip, knee, and ankle, respectively. Additionally, no kinematic differences were found between age groups of approximately 18 to 68 years old or between males and females in any of the three planes of movement. Results from Yum et al. {, 2021 #318} and Bini et al. {, 2010 #108} generally coincide with the latter reporting typical hip, knee, and ankle ROM of 45°, 75°, and 20°, respectively. Knee extension ROM appears to significantly change with increasing workloads at a constant cadence, but does not significantly change given a constant workload with increasing cadence {Fang, 2016 #52}. However, Fang et al. {, 2016 #52} reported an approximate knee extension ROM of 76° to 80° with approximately 5-6° of standard deviation across multiple workloads and cadences, similar to findings reported by Yum et al. {, 2021 #318}.

Pedaling on a cycle ergometer at 0, 120, and 240 W revealed that total work per revolution increases with workrate and that the proportions of hip and ankle positive work decreased as workload increased {Ericson, 1988 #29}. Positive knee extensor and knee flexor work increased across these workloads in compensation. Cycling at 120 W at 60 RPM revealed 28.8 Nm and 11.9 Nm of peak knee extension and flexion moments, respectively {Ericson, 1986 #5}. Importantly, Ericson et al. (150) reported that hip moments were significantly affected by

cadence and workrate, while knee moments were only affected by workrate. This finding suggests that workrate should be more carefully considered in populations where knee health, comfort, and performance is of concern, such as knee OA or knee joint arthroplasty populations. Using workloads from 0.5 to 2.5 kg at a constant 60 RPM resulted in an increase in peak knee abduction moment from 5.82 to 14.36 Nm with a concurrent knee extension moment increase from 11.61 to 37.16 Nm {Fang, 2016 #52}. Fang et al. {, 2016 #52} also found that medial PRF, vertical PRF, and knee extension ROM were all significantly different among the range of workloads. Knee extension ROM increased by approximately 3.5° from 0.5 to 2 kg while the abduction moment increased from -5.82 to -14.36 Nm. While recommendations for limiting frontal plane moments during activity do not exist, these results suggest that kinetics at the knee can greatly change with workload. In combination with potentially long-term cycling sessions, workload should be taken into account when recommending rehabilitation and exercise regimens to clinical populations.

An investigation of increasing workload at a constant cadence of 80 RPM in varying tibiofemoral alignment groups reported knee kinetics for the neutral alignment group for workloads of 0.5, 1.0, and 1.5 kg which is equivalent to approximately 40, 78, and 117 W with 80 RPM, respectively {Shen, 2018 #473}. Shen et al. {, 2018 #473} reported neutral alignment group peak knee extension moments of 23.5 to 42.5 Nm for workloads from 0.5 to 1.5 kg conditions, respectively, which are higher than those reported by Fang et al. {, 2016 #52} who reported 11.61 and 26.04 for those conditions. Peak knee abduction moments from Shen et al. {, 2018 #473} also reported neutral alignment group values of -4.4 to -8.7 Nm compared to the Fang et al. {, 2016 #52} reports of -5.82 to -10.18 Nm. No mention of alignment is present in the Fang et al. {, 2016 #52} paper, which may be a partial contributor to the difference seen, but it is

more likely that the RPM difference between the studies contributed to this difference. Additionally, since moments are not generally normalized during cycling, there is a potential effect of limb weights and general cardiovascular condition of participants to consider. However, values reported by Ericson et al. {, 1988 #29} at 120 W and 60 RPM are approximately similar to values from Shen et al. {, 2018 #473} at approximately 40 W and 80 RPM. The interaction of workload and cadence are complicated and may be further exacerbated by knee alignment of the participants, which can be unreported, making comparisons more difficult.

During simulated 40-km time trials, the dominant limb produces greater peak crank torque, and it was reported that increasing exercise intensity stages resulted in decreased kinetic pedaling asymmetries {Carpes, 2007 #362}. Conversely, when pedaling at ranges of 100 to 250 W, peak torque output asymmetries became larger in favor of the dominant limb as measured by a set of instrumented pedals {Bini, 2014 #363}. Caution is again warranted with these interpretations as they were done in a small set of competitive cyclists. Further research in time trial kinetic asymmetry supports work done by Bini et al. {, 2014 #363} and demonstrated increasing pedal asymmetries linked to improved performance in 500m sections of a 4-km time trial {Bini, 2015 #322}.

Crank Arms

Bicycle crank arm inertia varies from ergometers to road bicycles which tend to be low and high, respectively, because of gear ratios. With an increase in speed, higher gear ratios in road bikes are required to maintain a given cadence, which results in an increase in crank inertial load {Hansen, 2002 #28}. Understanding the effects of inertia during cycling is important for transferring the findings of laboratory or clinically based findings from ergometers to the general population that commutes, competes, or exercises recreationally with road bicycles. When

emulating low and high inertial loads, no significant differences in hip or knee joint moments for either flexion or extension were discovered while the ankle exhibited a significant small, increase in plantarflexion moment in the power phase with higher inertia {Fregly, 1996 #295}. Further, there were significant decreases in kinematic variability within and between cycles with greater inertia concurrent with significant unilateral torque variability increases. These findings suggest that cadence is more easily maintained with higher inertia crank loads. As a result, increased crank inertia may be beneficial for populations with neuromuscular coordination difficulties. To further support these conclusions, another study recruited competitive cyclists for low (2.4 kgm^2) and high (94.2 kgm^2) inertia conditions to investigate metabolic and kinetic effects during seven minute bouts of cycling {Edwards, 2007 #296}. Edwards et al. {, 2007 #296} concluded that, given constant cadence, no significant differences were found between inertia conditions for oxygen uptake, blood lactate concentration, mean peak crank torque, or mean minimum crank torque. The torques measured in this publication were a result of an ergometer equipped with an SRM system that is a combination of gears and flywheels.

The bicycle crank arm extends from the crank to which the pedal is attached via the pedal spindle, and common commercial lengths range from 170 to 175mm for small and large bike frames, respectively {Gatti, 2022 #334}. Previous literature investigating maximal cycling activities has shown no difference in relative hip, knee, or ankle joint powers as a result of crank arm length once pedal rate and pedal speed have been considered {Barratt, 2011 #286}. Barratt et al. {, 2011 #286} reported that a significant effect on hip and knee joint power was observed when comparing the extremes of the crank lengths (150 to 190mm) given constant pedaling rate. The proportion of total lower limb joint power produced by the knee was approximately $12\% \pm 8\%$ across all crank lengths. Hip and knee joint excursions were increased by approximately 10°

from 150 to 190mm crank arm lengths while the ankle changed by approximately 1° , but with higher variability. Joint angular velocities of the ankle, knee, and hip all increased with increased crank arm length as the pedal must make a larger arc in the same unit of time given a constant cadence. Barratt et al. (2011) reported joint angular velocities in degrees per second, but it should be noted that for other calculations such as power, radians should be used. In a study that recruited 15 trained cyclists, submaximal trials were performed at 90 RPM and a workrate of 240 W for crank lengths of 150, 165, 170, 175, and 190 mm {Barratt, 2016 #508}. Given a constant pedal speed, increasing crank lengths did not result in changes to hip, knee, or ankle flexion or extension moments. Peak hip extension power increased from an average of 161 ± 42 W to 201 ± 54 W for the 150mm and 190mm crank lengths, respectively, occurring around 90° in the pedal cycle. Peak hip flexion power increased from an average of 25 ± 27 W to 42 ± 27 W for the 150mm and 190mm crank lengths, respectively, occurring around 270° in the pedal cycle. Participants were given familiarization time with the 150mm and 190mm crank lengths as they are not standard lengths and were not given practice with the other crank lengths. Despite this, there seems to be a large standard deviation in the peak hip flexion power that is likely a result of using the participant's own bike measurements to determine the ergometer fit. Cyclists performing pedaling at 150, 200, and 250 W across crank lengths of approximately 167.5 to 180mm at 90 RPM exhibited no metabolic differences based on crank lengths {Ferrer-Roca, 2017 #292}. Crank length ranges were set adding or subtracting 5mm from the participant's preferred crank length, the range of which was 172.5-175mm. Further, Ferrer-Roca et al. {Ferrer-Roca, 2017 #292} reported that hip and knee range of motion significantly increased with longer crank arms. Knee range of motion increased from the shorter crank arm, subtracting 5mm from preferred length ($72.6^\circ \pm 3.9^\circ$) to the longer crank arm, adding 5mm to the preferred

length ($75.5^{\circ} \pm 3.6^{\circ}$), conditions. Another study recruited trained cyclists for maximal cycling trials using crank lengths of 120, 145, 170, 195, and 220mm to assess the effects on maximum power output {Martin, 2001 #30}. Optimal crank arm lengths were determined to be at the 145 and 170mm conditions and extending beyond those ranges decreased power production. A study that assessed crank arm length changes via a 30-second Wingate with lengths of 110, 145, 180, 230, and 265mm described significant ROM increases in the lower limb concurrent with crank arm length increases {Too, 2000 #294}. The optimal crank length from this study was determined to be 180mm, but this study used considerably larger crank length increments of 35mm than other literature using 5-20mm increments. As a result, common crank lengths of around 170mm were not included. It is likely that the “optimal” crank arm length was misidentified due to the lack of sensitivity in length changes.

While symmetric crank arm length changes have been investigated thoroughly, asymmetric changes in crank arm length have scarcely been studied and could be relevant to clinical populations where equivalent power production or joint loading is not possible or is undesirable between limbs. Three male cyclists with lower limb length differences equal to or greater than 10mm were recruited to perform stationary cycling tests with the shorter limb having a changing crank arm length of 170, 165, and 160mm while the contralateral limb had a consistent crank arm length of 170mm {Millour, 2020 #226}. Millour et al. {, 2020 #226} reported that knee ROM was significantly decreased in the shorter limb because of a lower peak knee extension angle from the shortened crank conditions. Additionally, perceived exertion during testing was lower while comfort and maximal power output improved with the asymmetric crank arm conditions. All three participants reported that the 160mm crank arm was the most comfortable for them compared to the 165 and 170mm options. A significant limitation

of this study was that the motion capture data was two dimensional and as such, no information of frontal or transverse plane changes or compensations could be obtained. When asymmetric crank arms have been utilized in transtibial amputee populations, similar ROM decreases are found at the hip and knee when using a crank arm 10mm shorter on the amputated side of the individual {Childers, 2014 #225}. Childers and Kogler {, 2014 #225} used custom made force pedals to conclude that while kinematic symmetry was closer with the asymmetric crank arm lengths, kinetic symmetry was not achieved. Specifically, pedal work, pedal force, and peak sagittal knee joint moment asymmetries were not significantly changed between the control and asymmetric crank arm conditions. In fact, the authors reported that the amputated limb had a significant decrease in knee joint moment which exacerbated the already existing kinetic asymmetry with the control limb.

Cycling in OA and TKA Populations

Knee OA Populations

Pre-TKA patients tend to have end-stage diagnosed knee OA as the cause of their TKA. While cycling for knee OA patients cannot provide permanent relief, eight-week interventions of cycle ergometry have shown decreases in perceived pain and increased in KOOS scores, six-minute walk tests, and time up-and-go tests {Kabiri, 2018 #498}. Older knee OA populations have been assessed during four cycling combinations at 75 or 100 W and at 60 or 90 RPM for interlimb asymmetry. It has been reported that knee OA groups demonstrated an approximately 10% power output difference, favoring their affected limb, over 60 crank cycles {Buddhadev, 2018 #89}. Healthy controls did not exhibit this discrepancy and only had an average of 0.1% power output difference between limbs. Importantly, Buddhadev et al. {, 2018 #89} described that asymmetry amount was unaffected by workload and cadence combinations described above.

However, given their observational nature, these data do not allow us to determine if the presence of knee OA causes, or is caused by, the kinetic asymmetry seen in this study.

The literature on observational cycling biomechanics in knee OA populations is sparse, but some acute intervention work has been conducted using lateral shoe wedges and toe-in angles {Gardner, 2016 #106; Fang, 2016 #52; Thompson, 2020 #291}. An initial study in a series of studies recruited participants with at least one knee with a radiographically diagnosed KL score of 1 or higher. At 60 RPM and 80 W, participants cycled with neutral, 5°, and 10° of toe-in angles {Fang, 2016 #52}. Knee flexion angle, knee adduction angle, and knee internal rotation moments were significantly different based on condition. Knee flexion angle decreased from 44.9 in the neutral condition to 41.6° in the 10° toe-in condition while the internal rotation moment of the knee changed from 8.0 to 6.1 Nm. KL score was found to be a significant predictor for peak KAbM across all three conditions. It is likely that the observed knee flexion angle decrease, and commensurate knee adduction angle change were a result of the toe angle shifting the knee to have more frontal and transverse behavior. The purpose of the study was to quantify changes in peak KAbM, and pain and it was concluded that only KAbM was significantly improved. In a secondary analysis, the same OA participant pool placed 5 and 10° lateral wedges on the ergometer pedal surface to additionally investigate KAbM and pain during cycling {Gardner, 2016 #106}. Peak vertical PRFs were significantly higher in the wedge conditions and only peak knee flexion angle and peak KAbM were significantly different based on wedge condition relative to neutral. While the authors reported that KAbM decreases were favorable, without an accompanying decrease in pain or adduction angle, the practical implementation may not be warranted on account of the potential concurrent increase in knee extensor moment.

TKA Populations

Cycling is an activity that reduces required weight-bearing loads of the knee which decreases tibiofemoral forces compared to gait while maintaining cardiovascular benefits such as lowered all-cause mortality, coronary heart disease risk, and obesity {Kutzner, 2012 #4; Logan, 2023 #216; Oja, 2011 #217}. Early in the TKA rehabilitation process, stationary cycling has been implemented to facilitate ROM increases, pain improvement, and patient reported outcome scores via surveys {Zech, 2015 #220}. Zech et al. {, 2015 #220} reported that knee ROM improved, pain decreased, and WOMAC scores for pain, stiffness, and function improved with an exercise regimen started approximately 2 weeks after TKA. This program included two to five instances of cycle ergometer training per week with an average Borg scale RPE intensity of 11.6. An analysis of covariance determined that these findings were not explained by either the average intensity or duration of therapy regimens. Rehabilitation protocols implementing slightly alternative cycling modalities such as those assisted by a flywheel have been investigated in patients 55 to 80 years old after unilateral TKA {Sanzo, 2021 #497}. No statistically significant changes were found in standard rehabilitation protocols or the additive cycling protocol for knee pain, knee extension ROM, knee strength, or blood biomarkers. However, 82% of the cycling participants reported they would be comfortable exerting themselves more during the exercise, which leads to the belief that a minimum level of exercise intensity may be necessary to elicit significant beneficial effects during rehabilitation.

Continuous x-ray images of ergometer cycling in patients at least six months after TKA provided evidence that stationary cycling is encouraged over activities such as golf in order to maintain acceptable loading of the implant's polyethylene insert {Hamai, 2008 #319}. Qualitatively, Hamai et al. {, 2008 #319} demonstrated that TKA implant flexion angle

excursion in the sagittal plane was favorable and that axial rotation was acceptable during cycling. The implant flexion angle was determined by the flexion (positive) or extension (negative) of the femoral component relative to the tibial component. Axial rotation of the implant from top dead center to bottom dead center was 6.7° and the maximal implant flexion angle was 100° over a complete crank cycle.

One of the first studies to utilize instrumented knee implants in cycling activities was done in 2008 and consisted of varying difficulties of cycling at 60, 70, 80, and 90 RPM {D'Lima, 2008 #433}. D'Lima et al. {, 2008 #433} measured a maximum knee flexion angle of 94.3° and an average total tibial force of 1.03 times bodyweight obtained from instrumented knee implants. Cadence had a unique effect of altering the knee flexion angle at which maximum tibial force was observed. Tibial force peaked at approximately 74° and 37° of knee flexion for the 60 and 90 RPM conditions, respectively. Although the point at which the peak knee contact forces during the power phase occurred were different, the magnitude of the peak at 60 and 90 RPM were not significantly different.

Two of the kinetic variables associated with tibiofemoral joint loads and radiographic OA are internal knee extension moment (KEM) and internal knee abduction moment (KAbM) {Richards, 2018 #14}. Richards et al. {, 2018 #14} recruited patients with radiographic medial knee OA and simulated normal and altered gait patterns using the modeling software AnyBody (AMS, AnyBody Technology, Aalborg, Denmark). The simulations results found the KEM was significantly associated with the medial contact force to total force ratio as opposed to solely the medial contact force. KAbM was a slightly less correlated variable for only the first peak of medial contact force and not for significant for the second peak. However, a strong ($R^2 = 0.86$) relationship between KAbM impulse and the medial to total contact force ratio was found.

Despite the decreased knee joint loading of cycling, TKA patients have shown significantly reduced peak KEMs of the replaced limb compared to the native limb on a cycle ergometer {Hummer, 2021 #200}. Hummer et al. {, 2021 #200} recruited TKA patients 6 to 18 months post unilateral TKA and found that peak KEM was 18.7 ± 5.2 and 24.7 ± 6.9 Nm for the replaced and non-replaced limb, respectively, for a work rate of 80 W at 80 RPM. Increases in peak vertical PRFs and knee extension ROM were found at all workrates for the non-replaced limb. KAbM was not found to be significantly different as a result of limb or work rate, however knee abduction ROM did demonstrate a limb effect. At 80 W and 80 RPM, the abduction ROM was -13.3 and -8.4° for the replaced and non-replaced limbs, respectively. The knee extension moment for the non-replaced limb was, on average, 24.7 Nm and 24.2 Nm for the 80 and 100 W conditions, respectively. Considering peak knee extension moment was not significantly affected by work rate, but peak plantarflexion and hip extension moments were, it suggests the presence of a compensatory mechanism for the increase in work rate.

Post-surgery quadriceps weakness or avoidance and ROM deficits may make certain cycling conditions difficult for this population exceedingly early in the rehabilitation process. Further, it is known that TKA patients have a quadriceps avoidance strategy even at 12 months post-surgery {Li, 2013 #436}. Therefore, difficulty encountered by a TKA recipient during cycling may lead to compensation via increased loading of the contralateral limb, which may be detrimental to the cartilage of the contralateral knee and could accelerate OA symptoms.

Asymmetries such as a lower peak KEM in the replaced limb is also present in this population with other tasks, such as stair climbing and ramp ascent {Yocum, 2022 #222;Thorsen, 2023 #230}. Decreased KEM during physical activities is likely a result of factors such as decreased quadriceps strength post-TKA which is further exacerbated by muscle

attenuation after surgery that does not recover at six months post-operation {Paravlic, 2022 #301}. Utilizing musculoskeletal modeling, it has been reported that peak medial tibiofemoral compressive force for the replaced limb compared to the non-replaced limb in TKA patients during stationary cycling is 23.5% lower {Hummer, 2022 #169}.

Short, cycle sprint testing has been implemented in TKA patients at 3-, 6- and 12-months post-operation and showed significantly decreased lower limb power output at 3 months and significantly greater power output at 12 months {Hurworth, 2021 #229}. Hurworth et al. {, 2021 #229} additionally provided Oxford Knee Scores and KOOS scores that improved for each follow-up. These findings show support for measuring muscle power in TKA populations as a dynamic value instead of maximum muscle force output on an isokinetic dynamometer, for example.

Limb Alignment Effects

Varus and valgus knee alignment is of particular concern to the TKA population after surgery because of its relationship to increased polyethylene insert wear and implant loosening, a common cause of revision surgery {Kutzner, 2017 #170;Gromov, 2014 #431;Ritter, 2011 #428;Collier, 2007 #429;Srivastava, 2012 #430}. Forces through the medial and lateral compartments of the knee relative to static knee alignment are, therefore, of increased interest to clinicians for minimizing poor surgical outcomes. Gait and activities of daily living have been studied *in vivo* to assess the forces within the compartments as measured by instrumented implants with strain gauges implanted in the tibial component. Gait analysis at a self-selected, comfortable speed was done on participants twelve months after receiving a TKA implant to assess total and compartmental force distributions {Halder, 2012 #432}. Halder et al. {, 2012 #432} reported that the two peaks of axial force, as measured by strain gauges embedded in the

tibial component, during stance were 215 and 266% bodyweight with the medial load percentage being 73% and 65% for the two peaks, respectively. An important result of this work was an associated 5% increase in medial load as a percentage of the total for each 1° of varus deviation from a neutral alignment. Medial loading exceeding 50% of total knee joint load has been seen in other work as well, notably in a study assessing *in vivo* knee joint forces during a variety of daily living tasks {Kutzner, 2017 #170}. Kutzner et al. {, 2017 #170} analyzed the mechanics of instrumented TKA recipients performing level walking, stair ascent and descent, one-legged standing, jogging, knee bends, standing and sitting, as well as ergometer cycling. This work presented medial force ratio as the percent of medial force from the total force for all the activities listed. Average medial force ratio values ranged from 42 to 77 for the knee bend and first peak of level walking, respectively, and ergometer cycling had an average medial force ratio of 58. The authors discuss implications for surgical decision making for implant alignment goals to reduce the force distributions. To do this, it was suggested to create a valgus alignment of 3 or 4°. The authors defend their position by stating that high medial force ratios following implants resulting in a varus malalignment should be avoided. In order to accomplish this, slight valgus alignments lead to a more equitable distribution of force between the medial and lateral compartments of the replaced knee. The authors cite clinical results suggesting that polyethylene wear between the compartments is equivalent with a tibiofemoral alignment of 3° valgus {Collier, 2007 #429}.

Knee alignment in cycling has been assessed at a consistent cadence of 80 RPM and varying workloads of 0.5, 1.0, and 1.5 kg of resistance on a Monark Ergometer {Shen, 2018 #473}. Shen et al. {, 2018 #473} collected data in 11 varus, 11 neutral, and 10 valgus participants and confirmed a significant difference in tibiofemoral alignment between the groups.

Similar to reports by Fang et al. {, 2016 #52}, Shen et al {, 2018 #473} described a majority of participants, approximately 72%, across all groups displaying an internal knee abduction moment while the remaining participants displayed an internal knee adduction moment during the cycling. Despite the association of varus alignment and internal knee abduction moments in OA populations during gait, this association was not found in cycling by Shen et al. {, 2018 #473} {Messier, 2014 #475}. It is likely that the lack of significant difference in the frontal plane moments between alignment groups is, at least partially, the result of a combination of fixed pedal width and cycling being a more constrained movement for the knee compared to gait. The authors posit that the fixed pedal width of 31 cm combined with decreased vertical reaction forces during cycling may have, in part, negated the effect of tibiofemoral alignment on KAbM. The peak knee extension moment during cycling was not significantly different between alignment groups and suggests further that cycling is a suitable alternative to walking for minimizing differences in biomechanics between alignment groups.

Literature reports of *in vivo* measurements of ergometer cycling combined with kinematic data and limb alignment are, to the author's knowledge, extremely limited. Telemetric data from instrumented knee implants during ergometer cycling do exist, and one such study investigated the forces experienced during cycling at 25, 35, 50, 60, 70, 85, and 95 W with a cadence of 40 RPM {Kutzner, 2012 #4}. No investigations of alignment were made, and the authors were only able to comment that mediolateral force distributions were likely to be equivalent to that seen in gait for comparable populations. Understanding the distribution of loading for the medial and lateral compartments of the knee and their relationship with tibiofemoral alignment has positive implications for implant longevity and health in TKA populations {Andriacchi, 1986 #495;Nham, 2023 #184}.

Musculoskeletal Modeling

Musculoskeletal modeling is a field that experimentally uses recorded kinematic data, kinetic data, and/or EMG signals to simulate human and animal movement to understand factors such as individual muscle force, joint reaction forces, and muscle activations. Simulation of dynamic tasks is important because it allows for information to be obtained that is not traditionally made available by laboratory experiments. For example, forces generated by specific muscles *in vivo* or contact forces *in vivo* between bones are either impossible to acquire or quite rare to record, respectively. Commercial software options for musculoskeletal modeling are available, with a popular option being the AnyBody Modeling System from AnyBody Technology, which has been applied to various research fields including exoskeletons, upper extremity modeling, and occupational back loading {Engelhardt, 2020 #480;Koblauch, 2016 #481;Zhang, 2021 #482}. While commercially available software can be powerful, it is also expensive, and the most common musculoskeletal simulation software used in academia is likely the open-source package OpenSim developed at Stanford University. OpenSim has allowed musculoskeletal modeling to flourish and for clinically relevant information to be disseminated quickly amongst the research community {Delp, 2007 #221}. Prior to the development of OpenSim, a predecessor known as SIMM (Software for Interactive Musculoskeletal Modeling), was developed and distributed in the hopes of reducing the need for complex programming to conduct musculoskeletal simulations {Delp, 1995 #442}.

OpenSim provides a graphical user interface for importing data such as biomechanical marker data, ground reaction forces, and EMG to perform inverse kinematics, inverse dynamics, forward dynamics, and joint reaction analyses. The software allows for adjustments of models, represented muscles, joints and respective movement capacities, and muscle fiber dynamics to be

included in simulations in attempts to provide physiologically relevant results. OpenSim has demonstrated a capacity to process a wide variety of data for research projects including triceps brachii functions, amputee populations, cervical spine models, and TKA implant recipients {Raveendranathan, 2023 #273;Barrett, 2021 #275;Kholinne, 2018 #274;Hummer, 2022 #169;Thorsen, 2021 #173}. In OpenSim, the model used is of the highest importance as it reflects the neuromuscular and musculoskeletal dynamics that will be investigated for the given research question. Necessary components of a model include bodies (segments), joints, constraints, forces, contact geometry, markers, and controllers. These components interact with each other to estimate muscle forces during a given motion, but the simulation accuracy depends highly on the model construction and parameter validity. Once the initial parameters and model have been established, OpenSim provides advantages over traditional mathematical calculations such as inverse dynamics. Inverse dynamics is traditionally implemented to calculate joint torques in the field of biomechanics outside of OpenSim, but one limitation is that the output is only the net joint torque with no knowledge of internal interactions such as cocontraction of agonist and antagonist musculature. Using OpenSim, given appropriately set parameters, it is possible to estimate the muscle activations and forces that contribute to net joint torques.

Muscle Force Estimation

Prior to the advent of computers and advanced musculoskeletal modeling software, the investigation of muscular forces and their corresponding effects on the skeleton were highly limited. Muscle dynamics at the microscopic level were obtained from *in vitro* analyses in animals and the closest approximation for human *in vivo* muscle activities and related forces came from EMG measurements done *in vitro* and then *in vivo* {Bekoff, 1977 #523;Bekoff, 1977 #523;Farina, 2024 #524;Kazamel, 2017 #525}. While these experiments were able to gather

knowledge that is invaluable, the estimation of individual muscle forces during dynamic tasks has become more viable with computational modeling advances.

Fiber Dynamics

The estimation of muscle forces is a complicated task plagued by difficulties in mathematical representation of fiber physiology, EMG acquisition, determination of DoF for joints of interest, and the muscle redundancy problem. The human body has a considerably higher number of muscles than joint DoF, such as flexion and extension. Therefore, it is complicated to decide the contribution of each muscle to experimentally obtained kinematics since a functionally infinite combination of muscle activations could contribute to the movement. This situation is referred to as the muscle redundancy problem. At the knee, for example, the four quadriceps and four hamstring muscles, perform knee flexion and extension, among other actions. For a given observed joint movement, it is not possible to determine how much of a contribution the vastus medialis has to that movement compared to one of the other quadricep or hamstrings muscles. As a result, we must create a mathematical framework to reasonably distill the overall force into multiple components. A common method to reasonably resolve this problem is using an optimization function that tries to minimize or maximize criterion such as muscle excitation levels, muscle oxygen usage, or approximations of muscular effort such as muscular fatigue {De Groote, 2016 #387; Seth, 2018 #368; Crowninshield, 1981 #401; Ackermann, 2010 #385}. Ackerman and van den Bogert {, 2010 #385} reported that effort cost functions resulted in simulation muscle use avoidance that caused high impact forces because of a lack of knee flexion during stance in gait, resulting in an overly stiff heel strike in extension. Even though muscle use avoidance is not often present in healthy populations, it has been observed in the quadriceps of TKA patients in gait {Hyodo, 2020 #233}. Soon after

surgery, neuromuscular deficits prevent quadriceps activation and as rehabilitation progresses, these are ameliorated, but not fully, even up to six months {Paravlic, 2022 #301}. Therefore, while a cost function incorporating neuromuscular deficits as discussed is incredibly difficult and problematic, it may be representative of certain patient populations.

Introducing more accurate non-linear physiological relationships such as tendon force-length, muscle force-length, and muscle force-velocity can provide more anatomical relevance for simulation results {De Groote, 2016 #387}. The quantification of these parameters for each muscle involved in a simulation can result in more physiologically grounded inputs for the chosen cost function. Contraction dynamics are most commonly modeled after the muscle-tendon unit model proposed by AV Hill {Hill, 1970 #389;Zajac, 1989 #388}. The Hill model consists of a tendon modeled as a nonlinear spring connected at some given pennation angle in series with a muscle unit comprised of a contractile element in parallel with a passive element. A series of equations are presented below that have been presented in previous literature to describe various components of contraction dynamics {De Groote, 2016 #387;Winters, 1995 #488}.

Equation 1 describes the calculation of tendon force (F_T).

$$F_T = F_M^0 f_t(\tilde{l}_T)$$

Equation 1: Calculation of tendon force.

Where F_M^0 is peak isometric force, f_t is the tendon force-length characteristic, and $\tilde{l}_T$ is the normalized tendon length relative to its slack length. Equation 2 defines the muscle force (F_M).

$$F_M = F_M^0 [a f_{act}(\tilde{l}_M) f_v(\tilde{v}_M) + f_{pas}(\tilde{l}_M)]$$

Equation 2: Calculation of muscle force.

Where ‘a’ determines activation dynamics from a value of 0 to 1, f_{act} is the active muscle force-length relationship, $\tilde{l}_M$ is the normalized muscle fiber length, f_v is the muscle force-velocity

representation, $\tilde{v}_M$ is normalized fiber velocity, and f_{pas} is the passive muscle force-length relationship. Equations 3, 4, and 5 describe muscle and tendon interactions.

$$l_{MT} = l_T + l_M \cos \alpha$$

Equation 3: Musculotendon length.

Where l_{MT} is the muscle-tendon length, l_T is the length of the tendon, l_M is the length of the muscle, and α is the pennation angle.

$$l_M \sin \alpha = l_M^0 \sin \alpha_0$$

Equation 4: Musculotendon length and pennation angle interaction.

Where l_M^0 is the optimal fiber length, and α_0 is the pennation angle at optimal fiber length.

$$F_T = F_M \cos \alpha$$

Equation 5: Muscle force accounting for pennation angle.

Given muscles' non-linear relationships regarding force-length and force-velocity, these formulas may be important for obtaining a more representative behavior for simulations.

Conflicting literature has been presented as to the importance of static versus dynamic optimizations regarding muscle force across different tasks. In gait tasks, solving the muscle redundancy problem with differing optimization techniques seems to produce equivalent results {Anderson, 2001 #526}. In this paper, two inverse dynamics based static solutions were implemented: one using pure force generators while the other included force-length-velocity parameters. These two solutions were compared to a forward dynamics solution. Findings from Anderson and Pandy {Anderson, 2001 #526} highlight the importance of understanding the relationship between the optimization method being used as well as the objective function. Performing a similar static and dynamic comparison in wheelchair propulsion simulations has produced results that do not align as closely between the static and dynamic techniques {Morrow, 2014 #527}. Specifically, deviations were shown in elbow moment, pronation-

supination motion, and hand rim forces during wheelchair propulsion when comparing static and dynamic optimization. Measurement errors in the upper body may have contributed to these differences, and the authors concluded that care should be taken to determine if static or dynamic optimization should be used on a task-by-task basis. Additional muscle parameters included in the equations above tend to have a more consistent conclusion regarding the effect on simulation accuracy, such as pennation angle. For example, variations of up to 2.1 bodyweights of knee joint contact force during a squatting movement have been shown to result from perturbations to musculotendon parameters such as maximum isometric force and pennation angle when comparing to *in vivo* Comprehensive Assessment of the Musculoskeletal System (CAMS) knee data {Hosseini Nasab, 2022 #399}.

Pennation angle and optimal fiber length have been established as good indicators for force production {Lieber, 2000 #464;Delp, 1990 #465;Zajac, 1989 #388}. The pennation angle of a muscle can be defined as the fiber angle relative to the longitudinal axis of the overall muscle. Pennation angle will adjust the amount of effective force provided to a tendon via the attached musculature, but a common assumption in musculotendon modeling is a consistent pennation angle through contraction and joint motion. Pennation angle has been measured through a surrogate fascicle plane orientation and angles of 6° and 25° during contraction have been measured in the vastus lateralis and medial gastrocnemius, respectively {Kawakami, 1998 #466;Fukunaga, 1997 #467}. The potential change in pennation angle of musculature during movements signifies a change in effective force produced by that muscle as the related joint moves, in turn affecting the muscle redundancy problem. Moreover, the capacity of a muscle to produce force is adjusted beyond force-length and force-velocity relationships as a result of the joint angle and pennation angle. However, accounting for this change in fiber angle on a general

or subject basis would be a difficult task, as muscle size has been shown to alter pennation angles in human triceps brachii muscles when comparing untrained individuals with trained bodybuilders {Kawakami, 1993 #489}. While bodybuilders are a small percentage of the population, the concept that a hypertrophied muscle has altered pennation angles may be of concern for testing with athletic populations. The inverse may also be relevant for atrophied muscle present in certain clinical populations, the elderly, and those who have been subjected to extended periods of bedrest. Pennation angle has also been shown to be significantly different in cadaveric human soleus musculature when measured superficially and deep {Sopher, 2017 #371}. If it is assumed that muscle force is proportional to the cosine of the pennation angle, the muscle force of the soleus would be underestimated by approximately 15% using surface level pennation angle as opposed to the deep pennation angle. Sopher et al. {, 2017 #372} reported that ankle contact forces were not significantly affected by the fiber pennation angle but theorized based on previous literature that the significance is likely more sensitive to muscles with larger pennation angles such as 23° {Zajac, 1989 #388}. The effects of a large pennation angle are therefore likely to be more of a concern during simulation in musculature around the knee where muscles such as the vastus medialis have had fiber angles measured on average as 36.7° {Engelina, 2014 #468}. Despite these anatomical measurements and theoretical guesses, assessment of musculotendon parameters in OpenSim modeling has shown that pennation angle is the least impactful variable for muscle force estimation of the variables tested {Chen, 2023 #376}.

Some of the following parameters have been perturbed to determine their sensitivity regarding simulation results: the representation of muscle force-length-velocity relationships, optimal fiber length, maximum fiber velocity, and tendon slack length. Ensuring that muscle

force-length and force-velocity components are included in a model is critical, as the force outputs are inherently linked to muscle fiber length and contraction velocity. The force-length curve represents how muscle force output changes with length, with the main concept being that a muscle has an optimal fiber length and once the muscle is outside of that length the force production decreases. The physiological basis of this relationship stems from sarcomeres being too overlapped when the muscle is shortened, or too far apart when the muscle is lengthened, to form an ideal number of cross bridges. The force-velocity relationship is not simply force-speed because muscle contraction directions can be concentric or eccentric, both of which generate different amounts of tension within a muscle. With a velocity of zero, the contraction is isometric, and a positive value represents an eccentric contraction that produces considerably more force than a concentric contraction of the same absolute value. The physiological theory that underpins eccentric contractions and their increased force output comes partially from the protein titin, which can act like a spring within muscle tissue. Titin is the largest known human protein and acts elastically by increasing stiffness as muscle force production increases {Hessel, 2017 #540}.

Setting a maximal fiber velocity is also important regarding the muscle's force-velocity curve because there needs to be end points for concentric and eccentric contractions, or else the model may produce essentially zero force for an extreme velocity concentric contraction, or incredibly high forces for the eccentric contraction. Generalization of the fiber's maximum velocity quickly becomes difficult due to the different fiber types present in a given muscle, an individual's exercise history, and the age of the individual. Specifically, maximal fiber velocity is considerably higher in fast oxidative and fast oxidative glycolytic fiber types compared to slow oxidative types {Fitts, 1991 #541}. Further, participation in exercise training that is not

endurance based has shown a shift in the slow oxidative fiber's maximal contraction velocity {Fitts, 1991 #541}. Age related changes in muscle fiber dynamics are well studied and demonstrate that fibers with large expressions of slow myosin heavy chain (MHC) I do not degrade extremely with age, but faster MHC II expression does decrease significantly {Grosicki, 2022 #542}. This is certainly not the only contributing factor to declining muscular power output with age but is an important consideration for musculoskeletal modeling purposes. When deciding which model to implement, it may not be appropriate to use a model with parameters entirely based on cadaveric data when simulating movements in high level athletes in their twenties, for example. Potential simulation errors are likely to occur in this case as the musculotendon actuators of a given model may not produce nearly enough force to complete static optimization (SO). As a result, model selection and the setting of parameters should be carefully considered before performing an experiment to ensure that as many parameters can be matched between the participant demographics and model as possible.

Tendon slack length is a critical parameter with using the Hill-type muscle model as it reflects the length of the tendon when the generation of force to change its length begins. However, Chen and Franklin {, 2023 #376} present Equation 6 to determine the tendon slack length.

$$l_{sla}^t = l^{mt}(q_o) - l_{opt}^f \cos \alpha_{opt}$$

Equation 6: Tendon slack length.

Where l_{sla}^t is the tendon slack length, $l^{mt}(q_o)$ is the musculotendon unit length at a predetermined joint position q_o , and α_{opt} is the pennation angle at optimal fiber length. Accurate representation of the tendon slack length is important due to its direct relationship with the force length curve and musculotendon unit length borrowed from Chen and Franklin {, 2023 #376}

(**Figure 10**). **Figure 10** shows example ankle joint positions and their relative positions on the force length curve, with the left side graph setting optimal fiber length at plantarflexion and the right-side graph setting the optimal fiber length at dorsiflexion. Chen and Franklin {, 2023 #376} discuss the difficulties in measuring optimal fiber length in relation to this curve, because, as shown in **Figure 10**, the contribution of active and passive components will change depending on which joint position is set as optimal for a given muscle.

Assessing the sensitivity of the previously discussed model parameters in the estimation of muscle forces and joint reaction forces has been conducted using gait simulations, predominately {Carbone, 2016 #531;Serrancolí, 2020 #532;Raikova, 2001 #528;Scovil, 2006 #529;Redl, 2007 #530;Xiao, 2010 #377}. Carbone et al. {Carbone, 2016 #531} determined that, in order, tendon slack length, maximum isometric force and optimal fiber length were the most sensitive parameters during gait while pennation angle was the least sensitive parameter. Muscles that are prime movers, such as the rectus femoris, soleus, and gastrocnemius, during gait were the most sensitive to the previously mentioned variables. During gait and running simulations, Scovil and Ronsky {, 2006 #529} reported that tendon parameters, such as slack length, were the most sensitive in muscle force estimation along with maximum isometric force and the force length curve implementation. The authors reported that the model was insensitive to the force velocity curve inclusion and muscle activation time constants. Results from Redl et al. {, 2007 #530} coincide with Carbone et al. {Carbone, 2016 #531} and Scovil and Ronsky {, 2006 #529} during gait simulations by concluding that muscle force estimates were most sensitive to tendon rest length as well as fiber length. Redl et al. {, 2007 #530} did also report that the muscle force estimates were least sensitive to PCSA. Xiao and Higginson {, 2010 #377} executed simulated gait trials and determined that tendon slack length as well as optimal fiber

length were critical to properly set for the ankle plantarflexors and knee extensors, specifically. Raikova and Prilutsky {, 2001 #528} used a nine muscle leg model to assess the effect of model parameters on two different cost functions to determine parameter sensitivity. The authors concluded that muscle moment arm offsets and PCSAs adjusted muscle force magnitudes during movement, but not the force pattern. A non-linear relationship was observed within each muscle and the authors mention that muscle forces are highly interlinked with the moment arms and PCSAs of all other muscles and joints in the model. Parameter sensitivity has also been assessed by using knee contact force during gait as the measure of change by Serrancolí et al. {, 2020 #532}. The authors reported that moment arm offsets alone were the most important variable for estimating medial and lateral contact forces when the other two options were optimal fiber length and tendon slack length. This conclusion is logical because an underestimated moment arm simply leads to an overestimated force output to meet the same desired torque. However, when utilizing multiple variables, medial contact forces were best estimated optimal fiber length, tendon slack length, and muscle moment arm offsets applied. When physiological parameters for muscle-tendon units are implemented, additional consideration should be given to the approximated locations of the muscle-tendon units. When comparing an unscaled model to a model developed from magnetic resonance images the largest relative differences in variability of muscle-tendon unit origin and insertion locations on the pelvis, femur, and tibia were in the sagittal plane {Bosmans, 2015 #373}. The authors specify that the variability seen in their results exceeds segmentation errors for all planes, thus providing support for the use of imaging for model personalization. Whether musculoskeletal modeling simulations benefit from highly complex estimations of musculotendon parameters has been debated {Chen, 2023 #376}. Chen and Franklin {, 2023 #376} posit that cadaveric and MRI data are not guaranteed to reflect

musculotendon parameters as they are reflective of muscle architecture which may not be the same as muscle biomechanics. Further, the authors mention that since Hill muscle model parameters are predominately nonlinear, positive developments in parameter accuracy may not lead to comparable simulation result accuracy.

Model Parameters

The uncertainty of model parameters of maximum isometric force, tendon slack length, pennation angle, origin and insertion points, and via points were perturbed by Hosseini Nasab et al. {, 2022 #399} to understand their effect on knee contact force using the Rajagopal model {Rajagopal, 2016 #183}. The overall findings were in line with postulations in previous work regarding the increased level of knee joint contact force when influential musculature is perpendicular to the tibial plateau, as is the case in deep flexion movements {Lloyd, 2001 #441}. Lloyd and Buchanan {Lloyd, 2001 #441} reported that participants used their quadriceps and hamstrings in an ineffective way to reduce contact forces in response to varus and valgus tasks. Specifically, since the hamstrings and quadriceps have a minor capacity to correct for varus and valgus moments, they require a relatively high force output to “correct” for those moments. Consequently, when participants displayed a high level of hamstring and quadriceps cocontraction during the varus and valgus tasks, which caused a considerable increase in knee joint contact forces. The most influential parameter on resulting knee contact force from the perturbed variables was found to be the maximum isometric force of the muscles included. This finding is mirrored in a sensitivity analysis of an OpenSim model of the upper limb in movement measured via intra-cortical pins {Blache, 2019 #469}. If maximal isometric force is the most important variable for muscle force estimation, it should be of great concern to consider the values applied to muscles in patient populations where muscle strength is likely altered. For

example, quadriceps weakness is mostly guaranteed in the affected limb after TKA and a generic model musculotendon parameters may not reflect this type of neuromuscular change accurately {Paravlic, 2022 #301; Mizner, 2005 #237; Li, 2013 #436}.

The maximum force production of a muscle is inherently tied to the PCSA, often more strongly, than the anatomical cross-sectional area (ACSA) {Powell, 1984 #487}. PCSA is calculated using the width of the fibers by measuring perpendicular to them as opposed to the ACSA, which uses the width of fibers based on the longitudinal axis of the muscle. As a result, in muscles with higher pennation angles, the gap between the absolute value of the PCSA and ACSA increases. It is possible to obtain estimates of PCSA through the use of MRI and an investigation has been conducted to determine the effect of participant specific PCSA values for an upper extremity model in maximum force tasks {Bolsterlee, 2015 #486}. Results indicated that PCSA scaling had a negligible effect ($< 1.5\%$) on glenohumeral joint contact forces for submaximal tasks and the need for PCSA scaling compared to uniform scaling was not strongly supported. Maximum isometric force is estimated using the product of PCSA and specific tension of a given muscle. Specific tension can be calculated as the maximum force produced by a muscle normalized to the PCSA, reported in N/cm^2 {O'Brien, 2010 #533}. Early values of specific tension in simulation used by Delp et al. {Delp, 1990 #465} were $61 \text{ N}/\text{cm}^2$ to match the physiological properties of young subjects. Specific tension originally derived from the PCSA of elderly cadavers has been “scaled” to match younger subjects *in vivo* data, but the result can be considerably larger joint moments compared to experimental results {Rajagopal, 2016 #183}. Chen and Franklin {Chen, 2023 #376} argue that specific tension should be measured for each muscle under isometric conditions, as the force produced per area of muscle will change dependent on the muscle’s length and velocity. Indeed, the reasoning for calculating a value for

each muscle is supported by findings that agonist-antagonist muscle groups such as the elbow and ankle have different specific tensions, as well as muscles within the same functional group {Buchanan, 1995 #534;Fukunaga, 1996 #535;Javidi, 2019 #536;Kawakami, 1994 #537}.

Understanding the interplay between specific tension and PCSA is important in the pursuit of more accurate simulation results. Finding relevant specific tension values for all populations in previous literature may not be possible, and scaling of the closest demographic's approximated values may be warranted. This is an important consideration in some clinical populations, for example, who have undergone surgical trauma that has caused temporary neuromuscular damage. In these populations, the force production to muscular area is more likely than not to be affected, leading to potential simulation discrepancies regarding specific tension.

Muscle force estimation accuracy can be further improved by the inclusion of time-based components of muscle physiology. Specifically, muscles have an inherent delay between their activation and the onset of force generation that should be accounted for mathematically. The cause of this delay stems from the time it takes a generated action potential to cause calcium ions to release from the sarcoplasmic reticulum to initiate cross-bridge formation. A delay is also present in the relaxation of the same muscle fibers as there is a longer delay required for the re-uptake of the calcium ions. The modeling of calcium dynamics in OpenSim is found below in Equation 7.

$$\frac{da}{dt} = \frac{u - a}{\tau(a, u)}$$

Equation 7: Calcium dynamics.

The purpose of this equation is to model a relationship between the rate of change in muscle activation (a) and muscle excitation (u). The denominator on the right side of Equation 7

represents a time constant for activation and deactivation, which is assumed to be 10 and 40 ms, respectively {Zajac, 1989 #388;Winters, 1990 #539}.

Research in rabbit mastication and EMG provided evidence for the implementation of a twitch response component in EMG-to-activation models for improved muscle force estimation {van Ruijven, 1990 #443}. A secondary difficulty of this time-based component is that processed EMG signals, such as those that have been rectified and smoothed, may create an inability to quantify the muscle's time delay {Guimaraes, 1995 #444}. While time-based variables do provide more physiological relevance, they are not always applicable depending on the method of computation, which is discussed later.

Simulation results are further influenced by muscle force estimates because the number of muscles and their associated maximum isometric force capacities between models are often different, leading to kinematic and kinetic variability for the same gait trial {Roelker, 2017 #370}. In terms of muscle isometric force, the gait2392 model has a total hamstring force producing capacity of 3398 N, while the Arnold model has a total of 2485.6 N {Arnold, 2010 #393}. Differences of approximately 912 N in these isometric force capacities can significantly affect muscle activation level, as seen in other musculature by Roelker et al. {Roelker, 2017 #370} in the gastrocnemius and rectus femoris between the gait 2392 and Arnold models. Gastrocnemius and rectus femoris activations in the Arnold model were 42-50% and 84-233% greater than the gait2392 model, respectively. Gastrocnemius peak forces were not reported by Roelker et al. {Roelker, 2017 #370}, but peak rectus femoris force was significantly higher in the Arnold model compared to the gait2392 model. Significantly higher muscle forces in the Arnold model were also seen in the Vasti muscles, soleus, and tibialis anterior. Some of the variation in muscle activations for the rectus femoris, and therefore muscle force, may be

accounted for by the peak isometric muscle force values for the different models. Gait2392 has the rectus femoris peak force at 1169 N, while the Arnold model has a value of 848.8 N.

Kinetic differences between the two models were also seen in the peak knee extension and second peak knee flexion moments. Model construction differences in joint geometry, specific constraints, and degrees of freedom are additional variables that feed into kinetic differences that may occur. When using static optimization, as implemented by Roelker et al. {Roelker, 2020 #375}, the resulting individual torques that are obtained by decomposing the net torque will also result in kinetic variations. By extrapolating, it is likely that with altered knee kinetics and surrounding musculature activation changes, that the knee contact forces would be significantly different as well. Indeed, estimates of vertical knee contact forces between models when compared to *in vivo* measurements of gait and sit-to-stand tasks have been shown to be different when comparing three separate lower limb models {Pelegrinelli, 2023 #382}.

Muscle Path and Moment Arm

When evaluating muscle forces and subsequent joint contact forces during a movement, proper modeling of muscle moment arms is important for more valid simulation results. A muscle moment arm represents a measure of instantaneous effectiveness a muscle has for contributing to motion at one or more joints over a range of configurations or lengths over a path which is considered frictionless and massless {Sherman, 2013 #397}. There are considerable challenges to modeling a muscle to have a mathematically accurate moment arm because many muscles follow nonlinear paths that may span multiple bones, surfaces, or joints {Garner, 2000 #396}. Garner and Pandy {Garner, 2000 #396} presented a method for muscle modeling that represents the muscle as a frictionless elastic band whose path is constrained by geometric objects such as cylinders and spheres. This concept of manipulating a muscle around objects to

represent an anatomical path is used in OpenSim and is referred to as a wrapping point.

Estimates of moment arms at the knee are complex as a result of the discrepancy between actual six rotational and translational DoF present in the human knee and what can be feasibly modeled. Specifically, since the moment arm is dependent on the axis of rotation within the knee joint, the allowable motions of a knee model will have an effect on where the axis is located during dynamic movement. Research utilizing the gait2392 model with a knee of one DoF and modified to three DoF in a single leg hopping task reported that interpretation of forces from biarticular muscles of the knee such as select hamstrings and the soleus are suspect when using one DoF {Mokhtarzadeh, 2014 #398}. Mokhtarzadeh et al. {Mokhtarzadeh, 2014 #398} reported decreased force output for the hamstrings, rectus femoris, and gastrocnemius with the use of a 1 DoF knee, while the soleus force was lower using the 3 DoF knee. The authors hypothesized that the decrease in the muscle forces seen in the 1 DoF knee could be a compounding result of the lack of excitation and activation dynamics as well as the muscle force combination solution was for only one plane. When solving for the net knee joint moment in multiple planes, the combinations of muscle forces become increasingly complex with respect to muscles that operate in more than one plane of movement or span more than one joint. Despite this theory, the soleus surprisingly displayed the opposite behavior of the hamstring and showed decreased force with the 3 DoF that is likely a result of the change in ankle kinetics. Specifically, the gastrocnemius generated considerably more of the ankle plantarflexor moment, causing the required contribution from the soleus to decrease. The authors conclude that the choice of knee joint DoF can have a significant impact on ankle musculature force when using SO.

Estimation of moment arms for simulation is complex because while movements of the knee and related tibiofemoral coordinate differences are easily measured via cadaver studies,

they are taken in unloaded conditions and therefore may not be transferable to dynamic tasks. External loading due to gravity as well as the contribution of knee musculature contractile forces affect the kinematics of the knee joint itself. Moment arms of muscles crossing the knee change during dynamic movements and work has been done to assess how muscle moment arms change during the single leg lunge and the seated knee extension {Navacchia, 2017 #372}. RMSE differences in knee moment arms between cadaveric and experimental kinematics reached a maximum of 9 mm. Using a Monte Carlo approach, the largest confidence bounds for a muscle's moment arm was tied to the semimembranosus during a lunge movement and the smallest was for the rectus femoris during knee extension. The authors conclude from their results that muscle forces and joint loads are likely to be altered when activity and/or subject specific kinematics are included. This highlights the limitation of using passive cadaveric data to model knee joint motions as opposed to an *in vivo* measurement of the specific activity of interest. Moreover, knee behavior from passive cadaveric knee extension is likely not representative of knee kinematics during *in vivo* dynamic movements such as a lunge. External loading of the knee is not insignificant, as transverse rotation and sagittal translation of the knee can be different when comparing knee kinematics in seated and weight-bearing knee extension activities {Myers, 2012 #458; Lafortune, 1992 #457}. Myers et al. {Myers, 2012 #458} used biplanar fluoroscopy to assess tibial translation during unloaded knee extensions, walking, landing, and maximum isometric knee extensions and found that the knee can constrain the tibia depending on the task demands in similar manners. These findings have important musculoskeletal modeling implications in that the movement of the contact points are mediated by soft tissue support as well as muscular contributions, both of which are essentially impossible to distill into individual contributions given some external load.

As with many OpenSim models, one DoF (flexion/extension) is a frequent limitation for tibiofemoral modeling accuracy. Not all muscles exhibit this simplification, as some models contain secondary movements that behave in conjunction with given flexion or extension amounts, often used for more thorough analysis of knee soft tissue connections {Marieswaran, 2018 #460}. Fluoroscopy and finite element modeling of the knee have shown approximately 5mm and 8-10mm of anteroposterior excursion of the medial and lateral contact points, respectively, in downhill or level gait trials {Farrokhi, 2014 #454;Liu, 2010 #455;Marouane, 2016 #456}. Marouane et al. {Marouane, 2016 #456} explicitly state that the moving contact point locations need to be considered when assessing compartmental contact forces in gait. The lack of this anteroposterior degree of freedom in many knee models is an inherent limitation of OpenSim joint reaction results, but the one degree of freedom knee joint allows for more easily conducted simulations compared to even more complex finite element models. The absence of anteroposterior movement is primarily due to a limitation in motion capture where movements of such a small scale are within the measurement error of the system.

Geometry

Discussion of model geometry within OpenSim leads to one of the assumptions that is made for larger scale simulation feasibility- the assumption that body segments are rigid bodies. While bone is a brittle material, computational methods such as finite element analysis (FEA) have provided insight into the effect of external loads and muscle force on bones and corresponding bone adaptations {Meslier, 2023 #490}. FEA works by assessing mechanical load and strain on an object that has been subdivided into x number of parts and the interactions between these hundreds or thousands of parts are calculated. It is understood that external loads and forces act on, and slightly deform human bones, but this aspect of physics is not always

required for all research questions. FEA is computationally expensive, and the modeling of a gait trial with over twenty segments and ninety musculotendon actuators is simply not feasible.

Therefore, OpenSim maintains the assumption that segments are rigid bodies. While the deformation of bones may not be modeled in OpenSim, modeling of soft tissue restraints and more complex joint movements has been presented.

Additional assumptions are made for joint movements, often being modeled with fewer degrees of freedom than what is true or sometimes even locked in many models at the metatarsophalangeal (MTP) joint and subtalar joint. In the case of the knee, a majority of the available models model the knee as a hinge joint with flexion and extension available even though it is known that the knee has more rotations and translations. While one DoF knee models are common, efforts have been made to develop more complex knee models that include more soft tissue support such as anterior and posterior cruciate ligament bundles or increasing the DoF to six {Xu, 2015 #459; Ozada, 2015 #462; Marieswaran, 2018 #460}. The absence of knee ligaments in modeling is likely a significant contributor to observed knee motion, or lack thereof, as modeling results have shown ACL peak forces during gait ranging from 0.2 to 1.7 BW {Shelburne, 2004 #461}. Xu et al. {Xu, 2015 #459} developed an OpenSim knee model that includes ten separate ligament segments, many of which are subsections of the collateral and cruciate ligaments. One complexity noted in this work was the determination of the zero-load length of the ligaments, as in, at what point are the ligaments considered to be in a resistive state or slack. An additional difficulty with ligament modeling is that models such as the one provided by Xu et al. {Xu, 2015 #459} modeled soft tissue as straight segments from origin to insertion, which is likely simplifying the path of *in vivo* knee ligaments and their reaction to dynamic movement. The model presented by Xu et al. {Xu, 2015 #459} was further modified by

Marieswaran et al. {Marieswaran, 2018 #460} to include joint capsules, additional ligaments, menisci, and femoral articular cartilage. This updated model's ligament strains were evaluated with passive movement from 0° to 120° of flexion, 40° of external to 30° of internal rotation, and -15° to 15° of adduction. Strain properties were not compared to an *in vivo* standard, so evaluation of the validity of the ligament properties is difficult. However, the authors note that a limitation involved with their modeling was that the soft tissue was considered linearly elastic, when true tissue behavior is non-linear and viscoelastic. The additional complexity introduced into modeling of the knee with the inclusion of multiple ligaments and other soft tissue elements may or may not increase the validity or practical usefulness of the model for a given research question. For research questions involving TKA, for example, one of the few knee ligaments of interest (or that remain from surgery) would be the PCL. Consideration should be given to whether soft tissue contributions are of interest to the research question or whether they can be omitted in favor of other outcome variables.

Of particular interest at the knee, the patella has been included in models by wrapping the quadriceps line of action over the patella and on to the tibia, or by wrapping the patellar ligament itself to the tibia for the Rajagopal and Arnold models, respectively {Rajagopal, 2016 #183;Arnold, 2010 #393}. Validation of resulting moment arms of muscles crossing the knee can be done by comparing to experimentally measured moment arms derived from MRI-based bone models {Blemker, 2005 #491}. Proper modeling of the patella and its interaction with the femur and tibia are of particular interest to researchers that investigate patellofemoral pain and loading {Wheatley, 2023 #492}.

OpenSim Modeling

Scaling

One of the first steps in producing a simulation in OpenSim consists of scaling the chosen model's anthropometry to the participant with the best approximation possible. Virtual markers closely mirroring experimentally placed markers are moved by the user onto the skeleton in OpenSim and the scaling determines the relative distance between a virtual marker and the true experimental marker, referred to as marker registration {Delp, 2007 #221}. The error between experimental and virtual markers can be weighted based on researcher confidence in the marker placement based on factors such as bony prominence or skin artifact. Specification of a greater weight for a given marker will decrease the error tolerance for marker registration and generally increase the agreement between an experimental and virtual marker.

Additionally, segment masses, muscle fiber and tendon slack lengths are scaled by percentage according to normative data defined by the specific model during this process. It is important to note that models are not scaled to normative data native to OpenSim, the parameters are determined by the creators of the original model. The model's base parameters are commonly based on cadaveric data that are not always specific to certain clinical populations of interest, but some modern models have been created that include cadaveric and young, healthy individuals' MRI data in hopes of attaining more accurate simulation results for a younger population {Rajagopal, 2016 #183}. Many small adjustments were made from the Arnold model to produce the model presented by Rajagopal et al. {Rajagopal, 2016 #183}, such as altering wrapping points from ellipsoid to cylindrical objects in efforts to reduce simulation time.

Scaling is a critical component for simulation accuracy and research using the popular gait2392 model demonstrated that errors in experimental lateral epicondyle marker placement on

the participant resulted in approximately 9° hip flexion angle changes and 0.15 Nm/kg knee flexion moment changes {Koller, 2021 #390}. Maximum muscle forces of the medial gastrocnemius as well as hip joint contact forces were also significantly altered by 46 and 72%, respectively. These findings highlight the importance of accuracy in marker placement and provide some justification for the development of more personalized musculoskeletal models utilizing medical imaging modalities that reduce or avoid researcher placement landmark identification errors {Schmid, 2009 #178;Smale, 2019 #180;Fernandez, 2023 #198}. Even when marker position errors are minimized to 2 cm of uncertainty, ankle plantarflexion angle, moment, and positive power at push off can be altered by 15.9°, 26.6 Nm, and 75.9 W, respectively, during healthy gait {Uchida, 2022 #391}. Uchida and Seth {Uchida, 2022 #391} also found large variation in the previous variables when segment uncertainty was also, at most, 2 cm. However, the variation seen in segment dimensions was less impactful than marker registration uncertainty.

Determination of a given joint's center of rotation is an additional, important consideration during scaling as the muscle moment arms and resulting joint reaction forces about that joint can be altered by generic or subject-specific joint centers {Song, 2020 #483}. Joint centers can be evaluated at varying levels of complexity, ranging from three-dimensional bone reconstruction and fluoroscopy to equations based on superficial anatomical landmark distances {Song, 2020 #483;Fiorentino, 2016 #484}. Proper joint center identification through medical images is critical for identifying kinetic patterns in clinical populations {Harris, 2017 #485}. Harris et al. {Harris, 2017 #485} was able to discover that lateralization of hip joint centers in patients with hip dysplasia resulted in greater medially directed joint reaction forces compared to healthy controls in gait trials. Further work in this patient population has revealed significantly

different hip muscle moment arm lengths and resulting hip joint reaction forces compared to *in vivo* data {Song, 2020 #483}.

Inverse Kinematics

Once a model has been scaled, the next step involves the relationship between measured kinematics and the virtual markers placed on the OpenSim model during a motion. In order to reproduce the measured kinematics of the subject data most accurately, a minimized least-squares problem (Eq. 8) is solved that is subject to certain joint constraints as defined by the model within OpenSim {Lu, 1999 #392}. Joint constraints are meant to reflect the physiological DoF present in joints such as the elbow, which can perform active flexion and extension.

$$\text{Squared Error} = \sum_{i=1}^{\text{markers}} \omega_i \left(\overrightarrow{x_i^{\text{sub}}} - \overrightarrow{x_i^{\text{mod}}} \right)^2 + \sum_{j=1}^{\text{joint angles}} \omega_j \left(\overrightarrow{\theta_j^{\text{sub}}} - \overrightarrow{\theta_j^{\text{mod}}} \right)^2$$

Equation 8: Minimized least squares for kinematics.

$\overrightarrow{x_i^{\text{sub}}}$ and $\overrightarrow{x_i^{\text{mod}}}$ are the three-dimensional marker coordinates for some i th marker or joint center for a given subject and model, $\overrightarrow{\theta_j^{\text{sub}}}$ and $\overrightarrow{\theta_j^{\text{mod}}}$ are the joint angle values of the subject and model at the j th joint, and ω_i and ω_j are separate weighting factors used to adjust coordinates and joint angles separately. Marker weights, called inverse kinematics (IK) Tasks, are also specified at this step to allow the user to decide how stringent the error term should be bounded. Coordinates, often joint angles as solved by the IK tool, can be prescribed or unprescribed which has to do with whether they will be evaluated using the formula. Prescribed coordinates might be the subtalar and MTP joints which, when locked, are assigned a value of 0° as opposed to unprescribed coordinates which are unknown and therefore solved by the IK tool. Although 0° is assigned, the actual units used by the IK solver in OpenSim for angles are radians. This is

important to note in case results from OpenSim are compared to another IK method that is custom or found in another piece of software.

Two of the potential sources of error in IK results arise from actual marker placement by a researcher and then the manual placement of the virtual marker on the OpenSim model through the user interface. In order to minimize the error in marker registration, at least one method has been proposed within the OpenSim framework that removes subjective user input in the placement of the virtual markers {Dunne, 2021 #494}. Dunne et al. {Dunne, 2021 #494} placed retroreflective markers on a robot with the capacity to walk in order to test some strategies for marker registration. Using the internally recorded information from the robot, the virtual markers were able to be placed exactly on the experimental markers for a total error of zero as a reference. An orientation registration method was also tested that used anatomical markers and segment reference frames were calculated using the experimental markers. Anatomical markers were used to generate reference frames, and then the InverseKinematicsSolver in OpenSim 4.1 was implemented to pose the model. From there, virtual markers were algorithmically placed on the model. The final registration method used human input to place the virtual markers on the model based on a photograph of the robot with markers attached. Dunne et al. {Dunne, 2021 #494} reported an average root mean square error (RMSE) decrease for joint angles and moments of 67% and eliminated variability in registration between different users. Removing user variability is ideal for larger projects and would assist with registration of markers on the same subject over multiple testing sessions to ensure consistency. However, this method was developed with a robot where segments properties such as geometry and mass, as well as joint angles, were known. While these results are promising for improving IK errors, this specific method may be difficult to generalize to human populations with varying segmental properties.

Joint Contact Forces

The JointReaction tool in OpenSim is designed to calculate the cumulative forces and moments that are transferred between two segments arising from forces acting on the model itself. Importantly, these aggregate forces additionally represent physiological parameters that are likely not included in the model such as cartilage interactions and absent ligamentous tissue. Joint reaction analyses at the knee have been used to investigate the effects of crouch gait and model developments have also allowed for medial and lateral tibiofemoral compartment loads to be differentiated {Steele, 2012 #413;Lerner, 2015 #171}. Steele et al. {Steele, 2012 #413} utilized a common objective function in SO (Eq. 9) that had weighting constants for individual muscles that were adjusted according to contact force data from an instrumented TKA in order to assess crouch gait in children with cerebral palsy.

$$\text{Objective Function} = \sum_{i=1}^N c_i a_i^2$$

Equation 9: Objective function including squared muscle activations.

In Equation 9, N corresponds to the muscles present in the model, a is the activation level that can range from 0 to 1, and c is the weight for each muscle used by Steele et al. {Steele, 2012 #413}. Weighting constants were allowed to range from an integer of one to ten and performed simulations for all combinations of weights and selected the output that matched the *in vivo* data most.

$$\overrightarrow{R_{knee}} = [M]_{tibia} \overrightarrow{a_{tibia}} - (\overrightarrow{R_{ankle}} + \sum \overrightarrow{F_{muscles}} + \overrightarrow{F_{gravity}})$$

Equation 10: Reaction force at the knee, Steele {, 2012 #413}.

Equation 10 represents the reaction force at the knee used by Steele et al. {Steele, 2012 #413}. $\overrightarrow{R_{knee}}$ is force from the femur expressed on the tibia, $[M]_{tibia}$ is the tibial inertial

properties matrix, $\overline{a_{tibia}}$ contains six dimensions of linear and angular accelerations for the tibia, and $\overline{F_{muscles}}$ and $\overline{F_{gravity}}$ represent muscle and gravitational force applied to the tibia. After the weighting constants for SO were changed, an average error of 0.02 times BW throughout the gait cycle was obtained. These results highlight the importance of considering weight constant adjustments for SO for questions regarding joint reactions.

Estimation of joint contact forces in joint arthroplasty populations is especially important for assessing rehabilitation and activities that are done to improve general health and quality of life while reducing negative outcomes such as implant loosening or revision surgeries. Standard models in OpenSim provide a generic frame for simulation data, but significant efforts have been made to develop models that can return specialized data for clinical populations. OpenSim musculoskeletal modeling result accuracy can be further enhanced by using subject specific metrics to adjust each model through medical imaging modalities such as x-rays or MRI {Davico, 2022 #199; Fernandez, 2023 #198; Gerus, 2013 #172; Smale, 2019 #180}. Without personalization, considerable differences in simulation results using generic models can occur {Correa, 2011 #181; Scheys, 2008 #182; Davico, 2022 #199}. One such model, the Lerner OpenSim model, allows parameters such as the hip-knee-angle (HKA), as well as medial and lateral contact points, to be set on a per subject basis to produce results more alike *in vivo* knee implant results compared to a generic model {Lerner, 2015 #171}. Lerner et al. {Lerner, 2015 #171} presented results showing 12.4% and 11.9% error for peak medial and lateral contact forces, respectively, when compared to true *in vivo* data. Conversely, a generic model produced 63.1% and 42.0% peak medial and lateral contact force errors {Lerner, 2015 #171}. Building off the components from each of the Lerner, Rajagopal, and Catelli models, Bedo et al. {Bedo, 2020 #248} presented an aggregated model that still utilizes HKA and contact locations of the knee

but is more suited for tasks requiring high levels of knee and hip flexion {Bedo, 2020 #248;Rajagopal, 2016 #183;Catelli, 2019 #277}. The Bedo model allows for greater hip and knee flexion maximums while retaining kinematic, kinetic, and muscle force accuracy.

Computational Methods

Static Optimization

SO is an optimization procedure that minimizes or maximizes a given cost function at individual frames of a given set of kinematics, making this a time-independent method. The cost function used can vary from the sum of squared muscle activations to metabolic cost but is often chosen to be the former. Results from the inverse kinematics tool can be used as input motions to calculate muscle excitations that could reproduce the joint moments seen in the input data. A critical component to understand regarding SO is that the simulation results do not have a time dependence, which is one of the most significant drawbacks of SO. Specifically, since the calculations are made on a frame-by-frame basis, the simulation does not consider the previous and proceeding frames; all calculations are instead instantaneous. Therefore, dynamic inconsistency is not strictly required in this optimization approach. SO is a computationally faster method for the estimation of muscle activations compared to other methods that will be discussed, but there have been efforts to improve the efficiency of computation via linearization of constraints or with the use of cubic splines {Michaud, 2020 #471}. Michaud and Begon {Michaud, 2020 #471} found that using their linearized or spline methods for calculating muscle excitations in a pure elbow flexion model was more accurate in terms of RMSE and additionally was at least 100 times faster than traditional SO. The implications of improved computational efficiency for simulations of this kind revolve around improving the ability to perform

simulations in real-time and provide relevant feedback to individuals in clinical or research settings.

While SO is computationally superior in speed relative to other methods that will be discussed, there are some additional drawbacks and considerations when choosing to utilize this method. Considering that SO operates on positions, velocities, and accelerations of body segments, the accuracy of the results will be tied to accuracy of the collected data. These kinematic measurement systems have inherent error, but efforts to reduce noise and quantify system errors have been published and improved upon over the years {Winter, 1974 #550;Ladin, 1989 #551}. When errors are present in these calculations, net joint torque estimations will be altered and muscle force estimations will therefore be incorrect {Anderson, 2001 #549}. Further, when velocities are quantified inaccurately, then the resulting muscle length and contraction velocity estimation will additionally be inaccurate. Despite these drawbacks, SO has been suggested as the recommended optimization technique for gait simulation when only muscle force estimates are of interest {Anderson, 2001 #526}. Anderson and Pandy {Anderson, 2001 #526} reported that joint contact force estimates at the hip, knee, and ankle between static and dynamic optimization methods were similar, despite the static solution including a method that modeled muscles as ideal force generators. However, these comparisons of joint reaction forces were mostly qualitative in nature over the entire time series for a cycle of gait. The authors present cases in which SO should be cautioned against such as when experimental kinematics are unavailable, activation dynamics are critical to the movement, or an obvious time-dependent optimization problem is not present.

Forward Dynamics

Forward dynamics requires inputs of muscle excitations which can be generated by the computed muscle control (CMC) tool in OpenSim to solve a set of differential equations defined by the musculoskeletal model in question. Forward dynamics, overall, is the process of calculating generated motion from a set of known forces and/or torques as opposed to inverse dynamics which estimates forces given known motions. It is important to note that forward dynamics, in its purest form, is only supplied with initial excitations and it begins to predict values according to the optimization problem from that point. In OpenSim, applying CMC muscle activation history will hamper the predictive benefit of implementing pure forward dynamics. Computational issues associated with forward dynamics include a soft requirement to have the interval of interest be short or simulations may accrue numerical errors during integration. However, with the advent of computing innovations like parallel processing, the extreme computational time to perform forward dynamics simulations has significantly decreased with modern processors. The differential equations that govern the model's dynamics require, at the base level, muscle activations and excitations to determine the related muscle fiber velocities. Once muscle fiber velocities are obtained, they are combined with muscle fiber lengths and activations to obtain muscle force. Muscle force multiplied by the respective moment arm will result in the net muscle torques that determine the resultant kinematics.

Pandy {Anderson, 2001 #549} presents a detailed discussion on inverse and forward dynamics, identifying the critical difference between the two methods being that forward dynamics solves one optimization for each cycle of movement. Forward dynamics has a primary advantage performing integration forward in time, which allows for the inclusion of time-based muscle physiology that is absent in SO. Forward dynamics, in theory, is enticing because it

solves motion problems based on a given optimization problem without kinematic constraints. However, that benefit also introduces difficulty in selecting and formulating an appropriate cost optimization problem, as the problem will be the prime driver of the simulation. The solution space for solving the optimization problem with forward dynamics given body segment motions, ground reaction forces, muscle forces, and muscle excitations is incredibly large, taking up thousands of hours of computational time in 2001 {Anderson, 2001 #526}. The more inherent movement constraints there are or well-defined movement goals there are, the smaller the solution space will be. For example, Pandy {Anderson, 2001 #549} discusses jumping and bicycle pedaling as good movements for this as the goal of jumping is to reach the highest distance, and pedaling involves a constrained movement with minimal need for balancing the upper body. The conclusion of Pandy {Anderson, 2001 #549} was that, given accurate measurements of external forces and segment motion, static optimization should be preferred over dynamic optimization for the estimation of muscle force. However, when investigating the effect of physiological parameter changes for a given task's execution, dynamic optimization should be chosen.

Electromyography in Modeling

The use of EMG recordings coupled with joint kinematics to understand muscular force, soft tissue support, and muscular contributions to specific movements such as varus or valgus at the knee requires a mathematical representation of many physiological factors to increase simulation accuracy. Specific complexities arise with the muscle redundancy problem, which manifests itself with participants exhibiting varying activation amounts and patterns in the same set of musculature to perform a given movement {Lloyd, 2001 #441}. In work by Lloyd and Buchanan {Lloyd, 2001 #441}, subjects instrumented externally with multiple load cells

performed a variety of knee movements and it was determined that the medial gastrocnemius had two separate patterns among participants regarding its contribution to varus and valgus moments. Further, the quadriceps and hamstrings were shown to have the greatest amount of muscular support during varus and valgus moments encountered by the knee. The authors observed that varus and valgus moment arms are greater with accompanying knee extension because of the musculature being more perpendicular to the tibial plateau. Predictions of flexion and extension knee joint moments using experimental EMG in isokinetic dynamometer, sidestepping, and running tasks resulted in a mean residual error of 12 Nm and an average R^2 of 0.91 when compared to moments calculated via inverse dynamics {Lloyd, 2003 #438}. When implementing an EMG driven simulation model, some musculature may be left out on account of the musculature being too deep for surface EMG or muscles in the area can be assumed to have minimal contribution to the motion(s) of interest. For example, Lloyd and Besier {Lloyd, 2003 #438} opted to neglect the plantaris and popliteus for two reasons: quite a small physiological cross-sectional area and assumed minimal contributions to knee flexion and extension.

While EMG measurements appear to improve simulation accuracy for joint moments, the improvement of simulation accuracy of EMG in conjunction with external, experimentally measured forces is less clear. Simulations of total, medial, and lateral contact forces performed with a subject-specific CAD modelled TKA compared to *in vivo* kinetics revealed that EMG and external loads were not good predictors of internal loading {Meyer, 2013 #445}. However, Meyer et al. {Meyer, 2013 #445} did report that medial-lateral force distributions were likely better predicted by the external knee adduction moment as opposed to contact force peaks. It should be noted that this work was not done in OpenSim, and results are therefore incompatible with OpenSim literature. However, EMG-driven literature has been presented using the

OpenSim platform, with extended compatibility available through the toolbox “Calibrated EMG-Informed NMS Modelling Toolbox,” or CEINMS {Pizzolato, 2015 #543;Romanato, 2022 #544}. CEINMS has allowed research using EMG-informed simulations to be conducted on various clinical populations such as individuals with cerebral palsy, Parkinson’s disease, and hip osteoarthritis {Davico, 2020 #545;Hoang, 2019 #546;Romanato, 2022 #544}.

Computed Muscle Control

CMC is a feedback-control method for determining muscle excitations that will drive a set of known kinematics that may or may not include external forces. Muscle excitations and muscle forces are also known as actuator controls and actuator forces, respectively, in a broader sense when applied to musculoskeletal modeling in OpenSim. The CMC algorithm available in the OpenSim Application Programming Interface (API) works by combining proportional-derivative control and static optimization. Initially, a set of desired accelerations need to be computed for the model’s coordinates to be driven to the experimental coordinates. One of the unique components of CMC compared to SO is that OpenSim computes accelerations at some defined time interval ahead of the current frame for muscle forces to be given time to change. It should be noted that the initial muscle states used as the starting values for CMC occur in the first 0.030 seconds of the trial of interest, and therefore the trial should be slightly extended before the desired simulation starting point.

Currently, OpenSim has two optimization formulas for users to choose from- slow target and fast target. The slow target formula minimizes a weighted sum of squared actuator controls with the addition of the sum of desired acceleration errors, referred to mostly as RRA. The fast target formula minimizes the sum of squared controls with the addition of an equality constraint that requires the difference between the model and desired accelerations to be within some user

defined tolerance, referred to mostly as CMC. Since the fast target formula has a constraint that must be met, it is possible that the simulation will fail due to the model's actuators being too weak. To account for this, reserve actuators can make up for the failure to meet certain accelerations within the simulation. The inability to produce enough force through the sum of individual muscle torques means the constraint meeting the net joint torque has been violated. These reserves are not strictly "anatomical," and their forces should be relatively low in order to discourage their use in optimization and should sufficiently penalize the simulation for their excessive use. It should be noted that the use of reserves within SO or CMC is not strictly necessary for simulation completion, and therefore the reserve values should be checked to ensure they are within an acceptable range.

Hicks et al. {Hicks, 2015 #414} present multiple case studies, including examples that showcase the reduction of reserve actuators as well as comparisons of muscle activations to experimentally measured EMG. This paper provides a multitude of guidelines for validating model results, parameters, and dependencies to be aware of, and determining if certain the results are "good enough" relative to previous literature. For example, Hicks et al. {Hicks, 2015 #414} discuss a case study where vertical and frontal plane residual forces and moments oscillate in a non-physiological manner, and use center of mass adjustments as well as minor joint angle changes to reduce these to more acceptable levels. A specific goal presented was to attain reserve joint torques that are less than 5% of net joint moments. Experimental EMG can also be compared to simulated activations to assess the timing of onset and offset of force generation, as well as peak muscle activity {Hamner, 2010 #449}. Investigating the activation levels of specific musculature can be critical when those muscles have a pronounced effect on the center of mass

movement. Failure to check these activations could result in faulty conclusions depending on the research question at hand.

A limitation of CMC, within the OpenSim user interface, compared to SO is that a limited objective function available in CMC, which is only to minimize the sum of squared muscle activations. SO and CMC additionally can rely on reserve actuators when a model actuator is unable to produce enough force to meet the simulation's requirements. One of the benefits of CMC implementation is that it allows for the inclusion of the temporal offset in force generation relative to muscle excitations. Further, CMC requires torque production to be generated over a previous set of time steps as opposed to pure static optimization that isolates individual frames. As a result, time-dependent criteria are not explicitly included, but since activation dynamics are used there is an increase in physiological relevance with CMC.

Optimization Method Comparison

SO and CMC methods have been compared in gait in the literature with reports of increased error in knee extension torques as well as increased muscle forces, activations, and cocontraction indices in CMC optimization {Roelker, 2020 #375}. Roelker et al. {Roelker, 2020 #375} utilized two different OpenSim models, gait 2392 and Full Body Running, to perform the comparison and reported that the changes in CMC outputs were additionally more sensitive to model choice compared to SO. An additional comparison of SO and CMC in gait and running trials concluded that SO was recommended for estimating muscle forces in human locomotion using previously published model parameters {Lin, 2012 #450;Anderson, 1999 #451}. Joint moments calculated for gait and running in all three planes via SO had less than 1 Nm of reserve torque difference from inverse dynamics calculations, while CMC was able to achieve the same residual value for the sagittal plane only by additionally adjusting the kinematics through the

residual reduction algorithm (RRA). RRA is designed to slightly adjust the torso center of mass and kinematics to obtain inverse kinematics that are more dynamically consistent with the experimentally measured ground reaction forces. The RRA tool does not use musculotendon actuators and modifies the kinematics slightly in attempts to minimize the dynamic inconsistencies that were present. Once kinematics have been altered, the torques at certain joints are also likely to deviate from what would be calculated from inverse dynamics. The authors discuss that the ability of SO to accurately reproduce joint moments while being approximately five times more efficient in computational load makes it a superior choice for modeling purposes. Additionally, the authors present that since forward dynamics simulations may be prone to compounding errors due to user-defined model parameters, SO should again be favored. However, the authors explicitly describe cases in which SO is not advised for estimating muscle forces and methods such as CMC should be considered. Specifically, activities that are highly time-dependent or ballistic in nature are not likely to have muscle activation patterns that are well matched by SO. The reason these activities may not be applicable to SO is that the cost functions are likely to be variables such as muscular power output or impulse that require the inclusion of time component to calculate.

A single leg triple hop test has also been experimentally measured and implemented in OpenSim with a 21 segment, 92 muscle model that was modified from a previously published model {Hamner, 2010 #449;Alvim, 2018 #448}. For a high force task such as this, the RRA tool was used. Alvim et al. {Alvim, 2018 #448} concluded that the RRA was necessary for the preparatory phase of the hopping, but considering the small residuals from SO and CMC, the additional time required to recursively apply the RRA should be considered.

Comparison of SO and CMC in human activities is difficult because the ground truth measurements of humans to compare simulation results to are often less direct than in animal models. An *in vivo* cat hindlimb validation experiment compared SO and CMC to experimentally measured EMG and muscle force recorded by buckle transducers on the tendons of the soleus, medial gastrocnemius, and tibialis anterior during free movement {Karabulut, 2020 #447}. The SO derived forces for the medial gastrocnemius, soleus, and tibialis anterior were more comparable to the *in vivo* measurements compared to CMC. RMSE between the SO and CMC muscle forces were significantly different in level walking for the medial gastrocnemius and tibialis anterior. For 30° and 45° incline walking, SO and CMC muscle forces were significantly different from all three measured muscles. To further investigate why CMC forces were often greater than SO forces, the authors performed an analysis on the passive component of muscle force contribution that is only present in CMC. The authors' conclusion was that CMC is more sensitive to anatomical musculoskeletal parameters such as tendon slack length. While CMC has the benefit of additional physiological parameters, if those additional parameters are not measured experimentally or previously validated, the accuracy of the simulation may be compromised or at least be more sensitive to musculotendon parameter values.

Knee Models

Usage and Validation

Modeling of the knee joint allows researchers and clinicians to acquire estimated magnitudes of knee joint contact forces during motion. Asymmetric knee contact forces are likely contributors to the progression of knee OA and quantification of forces before TKA in patients with end-stage OA are useful to describe the knee's kinetic behavior as a result of known cartilage wear {Zhao, 2007 #249}. Understanding knee contact forces in TKA patients is

important as unexpected magnitudes and proportions of force on the lateral and medial compartments may lead to implant loosening and mechanical failure which have been associated with costly revision surgeries {Andriacchi, 1986 #495;Nham, 2023 #184}. In lieu of expensive and uncommon instrumented knee implants used by Kutzner et al. {Kutzner, 2012 #4}, OpenSim allows for tibiofemoral reaction forces to be estimated from measured external kinematics and kinetics {Hummer, 2022 #169}. Popular models include the ‘Lower Limb Model 2010’, ‘Rajagopal2016 Full-Body Model’, ‘Lai2017 Full Upper and Lower Body Model’, and the model developed by Lerner et al. {Rajagopal, 2016 #183;Lerner, 2015 #171;Lai, 2017 #394;Arnold, 2010 #393}. The fifth Grand Challenge Competition to Predict *in vivo* Knee Loads has been used to compare TKA data between the Arnold, Rajagopal, and Lai models {Curreli, 2021 #160;Fregly, 2012 #395}. Curreli et al. {Curreli, 2021 #160} reported that correlations (R^2) between predicted and measured values of force transmitted to the knee joint were 0.22, 0.77, and 0.80 for the Arnold, Lai, and Rajagopal models, respectively. Explanations for these discrepancies are found in the differences between muscle architecture definitions and musculotendon dynamics parameter changes between the models. For example, the Arnold model uses older fiber dynamic characteristics such as force-length and force-velocity characterizations compared to Lai and Rajagopal who have updated these parameters and introduced additions to reduce unnecessary co-contraction of musculature due to increased passive forces.

Comparisons of available models designed with tibiofemoral contact force calculations in mind are limited, but literature comparing the Rajagopal, Bedo, and a further modified model has been published {Pelegriinelli, 2023 #382}. Pelegriinelli et al. {Pelegriinelli, 2023 #382} used SO as inputs for the joint reaction analysis and found that RMSE relative to *in vivo* contact forces

between the Rajagopal and Bedo models were inferior to the newly created model presented for the sit-to-stand trials. In gait tasks, all three models performed similarly for total knee and medial contact force magnitudes with muscle activation peaks matching corresponding EMG peaks. The authors note that vertical knee joint center position influenced contact forces in participants with varus deformities. Specifically, considering the medial and lateral compartment forces were calculated with respect to the knee varus and valgus moments, the knee joint center position adjusted the resulting contact forces. Lateral compartment force being lower than the medial compartment force during gait was seen, which has been documented in related research, even with point contact modelling of the knee {Saliba, 2017 #452}.

Subject Specific Models

Subject specific modeling can have a broad range of parameters, but the goal of each one is to theoretically increase the accuracy of a given simulation by introducing as much relevant information as possible to the model. In many cases, this can mean the inclusion of imaging data of bone geometry and joint centers, or it could mean adjusting muscle strength values based on experimentally recorded dynamometer data. Increasing the model's complexity is not a guaranteed improvement, and in fact has been cautioned when validating a model or simulation results {Hicks, 2015 #414}. However, there is a balance to be struck by including anatomically relevant parameters for the research question of interest while ensuring oversimplifications are not made elsewhere.

Subject specific knee models have the benefit of publicly available *in vivo* kinetic data for validation, and a variety of personalization methods have been published. Generic SO has been compared to personalization methods such as applying weighting constants to muscle groups at the knee, pattern search optimizations for weighting constants, and subject specific isometric

knee flexion and extension strength parameters {Knarr, 2015 #369}. The highest average RMSE of knee contact force measured in bodyweights was 0.67, 0.48, 0.41, and 0.4 for standard static optimization, previous literature muscle weights, optimized muscle weights, and strength specific adjustments, respectively. A highly relevant conclusion by Knarr and Higginson {Knarr, 2015 #369} was that the weighting constants were sensitive to the cost function and amount of data in the simulation. Specifically, the cost function was either minimization of RMSE or RMSE with the addition of peak errors and the data amount changed with a full gait trial or simply simulating the stance phase. Subject specific model parameter inclusion should be generalizable and since the optimized muscle constants were based on known *in vivo* loads, this method is unlikely to be used clinically.

The Lerner model has been used to investigate the sensitivity of mediolateral knee contact forces to alterations to tibiofemoral alignment and medial and lateral contact locations in gait trials of healthy and knee OA participants {Saliba, 2017 #452}. Monte Carlo analyses were performed to investigate the effects of simultaneous parameter changes and their subsequent effects on joint contact force. The perturbations were in the range of -10° to 10° and -10 to 10mm for the ranges of hip and knee angle, and medial and lateral contact point distances from the joint center, respectively. The medial and lateral contact point distance values were not used as absolutes, instead they were scaled by the knee width of the participant. The sensitivity analysis revealed that the perturbations resulted in up to three times bodyweight differences for the first and second contact force peaks as well as the mean stance phase values for the outcome measures. Perturbations of the HKA resulted in 5-6% bodyweight per degree (BW/deg) medial contact force and 3-6% lateral contact force BW/deg for the varus and valgus perturbations, respectively. Additionally, medial contact force was reduced by 2-4% bodyweight per mm

(BW/mm) and lateral contact force was increased by 1-3% BW/mm with increased medial contact point distance. Saliba et al. {Saliba, 2017 #452} also demonstrated that sensitivity to these perturbations vastly differ between participants that exhibited condylar unloading from those that did not. In this work, the presence of lateral condylar unloading during the simulation was met with a tensile force provided by a representative collateral ligament {Brandon, 2014 #453}. Despite parameter perturbation, no significant changes were detected in knee flexion moments or muscle forces. These findings suggest that the inclusion of subject specific HKAs and compartmental contact points are valuable for increasing the accuracy of simulated knee joint contact forces, especially for patient populations.

Creating subject specific geometry can alternatively consist of EMG-driven models with the inclusion of computed tomography scan bone geometries and fluoroscopic kinematics of a participant's instrumented knee during gait trials {Gerus, 2013 #172}. Comparisons were made for all permutations of generic or subject specific kinematics and segment geometry. Subject specific segments included the femur, tibia, fibula, and patella. Gerus et al. {Gerus, 2013 #172} found that subject specific geometry did improve RMSE in medial contact force estimates relative to *in vivo* measurements compared to the other models. Fluoroscopic kinematics did not result in a significant difference in RMSE for the estimation of either compartmental contact force, which provides some evidence that geometry adjustments may be sufficient for improving estimates without the difficulty of obtaining fluoroscopy measurements.

Additional simulation techniques such as Monte Carlo have been utilized to further investigate the relationship between kinematics and knee joint compressive forces. Gait analysis of the CAMS dataset using Monte Carlo simulations shows that internal knee joint compressive forces are more tied to muscle activation patterns as opposed to kinematic aberrations {Febrer-

Nafria, 2023 #320}. Additionally, Febrer-Nafria et al. {Febrer-Nafria, 2023 #320} reported that implant alignment had a significant kinematic effect in the frontal and transverse planes. Highly personalized models incorporating x-ray, MRI, and FEA modalities in two varus aligned unilateral TKA patients have shown that second peak medial plateau contact force during gait changes from approximately 85% to 68% of total contact force 6 months after TKA {Du, 2022 #321}. Knee joint contact force has further been investigated using gait simulations and a model with 23 DoF, 54 musculotendon actuators, with a knee consisting of one DoF to compare knee OA severity by group {Richards, 2010 #384}. Decreased overall loading of the knee is observed with increasing OA severity, however, only the second contact force peak was significantly different between OA severity groups. The authors postulate that increasing OA severity can be linked to compensations via decreased muscle force, slower gait speeds, and finally a resulting decrease in contact force.

Cycling Models

Musculoskeletal modeling in cycling populations has been performed using the standard gait2392 model as well as the Lerner model developed in 2015 {Lerner, 2015 #171;Hummer, 2022 #169;Thorsen, 2021 #290}. Both models consist of one DoF (flexion/extension) for the tibiofemoral joint. In a study consisting of 6–18-month post-operative TKA patients, the gait2392 model was used to assess medial and lateral compartmental loading during stationary cycling at 80 RPM with power settings of 80 and 100 W {Hummer, 2021 #200}. Hummer et al. {Hummer, 2022 #169} used a 3-way ANOVA to assess the interactions between knee compartment, limb, and workrate. The only significant interaction found was between compartment and limb. The increase in 20 W did not result in significant increases in the peak, medial, or lateral contact forces. One of the main conclusions from this work was that the

nonreplaced limb of TKA patients had a greater peak medial contact force compared to the lateral compartment. This finding may raise concerns about contralateral knee OA progression as the trend may be that these participants are loading their nonreplaced limb as a subconscious compensation mechanism during cycling.

An additional study investigated the effect of q-factor (pedal width) on medial compartment loading in stationary cycling using the Lerner model {Thorsen, 2021 #290}. Important findings from this study included the finding that regression analyses revealed that knee abduction moment and knee extension moments accounted for 87.4% of the variance in medial compartment contact force when q-factor was accounted for. The findings from Hummer et al. {Hummer, 2022 #169} show an approximately 21% increase in peak knee extension moment in the nonreplaced limb of TKA patients with no accompanying change in knee abduction moment between limbs. This finding aligns with reports by Thorsen et al. {Thorsen, 2021 #173} associating increased medial compartment loading with increased knee extension moment.

OpenSim has a supplemental tool called Moco that allows for state and control trajectories to be estimated using polynomial splines, a method known as direct collocation {Dembia, 2020 #476}. One of the presented advantages of Moco is the ability to solve problems with multiple objective functions and/or constraints that were traditionally limited to one function, as seen in CMC {Dembia, 2020 #476;Dembia, 2020 #476}. However, literature has been published where multiple objective functions have been combined during gait, specifically to minimize the sum of muscle activations and joint loads {Demers, 2014 #547}. An additional proposed advantage of Moco is the inclusion of muscle dynamics that are absent in SO, but without the computational cost of more complex simulation methodologies. Moco has been used

to estimate muscle forces and knee joint reaction forces in cycling data from 16 healthy participants who cycled for three minutes {Clancy, 2023 #434}. Participants completed the trials at self-selected cadences of 75-99 RPM and workloads set to generate a heart rate of 70-75% age-estimated maximum which was 40-216 W for the study. One of the objective functions used was to minimize the muscular effort during cycling, which consisted of a combination of the sum of squared muscle excitations and the sum of squared reserve and residual excitations for each DoF. The second objective function was designed to minimize knee forces by minimizing the sum of squared tibiofemoral forces. Results from Clancy et al. {Clancy, 2023 #434} may have actionable clinical rehabilitation impacts regarding muscle coordination during dynamic tasks. Specifically, altering coordination to increase soleus force and decrease gastrocnemii forces lead to reduced knee contact forces, which has also been shown to be trainable in one session using biofeedback during the late stance {Uhlrich, 2022 #477}. The addition of tibiofemoral force to the objective function resulted in an average peak contact force reduction during cycling that was closer to *in vivo* data when compared to an objective function only based on muscle excitations {Kutzner, 2012 #4}. Mean peak tibiofemoral force from Clancy et al. {Clancy, 2023 #434}, simulated, was approximately 1200 N while Kutzner et al. {Kutzner, 2012 #4}, *in vivo*, recorded ranges of 793 to 1520 N with average power settings of 137 and 120 W, respectively between the two studies. The results from Moco in the work from Clancy et al. {Clancy, 2023 #434}, however, did present some EMG timing discrepancies when compared to previous cycling literature {Ryan, 1992 #479}.

Chapter 3: Materials and Methods

Participants

Individuals with end-stage knee OA that received a TKA procedure at the Tennessee Orthopaedic Alliance (TOA) were recruited to participate in this study. The collection session was between 1-2 months post operation. Potential participants were identified by a TOA physical therapist and given an information sheet about the study. Participants were only given further study information if they were in alignment with the inclusion and exclusion criteria (Appendix A, **Table 2**) for participation. Contingent upon consent to release their patient health information to researchers, long standing and short frontal plane radiographs of the knee were obtained for use in the musculoskeletal modeling component of the study. Participants were compensated with \$90 after completion of the testing session and were given no compensation if the testing session was only partially completed.

An *a priori* power analysis to determine sample size for the study was conducted using GPower 3.1 (Heinrich Heine University Düsseldorf). To obtain a power of 0.80 with an α of 0.05, mean knee range of motion with an effect size of 0.979 from a previous study of submaximal ergometer cycling in a healthy population were used as input with a resulting sample size of 4 participants (submitted manuscript). This study was approved by the Institutional Review Board at the University of Tennessee with the study identifier UTK IRB-23-07760-XP. This study received financial support through a 2024 Matching Dissertation Grant from the International Society of Biomechanics. Prior to participation, all participants read, signed, and gave informed consent after as much time as they required.

Instrumentation

Three-dimensional (3D) kinematic data were recorded at 240 Hz using a twelve-camera motion capture system (Vicon Motion Analysis Inc., United Kingdom) for the cycling trials.

Retroreflective markers denoting anatomical landmarks of interest were placed on the acromion processes, iliac crests, greater trochanters, medial and lateral epicondyles of the femur, medial and lateral malleoli, 1st and 5th metatarsal heads, and the distal 2nd distal phalange. Semi-rigid, molded thermoplastic shells with four retroreflective markers attached were affixed via neoprene straps or harnesses to the trunk, thighs, and shanks to track segmental motions. Additional clusters were placed on the lateral heel of each shoe to track the foot segment, modeled as a rigid body.

For all cycling trials, an electromagnetically braked stationary ergometer (Excalibur Sport, Lode B.V., Groningen, Netherlands) was used. The crank arms of the ergometer were outfitted with custom-machined metal crank arms that allowed for asymmetric adjustments in length from 130-190mm on either side. Kinetic data were recorded with customized instrumented bike pedals fitted with two 3D force transducers (1200 Hz, Type 9027C, Kistler, Switzerland) coupled with two industrial charge amplifiers (Type 5073A and 5072A, Kistler, Switzerland). This instrumentation allowed for the recording of 3D forces and moments at the pedal-foot interface. Kinetic data were temporally synchronized to kinematic data through the Vicon Nexus software suite during cycling trials.

To record muscle activity during cycling, a wireless 16-channel EMG system (Trigno Wireless EMG, Delsys, Natick, MA, USA) was used with a 1926 Hz recording frequency. Sensors were placed according to SENIAM guidelines bilaterally on the following muscles: Rectus femoris, semitendinosus, medial gastrocnemius, and tibialis anterior. Prior to sensor placement, the participant's skin was shaved, rubbed with an abrasive preparation gel, and cleaned with an alcohol wipe. To prevent unwanted movement, all sensors were applied with double sided tape, wrapped in athletic pre-wrap, and taped together.

Experimental Protocol

Studies 1 – 3

TKA participants came to the lab for one testing session one to two months after their TKA procedure. The testing session included completion of the informed consent packet, participant health information release, basic demographic information, and the PARQ+. Participants were allowed to continue their participation in the study if they answered “no” to the first page of the PARQ+ survey in addition to informed consent signing. additionally, participants completed of the Tampa Scale of Kinesiophobia and the Knee Injury and Osteoarthritis Outcome Score surveys.

Participants were familiarized with the ergometer they would be using first, and a procedure for proper bike fit was conducted for each participant. The ergometer handlebars were adjusted to create a 90° hip angle while the saddle height was altered to create 25-30° of knee flexion with the pedal at bottom dead center. If participants were highly uncomfortable or unable to cycle with the amount of knee flexion, the range was increased to approximately 40°. The saddle fore-aft position was also adjusted to have the patella above the pedal spindle at the most forward pedal position as measured with a plumb bob placed on the superior patella.

Clinical testing was then conducted, which began with passive knee ROM testing with the participant lying supine on a patient table with an approximately 90° hip angle. The tibia was moved gently through its maximum ROM until the end of motion was felt by the researcher or the participant requested to stop. This procedure was completed three times, and an average of the measurements were taken. Knee circumference was then taken using a sewing tape measure around the widest aspect of the knee. This procedure was completed three times per limb, and an average of the measurements were taken. The final clinical test was the timed up-and-go (TUG)

test that involved participants standing from a chair and walking three meters around a cone and returning to the seat at a comfortable pace. Participants were not allowed to use their arms to push off or out of the seat.

EMG placement was conducted according to SENIAM guidelines. Functional movements were used to normalize EMG data in lieu of maximal contraction normalization methods to reduce strain on the participants. For each movement, three repetitions were conducted at a pace of 50 BPM as controlled by a metronome. A bodyweight squat movement was used to normalize the rectus femoris, semitendinosus, and anterior tibialis while the calf raise was used to normalize the medial gastrocnemius.

Retroreflective markers denoting anatomical landmarks of interest were placed on the acromion processes, iliac crests, greater trochanters, medial and lateral epicondyles of the femur, medial and lateral malleoli, 1st and 5th metatarsal heads, and the distal 2nd distal phalange. Semi-rigid, molded thermoplastic shells with four retroreflective markers attached were affixed via neoprene straps or harnesses to the trunk, thighs, and shanks to track segmental motions. Additional clusters were placed on the lateral heel of each shoe to track the foot segment, modeled as a rigid body.

For the cycling conditions (Appendix A, **Table 3**), participants cycled in a random order on the ergometer at 70 RPM and 70 Watts. Participants were given a three-minute warmup on the cycle ergometer and additional time was given for familiarization with the first asymmetric crank condition (C1 or C2) that was tested. Participants cycled for two minutes at each condition and data were recorded only during the final 10 seconds. From this section of data, the first five consecutive full crank revolutions were extracted and counted as separate trials for further data

analysis. Following each cycling condition, participants were asked to score their pain on a 1-10 numeric visual analog pain scale as well as a 6-20 Borg rating of perceived exertion (RPE).

3.4 Data Analysis

Studies 1 – 3

The software Visual 3D (C-Motion Inc., Germantown, MD, USA) was used to process 3D kinematics, kinetics, and EMG data exported from Vicon Nexus. Kinematic data were interpolated, and then kinematic and kinetic data were filtered using a 4th order zero-lag Butterworth filter with a cutoff frequency of 6 Hz. 3D kinematics were resolved with a Cardan sequence of X-Y-Z in the joint coordinate system. The inverse dynamics method in Visual 3D calculated relevant joint moments. Kinematic and kinetic variables were reported in accordance with the right-hand rule. Events of interest were manually selected or assessed using a custom Python (Python Software Foundation, Wilmington, DE) script to visualize each variable's time series curves for the cycling trials. For each variable of interest, five discrete data points were selected and averaged to obtain a final value for later statistical analysis.

Raw EMG data were band pass filtered (20-450Hz) by the collection hardware and digitally high pass filtered at 20 Hz to remove the zero offset. Filtered data were then full wave rectified before a moving root mean square function with a 91ms window size applied. The fully processed data were normalized to the maximum value seen for each muscle during a complete crank cycle for the respective limb. EMG data were additionally exported for use in musculoskeletal modeling validation.

Study 3

Musculoskeletal modeling was performed in the open-source software OpenSim (v4.5 OpenSim, SimTK, Stanford University) that performs a variety of simulation capabilities

including inverse kinematics, muscle force estimation, and joint reaction force calculations {Delp, 2007 #221}. Kinematic and kinetic cycling data were exported from Visual 3D and SIMM files representing each of the five trials per condition were used. Visual3D exports SIMM files with scaling and inverse kinematics done through Visual3D during the export process, allowing the user to move directly to Static Optimization and bypassing the Scaling tool within OpenSim. A publicly available musculoskeletal model, referred to as the Lerner model throughout, consisting of 18 segments, 12 DoF, and 92 muscle-tendon actuators was used in all simulations {Lerner, 2015 #171}. This model resolves medial and lateral compartment contact forces of the tibiofemoral joint while also allowing for subject specific tibiofemoral alignment and medial and lateral contact points to be specified by the user. The base Lerner model was adjusted with radiographically determined tibiofemoral alignment along with medial and lateral contact points of the replaced knee by modifying the relevant weld joints as described in the model's original paper {Lerner, 2015 #171}.

Once a fully informed subject-specific model was created for each subject, muscle activations and muscle forces were obtained through the OpenSim Static Optimization tool {Steele, 2012 #413}. The objective function of the static optimization performed minimizes the sum of squared muscle activations. Previously published guidelines for simulation quality assessment and results validation were used to check maximum reserve torque actuator values {Hicks, 2015 #414}. Finally, joint contact forces for the medial and lateral tibiofemoral compartments were calculated using the 'Joint Reaction Analysis' tool in OpenSim and were expressed relative to the tibia {Steele, 2012 #413}. Two extrema were identified for the total, medial compartment, and lateral compartment curves for each subject across the trials of a particular condition.

Statistical Analysis

Study One

Two-way 2x5 (Limb x Condition) repeated measures analysis of variance (ANOVA) with an α level of 0.05 were implemented for all variables of interest to investigate significant effects of crank arm length (IBM SPSS Statistics 29, Armonk, NY). Two one-way ANOVAs were used to assess pain and RPE across conditions. The Shapiro-Wilk test was used to check for normality, and the Greenhouse-Geisser correction used if sphericity was violated. Non-parametric ANOVAs were not used because of the exploratory and patient-population nature of this work. No statistical adjustments were made to post-hoc comparisons. Effect sizes were reported as partial eta squared (η_p^2) and interpreted as small ($\eta_p^2 < 0.06$), medium ($0.06 \leq \eta_p^2 < 0.15$), or large ($\eta_p^2 \geq 0.15$) {Cohen, 1988 #152; Lakens, 2013 #552}.

Study Two

For each analyzed variable, the discrete value for the replaced side was subtracted from the contralateral side and that difference was used as the input for the ANOVAs. Specifically, two-way 2x5 (Group x Condition) ANOVAs with an α level of 0.05 were implemented for all variables of interest to investigate significant interactions and/or effects of group or crank arm length condition. Furthermore, normality, adjustment, and effect size procedures are the same as for *Study One*.

Study Three

To assess the potential effects on TCF, MCF, and LCF, one-way (crank length: 3) repeated measures ANOVAs were used with an α level of 0.05 set as the significance threshold. Furthermore, normality, adjustment, and effect size procedures are the same as for *Study One*.

Chapter 4: Effects of Asymmetrical Crank Arm Length Changes on Knee Biomechanics of TKA Patients During Ergometer Cycling

Abstract

Ergometer cycling is a commonly used exercise modality for patients recovering from a total knee arthroplasty, but most ergometers use fixed, symmetrical crank arm lengths. Recovering patients often have difficulty with knee range of motion after surgery and asymmetrically adjusting the crank arm length for this population may accommodate for the stiff knee joint.

Purpose: To determine the biomechanical effects of asymmetrical crank arm usage at the replaced limb of knee arthroplasty patients 1-2 months after their surgery. **Methods:** Seven participants with total knee replacement engaged in five cycling conditions of varying asymmetrical and symmetrical crank length shortening at 140mm, 155mm and 170mm (70 W, 70 RPM). **Results:** Decreases in crank length, symmetrically and asymmetrically resulted in significant decreases in knee and hip range of motion (ROM) as well as peak knee extension power. Decreased peak knee extension moment (KEM) of replaced limbs compared to contralateral limbs was seen for some crank lengths. Pain levels were not increased compared to the pre-testing pain levels and between conditions of the replaced knees. **Conclusion:** Knee kinematics were the primarily altered variables in response to crank arm length shortening and changes in affected muscle lengths due to decreased ROM likely explain a lack of increased KEMs. Asymmetrical crank arm length shortening did not increase KEMs or pain for patients 1-2 months after total knee arthroplasty surgery and may be considered for total knee arthroplasty rehabilitation.

Keywords: knee, arthroplasty, cycling, crankarm length, rehabilitation

Introduction

One of the more prevalent orthopedic surgeries is the total knee arthroplasty (TKA) that is the most common avenue to remedy end-stage knee osteoarthritis (OA). The number of TKA

surgeries from the US based on Medicare data was approximately 480,000 in 2019 and has a projected annual growth of approximately 4.44% {Shichman, 2023 #224}. A primary concern in the functional recovery of TKA patients is the lack of adherence to exercise regimens which can result in stunted recovery and potential weight gain {Bakaa, 2022 #219}. Often, these surgeries are initially caused by knee OA due to excess body weight {Nicholls, 2012 #202;Kulkarni, 2016 #204}. One of the difficulties in early recovery from a TKA is obtaining functional knee range of motion (ROM) as preoperative capability has been shown to only be reached one year after surgery {Köglberger, 2022 #589}. The reduced knee ROM primarily stems from neuromuscular surgical damage as well as effusion of the knee joint region, causing joint stiffness and pain. While many post-operation exercise modalities are available, cycling is an activity that reduces weight-bearing loads of the knee compared to gait while still providing exercises and muscular strength improvements of lower limb {Kutzner, 2012 #4;Logan, 2023 #216;Oja, 2011 #217}. Early in the TKA rehabilitation process, stationary cycling has been implemented to improve early participation in exercise, knee ROM, pain reduction, and patient reported outcome scores with an exercise regimen started approximately 2 weeks after TKA {Zech, 2015 #220}.

Cycling modifications may be beneficial to accommodate these limitations, one such modification on a cycle ergometer is the crank length. Adjustment only of the replaced knee side crank arm length may be able to accommodate for the reduced ROM seen after surgery in this population. Previous literature suggests that symmetrical shortening of the crank arms decreases knee joint ROM and lower limb joint velocities {Barratt, 2011 #286}. In the TKA population, symmetrical shortening may not be ideal when the contralateral limb is able to perform the movement without the same ROM limitations. TKA patients have shown significantly reduced peak knee extension moment (KEM) at the replaced limb on a cycle ergometer even after six

months after surgery as well as during level walking, ramp and stair navigation and patients may find it functionally difficult to perform with the replaced limb shortly after surgery {Hummer, 2021 #200;Yocum, 2022 #222;Wen, 2019 #567}.

Asymmetric crank arm usage has shown evidence for equalization of kinematic but not knee or ankle kinetic asymmetries between limbs in individuals with transtibial amputations as well as reduced knee extension requirements during submaximal cycling {Childers, 2014 #225;Millour, 2020 #226}. However, there has been scarce research on the implementation of asymmetric crank arms, and no studies, to our knowledge, have examined applications in the TKA population, and their usage may improve outcomes in rehabilitation and encourage early exercise adoption {Zhang, 2025 #591}. Therefore, the purpose of this study is to examine the novel idea of using a pair of adjustable crank arms to examine asymmetrical shortening of crank length in accommodating for ROM difficulties and pain during cycling at the knee of patients in early post-TKA rehabilitation {Gaffney, 2016 #175;Hummer, 2021 #200;Kline, 2020 #176}. It is hypothesized that knee ROM, and peak KEM and knee extension power will be lower in the most recently replaced knee during asymmetrical crank length conditions when compared to the contralateral knee of patients with at least one TKA. Secondly, it was hypothesized that replaced knee pain and rating of perceived exertion (RPE) will be lower for asymmetrical crank length conditions compared to the symmetrical conditions.

Methods

Participants

Seven participants (4 female, 3 male, 67 ± 7 years, 1.7 ± 0.14 m, 95 ± 23.6 kg, 31.5 ± 4.6 BMI, 54 ± 4 days post-TKA) with at least one TKA (3 bilateral, 4 unilateral) were recruited to participate in this study (**Table 1**). The most recent implant designs for all patients included

Stryker Triathlon CR, Smith & Nephew Oxinium Journey II CR, or Zimmer Biomet Persona Narrow PS. All participants were from a local physical therapy clinic and operated on by the same surgeon at Tennessee Orthopaedic Alliance East. The comparisons made for this study considered the most recently operated on knee as the “replaced” knee and the contralaterally replaced knee was considered the “contralateral” side for bilateral TKA recipients. All participants completed testing between 1 and 2-months post-operation and were required to be between the ages of 50 and 75. Inclusion criteria included the capacity to reach at least 100° of knee flexion passively in the most recently replaced knee as measured and reported by the local physical therapy clinic. Exclusion criteria included those with a BMI over 40, neurological conditions that would prevent walking or cycling without external aid, total or partial lower limb amputation, cardiovascular disease, or diagnosed OA of the ankles, hips, or spine. Additionally, participants were excluded if they had received a contralateral TKA within 4 months of their most recent surgery date. Using data from an unpublished study, a power analysis (G*Power 3.1; Heinrich Heine University Dusseldorf, Dusseldorf, Germany) using sagittal knee ROM during submaximal asymmetric crank arm ergometer cycling in a young, healthy population was conducted with an effect size of 0.979 and an α level of 0.05 to achieve a power of 0.80 {Zhang, 2025 #591}. The analysis revealed a minimum sample size of 4 participants were needed. All participants signed the approved informed consent form before study participation approved by the university’s Institutional Review Board.

Instrumentation

An electromagnetically braked cycle ergometer (Excalibur Sport, Lode, Groningen, Netherlands) equipped with a visual cadence and power screen was used to carry out all cycling tasks. The Lode Ergometry Manager software package was used on an external laptop and

connected to the bike to control workrate. In order to adjust crank arm lengths asymmetrically, adjustable crank arms (Sport crank, Lode) were used with a minimum adjustment of 5mm from 90mm - 185mm. Recording of 3-dimensional pedal reaction forces (PRFs) was done through a pair of custom-built instrumented bike pedals outfitted with two 3-dimensional force sensors (Type 9027C, Kistler, Switzerland), and amplifiers (Type 5073A, Kistler, Switzerland) recording analog data at 1200 Hz. The bike's position was aligned with the laboratory coordinate system using a custom metal jig attached to an in-ground force platform. Prior to data collection, a bike fit procedure was carried out that included adjustment of the saddle height to elicit 25-30° of knee flexion in the most recently replaced limb with the pedal at bottom dead center. For participants where this range was too restrictive or painful, a position resulting in approximately 40° of knee flexion was used. The fore-aft location of the saddle was also manipulated to vertically align the replaced knee with the pedal's spindle with the crank arm at approximately 90° of a crank cycle. The anteroposterior position of the handlebars was adjusted to achieve a 90° between the trunk and thigh.

A 12-camera motion analysis system (240 Hz; Vicon Motion Analysis Inc., Oxford, UK) was used to capture optical data. During data collection, each participant was provided with a pair of standardized laboratory shoes (Zoom Pegasus 34, Nike). Participants were outfitted with retroreflective anatomical markers and segment tracking marker clusters mounted on semi-rigid thermoplastic shells. For the static trial, anatomical markers were placed bilaterally on the following bony landmarks: acromion process, iliac crest, anterior superior iliac spine, posterior superior iliac spine, greater trochanter, medial and lateral femoral epicondyles, medial and lateral malleoli, the heads of the 1st and 5th metatarsals, and the distal phalanx of the second toe. Marker clusters, each consisting of four non-coplanar markers on semi-rigid thermoplastic, were

attached with Velcro to neoprene straps on the trunk and bilaterally on the posterolateral thighs, shanks, and heels. Heel clusters were secured with double-sided tape to the posterolateral portion of the shoe's rigid heel. Two separate 2-marker clusters were placed on the posterior pelvis to track the pelvis segment. PRF and kinematic data were synchronized and collected using Vicon Nexus (Version 16.0; Vicon Motion Analysis Inc., Oxford, UK) on the same computer system .

Experimental Protocol

Participants engaged in one laboratory collection session. Prior to data collection, participants completed the informed consent form, patient health information release, physical activity readiness questionnaire (PARQ+), Knee Injury and Osteoarthritis Outcome Score (KOOS) survey, and the Tampa Scale of Kinesiophobia (TSK). The first clinical test was the timed up-and-go (TUG) test, where participants stood from a seated position, walked 3 meters, and returned to the seat; it was performed three times and averaged. Next, knee circumference and passive knee ROM were taken with participants laying supine on a treatment table with a goniometer.

Before the actual testing, the participant warmed up for 3 minutes on the cycle ergometer with symmetrical 170mm crank arms (**Figure 1**, all figures available in the appendix). The five cycling conditions were as follows for the replaced/contralateral limbs (in mm): C1 (140/170), C2 (155/170), C3 (170/170), C4 (155/155), C5 (140/140). The participants cycled in each condition for 2 minutes at 70 W with a requested cadence of 70 RPM. Data were only recorded for the final 10 seconds of each 2-minute bout. Each participant completed the standard crankarm length on both sides (C3) first before completing the remaining four conditions in a randomized order. Participants were asked to maintain 70 ± 2 RPM to the best of their ability and were given a rest period between each condition that was a minimum of 2-minute or as long as

needed. All participants were able to maintain a minimum of 65 RPM. The participants were asked to rate their knee pain on a 0-10 numerical visual analog scale and their rating of perceived exertion (RPE) from 6-20 using the Borg scale.

Data Analysis

For each condition of cycling, the 10-second trial was truncated into the first five consecutive crank cycles that consisted of a single full crank cycle for both legs. The kinematic and kinetic data were exported to Visual 3D (Version v2025.01.2, HAS-Motion Inc., Kingston, Ontario, Canada) for further processing and computations. Kinematic and kinetic data were filtered using a 4th order zero-lag Butterworth filter with a cutoff frequency of 6 Hz {Gardner, 2016 #106}. 3D kinematics were resolved with a Cardan sequence of X-Y-Z in the joint coordinate system {Grood, 1983 #576}. Inverse dynamics was used to calculate joints moments and expressed at the proximal segment of the joint, for example, the knee moments were expressed at the thigh. In order to determine segment masses, the default setting in Visual 3D was used that implemented regression equations developed by Dempster {Dempster, 1955 #604}. Kinematic and kinetic variables were reported in accordance with the right-hand rule. Discrete events of interest were manually assessed and selected using a custom Python script {Menke, 2024 #585} and each variable's time series curves were visualized for the cycling trials to ensure accuracy.

Statistical Analysis

All statistical analysis was done using the IBM Statistical Package for Social Sciences (SPSS, v29.0.0.0) with an α level set *a priori* to 0.05 for all tests as the significance threshold. To assess each selected variable, two-way 2 x 5 (Limb x Condition) repeated measures ANOVAs were used. The Greenhouse-Geisser correction was applied whenever assumptions of sphericity

were violated. Partial eta squared (η_p^2) was reported to assess effect size with the following classifications: small ($\eta_p^2 = 0.01$), medium ($\eta_p^2 = 0.06$), and large ($\eta_p^2 = 0.14$) {Lakens, 2013 #552}. When a significant main effect or interaction is discovered, post-hoc comparisons were made without adjustments of alpha level due to the small sample size present. Finally, two one-way ANOVAs were performed on the pain and RPE data across conditions to evaluate potential significant differences between crank length conditions. Dependent t-tests were performed on pre-testing pain levels and post-test pain levels for each cycling condition.

Results

ANOVAs were performed with selected variables from the knees and pedals (**Table 2**). A significant main effect of crank arm length was found for knee extension ROM ($p < 0.000$, $\eta_p^2 = 0.103$), peak knee extension angle ($p < 0.000$, $\eta_p^2 = 0.740$), and knee power ($p = 0.032$, $\eta_p^2 = 0.345$). Post-hoc comparisons showed that knee ROM was significantly decreased, relative to C3, between all compared combinations of the conditions. Specifically, knee ROM decreased as crank arm length decreased in both symmetrical and asymmetrical crank arm conditions (all $p \leq 0.001$, **Table 2**). An ensemble curve for the knee joint angles across the crank arm conditions is provided (**Figure 2**). Specifically, the knee adopted a more flexed peak position in response to shorter crank arm lengths in both asymmetrical and symmetrical conditions. Peak knee extension angle results showed a consistently more flexed knee on the replaced side with increasing amounts of peak knee flexion as crank arm length shortened (all $p < 0.020$). There was an interaction ($p = 0.017$) between limb and condition for knee extension angle which indicates that a more complex relationship between changing crank lengths and the limb of interest is present. KEM showed a significant limb main effect ($p = 0.025$, $\eta_p^2 = 0.595$) with post hoc comparisons showing significant differences between limbs at asymmetric 155mm ($p =$

0.045) crank arm length as well as symmetric 155mm ($p = 0.025$) and 140mm ($p = 0.048$) crank arm lengths.

Peak knee power displayed a significant condition main effect ($p = 0.032$, $\eta_p^2 = 0.345$) and a significant limb main effect ($p = 0.031$, $\eta_p^2 = 0.569$). Post hoc comparisons revealed that peak knee power showed a significant main limb effect ($p = 0.025$). Interaction effects between condition and limb were significant for vertical pedal reaction force (PRFz, $p = 0.043$, $\eta_p^2 = 0.326$) but did not result in any significant post-hoc comparisons.

Additionally, ANOVAs were conducted on specific ankle and hip variables (**Table 3**). Condition main effects for ankle ROM ($p = 0.003$, $\eta_p^2 = 0.468$) and hip ROM ($p < 0.000$, $\eta_p^2 = 0.930$) were significant. Post-hoc comparisons revealed that hip ROM was significantly different amongst all compared combinations of conditions (all $p < 0.005$, **Table 3**), similar to knee ROM. An ensemble curve for sagittal hip joint angle (**Figure 3**) and sagittal ankle joint angle (**Figure 4**) across the crank arm conditions is provided. Ankle ROM was only significantly different between the pairs: C1/C5, C2/C3, C3/C5. Hip ROM was consistently lower for the replaced side in all conditions while ankle ROM was consistently greater for the replaced side except for C3. No condition and limb interactions were significant for the ankle or hip variables.

For pain and RPE measurements, no significant differences were observed between conditions (**Table 4**). Dependent t-tests were performed comparing pre-testing pain levels to post-testing pain levels after each condition. No significant differences were found between pre- and any post-testing pains.

Discussion

The primary aim of this work was to quantify the effects of asymmetrical and symmetrical shortening of ergometer crank arms on knee biomechanical variables during

stationary cycling. The main hypothesis of decreased knee extension ROM, peak KEM, and peak knee extension power during asymmetric crank conditions compared to symmetric crank conditions was partially supported by the results of this study. Knee ROM did significantly decrease for asymmetrical and symmetrical crank arm conditions (155mm and 140mm, while knee extension power significantly decreased for the symmetrical shortened conditions only. However, peak KEM did not change significantly across asymmetrically and symmetrically shortened crank conditions.

Specifically, knee ROM of the replaced limbs decreased 6.2° over a total crank length shortening of 30mm in asymmetrical conditions and 9.7° for symmetrical conditions, indicating a greater knee ROM reduction from symmetrical shortening. In comparison, these values are quite close to the approximate 5° ROM reduction found in asymmetric crank arm usage by Childers, et al. {, 2014 #225}, however they were using 162mm crank arms for the affected limb and their patients had a transtibial amputation and the results may not be directly comparable to the TKA population. Minimal literature exists on this topic and comparisons to literature incorporating asymmetrical manipulations of the crank arms are difficult. A possible explanation for discrepancies between these studies lies in the bike fit procedure. While this study implemented a standardized procedure, Childers et al. {Childers, 2014 #225} used bike fit measurements from the participant's own bikes. These results suggest that both asymmetrical and symmetrical shorting of crankarms can achieve the purpose of knee ROM reduction to accommodate the reduced knee ROM for the patients with TKA. However, considering symmetrical shortening results in a greater knee ROM reduction, even shorter asymmetrical crank arm length may be needed to produce the same amount of knee ROM reduction. This finding is unique considering the replaced limb crank arm lengths are the same, at C1 and C5,

and suggests that there may be some kinematic compensation from the contralateral limb or from neighboring joints of the replaced limb. Specifically, the asymmetrical conditions do allow the contralateral limb to utilize the 170mm crank arm length to reduce the ROM required of the replaced knee.

Peak KEM surprisingly did not exhibit any significant condition differences or interaction, although it did produce a limb main effect, showing peak KEM being consistently greater in the contralateral limb compared to the replaced limb. A significantly lower KEM for the replaced side was seen asymmetrically at 155mm and symmetrically at 155mm and 140mm. Additionally, peak KEM values for the replaced and contralateral limbs at symmetrical 170mm crank lengths were approximately 17 Nm and 25 Nm, respectively, which are in line with peak KEMs reported by Hummer et al. {, 2021 #200} at 80 W and 80 RPM. However, it should be noted that Hummer et al. {Hummer, 2021 #200} recruited participants 6-18 months post-surgery compared to the present study's criteria of 1-2 months post-surgery. This finding demonstrates a diminished ability of the replaced limb to produce torque even at symmetrical crank lengths, which is expected so close to surgery. It is important to note that the contralateral side for this study contains a mixed group of unilateral and bilateral participants and limb differences should therefore be interpreted with caution. However, Yocum et al. {, 2024 #595} reported only a significant lower peak KEM between the most recently replaced limb of bilateral TKA participants with unilateral participants during level walking. Considering that cycling is an activity with lower weight bearing requirements than gait, it should be of interest to further investigate the moment differences that may be present between these patients with unilateral and bilateral replacements during different specific phases of the crank cycle.

Our KEM results provide further support for the findings by Hummer et al. {Hummer, 2021 #200} that even after extended time post-op, these participants are not completely recovering their most recently replaced limb's muscular capacity. Furthermore, the lack of KEM increases with decreased crankarms may be explained by the associated muscle length changes. As knee ROM changes, so does the length of the involved knee extensors during the pedal cycle and thus the muscular capacity is expected to change. Given reduced ROM of the knee during the cycle in shortened crank length conditions, the quadriceps can be assumed to be changing length proportionally and their force output considered altered. While it was expected that KEM would increase in response to the decreased lever arm, these data suggest that participants are compensating more kinematically at the knee instead. Shortened muscle lengths due to crank arm shortening is likely decreasing optimal muscle force production which aligns with a lack of KEM increase seen in the present study. Barratt et al. {, 2016 #508} investigated the ratio of leg length to crank arm length and determined via a regression equation that there was not a linear proportional change in power output relative to the leg length to crank length ratio. When altering crank lengths for patient populations, muscle length changes due to exercise interventions such as crank arm length adjustments should be considered as there may be a point that the crank length is too short. While the lever arm of the knee extensor musculature is shortened in response to crank arm shortening, the commensurate reduction in knee ROM seems to mitigate that effect as shown by a lack of widespread changes in KEM.

Peak knee extension power of the replaced limb decreased significantly from symmetrical 170mm crank lengths to symmetrical 155mm crank lengths. Unlike the peak KEM results, these reductions are most likely a results of increased knee angular velocities at these conditions. Peak KEM was found to be significantly lower in the replaced limb compared to the

contralateral limb regardless of condition. Qualitatively, the contralateral limb was approximately 5 - 7 Nm greater than the replaced limb across conditions. This trend of increased values for the contralateral limb is unsurprising given the proximity of these participants to their surgery. However, considering the replaced limb values align with Hummer et al. {Hummer, 2021 #200} who used patients much further from surgery, the evidence from the present study suggests a lack of adequate KEM rehabilitation over time. . The differences between limbs are consistent across symmetrical and asymmetrical shortening as well, which indicates that minimal, if any, compensations are occurring in peak knee extension power for this population.

The secondary hypotheses regarding an expected decrease in pain and RPE of the replaced limb for the asymmetrical crank arm conditions were not supported. Average pain of the replaced limb during all cycling conditions was approximately 1.5 out of 10 with pre-testing pain being an average of 2.8. While the crank arm shortening for the replaced side did not decrease pain or RPE, they also did not increase the pain, which indicates the asymmetrical change would not be detrimental from this aspect. The pain levels during cycling were quite low with averages of between 1 and 1.6 on a 0-10 scale across all crank conditions, which indicates a minor reduction in overall pain due to cycling as a modality and provide evidence for the safety of cycling for TKA patients. Considering asymmetric conditions did not increase pain, it may be considered a safe alternative to symmetrical crank arm cycling subject to further work with a greater sample size and alternative populations with lower extremity pathologies of imbalances. Indeed, a meta-analysis of stationary cycling studies concluded that cycling did improve pain scores compared to those who were not exercising {Luan, 2021 #596}. The RPE values found for this submaximal cycling are considered “light” on the Borg scale and approximately equivalent to the values seen by Millour et al. {, 2020 #226} during submaximal cycling with

crank arm lengths of 160-170mm. Low pain and RPE levels may also indicate that the resistance of 70 W and cadence of 70 RPM used in this study were not sufficient enough to alter RPE, but with a patient population early in recovery it is not advisable to considerably increase exercise difficulty.

Ankle and hip variables were included in this study to evaluate if compensatory behaviors occur proximally or distally to the knee as a result of the crank length changes. Hip extension ROM was the most broadly affected variable in this regard because it was significantly different between all condition combinations. Hip ROM decreased for the replaced and contralateral sides when moving from 170mm crank lengths to asymmetrical 140mm cranks, with a replaced side average reduction of 3.5° . In comparison, replaced knee ROM had an average reduction of 6.1° for these same conditions. For the ankle, replaced side ankle ROM was reduced only by an average 0.6° from 170mm and asymmetrical 140mm crank length for the replaced limb. Millour et al. {, 2020 #226} did not report significant change of average ankle ROM from 170mm to 160mm crank lengths. When assessing the ankle, it does not appear to provide more kinematic or muscular assistance as shown by the lack of differences across conditions. Additionally, the hip extension moment showed no significant changes across conditions or limbs, but given the broad changes seen in hip extension ROM it is likely that the hip is more responsive to crank arm length changes than the ankle in the present study. Similar to knee musculature length changes due to reduced ROM, it is likely it occurs also at the hip and causing unaltered joint moments. Future work may evaluate if these patterns are seen in participants that are trained to cycle asymmetrically over more sessions to evaluate potential chronic mechanical or neuromuscular adaptations to the movement and additionally longitudinal interventions for TKA rehabilitation.

When interpreting the results from this work, several limitations should be considered. Although a power analysis suggests a minimum sample size of 4 is sufficient, the sample size of 7 in this study is quite small and should be bolstered to increase statistical confidence in the conclusions. A second potential limitation comes from the bike fit standardization procedure that has been used in other published works. With this study's population being so close to their surgery date, acquiring the specified knee ROM for the bike fit was challenging due to stiffness, pain, and discomfort. While this issue would likely be avoided with a population that is further along in their recovery, there would be a loss in the value of investigating potential rehabilitation effects due to the extended amount of time that has passed. The chosen wattage of 70 W and cadence of 70 RPM may be too easy for some participants. Varying recreational activity levels will likely be different amongst participants with TKA and a form assessing physical activity participation or objective dynamometer assessment may be warranted for future work. It may be warranted to personalize cycling workrate between patients to maximize the efficacy of cycling as a rehabilitation modality. Finally, the use of contralateral in the scope of this study is difficult because some participants had contralateral TKAs, some had diagnosed OA of the contralateral side, and some did not have OA on the contralateral side. When making an asymmetrical change to the ergometer for future clinical research, it may be necessary to incorporate more control over the status of the contralateral limb within the subject pool.

Conclusion

TKA patients adjusted their replaced knee, hip, and ankle ROM as a result of asymmetrical and symmetrical crank arm shortening but did not increase KEM or pain during stationary ergometer cycling. These findings suggest that knee loading is not adversely affected by asymmetrical or symmetrical crank arm length shortening for the replaced limb. It was seen

that symmetrical shortening reduced knee ROM more than asymmetrical shortening to the same length, and a greater shortening may be needed asymmetrically to achieve the same ROM changes. Additionally, the results suggest that asymmetrical and symmetrical shortening of the crank arm for individuals 1-2 months post-TKA is not likely to affect the implant negatively and could be implemented to accommodate knee stiffness deficits seen soon after surgery. Further, a longitudinal study evaluating the efficacy of asymmetrical versus symmetrical shortening for beneficial rehabilitation outcomes is necessary. At this point, the research evidence seems to suggest that clinical practitioners may decide to use asymmetric or symmetrical crank arm reductions to accommodate for the reduced knee ROM and facilitate neuromuscular function recovery after TKA surgery and these may be accomplished with commercially available crankarm adapters or cycle ergometers with adjustable crankarms.

Chapter 5: Preliminary Investigation of Asymmetric Cycling Biomechanics in Early Unilateral and Bilateral Patients of TKA

Abstract

Ergometer cycling is a commonly prescribed exercise for total knee arthroplasty patients to reduce knee stiffness, often using symmetrical crank arms. It is unclear whether asymmetrical adjustments to the replaced side similarly affect unilateral and bilateral patients early in recovery.

Purpose: To quantify the biomechanical differences between the limbs of unilateral and bilateral total knee arthroplasty recipients and determine if this intervention affects the groups differently.

Methods: Three bilateral and four unilateral recipients of knee replacements were recruited 1-2 months after their surgery. All cycled in symmetrical and asymmetrical crank lengths of 170mm, 155mm, and 140mm with the asymmetrically shortened cranks on their recently replaced limb, at 70 W and 70 RPM. **Results:** A significant condition main effect was found for the peak knee extension angle. Cycling asymmetrically at 155mm crank arm lengths was significantly different from symmetrically at 170mm and 155mm crank arm lengths. No significant results were found for peak knee extension moment, knee range of motion, peak ankle plantarflexion moment, or peak hip extension moment. **Conclusion:** The use of asymmetric or symmetric shortening of the crank arms does not appear to change the difference of knee kinetic and other lower limb biomechanical variables between the limbs of unilateral or bilateral participants during cycling. This study provides initial evidence for the efficacy of this intervention in both patient groups.

Keywords: crankarm length, arthroplasty, unilateral TKA, bilateral TKA, cycling, rehabilitation

Introduction

Total knee arthroplasty (TKA) is a surgery for individuals with end-stage knee osteoarthritis (OA). Approximately 36% of unilateral TKA patients will undergo a contralateral replacement within 10 years of initial surgery and approximately 70% of Kellgren-Lawrence grade 4 radiographic signs are present {Sayeed, 2011 #592;Kellgren, 1957 #136}.

Biomechanical research comparing unilateral and bilateral TKA patients is scarce, but investigations into stair climbing have revealed that kinetic differences between the two groups are present {Yocum, 2022 #222}. Yocum et al. {, 2022 #222} reported that bilateral patients exhibited lower knee extension moments (KEM) for their initially replaced limb compared to the contralateral limb of unilateral participants. Furthermore, Yocum et al. {, 2024 #595} have also presented level walking data from the same participant group and determined that unilateral and bilateral TKA recipients display altered kinetics. Loading response KEM for the most recently replaced limb of bilateral participants was significantly lower than both limbs of unilateral participants. During push off, no significant differences were seen between limbs or groups which indicate differences only occur during the power generation phases of gait. Specifically, it was concluded that the contralateral limb of unilateral TKA participants was not a valid control limb and therefore unilateral and bilateral TKA participants should be evaluated separately instead of pooled together. That research suggests that patients with unilateral and bilateral TKAs may be adopting different movement strategies, however this conclusion has not been investigated in terms of cycling biomechanics.

A consistent issue in recovery from any TKA is the difficulty in obtaining a functional knee range of motion (ROM), and it has been shown that knee flexion capacity is a significant risk factor for undergoing a contralateral TKA {Okamoto, 2023 #593}. Reduced knee ROM is

likely after surgery due to surgical side effects and effusion causing stiffness, and some patients only recover their pre-operative knee ROM one year after surgery {Köglberger, 2022 #589}. One of the most common exercise modalities used to reduce knee stiffness and promote knee ROM recovery after surgery is ergometer cycling because it lowers weight bearing requirements of the knee compared to gait while maintaining and promoting muscular strength recovery and cardiovascular benefits {Kutzner, 2012 #4; Logan, 2023 #216; Oja, 2011 #217}. Stationary cycling starting 2 weeks after surgery has been able to improve knee ROM and patient-reported outcome scores while also decreasing pain in the replaced knee {Zech, 2015 #220}. In addition to the ROM deficits, unilateral TKA patients also exhibit decreased KEMs during cycling, and ramp navigation {Hummer, 2021 #200; Wen, 2019 #567}. These kinematic and kinetic differences suggest that knee recovery after surgery may not be optimized and more tailored exercise modalities should be investigated. Exercise interventions that can gradually accommodate for deficits and allow for strengthening of the knee and surrounding tissue are therefore of interest.

Adjustments of the cycle ergometer can be made to accommodate different clinical populations in their rehabilitation, such as adjustment of the seat height, pedal width, or crank arm length. Symmetrical crank arm length shortening has shown a decreased knee ROM requirement as well as lowered knee joint velocities in healthy participants {Barratt, 2011 #286}. These findings show promise for the TKA population that has difficulty in obtaining sufficient knee ROM due to stiffness and pain soon after surgery, but symmetrical shortening may not be ideal for the patients with unilateral TKA. Asymmetric crank arm shortening has only been investigated in transtibial amputee {Childers, 2014 #225} and limb-length difference {Millour, 2020 #226} populations to reduce knee ROM and improve comfort during cycling. In the TKA

population, the potential benefit of cycling with asymmetric crank arm shortening is to accommodate reduced knee ROM and pain of the replaced limb, while secondarily assisting in the development of the replaced knee's strength soon after surgery. To our knowledge, there has been minimal research regarding asymmetric crank arms, with no research involving the TKA population, unilateral or bilateral {Zhang, 2025 #591}. Currently, it is unknown if rehabilitation interventions should be the same for patients with unilateral or bilateral TKAs or if a more group-based approach is warranted.

The purpose of this study was to compare the differences in biomechanical variables of the knee and other lower limb joints between the limbs of participants with a unilateral or bilateral TKA in response to asymmetrical crank arm adjustment in stationary cycling. Previous literature has suggested that bilateral participants may be utilizing the hip and ankle more as compensatory assistance for the knee compared to unilateral participants, as reported by Yocum et al. {Yocum, 2022 #222} in stair descent tasks. It is hypothesized that the difference of peak KEM between limbs will be significantly lower and peak hip extension or peak ankle plantarflexion moment differences will be significantly greater in bilateral TKA participants compared to unilateral TKA participants for all cycling conditions.

Methods

Participants

Seven participants, 3 bilateral (3 male, 69±4 years, 1.9±0.05 m, 116±14.4 kg, 33.2±4.5 BMI, 52±3 days post-TKA) and 4 unilateral (4 female, 65±8 years, 1.6±0.05 m, 78.4±12.9 kg, 30.2±4.9 BMI, 56±3 days post-TKA) were recruited from the physical therapy clinic of Tennessee Orthopaedic Alliance East to participate in this study (**Table 1**, all tables available in the appendix). All participants had TKA operations performed by the same surgeon at the

Tennessee Orthopaedic Alliance East. The “replaced knee” was considered the most recently replaced knee for bilaterally replaced participants. Participants were required to be between the ages of 50 and 75 at the time of data collection and between 1- and 2-months post-TKA operation. Participants were required to have at least 100° of passive knee flexion ROM in the most recently replaced knee in order to complete the cycling tasks. Participants were excluded from participating in the study if they were diagnosed with OA of the ankles, hips, or spine as well as those who had a contralateral TKA within 4 months of their most recent surgery date. BMI over 40, neurological conditions preventing them from cycling or walking without external aid, partial or total lower limb amputation, and/or cardiovascular disease were also screened for as exclusionary criteria.

A power analysis (G*Power 3.1; Heinrich Heine University Dusseldorf, Dusseldorf, Germany) based on a repeated measures ANOVA of sagittal knee ROM during submaximal asymmetric crank arm ergometer cycling in a young, healthy population found that, with an effect size of 0.979, a minimum of four participants were needed with an α level of 0.05 to achieve a power of 0.80 {Zhang, 2025 #591}. All procedures, recruitment paperwork, and study material were approved by the university’s Institutional Review Board.

Instrumentation

Cycling tasks were performed on an electromagnetically braked cycle ergometer (Lode Excalibur Sport, Groningen, Netherlands). Workrate for the ergometer was controlled from an externally connected laptop using the Lode Ergometry Manager software and displayed on the bike’s screen. Manufacturer-built adjustable crank arms (Sport crank, Lode) were attached to the ergometer to allow for asymmetric adjustment of crank arm lengths in the smallest increment of 5mm. A pair of custom-built instrumented bike pedals was used with each pedal consisting of

two 3-dimensional force sensors (Type 9027C, Kistler, Switzerland), and amplifiers (Type 5073A, Kistler, Switzerland). These sensors (1200 Hz) were used to collect pedal reaction force (PRF) data during testing.

A custom metal jig was constructed to align the bike to an in-ground force platform to align the cycle ergometer with the lab coordinate system. A standardized bike fit procedure was conducted for each participant, which included an adjustment of the saddle's height to produce 25-30° of knee flexion while the replaced knee's limb was at bottom dead center of the pedal cycle. For some participants, this position resulted in too much pain, or their knee stiffness made the position uncomfortable and a saddle height resulting in approximately 40° flexion was used. The fore-aft position of the handlebars was set by measuring the angle between the thigh and trunk to achieve approximately 90°. The fore-aft position of the saddle determined with the replaced knee limb resting at approximately 90° of the pedal cycle, or the most forward pedal position, and a plumb bob was used to align the patella with the pedal spindle.

A 12-camera motion analysis system (Vantage 5 and Valkraie 8, 240 Hz; Vicon Motion Analysis Inc., Oxford, UK) was used to record optical data for kinematics. During collection, each participant was provided a pair of laboratory standard shoes (Zoom Pegasus 34, Nike). Participants were outfitted with retroreflective anatomical markers and segment tracking marker clusters mounted on semi-rigid thermoplastic shells. Anatomical markers for the static trial were placed bilaterally on the following bony landmarks: acromion process, iliac crest, anterior superior iliac spine, posterior superior iliac spine, greater trochanter, medial and lateral femoral epicondyle, medial and lateral malleoli, head of 1st and 5th metatarsals, and the distal phalanx on the second toe. Marker clusters to track segmental motion consisted of 4 markers each that were non-coplanar affixed with velcro to neoprene straps to the trunk and bilaterally to the thighs,

shanks, and heels. The pelvis was tracked with two separate 2-marker clusters placed on the posterior pelvis. PRF and kinematic data were integrated and simultaneously recorded using the Vicon Nexus (Version 16.0; Vicon Motion Analysis Inc., Oxford, UK) software on one computer.

Experimental Protocol

Participants in the study attended one laboratory data collection. Upon arrival, patient demographics were collected along with a patient health information release and physical activity readiness questionnaire (PARQ+)

Participants were given at least a 3-minute warmup at a 0 W and a self-selected pace. All conditions were conducted with a wattage setting of 70 W and a desired cadence of 70 revolutions per minute (RPM) for a total of 2 minutes per condition. Each participant completed first with symmetrical 170mm crank arms (C3) as a reference before continuing through the other four conditions in a random order. The additional cycling conditions, presented as crank lengths for the Replaced/Contralateral limbs (in mm): C1 (140/170), C2 (155/170), C4 (155/155), C5 (140/140). Participants were requested to maintain the 70 RPM cadence to the best of their ability during the 2-minute period and data were recorded during the final 10 seconds of each condition. Between conditions, participants were able to rest at least 2 minutes, or as long as necessary, while the crank arm lengths were changed. Following each cycling condition, participants were asked to rate their replaced knee pain from 0-10 on a numerical visual analog scale as well as their general rating of perceived exertion (RPE) from 6-20 using the Borg Scale.

Data Analysis

The first five consecutive crank cycles for the replaced and contralateral were extracted as from each 10-second collection for each test condition. Visual 3D (Version v2025.01.2, HAS-

Motion Inc., Kingston, Ontario, Canada) was used to process 3D kinematics and kinetics as well as EMG data and to compute derived variables. A 4th order zero-lag Butterworth filter with a cutoff frequency of 6 Hz {Gardner, 2016 #106} was used to filter kinematic and kinetic data. A X-Y-Z Cardan sequence was used to resolve kinematics in the joint coordinate system {Grood, 1983 #576}. To calculate joint moments, inverse dynamics were implemented and expressed relative to the proximal segment of a given joint, such as expressing knee moments at the thigh. Default Visual 3D settings were used to calculate segment masses from the regression equations provided by Dempster {Dempster, 1955 #604}. A custom Python script was used to visually identify discrete data points for variables of interest and averaged together to obtain final values for statistical inputs {Menke, 2024 #585}. Discrete variables were obtained for the recently replaced limb and the contralateral limb of each participant. A simple difference was calculated by subtracting the replaced limb's discrete value from the contralateral limb's value. A result close to zero indicates that the peak response from both limbs was similar. This work can be considered exploratory due to the small sample size. The simple difference was chosen over a symmetry index to first evaluate variables individually instead of an index value that would be compared across variables.

Statistical Analysis

Statistical testing was performed using the IBM Statistical Package for Social Sciences (SPSS, v29.0.0.0) with the significance threshold set *a priori* to 0.05. For all variables of interest, a difference was taken by subtracting the value of the most recently replaced side from the contralateral side (native knee for unilateral, replaced knee for bilateral) for each condition. A value closer to zero indicates a more symmetrical behavior between the limbs while a positive number indicates that the contralateral limb had a greater value. These differences were used as

the input for two-way (2x5: Group x Condition) repeated measures ANOVAs . When assumptions of sphericity were violated, the Greenhouse-Geisser correction was applied. Partial eta squared (η_p^2) was reported to assess effect size with the following groups: small ($\eta_p^2 = 0.01$), medium ($\eta_p^2 = 0.06$), and large ($\eta_p^2 = 0.14$) {Lakens, 2013 #552}.

Results

ANOVAs were conducted on a variety of pedal and knee variables (**Table 5**) between the unilateral and bilateral groups with the input values being the result of the contralateral limb minus the replaced limb as described in the Methods. A significant main effect of condition was found for peak knee extension angle (**Table 5, Figure 5**, $p=0.023$, all figures available in the appendix). The post-hoc comparisons showed significant differences between the asymmetrical 155mm and 140mm conditions ($p = 0.043$) where the difference value decreased, indicating less difference between limbs with symmetrical 170mm crank arms. Additionally, a significant difference between asymmetrical and symmetrical 155mm crank arms ($p = 0.001$) was found with a similar effect of a decreased difference between limbs.

No significant main effects or interactions were seen for PRF, knee extension ROM, peak KEM (**Figure 6**), or peak knee extension power. For the hip and ankle variables, no significant main effects of condition and group, and their interactions were observed for these variables (**Table 6**). No significant main effects for pain or RPE were found (**Table 4**).

Discussion

The primary purpose of this study was to evaluate potential differences in response to asymmetric crank arm usage in ergometer cycling between participants with unilateral or bilateral TKAs. The main hypothesis regarding a significantly lower difference in peak KEM and significantly larger differences in peak hip extension and ankle plantarflexion moments in

bilateral participants at any condition was not supported. Peak KEM differences between limbs for unilateral participants at asymmetrical 140mm crank arms was approximately 3.4 Nm while it was approximately 9.4 Nm for bilateral participants, however, these changes were not statistically significant. Interestingly, although peak KEM was not significantly different between groups, the difference values were quite stable within the unilateral group, approximately 3.5 Nm, across all conditions while the average values of the bilateral group fluctuated. Peak hip extension moments were similarly unaffected, which is likely due to the considerable variability of responses within groups. Peak vertical PRF differences did not display any significant crank length effects providing additional support for the KEM results. With a partial eta squared value of 0.350, the effect size indicates that there may be a relationship given greater statistical power and may be noted for analysis in larger studies.

While the variables included in the hypothesis were not significant, it was revealed that knee extension angle had a main condition effect with post-hoc comparisons revealing significant differences between asymmetrical 155mm and symmetrical 170mm as well as asymmetrical 155mm and symmetrical 155mm crank arms. Asymmetrical 155mm crank arms produced an approximate 4.5° and 5.3° difference for unilateral and bilateral groups, respectively, while symmetrical 155mm crank arms had a -0.12° and 3.6° difference. This is to be expected because the variable is based on the difference between limbs, which is expected to be higher for asymmetrical conditions of a given crank length. While this was a statistically significant change, it is difficult to consider this difference functionally meaningful with such a small change in magnitude {, 2014 #225}.

There has been minimal published work with asymmetrical crank arms, and especially minimal work primarily comparing the biomechanical profiles of unilateral and bilateral TKA

recipients, regardless of activity. Yocum et al. {Yocum, 2024 #595} presented a comparison of unilateral and bilateral TKA participants during level gait {, 2024 #595}. Bilateral participants were reported to have decreased knee extension moments, knee abduction moments, dorsiflexion moments, and vertical ground reaction force peaks compared to unilateral participants. Cycling appears to be an activity that is less sensitive to the differences in limbs between these two groups, as none of the previously mentioned variables from Yocum et al. {Yocum, 2024 #595} were significantly different during symmetrical cycling at standard crank lengths of 170mm. However, considering the high standard deviations seen in the present study, it may also be true that group members may have unique individual responses that would require a more tailored approach to rehabilitation programs. Individual responses to acute exercise interventions in cycling have been documented by Thorsen et al. {, 2021 #290} who investigated estimated compressive medial contact force changes as a result of pedal width adjustment in cycling. Specifically, it was reported that there were responder and non-responder groups regarding changing medial contact forces to pedal width adjustments during cycling. The non-responder group displayed a decreased peak medial contact force in the power phase of the cycle which indicates that participants may react to acute exercise interventions differently. It is critical to note that the population of that study were young, recreationally active adults and conclusions may not apply to the patient population.

Individual responses may not solely be tied to bilateral or unilateral groups and could be a result of other individual factors independent of group. Factors such as pain or RPE in this study were not found to be significantly different between conditions or groups, which rules them out for contributors to individual or group differences. Pain level averages for all conditions, regardless of group were between 1 and 2 out of 10, while RPE averages were

between 11 and 12. These low pain levels and light RPE scores indicate that perhaps the difficulty of the exercise was too low, but perceived difficulty thresholds and higher workrates for exercise may need to be evaluated for patient rehabilitation purposes in future studies. Improvement of results could be accomplished with a bolstered participant pool and therefore larger group sizes as well as determining if similar responder or non-responder groups are present in response to crank arm length shortening. Despite the lack of significant change between groups, it is important to note that large qualitative asymmetries do appear to exist in both groups even though those changes were not significantly altered by crank arm length adjustments. For example, KEM differences between the groups ranged from approximately 4–12 Nm when assessing all five conditions with the bilateral group having consistently higher values. Further, both groups had peak knee power differences that were in the range of 20-50 Nm/s with the positive value indicating more knee power was produced on the contralateral side. A larger sample size may assist in clarifying whether the bilateral participants do indeed demonstrate greater asymmetries in KEM and knee power. These findings are expected due to post-surgery neuromuscular damage and invite future work to consider how training can be conducted to maximize recovery and minimize the asymmetries.

Given the lack of significant differences between unilateral and bilateral groups of participants, we have reported evidence that there is no difference in the effect of asymmetric and symmetric crank arm shortening between groups. While no significant group main effect or interaction was found, the knee extension moment (**Table 5, Figure 6**) may produce a group effect given a greater study power on account of the potentially large effect size ($\eta_p^2 = 0.440$). Similarly, vertical PRF may produce a condition effect with more participants when assessing the effect size ($\eta_p^2 = 0.350$). This is a valuable finding from this work because it is not guaranteed

that these groups will respond to an asymmetrical or symmetrical change in a similar manner. To further cement this claim, future work should attempt to ensure a larger participant pool and consider following these two groups longitudinally to evaluate if their inter-limb differences change as they move away from their surgery date. It should also be noted that symmetrical behavior between limbs for either group may not be an inherently beneficial goal for functional recovery. Aljehani et al. {, 2019 #594} discuss the validity of using the contralateral limb as a control for unilateral TKA patients and reported that although there were kinematic differences between limbs, no kinetic differences were observed between limbs during self-selected speed gait.

When evaluating the current study's results, specific attention should be given to the limitations that are present. This study primarily aims to compare between participants with unilateral and bilateral TKAs, however in this study participants within each group were exclusively of one sex. All unilateral participants were female, while all bilateral participants were male. It is impossible to conjecture if the asymmetric crank arm effects would cause similar changes between male and female participants and should be considered for further research. Another significant limitation within this work is the sample size, being only 7 participants. Combined with a sex distribution split between TKA groups, this makes granular conclusions about the causation of results, or lack thereof, quite difficult. Due to the small sample size, a less complex ANOVA was used by comparing the differences between the limbs within each group. For a more robust sample size and results, future work should consider comparing across groups, limbs, and conditions. Finally, cycling with asymmetric crank arms is an unusual and strange activity for most participants, and it may be beneficial to give participants more time to acclimate to the movement before conducting data collections.

Conclusion

Unilateral and bilateral patients displayed minor peak knee extension angle differences between limbs when comparing symmetrical and asymmetrical crank arm shortening during ergometer cycling. No significant differences between groups were observed for any other measured kinematic or kinetic variable, which suggests that asymmetrical shortening of the crank arm for the most recently replaced knee will not affect one group or the other differently. This conclusion is in line with previous research in TKA participants and biomechanical symmetry during gait as well as research involving asymmetric crank arm usage in transtibial amputees. Practitioners may not need to adjust their recommendations for unilateral or bilateral TKA surgery recovery activities, however differences between limbs may vary greatly on a per-patient basis and should be considered regardless of TKA surgeries.

Chapter 6: Knee Contact Forces in TKA Patients During Ergometer Cycling with Asymmetrically Adjusted Crank Arm Lengths

Abstract

Musculoskeletal modeling allows for the estimation of internal joint and muscle forces that are difficult to acquire in-vivo. Application of modeling software to patient populations, such as those with total knee arthroplasty, may help to illuminate factors affecting rehabilitation and improve clinical recommendations for movement. **Purpose:** To quantify the internal contact forces of the replaced knee and peak muscle forces during asymmetrical crank arm ergometer cycling in individuals 1-2 months after receiving a total knee implant. **Methods:** Seven participants who received a cruciate retaining implant cycled at 70 W and 70 RPM for cycling conditions of symmetrical 170mm crank arms as well as shortened 155mm and 140mm crank arms for the replaced limb. Static optimization and joint reaction analysis were conducted using OpenSim and outputs were compared to experimental data for model validation. **Results:** Decreases in crank length led to significantly reduced total vertical contact force and vertical lateral compartment contact force within the replaced limb. Peak muscle group activations surrounding the knee and vertical pedal reaction forces were not significantly different between the cycling conditions. **Conclusion:** Findings from this work suggest that asymmetric crank arm usage can be implemented early in the rehabilitation process of total knee arthroplasty patients to improve knee stiffness while not adversely affecting estimated internal knee loading and altering knee muscle forces. Future work may endeavor to understand where decreased forces at the replaced knee are being displaced.

Keywords: arthroplasty, knee, opensim, contact force, cycling, rehabilitation

Introduction

Total knee arthroplasty (TKA) is an elective treatment commonly aimed at alleviating the symptoms of end-stage knee osteoarthritis (OA). One of the primary difficulties for TKA patients and their recovery is obtaining satisfactory knee range of motion (ROM), which has been shown to only be achieved at least one year after surgery {Köglberger, 2022 #589}. Increasing physical activity to help reduce knee joint stiffness and promote ROM recovery post-operation while maintaining or lowering pain at the knee joint during exercise is a beneficial avenue of study. While patients with at least one TKA tend to reduce the replaced knee joint load soon after surgery on their own, some patients report difficulty “pushing through exercises” due to post-surgical issues {Bakaa, 2022 #219}. This finding suggests that even further minimizing knee loads may help hesitant patients with their rehabilitation. Cycling is a commonly prescribed exercise modality for exercise in this population and has been shown to improve knee range of motion and reduce pain {Zech, 2015 #220}. During cycling, TKA patients have decreased tibiofemoral forces and exhibited decreased knee extension moments (KEM) of the replaced knee at least six months after surgery and it is possible that patients earlier in their rehabilitation would have difficulty accomplishing complete pedal cycles {Hummer, 2021 #200;Kutzner, 2012 #4}.

There are many bike modifications that elicit biomechanical changes, such as changing the width between the pedals {Thorsen, 2020 #289} or adjusting the length of the crank arms {Martin, 2001 #30}; but not all avenues have been exhausted. Adjustments to the length of both crank arms have been shown to decrease knee joint ROM and lower limb joint velocities {Barratt, 2011 #286}. Known ROM and kinetic deficits in the replaced knee of TKA patients leads to the question of what ergometer adjustments can be made unilaterally to accommodate

the joint movement limitations in the TKA demographic. Asymmetrical shortening of the crank arm length on the replaced knee side of TKA patients is a potential strategy to reduce the required knee ROM and muscular efforts of the replaced limb. With these decreased movement requirements, there may also be a potential to additionally decrease the joint loading within the replaced knee during cycling which is important to minimize the possibility of implant loosening or mechanical failure {Andriacchi, 1986 #495;Nham, 2023 #184}. Initial investigation into asymmetric crank arm usage in a healthy population has demonstrated that knee ROM significantly reduced from symmetrical 170mm crank arms to asymmetrical 140mm crank arms by approximately 13° {Zhang, 2024 #601}. Peak KEMs reported by Zhang and Brown {Zhang, 2024 #601} in this range of crank arm adjustments were not significantly affected despite the asymmetric shortening. Peak KEMs also seem unaffected for the replaced limb in the TKA population in the 170mm to 140mm crank arm range (Chapter 4).

Estimating knee joint loading is difficult to do with traditional biomechanical analyses but is made possible through a musculoskeletal modeling approach. In lieu of instrumented knee implants used by Kutzner et al. (2012), musculoskeletal modeling using modeling packages such as OpenSim allow for tibiofemoral forces to be estimated from measured external kinematics and kinetics {Hummer, 2022 #169;Delp, 2007 #221}. Further, tibiofemoral forces can be decomposed into their medial compartment force (MCF), lateral compartment force (LCF), and total contact force (TCF) using an altered model of the commonly used *gait2392* model {Lerner, 2015 #171}. Musculoskeletal modeling of the TKA population in ergometer cycling has shown that participants exhibit significantly lower MCF of the replaced limb compared to the non-replaced limb {Hummer, 2022 #169}. Identifying the behavior of the knee compartment forces during novel movements for patient populations is especially important to ensure that loading is

not significantly increased in an unwanted manner. While OpenSim modeling is incredibly helpful for estimating internal physiological behaviors, modeling output accuracy should be investigated in accordance with established guidelines to ensure accurate conclusions are drawn. {Hicks, 2015 #414}.

Therefore, the primary purpose of this study was to evaluate the effects of asymmetrically shortened crank arms during cycling on contact forces, peak knee muscle group forces, and peak individual knee muscle forces of the replaced knee of participants with TKA. It is hypothesized that the peak TCF of the replaced limb will be unchanged during the asymmetric crank length conditions. It is also hypothesized that the peak quadriceps muscle force will be significantly lower at shorter crank arm lengths compared to the symmetrical, control lengths, primarily due to the expected decrease in knee ROM.

Methods

Participants

A single local physical therapy clinic was enlisted to recruit participants (4 female, 3 male, 67 ± 7 years, 1.7 ± 0.14 m, 95 ± 23.6 kg, 31.5 ± 4.6 BMI, 54 ± 4 days post-TKA) with at least one TKA (3 bilateral, 4 unilateral) performed by the same surgeon at Tennessee Orthopaedic Alliance East. Inclusion criteria for the study included those who were between 1- and 2-months post-operation, between the ages of 50 and 75, the ability to flex the replaced knee to at least 100° , and the ability to complete cycling and gait tasks without the aid of an external assistive device. Exclusionary criteria included participants with a BMI over 40, diagnosed OA of the spine, hips, or ankles or those with a contralateral TKA within 4 months of the testing date. Based on a repeated measures ANOVA of knee extension ROM in submaximal ergometer cycling in a young, healthy population, a power analysis (G*Power 3.1; Heinrich Heine

University Dusseldorf, Dusseldorf, Germany) found that, with an effect size of 0.979, a minimum of four participants were needed with an alpha level of 0.05 to achieve a power of 0.80 {Zhang, 2025 #591}. The university Institutional Review Board approved all procedures and study material used for this work.

Instrumentation

Kinematic (240 Hz), kinetic (1200 Hz), and EMG (1200 Hz) data were fed into OpenSim were collected using a 12-camera motion analysis system (Vicon Motion Analysis Inc., Oxford, UK) running the Vicon Nexus software (v2.16). The cycling tasks were conducted with a stationary electromagnetically braked cycle ergometer (Lode Excalibur Sport, Groningen, Netherlands) outfitted with custom-built instrumented bike pedals comprised of two 3-dimensional force sensors (Type 9027C, Kistler, Switzerland), and amplifiers (Type 5073A, Kistler, Switzerland). A standardized bike fit procedure described by Thorsen et al. {Thorsen, 2021 #290} was conducted to minimize the effect of ergometer adjustments on observed biomechanical results. Manufacturer-built crank arms were used (Sport crank, Lode) to allow for unilateral crank arm adjustments in increments of 1mm from 90mm to 185mm. Kinematic and kinetic data were filtered using a 4th order zero-lag Butterworth filter with a cutoff frequency of 6 Hz {Gardner, 2016 #106}. EMG data from a sixteen-channel wireless system (Trigno Wireless, Delsys Inc., Natick, MA, USA) were collected during cycling and EMG sensors were applied bilaterally to the rectus femoris (RF), semitendinosus (ST), medial gastrocnemius (MG), and the tibialis anterior (TA) in accordance with Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) guidelines {Hermens, 2000 #349}.

Cycling bouts were conducted for two minutes each with a requested cadence of 70 revolutions per minute (RPM) at 70 Watts. Five separate cycling conditions were tested that

consisten of changing the crank length unilaterally or bilaterally in increments of 15mm. The conditions (C) were as follows for the Replaced/Contralateral limbs (in mm): C1 (140/170), C2 (155/170), C3 (170/170). Data were recorded for the final 10 seconds of each condition and truncated into five consecutive crank cycles for analysis.

Musculoskeletal Simulation

The software package OpenSim (v4.5) was used to complete all musculoskeletal modeling {Delp, 2007 #221}. Each trial for the three conditions was individually exported from Visual 3D (Version v2025.01.2, HAS-Motion Inc., Kingston, Ontario, Canada) as an MOT file which uses experimental marker data and inverse kinematics to calculate the appropriate model scaling factors. In effect, this bypasses having to use the Scaling Tool within OpenSim. Input force vectors were programmatically adjusted using Matlab (2023a, MathWorks, Natick, Massachusetts, United States) to be compatible with cycling vertical pedal reaction force (PRF) data. The Lerner model was used that allowed for decomposition of knee joint reaction forces into medial and lateral compartments. Specifically, the knee joint retains only one degree-of-freedom that allows for flexion and extension, but the knee joint is composed of two revolute joints that allow the respective contact forces to be estimated. The Lerner model is based on the *gait2392* model which contains 92 musculotendon actuators and 23 degrees-of-freedom. The following compressive forces were reported: medial compartment contact force (MCF), lateral compartment contact force (LCF), and total contact force (TCF) of the tibiofemoral joint for the replaced limb were computed.

The OpenSim Export process in Visual3D adjusts the segment lengths and mass appropriately using the internal Visual 3D inverse kinematics algorithm. The subtalar angles were set to zero. Kinematics and experimental pedal reaction forces were used to compute

inverse dynamics within the pedals. Static Optimization (SO) was performed on five consecutive pedal cycles of each condition and results averaged together within condition. The SO objective function for evaluating muscle activations and forces minimized the sum of squared muscle activations as was previously done for modeling with this patient demographic {Hummer, 2022 #169}. The muscle force-length-velocity was included by default in OpenSim for these computations. Muscle groups chosen were summed as the following individual muscles: Quadriceps (rectus femoris, vastus medialis, vastus lateralis, vastus intermedius), Hamstrings (semitendinosus, semimembranosus, biceps femoris long head, biceps femoris short head), and Gastrocnemius (medial and lateral gastrocnemius). The Joint Reaction Analysis Tool was then used to estimate the TCF, MCF, and LCF, all of which were expressed in the tibia reference frame. OpenSim was programmatically executed using Python (3.12).

Data Analysis

Simulation lengths were determined as the time, during each trial, it took for the replaced limb to complete a full cycle starting from top dead center to the same limb returning to top dead center. Only the most recently replaced limb, for bilateral patients, or the replaced limb for unilateral patients was analyzed. Resulting muscle activations, muscle forces, and peak joint reserve actuator torques were compiled, inspected and qualitatively compared to experimental EMG, when applicable, to confirm result validity according to previously published guidelines {Hicks, 2015 #414}. Peak residuals were not reported as force from the handlebars and seat are unable to be accounted for. Peak values for joint reaction forces, muscle activations, and muscle forces were obtained from the power phase (0° to 180° crank angle) {Hummer, 2022 #169}.

EMG data were band pass filtered (20-450 Hz) by the collection hardware and high pass filtered digitally at 20 Hz to remove any potential zero offset. Filtered data were then full wave

rectified before applying a moving root mean square function with a 91ms window size. EMG data were normalized to the maximum value for each muscle during a complete crank cycle for the replaced limb across all of the conditions.

Statistical Analysis

All statistical testing was performed in SPSS v29 (IBM, Armonk, NY). One-way repeated measures ANOVAs with a 1x3 (Replaced Limb x Condition) design were used to assess the potential effects on MCF, LCF, and TCF as well as peak muscle force and activation variables with an α level set *a priori* to 0.05 as the threshold for significance. The Greenhouse-Geisser correction was used for assumptions of sphericity violation while normality was assessed using the Shapiro-Wilk test. To assess effect size, partial eta squared (η_p^2) was reported with the following classifications: small ($\eta_p^2 = 0.01$), medium ($\eta_p^2 = 0.06$), and large ($\eta_p^2 = 0.14$) {Lakens, 2013 #552}.

Results

The ANOVA results showed significantly reduced peak TCF from 170 – 140 mm with a large effect size ($p = 0.002$, $\eta_p^2 = 0.660$, **Table 7**). Post hoc comparisons revealed significant decreased TCF at 140mm ($p = 0.001$) and 155mm ($p = 0.024$) compared to 170mm. The average peak TCF decrease from 170mm to 155mm was approximately 178 N while the decrease from 155mm to 140mm was 26.76 N. Peak LCF was also significantly reduced across conditions ($p = 0.011$, $\eta_p^2 = 0.660$) with a large effect size. Post hoc comparisons determined that a significant decrease in LCF occurred at 140mm ($p < 0.001$) and 155mm ($p = 0.030$). The average peak LCF decrease from 170mm to 140mm was approximately 117 N. MCF was not significantly affected by the asymmetric shortening of the crank arms. Peak pedal reaction force showed no significant condition effect.

No significant changes in peak muscle group or selected individual muscle forces were seen (**Table 8, Table 9**). Simulation activation patterns for the ST (**Figure 7**, all figures available in the appendix) and RF (**Figure 8**) were more poorly matched to their experimental counterparts compared to the MG (**Figure 9**) and TA (**Figure 10**) activations. Experimental activation levels were consistently higher for all muscle groups despite the pattern agreement, but the magnitude differences are to be expected when comparing normalized experimental data and activation data generated from static optimization in OpenSim. Peak lower limb joint reserve torque values were also assessed (**Table 10**).

Discussion

The primary purpose of this study was to investigate the effects of asymmetrical crank arm shortening usage in ergometer cycling on estimated knee joint contact forces, peak muscle group activation, and peak individual muscle activations of the most recently replaced knee of participants. The primary hypothesis of unchanged TCF of the replaced limb for asymmetrical conditions was rejected with a significant average reduction of 19.8% when changing from 170mm to 140mm crank arms. Approximately 57% of the 206 N TCF reduction seen in the change from 170mm to 140mm crank arms came from the lateral compartment which indicates more off-loading of the lateral compartment of the replaced limb. Previous investigation into contact force estimation in participants with TKA cycling at 80 W and 80 RPM reported an average TCF of 918.6 N for the replaced limb {Hummer, 2022 #169}, which is slightly lower than the 1036 N in the present study. Since kinetic variables are not typically normalized in cycling research, differences in bodyweight could be a potential contributor to the small differences observed in TCF. However, the average bodyweight of the participants from

Hummer et al. {, 2022 #169} was 94.1kg while the average bodyweight from the current study was 95kg.

Available data for comparison to in-vivo vertical contact force is limited, but the Orthoload {P., 2008 #602} dataset consisting of cycling data at 60 RPM and 75 W demonstrates a peak vertical contact force of 907 N, or 0.98 BW, which is similar to the average 0.92 BW of the peak TCF seen in our study. Other in vivo reports of stationary cycling at 60 RPM at an unspecified wattage reported peak tibial forces of approximately 1.03 BW {D'Lima, 2008 #433}. Results from the present study appear to closely match the Orthoload dataset and slightly lower than that presented by D'Lima et al. {D'Lima, 2008 #433}, but it is difficult to assess without an explicit workrate setting provided by the authors. Another potential confounding variable is the different methods for bike fitting the participants. In the present study, due to discomfort at the knee, all participants opted for a lower seat height eliciting approximately 40° of knee flexion than the 25-30° range suggested by a standard bike fitting procedure {Hummer, 2021 #200}. A decrease in seat height combined with shortened crank arms for the replaced limb makes comparisons to literature estimates or in vivo data difficult beyond the symmetrical 170mm condition.

The hypotheses regarding decreased peak muscle group and individual muscle forces were not confirmed. It was expected that a decreased TCF would be accompanied by decreased peak muscle forces of the quadriceps, but this relationship was not displayed in this work. We found a non-significant decrease in peak quadriceps force from 1101 N at symmetrical 170mm crank arms to 942 N at asymmetrical 155mm crank arms, to 932 N at asymmetrical 140mm crank arms. This finding is surprising because initial work on compressive tibiofemoral forces linked higher peak forces primarily to quadriceps force, although this conclusion was drawn

from participants with cerebral palsy during crouch gait {Steele, 2012 #413}. A potential reason for the lack of quadriceps peak force changes in this investigation may come from the patient population used. Specifically, the Lerner model's muscle parameters are from the gait2392 model which did not use individuals recovering from a TKA as the basis for physiological muscle parameters. The neuromuscular damage from the surgery in these participants will affect their capacity to produce force and adjustments to the parameters of the replaced side's knee musculature may be helpful in the future. Another primary driver of altered TCF in cycling is the PRF. Our PRF results showed no significant changes across the crankarm lengths which is in agreement of non-significant changes of peak KEM within the replaced limb (Chapter 4). When combined, these findings seem to suggest that there may be future value in investigating the contralateral limb's behavior during this asymmetrical intervention.

Once the crank arm lengths were shortened, the average MCF and LCF were qualitatively equivalent which demonstrates potential benefits of asymmetric shortening for balancing the loading of the knee compartments. Previous investigations into compartment specific loading of the replaced limb during symmetric crank arm cycling at 80 W and 80 RPM using OpenSim did not report any significant difference in peak contact forces between the lateral and medial compartments where an average of 456.9 N and 489.4 N for MCF and LCF, respectively were reported {Hummer, 2022 #169}. The results from Hummer et al. {, 2021 #200} contrast those of Thorsen et al. {, 2021 #290} who reported that participants cycling at 80 W and 80 RPM had an average of 639.1 N MCF and 882.9 N TCF, indicating an approximate 243.8 N for LCF. Tibiofemoral alignment has an impact on medial and lateral compartment contact forces {Lerner, 2015 #171;Bennett, 2018 #600} and those alignment values were not reported by Thorsen et al. {, 2021 #290}. In the present study, the average frontal plane alignment of the replaced limbs

estimated from the static trial of motion capture was $176^{\circ} \pm 2^{\circ}$, indicating a slight bias towards varus for most participants. Additionally, it must be emphasized that Thorsen et al. {, 2021 #290} examined healthy, young males while Hummer et al. {, 2022 #169} did indeed recruit participants with one TKA and their results are not directly transferrable. The discrepancy in loading across the compartments in these two populations across the three studies indicates a significant off-loading of the medial compartment in TKA participants, likely due to the alignment corrected through TKA procedures.

Considering that TCF was reduced in conjunction with asymmetrically shortened crank arms of 140mm, this study presents evidence to suggest that asymmetric crank arm intervention is safe for use by clinicians in populations recovering from TKA or managing knee OA symptoms. For patients of TKA, cycling with shortened crankarms may be beneficial to accommodate the reduced knee ROM during cycling exercise soon after surgery and may help to promote rehabilitation progression. Specifically, crank arm length can be asymmetrically adjusted in a graded manner to progress with the patient as they improve function after surgery.

The primary limitation of this study is the low sample size of 7 participants, potentially further exacerbated by the mixture of unilateral and bilateral participants. Another limitation comes from the choice of SO as the method of calculating excitations and activations. Specifically, SO is a method that works independent of time and therefore discounts the known force properties of muscle that are time-dependent {Erdemir, 2007 #597}. Additionally, the objective function designed to minimize the sum of squared muscle activations may not be the most appropriate for the TKA population. Clancy et al. {, 2023 #434} evaluated the performance of the current objective function as well as one that adds a term to minimize tibiofemoral joint forces and reported that the latter helped to improve matching EMG profiles and in-vivo

measurements. However, they only examined healthy participants in cycling, and it was still seen that the additional term for contact force minimization was beneficial. It may be argued that one of the primary goals of patients with TKA is to minimize contact force, and potentially pain as a result, more so than minimizing entire body muscular activation levels. EMG profiles for two of the four reported muscles, ST and RF, somewhat poorly matched the qualitative patterns of the modeled muscle activations. Biarticular muscle discrepancies between experimental and simulated activations may be expected in this population due to altered quadriceps and hamstring co-contraction behavior that has been documented in the knee and hip OA population {Zeni, 2010 #599;Hoang, 2019 #546}. However, the conclusions from those previously mentioned studies should be interpreted carefully as patients with a total joint replacement may not exhibit the same behavior as those with diagnosed OA only.

Conclusion

To our knowledge, the present study is the first to perform musculoskeletal modeling investigating the vertical knee joint contact forces and peak knee musculature activation patterns of patients with a recent TKA cycling with asymmetrically adjusted crank arms . The findings demonstrate evidence of a significant reduction in TCF and LCF when cycling with crank arms that are 15 and 30 mm shorter on the replaced limb when compared to the standard 170mm crank arms. Peak muscle activations of the quadriceps, hamstrings, and gastrocnemius of the replaced limb were not significantly altered due to crank arm length changes. The findings suggest the potential for asymmetric crank arm usage to help reduce knee stiffness earlier during the rehabilitation process of patients recovering from TKA and it may reduce overall vertical contact forces while maintaining muscular activity levels. Future work may evaluate if the patterns of contact forces shown here are consistent longitudinally as rehabilitation progresses and if an

objective function to minimize joint reaction forces of the knee is beneficial in estimating contact force benchmarks in the TKA population.

Chapter 7: Conclusions

The purpose of this dissertation was to comprehensively assess the biomechanical changes elicited by asymmetric crank arm shortening in participants who had recently undergone a TKA surgery. Chapter 4 examined the responses of all participants responded to asymmetric and symmetric crank arms 1-2 months after surgery and found that shortened crank arms reduced knee range of motion without an increase in peak knee extension moments. Additionally, peak knee extension power was significantly reduced from the control crank arm length of 170mm to 140mm. Importantly, pain and RPE were not significantly affected by any of the cycling conditions and supports the use of asymmetric crank arms clinically.

Chapter 5 provides valuable insight into the lack of differences in response to the acute crank arm interventions between unilateral and bilateral TKA patients and demonstrated that rehabilitation with this method should equally affect both groups. Specifically, there was no difference in the relationship between limbs of unilateral or bilateral TKA participants in highly important variables such as peak KEM, peak knee power, knee ROM, or hip and ankle peak powers. The lack of differences found between the limbs of these groups suggest that, in the early stages of rehabilitation, the groups may be considered quite similar in their response to an asymmetric, acute intervention such as the one implemented in this work.

Chapter 6 implemented musculoskeletal modeling to estimate total knee contact forces and muscle activations and found reduced total contact force and unchanged knee musculature activations in response to asymmetrically shortened crank arms for the replaced limb.

Given the evidence from these studies, preliminary conclusions are that asymmetric crank arm usage in patients of TKA recovering from surgery seem to be safe for patients of TKA during early stage of rehabilitation and may help improve rehabilitation outcomes. Asymmetric

crank arms allow for the gradual adjustment of length to accommodate the reduced knee ROM of the recently replaced knee without increasing knee joint load. Future work should strive to improve the sample size of participants as well as investigate patient responses to asymmetric crank arm use via longitudinal cycling interventions.

References

1. Varacallo M, Luo TD, Johanson NA. Total Knee Arthroplasty Techniques. In: StatPearls. Treasure Island (FL): StatPearls Publishing
Copyright © 2023, StatPearls Publishing LLC. 2023
2. Parsons T, Al-Jabri T, Clement ND, Maffulli N, Kader DF. Patella resurfacing during total knee arthroplasty is cost-effective and has lower re-operation rates compared to non-resurfacing. J Orthop Surg Res 16(1): 185, 2021
3. Shichman I, Roof M, Askew N, Nherera L, Rozell JC, Seyler TM, Schwarzkopf R. Projections and Epidemiology of Primary Hip and Knee Arthroplasty in Medicare Patients to 2040-2060. JB JS Open Access 8(1), 2023
4. Guo B, Qin S, Huang Y. [Research progress of knee-salvage treatment for knee osteoarthritis]. Zhongguo Xiu Fu Chong Jian Wai Ke Za Zhi 32(10): 1292, 2018
5. Hanna FS, Teichtahl AJ, Wluka AE, Wang Y, Urquhart DM, English DR, Giles GG, Cicuttini FM. Women have increased rates of cartilage loss and progression of cartilage defects at the knee than men: a gender study of adults without clinical knee osteoarthritis. Menopause 16(4): 666, 2009
6. O'Connor MI. Implant survival, knee function, and pain relief after TKA: are there differences between men and women? Clin Orthop Relat Res 469(7): 1846, 2011
7. Vina ER, Ran D, Ashbeck EL, Kwok CK. Widespread Pain Is Associated with Increased Risk of No Clinical Improvement After TKA in Women. Clin Orthop Relat Res 478(7): 1453, 2020
8. Mehta SP, Perruccio AV, Palaganas M, Davis AM. Do women have poorer outcomes following total knee replacement? Osteoarthritis Cartilage 23(9): 1476, 2015

9. Johnson AJ, Costa CR, Mont MA. Do we need gender-specific total joint arthroplasty? Clin Orthop Relat Res 469(7): 1852, 2011
10. Driban JB, Harkey MS, Barbe MF, Ward RJ, MacKay JW, Davis JE, Lu B, Price LL, Eaton CB, Lo GH, McAlindon TE. Risk factors and the natural history of accelerated knee osteoarthritis: a narrative review. BMC Musculoskelet Disord 21(1): 332, 2020
11. Arnold JB, Walters JL, Ferrar KE. Does Physical Activity Increase After Total Hip or Knee Arthroplasty for Osteoarthritis? A Systematic Review. J Orthop Sports Phys Ther 46(6): 431, 2016
12. Vissers MM, de Groot IB, Reijman M, Bussmann JB, Stam HJ, Verhaar JA. Functional capacity and actual daily activity do not contribute to patient satisfaction after total knee arthroplasty. BMC Musculoskelet Disord 11: 121, 2010
13. Mistry JB, Elmallah RD, Bhave A, Chughtai M, Cherian JJ, McGinn T, Harwin SF, Mont MA. Rehabilitative Guidelines after Total Knee Arthroplasty: A Review. J Knee Surg 29(3): 201, 2016
14. Papalia R, Campi S, Vorini F, Zampogna B, Vasta S, Papalia G, Fossati C, Torre G, Denaro V. The Role of Physical Activity and Rehabilitation Following Hip and Knee Arthroplasty in the Elderly. J Clin Med 9(5), 2020
15. Zech A, Hendrich S, Pfeifer K. Association Between Exercise Therapy Dose and Functional Improvements in the Early Postoperative Phase After Hip and Knee Arthroplasty: An Observational Study. Pm r 7(10): 1064, 2015

16. Su W, Zhou Y, Qiu H, Wu H. The effects of preoperative rehabilitation on pain and functional outcome after total knee arthroplasty: a meta-analysis of randomized controlled trials. *J Orthop Surg Res* 17(1): 175, 2022
17. Pellegrini CA, Chang RW, Dunlop DD, Conroy DE, Lee J, Van Horn L, Spring B, Cameron KA. Comparison of a Patient-Centered Weight Loss Program starting before versus after knee replacement: A pilot study. *Obes Res Clin Pract* 12(5): 472, 2018
18. Bade MJ, Struessel T, Dayton M, Foran J, Kim RH, Miner T, Wolfe P, Kohrt WM, Dennis D, Stevens-Lapsley JE. Early High-Intensity Versus Low-Intensity Rehabilitation After Total Knee Arthroplasty: A Randomized Controlled Trial. *Arthritis Care Res (Hoboken)* 69(9): 1360, 2017
19. Mayr HO, Reinhold M, Bernstein A, Suedkamp NP, Stoeckl A. Sports activity following total knee arthroplasty in patients older than 60 years. *J Arthroplasty* 30(1): 46, 2015
20. Oja P, Titze S, Bauman A, de Geus B, Krenn P, Reger-Nash B, Kohlberger T. Health benefits of cycling: a systematic review. *Scand J Med Sci Sports* 21(4): 496, 2011
21. Logan G, Somers C, Baker G, Connell H, Gray S, Kelly P, McIntosh E, Welsh P, Gray CM, Gill JMR. Benefits, risks, barriers, and facilitators to cycling: a narrative review. *Front Sports Act Living* 5: 1168357, 2023
22. Chavarrias M, Carlos-Vivas J, Collado-Mateo D, Pérez-Gómez J. Health Benefits of Indoor Cycling: A Systematic Review. *Medicina (Kaunas)* 55(8), 2019
23. Johan de Hartog J, Boogaard H, Nijland H, Hoek G. Do the health benefits of cycling outweigh the risks? *Environ Health Perspect* 118(8): 1109, 2010

24. Littman AJ, Kristal AR, White E. Effects of physical activity intensity, frequency, and activity type on 10-y weight change in middle-aged men and women. *Int J Obes (Lond)* 29(5): 524, 2005
25. Kutzner I, Heinlein B, Graichen F, Rohlmann A, Halder AM, Beier A, Bergmann G. Loading of the knee joint during ergometer cycling: telemetric in vivo data. *J Orthop Sports Phys Ther* 42(12): 1032, 2012
26. Kutzner I, Heinlein B, Graichen F, Bender A, Rohlmann A, Halder A, Beier A, Bergmann G. Loading of the knee joint during activities of daily living measured in vivo in five subjects. *J Biomech* 43(11): 2164, 2010
27. Lester D, Barber C, Sowers CB, Cyrus JW, Vap AR, Golladay GJ, Patel NK. Return to sport post-knee arthroplasty : an umbrella review for consensus guidelines. *Bone Jt Open* 3(3): 245, 2022
28. Hamai S, Miura H, Higaki H, Shimoto T, Matsuda S, Okazaki K, Iwamoto Y. Three-dimensional knee joint kinematics during golf swing and stationary cycling after total knee arthroplasty. *Journal of Orthopaedic Research* 26(12): 1556, 2008
29. Zhao D, Banks SA, D'Lima DD, Colwell CW, Jr., Fregly BJ. In vivo medial and lateral tibial loads during dynamic and high flexion activities. *J Orthop Res* 25(5): 593, 2007
30. Hummer E, Thorsen T, Weinhandl JT, Cates H, Zhang S. Knee joint biomechanics of patients with unilateral total knee arthroplasty during stationary cycling. *J Biomech* 115: 110111, 2021
31. Yocum DS, Valenzuela KA, Standifird TW, Cates HE, Zhang S. Altered biomechanics in bilateral total knee replacement patients during stair negotiation. *Knee* 34: 9, 2022

32. Paravlic AH, Meulenberg CJ, Drole K. The Time Course of Quadriceps Strength Recovery After Total Knee Arthroplasty Is Influenced by Body Mass Index, Sex, and Age of Patients: Systematic Review and Meta-Analysis. *Front Med (Lausanne)* 9: 865412, 2022
33. Barratt PR, Martin JC, Elmer SJ, Korff T. Effects of Pedal Speed and Crank Length on Pedaling Mechanics during Submaximal Cycling. *Med Sci Sports Exerc* 48(4): 705, 2016
34. Childers WL, Kogler GF. Symmetrical kinematics does not imply symmetrical kinetics in people with transtibial amputation using cycling model. *J Rehabil Res Dev* 51(8): 1243, 2014
35. Millour G, Duc S, Puel F, Bertucci WM. Effect of asymmetric crank arm lengths on performance-related variables in cyclists with an anatomical lower limb length discrepancy. *Sports Engineering* 23: 1, 2020
36. Hummer ET, Thorsen T, Weinhandl JT, Reinbolt JA, Cates H, Zhang S. Medial and Lateral Tibiofemoral Compressive Forces in Patients Following Unilateral Total Knee Arthroplasty During Stationary Cycling. *J Appl Biomech* 38(3): 179, 2022
37. Hurworth M, Evans JM, Gibbons R, Mackie KE, Edmondston SJ. Cycle Sprint Test for the Evaluation of Lower Limb Muscle Power After Total Knee Arthroplasty: A Proof-of-Concept Study. *Arthroplast Today* 9: 118, 2021
38. Lerner ZF, DeMers MS, Delp SL, Browning RC. How tibiofemoral alignment and contact locations affect predictions of medial and lateral tibiofemoral contact forces. *J Biomech* 48(4): 644, 2015
39. Thorsen T, Wen C, Zhang S. Are Medial and Lateral Tibiofemoral Compressive Forces Different in Uphill Compared to Level Walking for Patients Following Total Knee Arthroplasty? *J Biomech Eng* 143(10), 2021

40. Liebs TR, Herzberg W, Rüther W, Haasters J, Russlies M, Hassenpflug J. Ergometer cycling after hip or knee replacement surgery: a randomized controlled trial. *J Bone Joint Surg Am* 92(4): 814, 2010
41. Black AL, Clark AL. Sexual dimorphism in knee osteoarthritis: Biomechanical variances and biological influences. *J Orthop* 32: 104, 2022
42. Shao Y, Zhang C, Charron KD, Macdonald SJ, McCalden RW, Bourne RB. The fate of the remaining knee(s) or hip(s) in osteoarthritic patients undergoing a primary TKA or THA. *J Arthroplasty* 28(10): 1842, 2013
43. Nham FH, Patel I, Zalikha AK, El-Othmani MM. Epidemiology of primary and revision total knee arthroplasty: analysis of demographics, comorbidities and outcomes from the national inpatient sample. *Arthroplasty* 5(1): 18, 2023
44. Mak WK, Bin Abd Razak HR, Tan HA. Which Patients Require a Contralateral Total Knee Arthroplasty Within 5 Years of Index Surgery? *J Knee Surg* 33(10): 1029, 2020
45. Kulkarni K, Karssiens T, Kumar V, Pandit H. Obesity and osteoarthritis. *Maturitas* 89: 22, 2016
46. Nicholls AS, Kiran A, Javaid MK, Hart DJ, Spector TD, Carr AJ, Arden NK. Change in body mass index during middle age affects risk of total knee arthroplasty due to osteoarthritis: a 19-year prospective study of 1003 women. *Knee* 19(4): 316, 2012
47. Lui M, Jones CA, Westby MD. Effect of non-surgical, non-pharmacological weight loss interventions in patients who are obese prior to hip and knee arthroplasty surgery: a rapid review. *Syst Rev* 4: 121, 2015

48. Vaqar S, Graber M. Thromboembolic Event. In: StatPearls. Treasure Island (FL): StatPearls Publishing

Copyright © 2023, StatPearls Publishing LLC. 2023

49. Barahona M, Barahona MA, Navarro T, Chamorro P, Alegría A, Guzman M, Palet MJ.

Increase in Postoperative Body Mass Index in Patients After Total Knee Arthroplasty. *Cureus* 15(9): e46203, 2023

50. Dowsey MM, Liew D, Choong PF. Economic burden of obesity in primary total knee arthroplasty. *Arthritis Care Res (Hoboken)* 63(10): 1375, 2011

51. Kellgren JH, Lawrence JS. Radiological assessment of osteo-arthritis. *Ann Rheum Dis* 16(4): 494, 1957

52. Kohn MD, Sassoon AA, Fernando ND. Classifications in Brief: Kellgren-Lawrence Classification of Osteoarthritis. *Clinical Orthopaedics & Related Research* 474(8): 1886, 2016

53. Sharma L, Song J, Dunlop D, Felson D, Lewis CE, Segal N, Torner J, Cooke TD, Hietpas J, Lynch J, Nevitt M. Varus and valgus alignment and incident and progressive knee osteoarthritis. *Ann Rheum Dis* 69(11): 1940, 2010

54. Ginesin E, Norman D, Peskin B. Knee Alignment and Its Significance: Is It Really Different in Various Population Groups? *Isr Med Assoc J* 20(2): 109, 2018

55. Mitani Y. Gender-related differences in lower limb alignment, range of joint motion, and the incidence of sports injuries in Japanese university athletes. *J Phys Ther Sci* 29(1): 12, 2017

56. Wei J, Gross D, Lane NE, Lu N, Wang M, Zeng C, Yang T, Lei G, Choi HK, Zhang Y. Risk factor heterogeneity for medial and lateral compartment knee osteoarthritis: analysis of two prospective cohorts. *Osteoarthritis Cartilage* 27(4): 603, 2019

57. Modenese L, Montefiori E, Wang A, Wesarg S, Viceconti M, Mazzà C. Investigation of the dependence of joint contact forces on musculotendon parameters using a codified workflow for image-based modelling. *J Biomech* 73: 108, 2018
58. Duffaut CJ, Goldman J, Miller EM. Clinical Evaluation of the Knee Arthritis Patient. *Tech Vasc Interv Radiol* 26(1): 100876, 2023
59. Aljehani MS, Christensen JC, Snyder-Mackler L, Crenshaw J, Brown A, Zeni JA, Jr. Knee biomechanics and contralateral knee osteoarthritis progression after total knee arthroplasty. *Gait Posture* 91: 266, 2022
60. Laxafoss E, Jacobsen S, Gosvig KK, Sonne-Holm S. The alignment of the knee joint in relationship to age and osteoarthritis: the Copenhagen Osteoarthritis Study. *Skeletal Radiol* 42(4): 531, 2013
61. Everhart JS, Abouljoud MM, Poland SG, Flanigan DC. Medial compartment defects progress at a more rapid rate than lateral cartilage defects in older adults with minimal to moderate knee osteoarthritis (OA): data from the OA initiative. *Knee Surg Sports Traumatol Arthrosc* 27(8): 2401, 2019
62. Song SJ, Park CH, Bae DK. What to Know for Selecting Cruciate-Retaining or Posterior-Stabilized Total Knee Arthroplasty. *Clin Orthop Surg* 11(2): 142, 2019
63. LeDuc RC, Upadhyay D, Brown NM. Cruciate-Retaining Versus Cruciate-Substituting Total Knee Arthroplasty: A Meta-Analysis. *Indian J Orthop* 57(8): 1188, 2023
64. Li N, Tan Y, Deng Y, Chen L. Posterior cruciate-retaining versus posterior stabilized total knee arthroplasty: a meta-analysis of randomized controlled trials. *Knee Surg Sports Traumatol Arthrosc* 22(3): 556, 2014

65. Zelle J, Van der Zanden AC, De Waal Malefijt M, Verdonschot N. Biomechanical analysis of posterior cruciate ligament retaining high-flexion total knee arthroplasty. *Clin Biomech (Bristol, Avon)* 24(10): 842, 2009
66. Gray HA, Guan S, Young TJ, Dowsey MM, Choong PF, Pandy MG. Comparison of posterior-stabilized, cruciate-retaining, and medial-stabilized knee implant motion during gait. *J Orthop Res* 38(8): 1753, 2020
67. Broberg JS, Ndoja S, MacDonald SJ, Lanting BA, Teeter MG. Comparison of Contact Kinematics in Posterior-Stabilized and Cruciate-Retaining Total Knee Arthroplasty at Long-Term Follow-Up. *J Arthroplasty* 35(1): 272, 2020
68. Tsai TY, Liow MHL, Li G, Arauz P, Peng Y, Klemm C, Kwon YM. Bi-Cruciate Retaining Total Knee Arthroplasty Does Not Restore Native Tibiofemoral Articular Contact Kinematics During Gait. *J Orthop Res* 37(9): 1929, 2019
69. Boese CK, Ebohon S, Ries C, De Faoite D. Bi-cruciate retaining total knee arthroplasty: a systematic literature review of clinical outcomes. *Arch Orthop Trauma Surg* 141(2): 293, 2021
70. Arnout N, Victor J, Vermue H, Pringels L, Bellemans J, Verstraete MA. Knee joint laxity is restored in a bi-cruciate retaining TKA-design. *Knee Surg Sports Traumatol Arthrosc* 28(9): 2863, 2020
71. Yang K, Sohn G, Sambandam S. Cruciate-Retaining Total Knee Arthroplasty: Current Concepts Review. *Cureus* 15(8): e43813, 2023
72. Movassaghi K, Patel A, Ghulam-Jelani Z, Levine BR. Modern Total Knee Arthroplasty Bearing Designs and the Role of the Posterior Cruciate Ligament. *Arthroplast Today* 21: 101130, 2023

73. Putman S, Boureau F, Girard J, Migaud H, Pasquier G. Patellar complications after total knee arthroplasty. *Orthop Traumatol Surg Res* 105(1s): S43, 2019
74. Volkmar AJ, Elrod R, Vickery JW, Yang CC, Polkowski GG, Martin JR. Cruciate Substituting Implants in Primary Total Knee Arthroplasty. *Adv Orthop* 2022: 2676715, 2022
75. Sabatini L, Barberis L, Camazzola D, Centola M, Capella M, Bistolfi A, Schiraldi M, Massè A. Bicruciate-retaining total knee arthroplasty: What's new? *World J Orthop* 12(10): 732, 2021
76. Blakeney W, Clément J, Desmeules F, Hagemeister N, Rivière C, Vendittoli PA. Kinematic alignment in total knee arthroplasty better reproduces normal gait than mechanical alignment. *Knee Surg Sports Traumatol Arthrosc* 27(5): 1410, 2019
77. Risitano S, Cacciola G, Sabatini L, Capella M, Bosco F, Giustra F, Massè A, Vaishya R. Restricted kinematic alignment in primary total knee arthroplasty: A systematic review of radiographic and clinical data. *J Orthop* 33: 37, 2022
78. Williams D, Metcalfe A, Madete J, Whatling G, Kempshall P, Forster M, Lyons K, Holt C. The relationship between alignment, function and loading in total knee replacement: In-vivo analysis of a unique patient population. *J Biomech* 112: 110042, 2020
79. Rivière C, Iranpour F, Auvinet E, Howell S, Vendittoli PA, Cobb J, Parratte S. Alignment options for total knee arthroplasty: A systematic review. *Orthop Traumatol Surg Res* 103(7): 1047, 2017
80. Vendittoli PA, Martinov S, Blakeney WG. Restricted Kinematic Alignment, the Fundamentals, and Clinical Applications. *Front Surg* 8: 697020, 2021
81. Moreland JR, Bassett LW, Hanker GJ. Radiographic analysis of the axial alignment of the lower extremity. *J Bone Joint Surg Am* 69(5): 745, 1987

82. Hohman DW, Jr., Nodzo SR, Phillips M, Fitz W. The implications of mechanical alignment on soft tissue balancing in total knee arthroplasty. *Knee Surg Sports Traumatol Arthrosc* 23(12): 3632, 2015
83. Diamond LE, Wrigley TV, Bennell KL, Hinman RS, O'Donnell J, Hodges PW. Hip joint biomechanics during gait in people with and without symptomatic femoroacetabular impingement. *Gait Posture* 43: 198, 2016
84. Dargel J, Pennig L, Schnurr C, Boese CK, Eysel P, Oppermann J. [Should we use hip-ankle radiographs to assess the coronal alignment after total knee arthroplasty?]. *Orthopade* 45(7): 591, 2016
85. Skyttä ET, Lohman M, Tallroth K, Remes V. Comparison of standard anteroposterior knee and hip-to-ankle radiographs in determining the lower limb and implant alignment after total knee arthroplasty. *Scand J Surg* 98(4): 250, 2009
86. Kumar N, Yadav C, Raj R, Anand S. How to interpret postoperative X-rays after total knee arthroplasty. *Orthop Surg* 6(3): 179, 2014
87. Aït-Si-Selmi T, Marie-Hardy L, O'Loughlin PF, Kobayashi K, Müller JH, Saffarini M, Bonnin MP. Patellar facet ratio affects knee pain, stair climbing and stair descent after TKA without patellar resurfacing. *Knee Surg Sports Traumatol Arthrosc* 28(12): 3796, 2020
88. Miozzari HH, Armand S, Turcot K, Lübbecke A, Bonnefoy-Mazure A. Gait Analysis 1 Year after Primary TKA: No Difference between Gap Balancing and Measured Resection Technique. *J Knee Surg* 34(8): 898, 2021

89. Dossett HG, Arthur JR, Makovicka JL, Mara KC, Bingham JS, Clarke HD, Spangehl MJ. A Randomized Controlled Trial of Kinematically and Mechanically Aligned Total Knee Arthroplasties: Long-Term Follow-Up. *J Arthroplasty* 38(6s): S209, 2023
90. Miura T, Takahashi T, Watanabe J, Kataoka Y, Ae R, Saito H, Takeshita K, Miyakoshi N. Postoperative clinical outcomes for kinematically, restricted kinematically, or mechanically aligned total knee arthroplasty: a systematic review and network meta-analysis of randomized controlled trials. *BMC Musculoskelet Disord* 24(1): 322, 2023
91. Gao ZX, Long NJ, Zhang SY, Yu W, Dai YX, Xiao C. Comparison of Kinematic Alignment and Mechanical Alignment in Total Knee Arthroplasty: A Meta-analysis of Randomized Controlled Clinical Trials. *Orthop Surg* 12(6): 1567, 2020
92. Bannuru RR, Osani MC, Vaysbrot EE, Arden NK, Bennell K, Bierma-Zeinstra SMA, Kraus VB, Lohmander LS, Abbott JH, Bhandari M, Blanco FJ, Espinosa R, Haugen IK, Lin J, Mandl LA, Moilanen E, Nakamura N, Snyder-Mackler L, Trojian T, Underwood M, McAlindon TE. OARSI guidelines for the non-surgical management of knee, hip, and polyarticular osteoarthritis. *Osteoarthritis Cartilage* 27(11): 1578, 2019
93. Kolasinski SL, Neogi T, Hochberg MC, Oatis C, Guyatt G, Block J, Callahan L, Copenhaver C, Dodge C, Felson D, Gellar K, Harvey WF, Hawker G, Herzig E, Kwoh CK, Nelson AE, Samuels J, Scanzello C, White D, Wise B, Altman RD, DiRenzo D, Fontanarosa J, Giradi G, Ishimori M, Misra D, Shah AA, Shmagel AK, Thoma LM, Turgunbaev M, Turner AS, Reston J. 2019 American College of Rheumatology/Arthritis Foundation Guideline for the Management of Osteoarthritis of the Hand, Hip, and Knee. *Arthritis Care Res (Hoboken)* 72(2): 149, 2020

94. Mooiweer Y, Stevens M, van den Akker-Scheek I. Being active with a total hip or knee prosthesis: a systematic review into physical activity and sports recommendations and interventions to improve physical activity behavior. *Eur Rev Aging Phys Act* 19(1): 7, 2022
95. Vu-Han TL, Gwinner C, Perka C, Hardt S. Recommendations for Patients with High Return to Sports Expectations after TKA Remain Controversial. *J Clin Med* 10(1), 2020
96. Beswick AD, Wylde V, Gooberman-Hill R, Blom A, Dieppe P. What proportion of patients report long-term pain after total hip or knee replacement for osteoarthritis? A systematic review of prospective studies in unselected patients. *BMJ Open* 2(1): e000435, 2012
97. Thong ISK, Jensen MP, Miró J, Tan G. The validity of pain intensity measures: what do the NRS, VAS, VRS, and FPS-R measure? *Scand J Pain* 18(1): 99, 2018
98. Franchignoni F, Giordano A, Ferriero G, Monticone M. Measurement precision of the Pain Catastrophizing Scale and its short forms in chronic low back pain. *Sci Rep* 12(1): 12042, 2022
99. Burns LC, Ritvo SE, Ferguson MK, Clarke H, Seltzer Z, Katz J. Pain catastrophizing as a risk factor for chronic pain after total knee arthroplasty: a systematic review. *J Pain Res* 8: 21, 2015
100. Ong WJ, Kwan YH, Lim ZY, Thumboo J, Yeo SJ, Yeo W, Wong SB, Leung YY. Measurement properties of Pain Catastrophizing Scale in patients with knee osteoarthritis. *Clin Rheumatol* 40(1): 295, 2021
101. Birch S, Stilling M, Mechlenburg I, Hansen TB. The association between pain catastrophizing, physical function and pain in a cohort of patients undergoing knee arthroplasty. *BMC Musculoskelet Disord* 20(1): 421, 2019

102. Kazarian GS, Anthony CA, Lawrie CM, Barrack RL. The Impact of Psychological Factors and Their Treatment on the Results of Total Knee Arthroplasty. *J Bone Joint Surg Am* 103(18): 1744, 2021
103. Güney-Deniz H, Irem Kınıklı G, Çağlar Ö, Atilla B, Yüksel İ. Does kinesiophobia affect the early functional outcomes following total knee arthroplasty? *Physiother Theory Pract* 33(6): 448, 2017
104. Yan Z, Wu Y, Liu M, Wang X, Wang J, Wang Z, Wu S, Luan X. Heterogeneous trajectories of kinesiophobia and their effects on rehabilitation outcomes after total knee arthroplasty: a prospective cohort study. *J Orthop Surg Res* 18(1): 449, 2023
105. Bakaa N, Chen LH, Carlesso L, Richardson J, Shanthanna H, Macedo L. Understanding barriers and facilitators of exercise adherence after total-knee arthroplasty. *Disabil Rehabil* 44(21): 6348, 2022
106. Swanson EA, Schmalzried TP, Dorey FJ. Activity recommendations after total hip and knee arthroplasty: a survey of the American Association for Hip and Knee Surgeons. *J Arthroplasty* 24(6 Suppl): 120, 2009
107. McConnell S, Kolopack P, Davis AM. The Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC): a review of its utility and measurement properties. *Arthritis Rheum* 45(5): 453, 2001
108. Nilsson AK, Lohmander LS, Klässbo M, Roos EM. Hip disability and osteoarthritis outcome score (HOOS)--validity and responsiveness in total hip replacement. *BMC Musculoskelet Disord* 4: 10, 2003

109. Culliton SE, Bryant DM, MacDonald SJ, Hibbert KM, Chesworth BM. Validity and Internal Consistency of the New Knee Society Knee Scoring System. *Clin Orthop Relat Res* 476(1): 77, 2018
110. Daly DJ, Cavanagh PR. Asymmetry in bicycle ergometer pedalling. *Med Sci Sports* 8(3): 204, 1976
111. Soden PD, Adeyefa BA. Forces applied to a bicycle during normal cycling. *J Biomech* 12(7): 527, 1979
112. Montalvo-Pérez A, Alejo LB, Valenzuela PL, Castellanos M, Gil-Cabrera J, Talavera E, Lucia A, Barranco-Gil D. Validity of the Favero Assioma Duo Power Pedal System for Measuring Power Output and Cadence. *Sensors* 21(7): 2277, 2021
113. Rodríguez-Rielves V, Lillo-Beviá JR, Buendía-Romero Á, Martínez-Cava A, Hernández-Belmonte A, Courel-Ibáñez J, Pallarés JG. Are the Assioma Favero Power Meter Pedals a Reliable Tool for Monitoring Cycling Power Output? *Sensors* 21(8): 2789, 2021
114. Fang Y, Fitzhugh EC, Crouter SE, Gardner JK, Zhang S. Effects of Workloads and Cadences on Frontal Plane Knee Biomechanics in Cycling. *Med Sci Sports Exerc* 48(2): 260, 2016
115. Bini R, Diefenthaler F, Carpes F. Determining force and power in cycling: A review of methods and instruments for pedal force and crank torque measurements. *INTERNATIONAL SPORTMED JOURNAL* 15: 96, 2014
116. Bini RR, Tamborindeguy AC, Mota CB. Effects of saddle height, pedaling cadence, and workload on joint kinetics and kinematics during cycling. *J Sport Rehabil* 19(3): 301, 2010

117. Hubenig LR, Game AB, Kennedy MD. Effect of different bicycle body positions on power output in aerobically trained females. *Res Sports Med* 19(4): 245, 2011
118. Bini RR, Hume PA, Lanferdini FJ, Vaz MA. Effects of body positions on the saddle on pedalling technique for cyclists and triathletes. *Eur J Sport Sci* 14 Suppl 1: S413, 2014
119. Ericson MO, Nisell R. Patellofemoral joint forces during ergometric cycling. *Phys Ther* 67(9): 1365, 1987
120. Bini R, Hume PA, Croft JL. Effects of bicycle saddle height on knee injury risk and cycling performance. *Sports Med* 41(6): 463, 2011
121. Hamley EJ, Thomas V. Physiological and postural factors in the calibration of the bicycle ergometer. *J Physiol* 191(2): 55p, 1967
122. Holmes JC, Pruitt AL, Whalen NJ. Lower extremity overuse in bicycling. *Clin Sports Med* 13(1): 187, 1994
123. Ferrer-Roca V, Bescós R, Roig A, Galilea P, Valero O, García-López J. Acute effects of small changes in bicycle saddle height on gross efficiency and lower limb kinematics. *J Strength Cond Res* 28(3): 784, 2014
124. Delp SL, Loan JP, Hoy MG, Zajac FE, Topp EL, Rosen JM. An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. *IEEE Trans Biomed Eng* 37(8): 757, 1990
125. Peveler WW, Green JM. Effects of saddle height on economy and anaerobic power in well-trained cyclists. *J Strength Cond Res* 25(3): 629, 2011
126. Sanderson DJ, Amoroso AT. The influence of seat height on the mechanical function of the triceps surae muscles during steady-rate cycling. *J Electromyogr Kinesiol* 19(6): e465, 2009

127. Jorge M, Hull ML. Analysis of EMG measurements during bicycle pedalling. *J Biomech* 19(9): 683, 1986
128. Bini RR, Hume PA, Lanferdini FJ, Vaz MA. Effects of moving forward or backward on the saddle on knee joint forces during cycling. *Phys Ther Sport* 14(1): 23, 2013
129. Menard M, Domalain M, Decatoire A, Lacouture P. Influence of saddle setback on pedalling technique effectiveness in cycling. *Sports Biomech* 15(4): 462, 2016
130. Menard M, Domalain M, Decatoire A, Lacouture P. Influence of saddle setback on knee joint forces in cycling. *Sports Biomech* 19(2): 245, 2020
131. Bini R, Daly L, Kingsley M. Changes in body position on the bike during seated sprint cycling: Applications to bike fitting. *Eur J Sport Sci* 20(1): 35, 2020
132. Skovereng K, Aasvold LO, Ettema G. On the effect of changing handgrip position on joint specific power and cycling kinematics in recreational and professional cyclists. *PLoS One* 15(8): e0237768, 2020
133. Holliday W, Swart J. Performance variables associated with bicycle configuration and flexibility. *J Sci Med Sport* 24(3): 312, 2021
134. Cruz CF, Bankoff AD. Electromyography in cycling: difference between clipless pedal and toe clip pedal. *Electromyogr Clin Neurophysiol* 41(4): 247, 2001
135. Ostler LM, Betts JA, Gore CJ. Gross cycling efficiency is not altered with and without toe-clips. *J Sports Sci* 26(1): 47, 2008
136. Boyd TF, Neptune RR, Hull ML. Pedal and knee loads using a multi-degree-of-freedom pedal platform in cycling. *J Biomech* 30(5): 505, 1997

137. Wootten D, Hull ML. Design and Evaluation of a Multi-Degree-of-Freedom Foot/Pedal Interface for Cycling. *International Journal of Sport Biomechanics* 8(2): 152, 1992
138. Hermens HJ, Freriks B, Disselhorst-Klug C, Rau G. Development of recommendations for SEMG sensors and sensor placement procedures. *J Electromyogr Kinesiol* 10(5): 361, 2000
139. Talib I, Sundaraj K, Lam CK, Hussain J, Ali MA. A review on crosstalk in myographic signals. *Eur J Appl Physiol* 119(1): 9, 2019
140. Arauz P, Klemm C, Limmahakhun S, An S, Kwon YM. Stair Climbing and High Knee Flexion Activities in Bi-Cruciate Retaining Total Knee Arthroplasty: In Vivo Kinematics and Articular Contact Analysis. *J Arthroplasty* 34(3): 570, 2019
141. De Luca CJ. The Use of Surface Electromyography in Biomechanics. *Journal of Applied Biomechanics* 13(2): 135, 1997
142. Ptazkowski K, Włodarczyk P, Paprocka-Borowicz M. The Relationship Between The Electromyographic Activity Of Rectus And Oblique Abdominal Muscles And Bioimpedance Body Composition Analysis - A Pilot Observational Study. *Diabetes Metab Syndr Obes* 12: 2033, 2019
143. Basmajian JV, Luca CJD. *Muscles Alive : their functions revealed by electromyography*. Baltimore: Williams & Wilkins Baltimore, 1985
144. Dorel S, Couturier A, Hug F. Intra-session repeatability of lower limb muscles activation pattern during pedaling. *J Electromyogr Kinesiol* 18(5): 857, 2008
145. Marsh AP, Martin PE. The relationship between cadence and lower extremity EMG in cyclists and noncyclists. *Med Sci Sports Exerc* 27(2): 217, 1995

146. Ericson M. On the biomechanics of cycling. A study of joint and muscle load during exercise on the bicycle ergometer. *Scand J Rehabil Med Suppl* 16: 1, 1986
147. Hunter AM, St Clair Gibson A, Lambert M, Noakes TD. Electromyographic (EMG) normalization method for cycle fatigue protocols. *Med Sci Sports Exerc* 34(5): 857, 2002
148. Fernández-Peña E, Lucertini F, Ditroilo M. A maximal isokinetic pedalling exercise for EMG normalization in cycling. *J Electromyogr Kinesiol* 19(3): e162, 2009
149. Tabasi A, Brouwer NP, Kingma I, van Dijk W, de Looze MP, Moya-Esteban A, Kooij HVD, van Dieën JH. The effect of back muscle fatigue on EMG and kinematics based estimation of low-back loads and active moments during manual lifting tasks. *J Electromyogr Kinesiol* 73: 102815, 2023
150. Hautier CA, Arsac LM, Deghdegh K, Souquet J, Belli A, Lacour JR. Influence of fatigue on EMG/force ratio and cocontraction in cycling. *Med Sci Sports Exerc* 32(4): 839, 2000
151. Suzuki S, Watanabe S, Homma S. EMG activity and kinematics of human cycling movements at different constant velocities. *Brain Res* 240(2): 245, 1982
152. Hug F, Dorel S. Electromyographic analysis of pedaling: A review. *Journal of Electromyography and Kinesiology* 19(2): 182, 2009
153. Holliday W, Theo R, Fisher J, Swart J. Cycling: joint kinematics and muscle activity during differing intensities. *Sports Biomech* 22(5): 660, 2023
154. Wilkinson RD, Lichtwark GA, Cresswell AG. The Mechanics of Seated and Nonseated Cycling at Very-High-Power Output: A Joint-Level Analysis. *Med Sci Sports Exerc* 52(7): 1585, 2020

155. Aasvold LO, Ettema G, Skovereng K. Joint specific power production in cycling: The effect of cadence and intensity. *PLoS One* 14(2): e0212781, 2019
156. Lucía A, Hoyos J, Chicharro JL. Preferred pedalling cadence in professional cycling. *Med Sci Sports Exerc* 33(8): 1361, 2001
157. Damm P, Dymke J, Bender A, Duda G, Bergmann G. In vivo hip joint loads and pedal forces during ergometer cycling. *J Biomech* 60: 197, 2017
158. Mornieux G, Guenette JA, Sheel AW, Sanderson DJ. Influence of cadence, power output and hypoxia on the joint moment distribution during cycling. *Eur J Appl Physiol* 102(1): 11, 2007
159. Bini RR, Hume PA. Relationship between pedal force asymmetry and performance in cycling time trial. *J Sports Med Phys Fitness* 55(9): 892, 2015
160. Gale T, Anderst W. Asymmetry in healthy adult knee kinematics revealed through biplane radiography of the full gait cycle. *J Orthop Res* 37(3): 609, 2019
161. Hunt MA, Sanderson DJ, Moffet H, Inglis JT. Interlimb asymmetry in persons with and without an anterior cruciate ligament deficiency during stationary cycling. *Arch Phys Med Rehabil* 85(9): 1475, 2004
162. Smak W, Neptune RR, Hull ML. The influence of pedaling rate on bilateral asymmetry in cycling. *J Biomech* 32(9): 899, 1999
163. Yum H, Kim H, Lee T, Park MS, Lee SY. Cycling kinematics in healthy adults for musculoskeletal rehabilitation guidance. *BMC Musculoskelet Disord* 22(1): 1044, 2021
164. Ericson MO. Mechanical muscular power output and work during ergometer cycling at different work loads and speeds. *Eur J Appl Physiol Occup Physiol* 57(4): 382, 1988

165. Ericson MO, Bratt A, Nisell R, Németh G, Ekholm J. Load moments about the hip and knee joints during ergometer cycling. *Scand J Rehabil Med* 18(4): 165, 1986
166. Shen G, Zhang S, Bennett HJ, Martin JC, Crouter SE, Fitzhugh EC. Effects of Knee Alignments and Toe Clip on Frontal Plane Knee Biomechanics in Cycling. *J Sports Sci Med* 17(2): 312, 2018
167. Carpes FP, Rossato M, Faria IE, Bolli Mota C. Bilateral pedaling asymmetry during a simulated 40-km cycling time-trial. *J Sports Med Phys Fitness* 47(1): 51, 2007
168. Bini RR, Hume PA. Assessment of bilateral asymmetry in cycling using a commercial instrumented crank system and instrumented pedals. *Int J Sports Physiol Perform* 9(5): 876, 2014
169. Hansen EA, Jørgensen LV, Jensen K, Fregly BJ, Sjøgaard G. Crank inertial load affects freely chosen pedal rate during cycling. *J Biomech* 35(2): 277, 2002
170. Fregly BJ, Zajac FE, Dairaghi CA. Crank inertial load has little effect on steady-state pedaling coordination. *J Biomech* 29(12): 1559, 1996
171. Edwards LM, Jobson SA, George SR, Day SH, Nevill AM. The effect of crank inertial load on the physiological and biomechanical responses of trained cyclists. *J Sports Sci* 25(11): 1195, 2007
172. Gatti AA, Keir PJ, Noseworthy MD, Beauchamp MK, Maly MR. Equations to Prescribe Bicycle Saddle Height based on Desired Joint Kinematics and Bicycle Geometry. *Eur J Sport Sci* 22(3): 344, 2022
173. Barratt PR, Korff T, Elmer SJ, Martin JC. Effect of crank length on joint-specific power during maximal cycling. *Med Sci Sports Exerc* 43(9): 1689, 2011

174. Ferrer-Roca V, Rivero-Palomo V, Ogueta-Alday A, Rodríguez-Marroyo JA, García-López J. Acute effects of small changes in crank length on gross efficiency and pedalling technique during submaximal cycling. *J Sports Sci* 35(14): 1328, 2017
175. Martin JC, Spirduso WW. Determinants of maximal cycling power: crank length, pedaling rate and pedal speed. *Eur J Appl Physiol* 84(5): 413, 2001
176. Too D, Landwer GE. The effect of pedal crank arm length on joint angle and power production in upright cycle ergometry. *J Sports Sci* 18(3): 153, 2000
177. Kabiri S, Halabchi F, Angoorani H, Yekaninejad S. Comparison of three modes of aerobic exercise combined with resistance training on the pain and function of patients with knee osteoarthritis: A randomized controlled trial. *Phys Ther Sport* 32: 22, 2018
178. Buddhadev HH, Crisafulli DL, Suprak DN, San Juan JG. Individuals With Knee Osteoarthritis Demonstrate Interlimb Asymmetry in Pedaling Power During Stationary Cycling. *J Appl Biomech* 34(4): 306, 2018
179. Gardner JK, Klipple G, Stewart C, Asif I, Zhang S. Acute effects of lateral shoe wedges on joint biomechanics of patients with medial compartment knee osteoarthritis during stationary cycling. *J Biomech* 49(13): 2817, 2016
180. Thompson RL, Gardner JK, Zhang S, Reinbolt JA. Lower-limb joint reaction forces and moments during modified cycling in healthy controls and individuals with knee osteoarthritis. *Clin Biomech (Bristol, Avon)* 71: 167, 2020
181. Sanzo P, Niccoli S, Droll K, Puskas D, Cullinan C, Lees SJ. The effects of exercise and active assisted cycle ergometry in post-operative total knee arthroplasty patients - a randomized controlled trial. *J Exp Orthop* 8(1): 41, 2021

182. D'Lima DD, Steklov N, Patil S, Colwell CW, Jr. The Mark Coventry Award: in vivo knee forces during recreation and exercise after knee arthroplasty. *Clin Orthop Relat Res* 466(11): 2605, 2008
183. Richards RE, van den Noort JC, van der Esch M, Booij MJ, Harlaar J. Effect of real-time biofeedback on peak knee adduction moment in patients with medial knee osteoarthritis: Is direct feedback effective? *Clin Biomech (Bristol, Avon)* 57: 150, 2018
184. Li K, Ackland DC, McClelland JA, Webster KE, Feller JA, de Steiger R, Pandy MG. Trunk muscle action compensates for reduced quadriceps force during walking after total knee arthroplasty. *Gait Posture* 38(1): 79, 2013
185. Thorsen T, Wen C, Porter J, Reinbolt JA, Weinhandl JT, Zhang S. Do Interlimb Knee Joint Loading Asymmetries Persist throughout Stance during Uphill Walking Following Total Knee Arthroplasty? *Int J Environ Res Public Health* 20(14), 2023
186. Kutzner I, Bender A, Dymke J, Duda G, von Roth P, Bergmann G. Mediolateral force distribution at the knee joint shifts across activities and is driven by tibiofemoral alignment. *Bone Joint J* 99-b(6): 779, 2017
187. Gromov K, Korchi M, Thomsen MG, Husted H, Troelsen A. What is the optimal alignment of the tibial and femoral components in knee arthroplasty? *Acta Orthop* 85(5): 480, 2014
188. Ritter MA, Davis KE, Meding JB, Pierson JL, Berend ME, Malinzak RA. The effect of alignment and BMI on failure of total knee replacement. *J Bone Joint Surg Am* 93(17): 1588, 2011

189. Collier MB, Engh CA, Jr., McAuley JP, Engh GA. Factors associated with the loss of thickness of polyethylene tibial bearings after knee arthroplasty. *J Bone Joint Surg Am* 89(6): 1306, 2007
190. Srivastava A, Lee GY, Steklov N, Colwell CW, Jr., Ezzet KA, D'Lima DD. Effect of tibial component varus on wear in total knee arthroplasty. *Knee* 19(5): 560, 2012
191. Halder A, Kutzner I, Graichen F, Heinlein B, Beier A, Bergmann G. Influence of limb alignment on mediolateral loading in total knee replacement: in vivo measurements in five patients. *J Bone Joint Surg Am* 94(11): 1023, 2012
192. Messier SP, Pater M, Beavers DP, Legault C, Loeser RF, Hunter DJ, DeVita P. Influences of alignment and obesity on knee joint loading in osteoarthritic gait. *Osteoarthritis Cartilage* 22(7): 912, 2014
193. Andriacchi TP, Stanwyck TS, Galante JO. Knee biomechanics and total knee replacement. *J Arthroplasty* 1(3): 211, 1986
194. Engelhardt L, Melzner M, Havelkova L, Fiala P, Christen P, Dendorfer S, Simon U. A new musculoskeletal AnyBody™ detailed hand model. *Comput Methods Biomech Biomed Engin*: 1, 2020
195. Koblauch H. Low back load in airport baggage handlers. *Dan Med J* 63(4), 2016
196. Zhang L, Liu Y, Wang R, Smith C, Gutierrez-Farewik EM. Modeling and Simulation of a Human Knee Exoskeleton's Assistive Strategies and Interaction. *Front Neurorobot* 15: 620928, 2021

197. Delp SL, Anderson FC, Arnold AS, Loan P, Habib A, John CT, Guendelman E, Thelen DG. OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE Trans Biomed Eng* 54(11): 1940, 2007
198. Delp SL, Loan JP. A graphics-based software system to develop and analyze models of musculoskeletal structures. *Comput Biol Med* 25(1): 21, 1995
199. Raveendranathan V, Kooiman VGM, Carloni R. Musculoskeletal model of osseointegrated transfemoral amputees in OpenSim. *PLoS One* 18(9): e0288864, 2023
200. Barrett JM, McKinnon CD, Dickerson CR, Callaghan JP. An Electromyographically Driven Cervical Spine Model in OpenSim. *J Appl Biomech* 37(5): 481, 2021
201. Kholinne E, Zulkarnain RF, Sun YC, Lim S, Chun JM, Jeon IH. The different role of each head of the triceps brachii muscle in elbow extension. *Acta Orthop Traumatol Turc* 52(3): 201, 2018
202. Bekoff A, Betz W. Properties of isolated adult rat muscle fibres maintained in tissue culture. *J Physiol* 271(2): 537, 1977
203. Farina D, Gandevia S. The neural control of movement: a century of in vivo motor unit recordings is the legacy of Adrian and Bronk. *J Physiol* 602(2): 281, 2024
204. Kazamel M, Warren PP. History of electromyography and nerve conduction studies: A tribute to the founding fathers. *J Clin Neurosci* 43: 54, 2017
205. De Groote F, Kinney AL, Rao AV, Fregly BJ. Evaluation of Direct Collocation Optimal Control Problem Formulations for Solving the Muscle Redundancy Problem. *Ann Biomed Eng* 44(10): 2922, 2016

206. Seth A, Hicks JL, Uchida TK, Habib A, Dembia CL, Dunne JJ, Ong CF, DeMers MS, Rajagopal A, Millard M, Hamner SR, Arnold EM, Yong JR, Lakshmikanth SK, Sherman MA, Ku JP, Delp SL. OpenSim: Simulating musculoskeletal dynamics and neuromuscular control to study human and animal movement. *PLoS Comput Biol* 14(7): e1006223, 2018
207. Crowninshield RD, Brand RA. A physiologically based criterion of muscle force prediction in locomotion. *J Biomech* 14(11): 793, 1981
208. Ackermann M, van den Bogert AJ. Optimality principles for model-based prediction of human gait. *J Biomech* 43(6): 1055, 2010
209. Hyodo K, Kanamori A, Kadone H, Takahashi T, Kajiwarra M, Yamazaki M. Gait Analysis Comparing Kinematic, Kinetic, and Muscle Activation Data of Modern and Conventional Total Knee Arthroplasty. *Arthroplast Today* 6(3): 338, 2020
210. Hill AV. First and last experiments in muscle mechanics: Cambridge University Press, 1970
211. Zajac FE. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. *Crit Rev Biomed Eng* 17(4): 359, 1989
212. Winters JM. An improved muscle-reflex actuator for use in large-scale neuro-musculoskeletal models. *Ann Biomed Eng* 23(4): 359, 1995
213. Anderson FC, Pandy MG. Static and dynamic optimization solutions for gait are practically equivalent. *J Biomech* 34(2): 153, 2001
214. Morrow MM, Rankin JW, Neptune RR, Kaufman KR. A comparison of static and dynamic optimization muscle force predictions during wheelchair propulsion. *J Biomech* 47(14): 3459, 2014

215. Hosseini Nasab SH, Smith CR, Maas A, Vollenweider A, Dymke J, Schütz P, Damm P, Trepczynski A, Taylor WR. Uncertainty in Muscle-Tendon Parameters can Greatly Influence the Accuracy of Knee Contact Force Estimates of Musculoskeletal Models. *Front Bioeng Biotechnol* 10: 808027, 2022
216. Lieber RL, Fridén J. Functional and clinical significance of skeletal muscle architecture. *Muscle Nerve* 23(11): 1647, 2000
217. Kawakami Y, Ichinose Y, Fukunaga T. Architectural and functional features of human triceps surae muscles during contraction. *J Appl Physiol* (1985) 85(2): 398, 1998
218. Fukunaga T, Ichinose Y, Ito M, Kawakami Y, Fukashiro S. Determination of fascicle length and pennation in a contracting human muscle in vivo. *J Appl Physiol* (1985) 82(1): 354, 1997
219. Kawakami Y, Abe T, Fukunaga T. Muscle-fiber pennation angles are greater in hypertrophied than in normal muscles. *J Appl Physiol* (1985) 74(6): 2740, 1993
220. Sopher RS, Amis AA, Davies DC, Jeffers JR. The influence of muscle pennation angle and cross-sectional area on contact forces in the ankle joint. *J Strain Anal Eng Des* 52(1): 12, 2017
221. Navacchia A, Kefala V, Shelburne KB. Dependence of Muscle Moment Arms on In Vivo Three-Dimensional Kinematics of the Knee. *Ann Biomed Eng* 45(3): 789, 2017
222. Engelina S, Robertson CJ, Moggridge J, Killingback A, Adds P. Using ultrasound to measure the fibre angle of vastus medialis oblique: A cadaveric validation study. *The Knee* 21(1): 107, 2014
223. Chen Z, Franklin DW. Musculotendon Parameters in Lower Limb Models: Simplifications, Uncertainties, and Muscle Force Estimation Sensitivity. *Ann Biomed Eng* 51(6): 1147, 2023

224. Hessel AL, Lindstedt SL, Nishikawa KC. Physiological Mechanisms of Eccentric Contraction and Its Applications: A Role for the Giant Titin Protein. *Front Physiol* 8: 70, 2017
225. Fitts RH, McDonald KS, Schluter JM. The determinants of skeletal muscle force and power: their adaptability with changes in activity pattern. *J Biomech* 24 Suppl 1: 111, 1991
226. Grosicki GJ, Zepeda CS, Sundberg CW. Single muscle fibre contractile function with ageing. *J Physiol* 600(23): 5005, 2022
227. Carbone V, van der Krogt MM, Koopman H, Verdonschot N. Sensitivity of subject-specific models to Hill muscle-tendon model parameters in simulations of gait. *J Biomech* 49(9): 1953, 2016
228. Serrancolí G, Kinney AL, Fregly BJ. Influence of musculoskeletal model parameter values on prediction of accurate knee contact forces during walking. *Med Eng Phys* 85: 35, 2020
229. Raikova RT, Prilutsky BI. Sensitivity of predicted muscle forces to parameters of the optimization-based human leg model revealed by analytical and numerical analyses. *J Biomech* 34(10): 1243, 2001
230. Scovil CY, Ronsky JL. Sensitivity of a Hill-based muscle model to perturbations in model parameters. *J Biomech* 39(11): 2055, 2006
231. Redl C, Gfoehler M, Pandy MG. Sensitivity of muscle force estimates to variations in muscle-tendon properties. *Hum Mov Sci* 26(2): 306, 2007
232. Xiao M, Higginson J. Sensitivity of estimated muscle force in forward simulation of normal walking. *J Appl Biomech* 26(2): 142, 2010

233. Bosmans L, Valente G, Wesseling M, Van Campen A, De Groote F, De Schutter J, Jonkers I. Sensitivity of predicted muscle forces during gait to anatomical variability in musculotendon geometry. *J Biomech* 48(10): 2116, 2015
234. Rajagopal A, Dembia CL, DeMers MS, Delp DD, Hicks JL, Delp SL. Full-Body Musculoskeletal Model for Muscle-Driven Simulation of Human Gait. *IEEE Trans Biomed Eng* 63(10): 2068, 2016
235. Lloyd DG, Buchanan TS. Strategies of muscular support of varus and valgus isometric loads at the human knee. *J Biomech* 34(10): 1257, 2001
236. Blache Y, Michaud B, Rogowski I, Monteil K, Begon M. Sensitivity of Shoulder Musculoskeletal Model Predictions to Muscle-Tendon Properties. *IEEE Trans Biomed Eng* 66(5): 1309, 2019
237. Mizner RL, Petterson SC, Snyder-Mackler L. Quadriceps strength and the time course of functional recovery after total knee arthroplasty. *J Orthop Sports Phys Ther* 35(7): 424, 2005
238. Powell PL, Roy RR, Kanim P, Bello MA, Edgerton VR. Predictability of skeletal muscle tension from architectural determinations in guinea pig hindlimbs. *J Appl Physiol Respir Environ Exerc Physiol* 57(6): 1715, 1984
239. Bolsterlee B, Vardy AN, van der Helm FC, Veeger HE. The effect of scaling physiological cross-sectional area on musculoskeletal model predictions. *J Biomech* 48(10): 1760, 2015
240. O'Brien TD, Reeves ND, Baltzopoulos V, Jones DA, Maganaris CN. In vivo measurements of muscle specific tension in adults and children. *Exp Physiol* 95(1): 202, 2010
241. Buchanan TS. Evidence that maximum muscle stress is not a constant: differences in specific tension in elbow flexors and extensors. *Med Eng Phys* 17(7): 529, 1995

242. Fukunaga T, Roy RR, Shellock FG, Hodgson JA, Edgerton VR. Specific tension of human plantar flexors and dorsiflexors. *J Appl Physiol* (1985) 80(1): 158, 1996
243. Javidi M, McGowan CP, Lin DC. The Contributions of Individual Muscle-Tendon Units to the Plantarflexor Group Force-Length Properties. *Ann Biomed Eng* 47(11): 2168, 2019
244. Kawakami Y, Nakazawa K, Fujimoto T, Nozaki D, Miyashita M, Fukunaga T. Specific tension of elbow flexor and extensor muscles based on magnetic resonance imaging. *Eur J Appl Physiol Occup Physiol* 68(2): 139, 1994
245. Winters JM. Multiple Muscle Systems: Biomechanics and Movement Organization: Springer, 1990
246. van Ruijven LJ, Weijs WA. A new model for calculating muscle forces from electromyograms. *Eur J Appl Physiol Occup Physiol* 61(5-6): 479, 1990
247. Guimaraes AC, Herzog W, Allinger TL, Zhang YT. The EMG-force relationship of the cat soleus muscle and its association with contractile conditions during locomotion. *J Exp Biol* 198(Pt 4): 975, 1995
248. Roelker SA, Caruthers EJ, Baker RK, Pelz NC, Chaudhari AMW, Siston RA. Interpreting Musculoskeletal Models and Dynamic Simulations: Causes and Effects of Differences Between Models. *Ann Biomed Eng* 45(11): 2635, 2017
249. Arnold EM, Ward SR, Lieber RL, Delp SL. A model of the lower limb for analysis of human movement. *Ann Biomed Eng* 38(2): 269, 2010
250. Roelker SA, Caruthers EJ, Hall RK, Pelz NC, Chaudhari AMW, Siston RA. Effects of Optimization Technique on Simulated Muscle Activations and Forces. *J Appl Biomech* 36(4): 259, 2020

251. Pelegrinelli ARM, Catelli DS, Kowalski E, Lamontagne M, Moura FA. Comparing three generic musculoskeletal models to estimate the tibiofemoral reaction forces during gait and sit-to-stand tasks. *Med Eng Phys* 122: 104074, 2023
252. Sherman MA, Seth A, Delp SL. WHAT IS A MOMENT ARM? CALCULATING MUSCLE EFFECTIVENESS IN BIOMECHANICAL MODELS USING GENERALIZED COORDINATES. *Proc ASME Des Eng Tech Conf* 2013, 2013
253. Garner BA, Pandy MG. The Obstacle-Set Method for Representing Muscle Paths in Musculoskeletal Models. *Comput Methods Biomech Biomed Engin* 3(1): 1, 2000
254. Mokhtarzadeh H, Perraton L, Fok L, Muñoz MA, Clark R, Pivonka P, Bryant AL. A comparison of optimisation methods and knee joint degrees of freedom on muscle force predictions during single-leg hop landings. *J Biomech* 47(12): 2863, 2014
255. Myers CA, Torry MR, Shelburne KB, Giphart JE, LaPrade RF, Woo SL, Steadman JR. In vivo tibiofemoral kinematics during 4 functional tasks of increasing demand using biplane fluoroscopy. *Am J Sports Med* 40(1): 170, 2012
256. Lafortune MA, Cavanagh PR, Sommer HJ, 3rd, Kalenak A. Three-dimensional kinematics of the human knee during walking. *J Biomech* 25(4): 347, 1992
257. Marieswaran M, Sikidar A, Goel A, Joshi D, Kalyanasundaram D. An extended OpenSim knee model for analysis of strains of connective tissues. *Biomed Eng Online* 17(1): 42, 2018
258. Farrokhi S, Voycheck CA, Klatt BA, Gustafson JA, Tashman S, Fitzgerald GK. Altered tibiofemoral joint contact mechanics and kinematics in patients with knee osteoarthritis and episodic complaints of joint instability. *Clinical Biomechanics* 29(6): 629, 2014

259. Liu F, Kozanek M, Hosseini A, Van de Velde SK, Gill TJ, Rubash HE, Li G. In vivo tibiofemoral cartilage deformation during the stance phase of gait. *Journal of Biomechanics* 43(4): 658, 2010
260. Marouane H, Shirazi-Adl A, Adouni M. Alterations in knee contact forces and centers in stance phase of gait: A detailed lower extremity musculoskeletal model. *Journal of Biomechanics* 49(2): 185, 2016
261. Meslier QA, Shefelbine SJ. Using Finite Element Modeling in Bone Mechanoadaptation. *Curr Osteoporos Rep* 21(2): 105, 2023
262. Xu H, Bloswick D, Merryweather A. An improved OpenSim gait model with multiple degrees of freedom knee joint and knee ligaments. *Comput Methods Biomech Biomed Engin* 18(11): 1217, 2015
263. Ozada N. Effect of six degrees of freedom knee kinematics on ligament length and moment arm in an intact knee model. *Technol Health Care* 23(4): 485, 2015
264. Shelburne KB, Pandy MG, Anderson FC, Torry MR. Pattern of anterior cruciate ligament force in normal walking. *J Biomech* 37(6): 797, 2004
265. Blemker SS, Delp SL. Three-dimensional representation of complex muscle architectures and geometries. *Ann Biomed Eng* 33(5): 661, 2005
266. Wheatley BB, Chacras NA, Seeley MA. Patellofemoral joint load and knee abduction/adduction moment are sensitive to variations in femoral version and individual muscle forces. *J Orthop Res* 41(3): 570, 2023
267. Koller W, Baca A, Kainz H. Impact of scaling errors of the thigh and shank segments on musculoskeletal simulation results. *Gait Posture* 87: 65, 2021

268. Schmid J, Sandholm A, Chung F, Thalmann D, Delingette H, Magnenat-Thalmann N. Musculoskeletal Simulation Model Generation from MRI Data Sets and Motion Capture Data. In: Magnenat-Thalmann N, Zhang JJ, Feng DD, eds. Recent Advances in the 3D Physiological Human. London: Springer London. 3. 2009
269. Smale KB, Conconi M, Sancisi N, Krogsgaard M, Alkjaer T, Parenti-Castelli V, Benoit DL. Effect of implementing magnetic resonance imaging for patient-specific OpenSim models on lower-body kinematics and knee ligament lengths. *J Biomech* 83: 9, 2019
270. Fernandez J, Shim V, Schneider M, Choisne J, Handsfield G, Yeung T, Zhang J, Hunter P, Besier T. A Narrative Review of Personalized Musculoskeletal Modeling Using the Physiome and Musculoskeletal Atlas Projects. *J Appl Biomech* 39(5): 304, 2023
271. Uchida TK, Seth A. Conclusion or Illusion: Quantifying Uncertainty in Inverse Analyses From Marker-Based Motion Capture due to Errors in Marker Registration and Model Scaling. *Front Bioeng Biotechnol* 10: 874725, 2022
272. Song K, Gaffney BMM, Shelburne KB, Pascual-Garrido C, Clohisy JC, Harris MD. Dysplastic hip anatomy alters muscle moment arm lengths, lines of action, and contributions to joint reaction forces during gait. *J Biomech* 110: 109968, 2020
273. Fiorentino NM, Kutschke MJ, Atkins PR, Foreman KB, Kapron AL, Anderson AE. Accuracy of Functional and Predictive Methods to Calculate the Hip Joint Center in Young Non-pathologic Asymptomatic Adults with Dual Fluoroscopy as a Reference Standard. *Ann Biomed Eng* 44(7): 2168, 2016

274. Harris MD, MacWilliams BA, Bo Foreman K, Peters CL, Weiss JA, Anderson AE. Higher medially-directed joint reaction forces are a characteristic of dysplastic hips: A comparative study using subject-specific musculoskeletal models. *J Biomech* 54: 80, 2017
275. Lu TW, O'Connor JJ. Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. *J Biomech* 32(2): 129, 1999
276. Dunne JJ, Uchida TK, Besier TF, Delp SL, Seth A. A marker registration method to improve joint angles computed by constrained inverse kinematics. *PLoS One* 16(5): e0252425, 2021
277. Steele KM, Demers MS, Schwartz MH, Delp SL. Compressive tibiofemoral force during crouch gait. *Gait Posture* 35(4): 556, 2012
278. Davico G, Lloyd DG, Carty CP, Killen BA, Devaprakash D, Pizzolato C. Multi-level personalization of neuromusculoskeletal models to estimate physiologically plausible knee joint contact forces in children. *Biomech Model Mechanobiol* 21(6): 1873, 2022
279. Gerus P, Sartori M, Besier TF, Fregly BJ, Delp SL, Banks SA, Pandy MG, D'Lima DD, Lloyd DG. Subject-specific knee joint geometry improves predictions of medial tibiofemoral contact forces. *J Biomech* 46(16): 2778, 2013
280. Correa TA, Baker R, Graham HK, Pandy MG. Accuracy of generic musculoskeletal models in predicting the functional roles of muscles in human gait. *J Biomech* 44(11): 2096, 2011
281. Scheys L, Van Campenhout A, Spaepen A, Suetens P, Jonkers I. Personalized MR-based musculoskeletal models compared to rescaled generic models in the presence of increased femoral anteversion: effect on hip moment arm lengths. *Gait Posture* 28(3): 358, 2008

282. Bedo BLS, Catelli DS, Lamontagne M, Santiago PRP. A custom musculoskeletal model for estimation of medial and lateral tibiofemoral contact forces during tasks with high knee and hip flexions. *Comput Methods Biomech Biomed Engin* 23(10): 658, 2020
283. Catelli DS, Wesseling M, Jonkers I, Lamontagne M. A musculoskeletal model customized for squatting task. *Comput Methods Biomech Biomed Engin* 22(1): 21, 2019
284. Michaud B, Begon M. Two efficient static optimization algorithms that account for muscle-tendon equilibrium: approaching the constraint Jacobian via a constant or a cubic spline function. *Comput Methods Biomech Biomed Engin*: 1, 2020
285. Winter DA, Sidwall HG, Hobson DA. Measurement and reduction of noise in kinematics of locomotion. *J Biomech* 7(2): 157, 1974
286. Ladin Z, Flowers WC, Messner W. A quantitative comparison of a position measurement system and accelerometry. *J Biomech* 22(4): 295, 1989
287. Anderson FC, Pandy MG. Dynamic optimization of human walking. *J Biomech Eng* 123(5): 381, 2001
288. Lloyd DG, Besier TF. An EMG-driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo. *J Biomech* 36(6): 765, 2003
289. Meyer AJ, D'Lima DD, Besier TF, Lloyd DG, Colwell CW, Jr., Fregly BJ. Are external knee load and EMG measures accurate indicators of internal knee contact forces during gait? *J Orthop Res* 31(6): 921, 2013
290. Pizzolato C, Lloyd DG, Sartori M, Ceseracciu E, Besier TF, Fregly BJ, Reggiani M. CEINMS: A toolbox to investigate the influence of different neural control solutions on the

- prediction of muscle excitation and joint moments during dynamic motor tasks. *J Biomech* 48(14): 3929, 2015
291. Romanato M, Volpe D, Guiotto A, Spolaor F, Sartori M, Sawacha Z. Electromyography-informed modeling for estimating muscle activation and force alterations in Parkinson's disease. *Comput Methods Biomech Biomed Engin* 25(1): 14, 2022
292. Davico G, Pizzolato C, Lloyd DG, Obst SJ, Walsh HPJ, Carty CP. Increasing level of neuromusculoskeletal model personalisation to investigate joint contact forces in cerebral palsy: A twin case study. *Clin Biomech (Bristol, Avon)* 72: 141, 2020
293. Hoang HX, Diamond LE, Lloyd DG, Pizzolato C. A calibrated EMG-informed neuromusculoskeletal model can appropriately account for muscle co-contraction in the estimation of hip joint contact forces in people with hip osteoarthritis. *J Biomech* 83: 134, 2019
294. Hicks JL, Uchida TK, Seth A, Rajagopal A, Delp SL. Is my model good enough? Best practices for verification and validation of musculoskeletal models and simulations of movement. *J Biomech Eng* 137(2): 020905, 2015
295. Hamner SR, Seth A, Delp SL. Muscle contributions to propulsion and support during running. *J Biomech* 43(14): 2709, 2010
296. Lin YC, Dorn TW, Schache AG, Pandy MG. Comparison of different methods for estimating muscle forces in human movement. *Proc Inst Mech Eng H* 226(2): 103, 2012
297. Anderson FC, Pandy MG. A Dynamic Optimization Solution for Vertical Jumping in Three Dimensions. *Comput Methods Biomech Biomed Engin* 2(3): 201, 1999
298. Alvim FC, Lucareli PRG, Menegaldo LL. Predicting muscle forces during the propulsion phase of single leg triple hop test. *Gait Posture* 59: 298, 2018

299. Karabulut D, Dogru SC, Lin YC, Pandy MG, Herzog W, Arslan YZ. Direct Validation of Model-Predicted Muscle Forces in the Cat Hindlimb During Locomotion. *J Biomech Eng* 142(5), 2020
300. Lai AKM, Arnold AS, Wakeling JM. Why are Antagonist Muscles Co-activated in My Simulation? A Musculoskeletal Model for Analysing Human Locomotor Tasks. *Ann Biomed Eng* 45(12): 2762, 2017
301. Curreli C, Di Puccio F, Davico G, Modenese L, Viceconti M. Using Musculoskeletal Models to Estimate in vivo Total Knee Replacement Kinematics and Loads: Effect of Differences Between Models. *Front Bioeng Biotechnol* 9: 703508, 2021
302. Fregly BJ, Besier TF, Lloyd DG, Delp SL, Banks SA, Pandy MG, D'Lima DD. Grand challenge competition to predict in vivo knee loads. *J Orthop Res* 30(4): 503, 2012
303. Saliba CM, Brandon SCE, Deluzio KJ. Sensitivity of medial and lateral knee contact force predictions to frontal plane alignment and contact locations. *J Biomech* 57: 125, 2017
304. Knarr BA, Higginson JS. Practical approach to subject-specific estimation of knee joint contact force. *J Biomech* 48(11): 2897, 2015
305. Brandon SCE, Miller RH, Thelen DG, Deluzio KJ. Selective lateral muscle activation in moderate medial knee osteoarthritis subjects does not unload medial knee condyle. *Journal of Biomechanics* 47(6): 1409, 2014
306. Febrer-Nafria M, Dreyer MJ, Maas A, Taylor WR, Smith CR, Hosseini Nasab SH. Knee kinematics are primarily determined by implant alignment but knee kinetics are mainly influenced by muscle coordination strategy. *J Biomech* 161: 111851, 2023

307. Du M, Sun J, Liu Y, Wang Y, Yan S, Zeng J, Zhang K. Tibio-Femoral Contact Force Distribution of Knee Before and After Total Knee Arthroplasty: Combined Finite Element and Gait Analysis. *Orthop Surg* 14(8): 1836, 2022
308. Richards C, Higginson JS. Knee contact force in subjects with symmetrical OA grades: differences between OA severities. *J Biomech* 43(13): 2595, 2010
309. Thorsen T, Hummer E, Reinbolt J, Weinhandl JT, Zhang S. Increased Q-factor increases medial compartment knee joint contact force during cycling. *J Biomech* 118: 110271, 2021
310. Dembia CL, Bianco NA, Falisse A, Hicks JL, Delp SL. OpenSim Moco: Musculoskeletal optimal control. *PLoS Comput Biol* 16(12): e1008493, 2020
311. Demers MS, Pal S, Delp SL. Changes in tibiofemoral forces due to variations in muscle activity during walking. *J Orthop Res* 32(6): 769, 2014
312. Clancy CE, Gatti AA, Ong CF, Maly MR, Delp SL. Muscle-driven simulations and experimental data of cycling. *Sci Rep* 13(1): 21534, 2023
313. Uhlich SD, Jackson RW, Seth A, Kolesar JA, Delp SL. Muscle coordination retraining inspired by musculoskeletal simulations reduces knee contact force. *Sci Rep* 12(1): 9842, 2022
314. Ryan MM, Gregor RJ. EMG profiles of lower extremity muscles during cycling at constant workload and cadence. *J Electromyogr Kinesiol* 2(2): 69, 1992
315. Cohen J. *Statistical Power Analysis for the Behavioral Sciences*. New York: Routledge, 1988
316. Lakens D. Calculating and reporting effect sizes to facilitate cumulative science: a practical primer for t-tests and ANOVAs. *Front Psychol* 4: 863, 2013

317. Köglberger P, Wurm A, Coraça-Huber D, Krismer M, Oberaigner W, Liebensteiner M. High range of motion in the first ten postoperative days after TKA does not predict superior outcome in the long run. *Arch Orthop Trauma Surg* 142(10): 2497, 2022
318. Wen C, Cates HE, Zhang S. Is knee biomechanics different in uphill walking on different slopes for older adults with total knee replacement? *J Biomech* 89: 40, 2019
319. Zhang S, Brown S, Menke W, Ovispo-Martinez A, Lu TW. Effects of Unilateral Crankarm Length Adjustments on Hip and Ankle Kinematics during Stationary Cycling. In: American College of Sports Medicine Annual Meeting. Atlanta, Georgia, USA. 2025
320. Gaffney BM, Harris MD, Davidson BS, Stevens-Lapsley JE, Christiansen CL, Shelburne KB. Multi-Joint Compensatory Effects of Unilateral Total Knee Arthroplasty During High-Demand Tasks. *Ann Biomed Eng* 44(8): 2529, 2016
321. Kline PW, Jacobs CA, Duncan ST, Noehren B. Step descent strategy is altered bilaterally despite unilateral muscle strength impairment after total knee arthroplasty. *Knee Surg Sports Traumatol Arthrosc* 28(5): 1508, 2020
322. Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J Biomech Eng* 105(2): 136, 1983
323. Menke W. Biomechanics Toolbox. In. 2024
324. Yocum D, Ovispo-Martinez A, Valenzuela KA, Wen C, Cates H, Zhang S. Investigation of Biomechanical Differences in Level Walking between Patients with Bilateral and Unilateral Total Knee Replacements. *Bioengineering (Basel)* 11(8), 2024
325. Luan L, Bousie J, Pranata A, Adams R, Han J. Stationary cycling exercise for knee osteoarthritis: A systematic review and meta-analysis. *Clin Rehabil* 35(4): 522, 2021

326. Sayeed SA, Sayeed YA, Barnes SA, Pagnano MW, Trousdale RT. The risk of subsequent joint arthroplasty after primary unilateral total knee arthroplasty, a 10-year study. *J Arthroplasty* 26(6): 842, 2011
327. Okamoto T, Yamanashi Y, Ikemoto T, Miyagawa H, Ishida T, Akao M, Takata T, Kato T, Kobayakawa K, Deie M. Risk factors for contralateral total knee arthroplasty after unilateral total knee arthroplasty. *J Orthop Sci* 28(6): 1311, 2023
328. Aljehani M, Madara K, Snyder-Mackler L, Christiansen C, Zeni JA, Jr. The contralateral knee may not be a valid control for biomechanical outcomes after unilateral total knee arthroplasty. *Gait Posture* 70: 179, 2019
329. Thorsen T, Strohacker K, Weinhandl JT, Zhang S. Increased Q-Factor increases frontal-plane knee joint loading in stationary cycling. *J Sport Health Sci* 9(3): 258, 2020
330. Zhang SaB, S. Impacts of asymmetric crankarm lengths on knee biomechanics during cycling. In: 2024 Annual Meeting of American Society of Biomechanics. Madison, WI. 2024
331. P. BGaD. Orthoload. In. Berlin Institute of Health at Charité Universitätsmedizin Berlin. 2008
332. Bennett HJ, Weinhandl JT, Fleenor K, Zhang S. Frontal Plane Tibiofemoral Alignment is Strongly Related to Compartmental Knee Joint Contact Forces and Muscle Control Strategies During Stair Ascent. *J Biomech Eng* 140(6), 2018
333. Erdemir A, McLean S, Herzog W, van den Bogert AJ. Model-based estimation of muscle forces exerted during movements. *Clin Biomech (Bristol)* 22(2): 131, 2007

334. Zeni JA, Rudolph K, Higginson JS. Alterations in quadriceps and hamstrings coordination in persons with medial compartment knee osteoarthritis. *J Electromyogr Kinesiol* 20(1): 148, 2010

Appendices

Appendix A: Tables

Table 1: Participant Demographics Information (mean $\pm$ std)

Variable	All Post	Unilateral	Bilateral
Age (years)	67 $\pm$ 7	65 $\pm$ 8	69 $\pm$ 3
Height (cm)	172 $\pm$ 14	160 $\pm$ 5	190 $\pm$ 5
Mass (kg)	94.6 $\pm$ 23.6	78.4 $\pm$ 12.9	116.1 $\pm$ 14.4
BMI (kg/m ²)	31.5 $\pm$ 4.6	30.2 $\pm$ 4.9	33.2 $\pm$ 4.5
Days from Most Recent Surgery	54 $\pm$ 4	55 $\pm$ 3	52 $\pm$ 3
Days from Contralateral Surgery	-	-	638 $\pm$ 402
KOOS Total (%)	65.0 $\pm$ 16.5	65.5 $\pm$ 19.6	64.3 $\pm$ 15.5
KOOS Quality of Life (%)	49.3 $\pm$ 25.7	45.5 $\pm$ 20.6	54.3 $\pm$ 35.6
KOOS Pain (%)	71.9 $\pm$ 11.9	70.3 $\pm$ 13.9	74.0 $\pm$ 11.1
KOOS Activities of Daily Living (%)	78.7 $\pm$ 12.3	80.0 $\pm$ 12.7	77.0 $\pm$ 14.2
KOOS Sports and Recreation (%)	60.0 $\pm$ 38.5	40.0 $\pm$ 18	60.0 $\pm$ 38.5
KOOS Symptoms and Stiffness (%)	72.5 $\pm$ 16.7	76.3 $\pm$ 2.3	72.5 $\pm$ 16.7

Table 2: Mean Results for Peak Pedal Reaction Force and Peak Knee Joint Angle, ROM and Kinetic Variables (mean $\pm$ std)

Variable	Limb	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)	Cond. p-value (η_p^2)	Limb p-value (η_p^2)	Inter. p-value (η_p^2)
PRFz (N)	Repl.	227.51 $\pm$ 47.82	225.22 $\pm$ 47.49	230.29 $\pm$ 42.00	216.99 $\pm$ 36.89	220.47 $\pm$ 31.77	0.192 (0.217)	0.170 (0.288)	0.043 (0.326)
	Contr.	238.41 $\pm$ 50.29	219.33 $\pm$ 42.51	250.96 $\pm$ 68.85	236.93 $\pm$ 55.21	254.55 $\pm$ 56.47			
Knee Extension ROM ($^\circ$) ^{a,b,c,α,β, γ, δ, λ}	Repl.	55.27 $\pm$ 8.12	58.39 $\pm$ 6.39	61.45 $\pm$ 6.83	57.75 $\pm$ 6.26	51.67 $\pm$ 6.30	< 0.000 (0.103)	0.436 (0.122)	0.090 (0.043)
	Contr.	60.5 $\pm$ 5.75	61.71 $\pm$ 4.20	64.1 $\pm$ 6.57	57.56 $\pm$ 4.73	53.58 $\pm$ 5.53			
Knee Extension Angle ($^\circ$) ^{b,c,α,β, γ, δ}	Repl.	-50.13 $\pm$ 8.96	-48.55 $\pm$ 6.43	-45.18 $\pm$ 7.04	-48.34 $\pm$ 7.51	-51.71 $\pm$ 6.99	< 0.000 (0.740)	0.062 (0.467)	0.017 (0.394)
	Contr.	-43.83 $\pm$ 8.53	-43.70 $\pm$ 7.71	-42.63 $\pm$ 9.56	-46.87 $\pm$ 9.20	-49.26 $\pm$ 9.14			
Knee Extension Moment (Nm)	Repl.	16.32 $\pm$ 6.13	16.02 $\pm$ 5.74 ⁰	17.19 $\pm$ 5.23	16.08 $\pm$ 7.60 ⁰	16.01 $\pm$ 3.84 ⁰	0.523 (0.121)	0.025 (0.595)	0.723 (0.079)
	Contr.	22.29 $\pm$ 9.89	21.55 $\pm$ 6.65	25.13 $\pm$ 11.81	22.53 $\pm$ 9.68	24.84 $\pm$ 9.52			
Knee Abduction Moment (Nm)	Repl.	-2.99 $\pm$ 3.73	-2.94 $\pm$ 3.79	-2.64 $\pm$ 3.33	-3.18 $\pm$ 3.55	-2.97 $\pm$ 4.11	0.496 (0.149)	0.470 (0.109)	0.572 (0.130)
	Contr.	-5.13 $\pm$ 4.73	-3.95 $\pm$ 6.42	-4.11 $\pm$ 5.46	-3.92 $\pm$ 4.78	-4.96 $\pm$ 6.46			
Knee Power (Nm/s) ^{α}	Repl.	58.86 $\pm$ 28.44	59.08 $\pm$ 20.76 ⁰	69.13 $\pm$ 17.62	51.25 $\pm$ 17.51 ⁰	54.80 $\pm$ 14.06	0.032 (0.345)	0.031 (0.569)	0.964 (0.023)
	Contr.	84.78 $\pm$ 35.88	83.94 $\pm$ 23.78	100.79 $\pm$ 45.52	80.43 $\pm$ 30.92	80.91 $\pm$ 27.86			

a: significant difference between C1 and C2, b: significant difference between C1 and C3, c: significant difference between C2 and C3,
 α : significant difference between C3 and C4, β : significant difference between C3 and C5, γ : significant difference between C4 and C5,
 δ : significant difference between C1 and C5, λ : significant difference between C2 and C4, θ : significant difference between Replaced and Contralateral at current condition
 η_p^2 : partial eta squared, Repl.: Replaced side, Contr.: Contralateral side, Cond.: Condition, Inter.: Interaction

Table 3: Mean Results for ROM and Peak Moment Variables for the Ankle and Hip (mean $\pm$ std)

Variable	Limb	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)	Cond. p-value (η_p^2)	Limb p-value (η_p^2)	Inter. p-value (η_p^2)
Ankle Plantarflexion ROM ($^\circ$) ^{δ, c, β}	Repl.	14.78 $\pm$ 8.95	15.81 $\pm$ 7.72	15.43 $\pm$ 6.78	14.29 $\pm$ 8.00	11.21 $\pm$ 4.83	0.003 (0.468)	0.733 (0.021)	0.136 (0.244)
	Contr.	11.89 $\pm$ 6.56	13.72 $\pm$ 3.39	18.60 $\pm$ 5.16	11.85 $\pm$ 2.32	9.74 $\pm$ 3.83			
Ankle Plantarflexion Moment (Nm)	Repl.	-17.35 $\pm$ 10.98	-17.93 $\pm$ 10.05	-18.88 $\pm$ 11.31	-17.5 $\pm$ 10.22	-17.33 $\pm$ 10.02	0.501 (0.154)	0.512 (0.081)	0.483 (0.012)
	Contr.	-19.51 $\pm$ 10.84	-17.36 $\pm$ 8.12	-20.57 $\pm$ 10.63	-19.82 $\pm$ 10.65	-19.55 $\pm$ 11.17			
Hip Extension ROM ($^\circ$) ^{$a, b, c, \alpha, \beta, \gamma, \delta, \lambda$}	Repl.	34.87 $\pm$ 6.10	36.49 $\pm$ 6.70	38.44 $\pm$ 7.52	35.6 $\pm$ 6.00	32.79 $\pm$ 6.14	< 0.000 (0.930)	0.288 (0.184)	0.433 (0.128)
	Contr.	40.97 $\pm$ 9.55	42.05 $\pm$ 8.97	43.91 $\pm$ 12.13	39.67 $\pm$ 9.47	36.73 $\pm$ 9.99			
Hip Extension Moment (Nm)	Repl.	-0.62 $\pm$ 9.05	1.98 $\pm$ 6.68	-6.12 $\pm$ 12.39	1.02 $\pm$ 7.62	0.00 $\pm$ 11.46	0.362 (0.174)	0.158 (0.354)	0.885 (0.054)
	Contr.	7.11 $\pm$ 12.32	5.96 $\pm$ 11.19	0.86 $\pm$ 22.34	2.92 $\pm$ 13.02	2.55 $\pm$ 6.9			

a: significant difference between C1 and C2, b: significant difference between C1 and C3, c: significant difference between C2 and C3,
 α : significant difference between C3 and C4, β : significant difference between C3 and C5, γ : significant difference between C4 and C5,
 δ : significant difference between C1 and C5, λ : significant difference between C2 and C4, θ : significant difference between Replaced and Contralateral at current condition
 η_p^2 : partial eta squared, Repl.: Replaced side, Contr.: Contralateral side, Cond.: Condition, Inter.: Interaction

Table 4: Mean Results for Pain and RPE of the Replaced Limb During Cycling (mean $\pm$ std)

Variable	Limb	Pre-Test	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)	Cond. p-value (η_p^2)
Pain (1-10)	Repl.	2.8 $\pm$ 1.0	1.5 $\pm$ 1.7	1.4 $\pm$ 1.6	1.6 $\pm$ 2.0	1.2 $\pm$ 1.4	1.3 $\pm$ 1.7	0.593 (0.106)
RPE (6-20)	-	-	11.4 $\pm$ 0.8	11.0 $\pm$ 2.2	11.0 $\pm$ 3.2	11.6 $\pm$ 1.3	11.1 $\pm$ 1.6	0.922 (0.002)

Repl.: Replaced side, Cond.: Condition, RPE: Rating of Perceived Exertion

Table 5: Mean Results for Differences between Limbs of Peak Pedal Reaction Force, Knee Kinematic, and Knee Kinetic Variables (mean $\pm$ std)

Variable	Group	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)	Cond. p-value (η_p^2)	Group p-value (η_p^2)	Inter. p-value (η_p^2)
PRFz (N)	U	9.12 $\pm$ 40.82	-1.90 $\pm$ 17.36	15.67 $\pm$ 34.85	11.65 $\pm$ 23.88	32.31 $\pm$ 14.12	0.061	0.800	0.856
	B	13.28 $\pm$ 51.1	-11.22 $\pm$ 47.48	27.34 $\pm$ 32.71	30.99 $\pm$ 41.54	36.44 $\pm$ 56.26	(0.350)	(0.014)	(0.061)
Knee Extension ROM ($^\circ$)	U	5.53 $\pm$ 13.38	3.16 $\pm$ 7.89	4.01 $\pm$ 13.26	-0.38 $\pm$ 9.48	3.25 $\pm$ 12.88	0.131	0.862	0.793
	B	4.82 $\pm$ 9.24	3.52 $\pm$ 4.63	0.82 $\pm$ 2.18	0.05 $\pm$ 1.61	0.11 $\pm$ 2.2	(0.287)	(0.007)	(0.077)
Knee Extension Angle ($^\circ$) ^{c,λ}	U	7.76 $\pm$ 5.73	4.5 $\pm$ 3.44	2.89 $\pm$ 6.32	-0.12 $\pm$ 4.01	1.90 $\pm$ 6.29	0.023	0.928	0.159
	B	4.35 $\pm$ 9.36	5.30 $\pm$ 1.73	2.10 $\pm$ 1.06	3.6 $\pm$ 1.63	3.19 $\pm$ 2.66	(0.418)	(0.002)	(0.270)
Knee Extension Moment (Nm)	U	3.39 $\pm$ 4.23	3.86 $\pm$ 7.26	3.83 $\pm$ 8.45	3.39 $\pm$ 4.27	3.78 $\pm$ 4.25	0.647	0.104	0.682
	B	9.43 $\pm$ 10.05	7.74 $\pm$ 2.91	13.4 $\pm$ 8.87	10.53 $\pm$ 5.22	15.57 $\pm$ 10.97	(0.112)	(0.440)	(0.104)
Knee Abduction Moment (Nm)	U	-2.28 $\pm$ 2.71	-1.45 $\pm$ 3.86	-1.95 $\pm$ 3.38	-1.22 $\pm$ 1.85	-2.19 $\pm$ 2.45	0.704	0.468	0.987
	B	0.81 $\pm$ 6.52	1.72 $\pm$ 8.96	1.63 $\pm$ 7.21	1.58 $\pm$ 5.99	0.37 $\pm$ 8.78	(0.099)	(0.110)	(0.016)
Knee Extension Power (Nm/s)	U	21.92 $\pm$ 12.97	21.37 $\pm$ 28.6	16.66 $\pm$ 32.46	12.29 $\pm$ 14.02	11.42 $\pm$ 12.09	0.911	0.231	0.410
	B	31.24 $\pm$ 53.34	29.52 $\pm$ 18.45	51.65 $\pm$ 41.3	51.7 $\pm$ 36.13	45.68 $\pm$ 36.75	(0.046)	(0.271)	(0.172)

a: significant difference between C1 and C2, b: significant difference between C1 and C3, c: significant difference between C2 and C3,
 α : significant difference between C3 and C4, β : significant difference between C3 and C5, γ : significant difference between C4 and C5,
 δ : significant difference between C1 and C5, λ : significant difference between C2 and C4,
 η_p^2 : partial eta squared, G: Group, U: Unilateral group, B: Bilateral group, Cond.: Condition, Inter.: Interaction

Table 6: Mean Results for Differences between Limbs of ROM and Peak Moment Variable Differences for the Ankle and Hip (mean $\pm$ std)

Variable	Group	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)	Cond. p-value (η_p^2)	Group p-value (η_p^2)	Inter. p-value (η_p^2)
Ankle Plantarflexion ROM (°)	U	8.42 $\pm$ 14.14	11.87 $\pm$ 17.03	7.68 $\pm$ 25.70	10.06 $\pm$ 12.73	5.24 $\pm$ 8.93	0.926 (0.042)	0.507 (0.093)	0.954 (0.032)
	B	1.30 $\pm$ 15.79	1.19 $\pm$ 13.57	1.05 $\pm$ 7.15	2.60 $\pm$ 9.61	1.27 $\pm$ 9.64			
Ankle Plantarflexion Moment (Nm)	U	-3.12 $\pm$ 4.30	-1.02 $\pm$ 1.39	-3.48 $\pm$ 6.01	-3.54 $\pm$ 2.59	-3.01 $\pm$ 2.69	0.541 (0.138)	0.563 (0.071)	0.974 (0.023)
	B	-0.87 $\pm$ 13.30	2.70 $\pm$ 9.63	0.68 $\pm$ 5.23	-0.67 $\pm$ 12.51	-1.16 $\pm$ 8.28			
Hip Extension ROM (°)	U	-7.77 $\pm$ 13.20	-7.17 $\pm$ 13.42	-8.31 $\pm$ 19.80	-6.05 $\pm$ 14.26	-6.26 $\pm$ 16.32	0.540 (0.138)	0.625 (0.051)	0.885 (0.054)
	B	-3.88 $\pm$ 6.58	-3.42 $\pm$ 4.96	-1.70 $\pm$ 3.91	-1.44 $\pm$ 3.93	-0.87 $\pm$ 4.34			
Hip Extension Moment (Nm)	U	5.93 $\pm$ 15.97	5.08 $\pm$ 4.92	2.24 $\pm$ 24.99	-1.20 $\pm$ 4.18	-0.41 $\pm$ 6.31	0.949 (0.034)	0.259 (0.245)	0.906 (0.048)
	B	9.23 $\pm$ 10.39	4.25 $\pm$ 9.17	11.45 $\pm$ 9.04	9.15 $\pm$ 11.78	6.59 $\pm$ 5.51			

a: significant difference between C1 and C2, b: significant difference between C1 and C3, c: significant difference between C2 and C3,
 α : significant difference between C3 and C4, β : significant difference between C3 and C5, γ : significant difference between C4 and C5,
 δ : significant difference between C1 and C5, λ : significant difference between C2 and C4,
 η_p^2 : partial eta squared, G: Group, U: Unilateral group, B: Bilateral group, Cond.: Condition, Inter.: Interaction

Table 7: Mean Peak Knee Contact Forces and Peak Vertical Pedal Reaction Force (mean $\pm$ std)

Variable	C1 (140/170)	C2 (155/170)	C3 (170/170)	p-value (η_p^2)
Total Contact Force (N) ^{b,c}	830.99 $\pm$ 301.94	857.75 $\pm$ 282.16	1036.28 $\pm$ 264.34	0.002 (0.660)
Medial Contact Force (N)	432.71 $\pm$ 203.15	447.01 $\pm$ 153.66	502.53 $\pm$ 168.63	0.082 (0.341)
Lateral Contact Force (N) ^{b,c}	434.20 $\pm$ 148.92	446.81 $\pm$ 180.72	551.79 $\pm$ 129.60	0.011 (0.660)
PRF (N)	227.51 $\pm$ 47.82	225.22 $\pm$ 47.49	230.29 $\pm$ 42.00	0.876 (0.022)

a: significant difference between C1 and C2, b: significant difference between C1 and C3, c: significant difference between C2 and C3, η_p^2 : partial eta squared, G: Group, U: Unilateral group, B: Bilateral group, Cond.: Condition, Inter.: Interaction

Table 8: Mean Peak Muscle Group Forces (mean $\pm$ std)

Variable	C1 (140/170)	C2 (155/170)	C3 (170/170)	p-value (η_p^2)
Quadriceps Force (N)	932.17 $\pm$ 321.10	942.80 $\pm$ 395.74	1101.88 $\pm$ 487.82	0.151 (0.270)
Hamstrings Force (N)	288.57 $\pm$ 65.91	321.00 $\pm$ 72.58	353.50 $\pm$ 197.52	0.428 (0.113)
Gastrocnemius Force (N)	192.72 $\pm$ 193.92	181.86 $\pm$ 166.67	234.10 $\pm$ 211.55	0.101 (0.318)

a: significant difference between C1 and C2, b: significant difference between C1 and C3, c: significant difference between C2 and C3, η_p^2 : partial eta squared, G: Group, U: Unilateral group, B: Bilateral group, Cond.: Condition, Inter.: Interaction

Table 9: Mean Peak Individual Muscle Forces (mean $\pm$ std)

Variable	C1 (140/170)	C2 (155/170)	C3 (170/170)	p-value (η_p^2)
Rectus Femoris Force (N)	219.26 $\pm$ 68.83	199.27 $\pm$ 87.98	190.19 $\pm$ 105.97	0.685 (0.061)
Semitendinosus Force (N)	17.21 $\pm$ 10.14	17.15 $\pm$ 12.57	23.10 $\pm$ 23.22	0.327 (0.161)
Medial Gastrocnemius Force (N)	156.20 $\pm$ 158.66	147.65 $\pm$ 136.83	187.80 $\pm$ 171.73	0.136 (0.283)
Tibialis Anterior Force (N)	61.19 $\pm$ 56.99	63.80 $\pm$ 62.08	54.10 $\pm$ 54.04	0.428 (0.132)

a: significant difference between C1 and C2, b: significant difference between C1 and C3, c: significant difference between C2 and C3, η_p^2 : partial eta squared, G: Group, U: Unilateral group, B: Bilateral group, Cond.: Condition, Inter.: Interaction

Table 10: Mean Peak Residuals and Peak Moments (mean $\pm$ std)

Variable	C1 (140/170)	C2 (155/170)	C3 (170/170)	p-value (η_p^2)
AP Force (N)	0.08 $\pm$ 0.10	0.07 $\pm$ 0.08	0.07 $\pm$ 0.04	0.855 (0.026)
SI Force (N)	0.12 $\pm$ 0.10	0.12 $\pm$ 0.11	0.12 $\pm$ 0.13	0.971 (0.000)
Sagittal Moment (Nm)	4.97 $\pm$ 1.97	5.29 $\pm$ 2.63	5.97 $\pm$ 2.42	0.345 (0.163)

AP: anteroposterior, SI: superoinferior, ML: mediolateral, η_p^2 : partial eta squared

Table 11: Participant Qualitative Demographics (mean $\pm$ std)

Participant	Recently Replaced Limb	Implant Brand	Implant Style
1*	Right	Stryker	Triathlon CR
2*	Left	Stryker	Triathlon CR
3	Left	Stryker	Triathlon CR
4	Left	Stryker	Triathlon CR
5	Left	Smith & Nephew	Oxinium Journey II CR
6	Right	Zimmer Biomet	Persona Narrow PS
7*	Right	Stryker	Triathlon CR

*: Indicates bilateral TKA participant, CR: Cruciate Retaining, PS: Posterior Stabilized

Table 12: Participant Quantitative Demographics (mean $\pm$ std)

Participant	Sex	Age (years)	Height (cm)	Mass (kg)	BMI (kg/m ²)	Initial Bilateral Surgery (Days)	Time From Recent Surgery (Days)
1*	Male	66	182	121.6	36.7	476	51
2*	Male	69	191	127	34.8	335	49
3	Female	69	165	93.5	34.3	-	55
4	Female	73	160	62.7	24.5	-	51
5	Female	54	155	82.2	34.2	-	58
6	Female	63	165	75.3	27.6	-	58
7*	Male	73	188	99.8	28.2	1096	55
Post Avg.		67 $\pm$ 7	172 $\pm$ 14	94.6 $\pm$ 23.6	31.5 $\pm$ 4.6	-	54 $\pm$ 4
Uni. Avg.		65 $\pm$ 8	160 $\pm$ 5	78.4 $\pm$ 12.9	30.2 $\pm$ 4.9	-	55 $\pm$ 3
Bi. Avg.		69 $\pm$ 3	190 $\pm$ 5	116.1 $\pm$ 14.4	33.2 $\pm$ 4.5	638 $\pm$ 402	52 $\pm$ 3

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 13: Study Inclusion and Exclusion Criteria

Inclusion Criteria	
•	Ages: 50-75
•	Willingness to attend a testing session occurring 1-2 months after arthroplasty
•	Knee flexion range of motion of at least 100°
•	Male or Female
•	Demonstrated asymmetry in their lower limbs as measured by passive knee flexion ROM in clinic with a goniometer
•	And/or flexion/extension contracture of involved side pre-surgery
•	OPTIONAL: Use of assistive device for walking/daily activities for operational side pre-surgery
Exclusion Criteria	
•	Diagnosed osteoarthritis of ankles, hips, or spine
•	No contralateral TKA within 4 months of study participation date
•	BMI value greater than 40
•	Neurological conditions that prevent the patients from cycling or walking without aids
•	Inability to maintain upright posture during cycling (as in, hip hiking or compensation during pedaling)
•	Pregnant or Nursing Women
•	Presence of partial or total limb amputation
•	Cardiovascular Disease or other risk factor as identified by the PARQ+

Table 14: Description of Cycling Crank Arm Length Conditions

Limb Crank Length	Cycling Condition				
	C1	C2	C3	C4	C5
Replaced Limb (mm)	140	155	170	155	140
Contralateral Limb (mm)	170	170	170	155	140

Table 15: Survey Scores (mean ± std)

Survey	Participant							Post Avg.	Uni. Avg.	Bi. Avg.
	1*	2*	3	4	5	6	7*			
KOOS Total (%)	58	53	89	72	43	58	85	65.0 ± 16.5	65.5 ± 19.6	64.3 ± 15.5
QOL (%)	44	25	69	50	19	44	94	49.3 ± 25.7	45.5 ± 20.6	54.3 ± 35.6
Pain (%)	72	64	86	67	53	75	86	71.9 ± 11.9	70.3 ± 13.9	74.0 ± 11.1
ADL (%)	72	66	96	81	65	78	93	78.7 ± 12.3	80.0 ± 12.7	77.0 ± 14.2
Sport/Rec (%)	25	35	100	85	20	35	60	51.4 ± 31.1	60.0 ± 38.5	40.0 ± 18
Sym/Stiff (%)	79	75	93	79	57	61	75	74.1 ± 12	72.5 ± 16.7	76.3 ± 2.3
TSK (%)	51	43	47	41	57	50	31	45.7 ± 8.4	48.8 ± 6.7	41.7 ± 10.1

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral, KOOS: Knee Injury and Osteoarthritis Outcome Score, QOL: Quality of Life, ADL: Activities of Daily Living, Sym/Stiff: Symptoms and Stiffness

Table 16: Clinical Metrics and Measurements (mean $\pm$ std)

Participant	TUG Time (s)	TUG Pain	Replaced Circ. (cm)	Contralateral Circ. (cm)	Replaced ROM (°)	Contralateral ROM (°)	Gait Speed (m/s)
1*	13.45	3	50.8	48.3	106	106	1.17
2*	9.22	1	48.3	45.7	121	118	1.33
3	9.38	0	50.6	51.2	122	120	1.08
4	6.31	2	41.9	37.9	134	142	1.36
5	9.46	0.5	42.3	39.6	123	131	1.30
6	6.72	2.5	40.9	36	134	154	1.60
7*	6.94	0	45.5	42.1	139	122	1.63
Post Avg.	8.8 $\pm$ 2.5	1.3 $\pm$ 1.2	45.8 $\pm$ 4.2	43 $\pm$ 5.6	125.6 $\pm$ 11.1	127.6 $\pm$ 16.1	1.35 $\pm$ 0.2
Uni. Avg.	8.0 $\pm$ 1.7	1.3 $\pm$ 1.2	43.9 $\pm$ 4.5	41.2 $\pm$ 6.8	128.3 $\pm$ 6.7	136.8 $\pm$ 14.6	1.33 $\pm$ 0.21
Bi. Avg.	9.9 $\pm$ 3.3	1.3 $\pm$ 1.5	48.2 $\pm$ 2.7	45.4 $\pm$ 3.1	122 $\pm$ 16.5	115.3 $\pm$ 8.3	1.38 $\pm$ 0.23

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral. TUG: Timed Up and Go

Table 17: Pain in the Replaced Knee during Cycling (mean $\pm$ std)

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	3.5	3.5	3.5	3.5	3.5
2*	0	0	0	0.5	0
3	2	2	2	2.5	2
4	0	0	0	0	0
5	4	3.5	5	2	3.5
6	0	0	0	0	0
7*	1	0.5	0.5	0	0
Post Avg.	1.5 $\pm$ 1.7	1.4 $\pm$ 1.6	1.6 $\pm$ 2	1.2 $\pm$ 1.4	1.3 $\pm$ 1.7
Uni. Avg.	1.5 $\pm$ 1.9	1.4 $\pm$ 1.7	1.8 $\pm$ 2.4	1.1 $\pm$ 1.3	1.4 $\pm$ 1.7
Bi. Avg.	1.5 $\pm$ 1.8	1.3 $\pm$ 1.9	1.3 $\pm$ 1.9	1.3 $\pm$ 1.9	1.2 $\pm$ 2

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 18: Rating of Perceived Exertion (RPE, Borg Scale) during Cycling (mean $\pm$ std)

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	11	11	11	11	11
2*	11	11	11	11	11
3	11	7	7	12	11
4	12	14	17	14	14
5	13	13	8	11	12
6	11	10	12	12	9.5
7*	11	11	11	10	9
Post Avg.	11.4 $\pm$ 0.8	11 $\pm$ 2.2	11 $\pm$ 3.2	11.6 $\pm$ 1.3	11.1 $\pm$ 1.6
Uni. Avg.	11.8 $\pm$ 1	11 $\pm$ 3.2	11 $\pm$ 4.5	12.3 $\pm$ 1.3	11.6 $\pm$ 1.9
Bi. Avg.	11 $\pm$ 0	11 $\pm$ 0	11 $\pm$ 0	10.7 $\pm$ 0.6	10.3 $\pm$ 1.2

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 19: Peak Vertical Pedal Reaction Force (N, mean $\pm$ std)

Participant	Limb	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	R	307.06 $\pm$ 13.56	304.68 $\pm$ 9.75	311.22 $\pm$ 9.4	277.25 $\pm$ 9.84	267.75 $\pm$ 11.21
	C	310.95 $\pm$ 7.13	289.0 $\pm$ 11.73	369.47 $\pm$ 12.1	322.16 $\pm$ 13.28	345.35 $\pm$ 12.48
2*	R	219.02 $\pm$ 12.34	196.72 $\pm$ 2.63	237.41 $\pm$ 6.15	220.49 $\pm$ 3.41	244.29 $\pm$ 18.45
	C	287.44 $\pm$ 31.04	235.05 $\pm$ 10.84	268.1 $\pm$ 20.76	284.27 $\pm$ 8.77	303.67 $\pm$ 13.45
3	R	280.51 $\pm$ 7.63	270.51 $\pm$ 15.49	237.43 $\pm$ 13.03	251.25 $\pm$ 4.41	239.05 $\pm$ 7.07
	C	239.63 $\pm$ 5.8	249.19 $\pm$ 19.05	270.31 $\pm$ 7.22	245.04 $\pm$ 6.26	269.51 $\pm$ 12.02
4	R	194.97 $\pm$ 11.27	186.13 $\pm$ 10.51	175.43 $\pm$ 8.17	180.86 $\pm$ 11.02	183.34 $\pm$ 10.73
	C	196.21 $\pm$ 10.1	175.83 $\pm$ 5.63	151.03 $\pm$ 6.15	193.19 $\pm$ 14.45	210.95 $\pm$ 11.27
5	R	201.83 $\pm$ 5.14	206.21 $\pm$ 4.7	230.99 $\pm$ 7.87	207.0 $\pm$ 6.52	203.14 $\pm$ 2.12
	C	259.42 $\pm$ 14.26	224.1 $\pm$ 2.65	230.89 $\pm$ 6.6	252.18 $\pm$ 14.68	255.33 $\pm$ 8.27
6	R	176.88 $\pm$ 7.75	176.69 $\pm$ 26.66	216.08 $\pm$ 12.1	173.76 $\pm$ 32.17	185.41 $\pm$ 3.8
	C	195.41 $\pm$ 21.94	182.82 $\pm$ 19.73	270.38 $\pm$ 20.34	169.04 $\pm$ 38.93	204.39 $\pm$ 8.84
7*	R	212.29 $\pm$ 12.89	235.62 $\pm$ 17.0	203.47 $\pm$ 19.5	208.35 $\pm$ 10.55	220.31 $\pm$ 8.72
	C	179.81 $\pm$ 9.0	179.32 $\pm$ 8.71	196.57 $\pm$ 6.09	192.63 $\pm$ 7.39	192.63 $\pm$ 7.39
Post Avg.	R	227.51 $\pm$ 44.28	225.22 $\pm$ 43.96	230.29 $\pm$ 38.88	216.99 $\pm$ 34.15	220.47 $\pm$ 29.41
	C	238.41 $\pm$ 46.56	219.33 $\pm$ 39.35	250.96 $\pm$ 63.74	236.93 $\pm$ 51.11	254.55 $\pm$ 52.28
Uni. Avg.	R	213.55 $\pm$ 45.87	209.88 $\pm$ 42.25	214.98 $\pm$ 27.84	203.22 $\pm$ 35.07	202.74 $\pm$ 25.79
	C	222.67 $\pm$ 32.05	207.98 $\pm$ 34.76	230.65 $\pm$ 56.25	214.86 $\pm$ 40.3	235.04 $\pm$ 32.25
Bi. Avg.	R	246.12 $\pm$ 52.88	245.67 $\pm$ 54.68	250.7 $\pm$ 55.09	235.36 $\pm$ 36.78	244.12 $\pm$ 23.72
	C	259.4 $\pm$ 69.92	234.46 $\pm$ 54.84	278.05 $\pm$ 86.88	266.35 $\pm$ 66.6	280.55 $\pm$ 78.94

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral, R: Replaced, C: Contralateral

Table 20: Sagittal Knee ROM ($^{\circ}$, mean $\pm$ std)

Participant	Limb	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	R	47.5 $\pm$ 0.2	50.6 $\pm$ 0.1	54.5 $\pm$ 0.1	50.7 $\pm$ 0.2	46.9 $\pm$ 0.1
	C	56.2 $\pm$ 0.1	56.9 $\pm$ 0.1	57.4 $\pm$ 0.1	52.6 $\pm$ 0.4	47.0 $\pm$ 0.3
2*	R	58.7 $\pm$ 0.2	58.8 $\pm$ 0.3	61.3 $\pm$ 0.3	56.2 $\pm$ 0.1	52.0 $\pm$ 0.2
	C	53.0 $\pm$ 0.1	56.9 $\pm$ 0.1	59.9 $\pm$ 0.3	55.6 $\pm$ 0.2	49.9 $\pm$ 0.2
3	R	62.7 $\pm$ 0.2	64.3 $\pm$ 0.1	67.5 $\pm$ 0.1	63.2 $\pm$ 0.1	56.5 $\pm$ 0.4
	C	55.5 $\pm$ 0.2	61.3 $\pm$ 0.2	62.2 $\pm$ 0.2	55.0 $\pm$ 0.1	50.8 $\pm$ 0.2
4	R	68.5 $\pm$ 0.5	68.8 $\pm$ 0.3	72.9 $\pm$ 0.2	69.1 $\pm$ 0.3	62.1 $\pm$ 0.9
	C	63.8 $\pm$ 0.3	64.9 $\pm$ 0.2	67.2 $\pm$ 0.2	61.0 $\pm$ 0.4	56.1 $\pm$ 0.1
5	R	48.7 $\pm$ 0.8	54.8 $\pm$ 0.6	54.0 $\pm$ 0.5	54.3 $\pm$ 1.3	42.6 $\pm$ 1.2
	C	67.6 $\pm$ 0.2	66.9 $\pm$ 0.3	76.7 $\pm$ 0.3	65.6 $\pm$ 0.5	64.1 $\pm$ 1.0
6	R	51.4 $\pm$ 0.7	58.3 $\pm$ 0.5	61.5 $\pm$ 0.8	56.3 $\pm$ 0.6	51.3 $\pm$ 0.3
	C	66.6 $\pm$ 0.6	65.6 $\pm$ 1.6	65.9 $\pm$ 0.7	59.7 $\pm$ 1.5	54.4 $\pm$ 0.6
7*	R	49.4 $\pm$ 0.3	53.2 $\pm$ 0.3	58.4 $\pm$ 0.1	54.4 $\pm$ 0.2	50.3 $\pm$ 0.2
	C	60.8 $\pm$ 0.1	59.3 $\pm$ 0.3	59.4 $\pm$ 0.2	53.3 $\pm$ 0.3	52.6 $\pm$ 0.4
Post Avg.	R	55.27 $\pm$ 0.42	58.39 $\pm$ 0.33	61.45 $\pm$ 0.30	57.75 $\pm$ 0.40	51.67 $\pm$ 0.47
	C	60.50 $\pm$ 0.23	61.71 $\pm$ 0.40	64.10 $\pm$ 0.28	57.56 $\pm$ 0.49	53.58 $\pm$ 0.39
Uni. Avg.	R	57.84 $\pm$ 0.56	61.55 $\pm$ 0.39	63.98 $\pm$ 0.40	60.73 $\pm$ 0.58	53.12 $\pm$ 0.69
	C	63.37 $\pm$ 0.30	64.71 $\pm$ 0.56	68.00 $\pm$ 0.34	60.35 $\pm$ 0.63	56.37 $\pm$ 0.46
Bi. Avg.	R	51.85 $\pm$ 0.22	54.18 $\pm$ 0.26	58.08 $\pm$ 0.17	53.79 $\pm$ 0.16	49.75 $\pm$ 0.18
	C	56.67 $\pm$ 0.13	57.70 $\pm$ 0.18	58.90 $\pm$ 0.21	53.84 $\pm$ 0.30	49.86 $\pm$ 0.30

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi: Bilateral, R: Replaced, C: Contralateral

Table 21: Peak Knee Extension Angle ($^{\circ}$, mean $\pm$ std)

Participant	Limb	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	R	-53.91 $\pm$ 0.28	-51.3 $\pm$ 0.31	-47.59 $\pm$ 0.47	-51.71 $\pm$ 0.55	-54.86 $\pm$ 0.53
	C	-44.87 $\pm$ 0.5	-45.37 $\pm$ 0.24	-45.22 $\pm$ 0.21	-46.71 $\pm$ 0.34	-50.89 $\pm$ 0.28
2*	R	-36.08 $\pm$ 0.28	-41.31 $\pm$ 0.42	-38.39 $\pm$ 0.41	-40.68 $\pm$ 0.37	-42.69 $\pm$ 0.44
	C	-42.51 $\pm$ 0.83	-37.96 $\pm$ 1.22	-37.47 $\pm$ 0.76	-38.87 $\pm$ 1.09	-42.46 $\pm$ 0.31
3	R	-50.49 $\pm$ 0.28	-47.07 $\pm$ 0.89	-42.11 $\pm$ 0.81	-49.64 $\pm$ 0.27	-53.35 $\pm$ 0.41
	C	-48.55 $\pm$ 0.09	-44.6 $\pm$ 0.51	-43.72 $\pm$ 0.55	-53.38 $\pm$ 0.36	-55.31 $\pm$ 0.35
4	R	-47.82 $\pm$ 0.6	-46.9 $\pm$ 0.52	-43.66 $\pm$ 0.3	-46.33 $\pm$ 0.47	-48.0 $\pm$ 0.34
	C	-41.85 $\pm$ 0.47	-43.56 $\pm$ 0.56	-43.44 $\pm$ 0.21	-48.43 $\pm$ 0.42	-49.77 $\pm$ 0.37
5	R	-45.61 $\pm$ 0.2	-41.83 $\pm$ 0.9	-38.21 $\pm$ 0.25	-39.03 $\pm$ 0.26	-47.61 $\pm$ 0.36
	C	-30.0 $\pm$ 0.42	-32.2 $\pm$ 0.21	-25.95 $\pm$ 0.33	-33.56 $\pm$ 0.69	-36.37 $\pm$ 0.35
6	R	-65.73 $\pm$ 0.83	-59.94 $\pm$ 2.16	-58.52 $\pm$ 0.72	-61.56 $\pm$ 0.83	-64.68 $\pm$ 0.72
	C	-58.22 $\pm$ 0.48	-57.37 $\pm$ 1.68	-57.8 $\pm$ 0.46	-61.7 $\pm$ 0.72	-64.59 $\pm$ 0.45
7*	R	-51.26 $\pm$ 0.36	-51.48 $\pm$ 0.16	-47.81 $\pm$ 0.34	-49.41 $\pm$ 0.36	-50.79 $\pm$ 0.21
	C	-40.81 $\pm$ 0.14	-44.86 $\pm$ 0.27	-44.81 $\pm$ 0.34	-45.42 $\pm$ 0.23	-45.42 $\pm$ 0.23
Post Avg.	R	-50.13 $\pm$ 8.3	-48.55 $\pm$ 5.96	-45.18 $\pm$ 6.51	-48.34 $\pm$ 6.95	-51.71 $\pm$ 6.48
	C	-43.83 $\pm$ 7.89	-43.7 $\pm$ 7.14	-42.63 $\pm$ 8.85	-46.87 $\pm$ 8.52	-49.26 $\pm$ 8.47
Uni. Avg.	R	-52.41 $\pm$ 9.1	-48.94 $\pm$ 7.73	-45.62 $\pm$ 8.9	-49.14 $\pm$ 9.39	-53.41 $\pm$ 7.96
	C	-44.66 $\pm$ 11.86	-44.43 $\pm$ 10.29	-42.73 $\pm$ 13.04	-49.27 $\pm$ 11.82	-51.51 $\pm$ 11.8
Bi. Avg.	R	-47.08 $\pm$ 9.62	-48.03 $\pm$ 5.82	-44.6 $\pm$ 5.38	-47.27 $\pm$ 5.82	-49.45 $\pm$ 6.2
	C	-42.73 $\pm$ 2.04	-42.73 $\pm$ 4.14	-42.5 $\pm$ 4.36	-43.67 $\pm$ 4.2	-46.26 $\pm$ 4.28

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi: Bilateral, R: Replaced, C: Contralateral

Table 22: Peak Knee Extension Moment (Nm, mean $\pm$ std)

Participant	Limb	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	R	19.96 $\pm$ 1.53	19.24 $\pm$ 1.58	26.05 $\pm$ 0.15	29.93 $\pm$ 0.99	13.42 $\pm$ 1.15
	C	40.36 $\pm$ 1.0	30.27 $\pm$ 3.41	46.77 $\pm$ 0.88	39.52 $\pm$ 2.61	41.55 $\pm$ 3.38
2*	R	27.41 $\pm$ 3.81	19.65 $\pm$ 1.7	15.53 $\pm$ 2.02	21.95 $\pm$ 1.12	22.34 $\pm$ 3.68
	C	28.06 $\pm$ 3.56	25.15 $\pm$ 3.06	31.48 $\pm$ 2.74	27.79 $\pm$ 0.9	30.21 $\pm$ 1.77
3	R	16.32 $\pm$ 1.18	23.9 $\pm$ 1.25	14.03 $\pm$ 1.81	17.72 $\pm$ 0.4	20.4 $\pm$ 0.94
	C	21.81 $\pm$ 0.8	23.37 $\pm$ 0.83	24.24 $\pm$ 2.08	19.61 $\pm$ 1.14	24.34 $\pm$ 1.45
4	R	12.87 $\pm$ 1.42	11.3 $\pm$ 0.98	14.79 $\pm$ 1.38	10.83 $\pm$ 1.22	13.67 $\pm$ 0.82
	C	9.94 $\pm$ 0.59	9.37 $\pm$ 0.46	17.0 $\pm$ 1.06	11.09 $\pm$ 1.4	11.63 $\pm$ 0.76
5	R	7.82 $\pm$ 1.33	6.97 $\pm$ 0.23	10.76 $\pm$ 0.8	9.61 $\pm$ 0.88	13.28 $\pm$ 0.94
	C	13.89 $\pm$ 2.53	21.1 $\pm$ 0.93	21.14 $\pm$ 1.62	19.31 $\pm$ 2.41	21.36 $\pm$ 1.46
6	R	15.6 $\pm$ 1.71	13.58 $\pm$ 4.23	16.94 $\pm$ 1.38	11.97 $\pm$ 7.61	12.98 $\pm$ 2.86
	C	20.5 $\pm$ 3.0	17.36 $\pm$ 2.98	9.46 $\pm$ 3.75	13.67 $\pm$ 4.04	18.11 $\pm$ 1.45
7*	R	14.24 $\pm$ 1.84	17.53 $\pm$ 1.38	22.26 $\pm$ 1.85	10.53 $\pm$ 1.17	15.98 $\pm$ 0.77
	C	21.5 $\pm$ 3.43	24.23 $\pm$ 7.48	25.8 $\pm$ 1.41	26.69 $\pm$ 3.65	26.69 $\pm$ 3.65
Post Avg.	R	16.32 $\pm$ 5.67	16.02 $\pm$ 5.32	17.19 $\pm$ 4.84	16.08 $\pm$ 7.03	16.01 $\pm$ 3.55
	C	22.29 $\pm$ 9.16	21.55 $\pm$ 6.16	25.13 $\pm$ 10.94	22.53 $\pm$ 8.96	24.84 $\pm$ 8.81
Uni. Avg.	R	13.15 $\pm$ 3.85	13.94 $\pm$ 7.19	14.13 $\pm$ 2.56	12.53 $\pm$ 3.59	15.08 $\pm$ 3.56
	C	16.54 $\pm$ 5.6	17.8 $\pm$ 6.14	17.96 $\pm$ 6.4	15.92 $\pm$ 4.22	18.86 $\pm$ 5.45
Bi. Avg.	R	20.54 $\pm$ 6.6	18.81 $\pm$ 1.12	21.28 $\pm$ 5.33	20.8 $\pm$ 9.75	17.25 $\pm$ 4.59
	C	29.97 $\pm$ 9.57	26.55 $\pm$ 3.25	34.68 $\pm$ 10.85	31.33 $\pm$ 7.11	32.82 $\pm$ 7.77

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi: Bilateral, R: Replaced, C: Contralateral

Table 23: Peak Knee Abduction Moment (Nm, mean $\pm$ std)

Participant	Limb	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	R	-9.51 $\pm$ 1.12	-7.55 $\pm$ 0.41	-8.97 $\pm$ 0.81	-10.34 $\pm$ 0.71	-8.97 $\pm$ 0.77
	C	-3.36 $\pm$ 1.17	-2.1 $\pm$ 0.75	-3.36 $\pm$ 1.31	-6.04 $\pm$ 1.04	-4.73 $\pm$ 1.28
2*	R	-7.41 $\pm$ 1.09	-7.5 $\pm$ 1.56	-7.47 $\pm$ 2.0	-8.0 $\pm$ 1.73	-8.1 $\pm$ 0.79
	C	-13.87 $\pm$ 1.87	-16.0 $\pm$ 0.78	-14.17 $\pm$ 1.47	-13.29 $\pm$ 0.57	-17.78 $\pm$ 1.32
3	R	-1.22 $\pm$ 0.7	-0.13 $\pm$ 0.44	-2.18 $\pm$ 0.17	-1.4 $\pm$ 0.57	-0.86 $\pm$ 0.48
	C	-5.24 $\pm$ 0.59	-4.55 $\pm$ 0.33	-5.13 $\pm$ 0.48	-4.19 $\pm$ 0.49	-4.75 $\pm$ 0.4
4	R	0.72 $\pm$ 0.45	0.43 $\pm$ 0.29	0.02 $\pm$ 0.24	-0.04 $\pm$ 0.25	0.88 $\pm$ 0.08
	C	-4.44 $\pm$ 0.79	-3.86 $\pm$ 0.07	-1.71 $\pm$ 0.07	-2.35 $\pm$ 0.25	-3.39 $\pm$ 0.34
5	R	-1.28 $\pm$ 0.69	-1.49 $\pm$ 0.43	-1.7 $\pm$ 0.91	-1.33 $\pm$ 0.53	-1.52 $\pm$ 0.34
	C	-1.4 $\pm$ 0.74	2.29 $\pm$ 0.91	0.8 $\pm$ 0.99	0.03 $\pm$ 0.87	-0.47 $\pm$ 0.6
6	R	-0.75 $\pm$ 0.52	-0.91 $\pm$ 1.38	1.2 $\pm$ 0.38	-0.9 $\pm$ 1.11	0.01 $\pm$ 0.29
	C	-0.55 $\pm$ 2.8	-1.79 $\pm$ 1.03	-4.4 $\pm$ 0.89	-2.02 $\pm$ 1.65	-1.64 $\pm$ 0.42
7*	R	-8.0 $\pm$ 1.17	-8.01 $\pm$ 0.67	-5.69 $\pm$ 0.77	-7.42 $\pm$ 0.35	-8.25 $\pm$ 0.32
	C	-5.26 $\pm$ 1.62	0.2 $\pm$ 0.56	-0.03 $\pm$ 0.54	-1.7 $\pm$ 1.01	-1.7 $\pm$ 1.01
Post Avg.	R	-3.92 $\pm$ 3.89	-3.59 $\pm$ 3.59	-3.54 $\pm$ 3.59	-4.2 $\pm$ 3.91	-3.83 $\pm$ 4.06
	C	-4.88 $\pm$ 4.04	-3.69 $\pm$ 5.47	-4.00 $\pm$ 4.99	-4.22 $\pm$ 4.11	-4.92 $\pm$ 5.46
Uni. Avg.	R	-0.63 $\pm$ 0.93	-0.52 $\pm$ 0.85	-0.66 $\pm$ 1.56	-0.92 $\pm$ 0.63	-0.37 $\pm$ 1.04
	C	-2.91 $\pm$ 2.28	-1.98 $\pm$ 3.08	-2.61 $\pm$ 2.71	-2.13 $\pm$ 1.73	-2.56 $\pm$ 1.89
Bi. Avg.	R	-8.31 $\pm$ 1.08	-7.69 $\pm$ 0.28	-7.38 $\pm$ 1.64	-8.59 $\pm$ 1.55	-8.44 $\pm$ 0.47
	C	-7.5 $\pm$ 5.6	-5.97 $\pm$ 8.76	-7.1 $\pm$ 10.0	-7.01 $\pm$ 5.86	-8.07 $\pm$ 8.54

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi: Bilateral, R: Replaced, C: Contralateral

Table 24: Peak Knee Power (Nm/s, mean $\pm$ std)

Participant	Limb	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	R	63.79 $\pm$ 5.57	65.59 $\pm$ 5.25	97.51 $\pm$ 1.98	42.99 $\pm$ 10.89	40.78 $\pm$ 3.44
	C	153.29 $\pm$ 3.08	114.56 $\pm$ 12.8	183.4 $\pm$ 2.75	135.41 $\pm$ 10.14	128.8 $\pm$ 10.09
2*	R	117.61 $\pm$ 17.57	75.48 $\pm$ 7.53	64.53 $\pm$ 8.79	83.89 $\pm$ 4.04	77.13 $\pm$ 11.68
	C	102.41 $\pm$ 14.55	102.81 $\pm$ 14.19	127.79 $\pm$ 13.78	107.33 $\pm$ 6.94	104.11 $\pm$ 4.91
3	R	59.58 $\pm$ 4.41	89.76 $\pm$ 3.71	63.83 $\pm$ 7.88	67.44 $\pm$ 1.48	71.57 $\pm$ 4.03
	C	88.03 $\pm$ 3.5	92.05 $\pm$ 2.46	96.08 $\pm$ 6.43	68.99 $\pm$ 4.56	78.09 $\pm$ 4.69
4	R	46.09 $\pm$ 4.86	41.2 $\pm$ 3.99	53.09 $\pm$ 4.77	41.18 $\pm$ 5.03	46.09 $\pm$ 3.29
	C	48.8 $\pm$ 1.58	42.09 $\pm$ 2.1	69.54 $\pm$ 4.84	47.98 $\pm$ 5.85	42.67 $\pm$ 2.2
5	R	27.69 $\pm$ 5.25	27.96 $\pm$ 0.86	49.37 $\pm$ 4.02	40.81 $\pm$ 4.3	48.66 $\pm$ 3.5
	C	53.45 $\pm$ 10.87	90.18 $\pm$ 6.97	95.75 $\pm$ 6.51	73.72 $\pm$ 8.34	70.08 $\pm$ 6.53
6	R	52.63 $\pm$ 8.16	52.65 $\pm$ 20.37	67.55 $\pm$ 5.49	45.21 $\pm$ 25.25	44.98 $\pm$ 10.68
	C	83.4 $\pm$ 6.31	72.71 $\pm$ 17.13	39.1 $\pm$ 16.71	53.11 $\pm$ 18.38	66.12 $\pm$ 5.56
7*	R	44.62 $\pm$ 6.51	60.89 $\pm$ 5.26	88.07 $\pm$ 8.06	37.21 $\pm$ 5.28	54.41 $\pm$ 2.55
	C	64.04 $\pm$ 11.6	73.15 $\pm$ 14.26	93.85 $\pm$ 7.22	76.46 $\pm$ 8.01	76.46 $\pm$ 8.01
Post Avg.	R	58.86 $\pm$ 26.33	59.08 $\pm$ 19.22	69.13 $\pm$ 16.32	51.25 $\pm$ 16.21	54.8 $\pm$ 13.02
	C	84.78 $\pm$ 33.21	83.94 $\pm$ 22.01	100.79 $\pm$ 42.14	80.43 $\pm$ 28.63	80.91 $\pm$ 25.8
Uni. Avg.	R	46.5 $\pm$ 13.69	52.89 $\pm$ 26.57	58.46 $\pm$ 8.62	48.66 $\pm$ 12.68	52.82 $\pm$ 12.59
	C	68.42 $\pm$ 20.15	74.26 $\pm$ 23.15	75.12 $\pm$ 27.04	60.95 $\pm$ 12.35	64.24 $\pm$ 15.22
Bi. Avg.	R	75.34 $\pm$ 37.84	67.32 $\pm$ 7.45	83.37 $\pm$ 16.98	54.7 $\pm$ 25.45	57.44 $\pm$ 18.36
	C	106.58 $\pm$ 44.77	96.84 $\pm$ 21.34	135.01 $\pm$ 45.21	106.4 $\pm$ 29.49	103.12 $\pm$ 26.18

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi: Bilateral, R: Replaced, C: Contralateral

Table 25: Sagittal Ankle ROM ($^{\circ}$, mean $\pm$ std)

Participant	Limb	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	R	7.7 $\pm$ 0.2	8.0 $\pm$ 0.7	9.1 $\pm$ 0.5	7.7 $\pm$ 0.9	6.6 $\pm$ 0.3
	C	14.6 $\pm$ 0.6	14.8 $\pm$ 0.3	15.0 $\pm$ 0.2	13.1 $\pm$ 0.6	10.2 $\pm$ 0.7
2*	R	30.1 $\pm$ 2.0	30.8 $\pm$ 1.0	28.2 $\pm$ 1.5	25.4 $\pm$ 1.5	21.4 $\pm$ 2.7
	C	10.6 $\pm$ 0.6	13.9 $\pm$ 0.3	19.8 $\pm$ 0.4	12.1 $\pm$ 0.7	9.1 $\pm$ 0.3
3	R	19.9 $\pm$ 0.6	18.8 $\pm$ 0.5	17.3 $\pm$ 0.6	23.6 $\pm$ 1.0	11.3 $\pm$ 1.1
	C	2.7 $\pm$ 0.4	7.6 $\pm$ 0.5	17.2 $\pm$ 0.2	8.8 $\pm$ 0.7	5.1 $\pm$ 0.7
4	R	20.9 $\pm$ 2.8	12.8 $\pm$ 1.5	8.9 $\pm$ 1.3	18.5 $\pm$ 2.2	11.8 $\pm$ 4.2
	C	7.1 $\pm$ 0.9	11.4 $\pm$ 1.0	12.7 $\pm$ 1.9	10.3 $\pm$ 1.8	6.9 $\pm$ 0.6
5	R	6.7 $\pm$ 4.5	17.9 $\pm$ 3.1	17.6 $\pm$ 2.3	8.2 $\pm$ 7.1	8.6 $\pm$ 5.4
	C	8.7 $\pm$ 0.4	18.2 $\pm$ 0.8	28.0 $\pm$ 0.3	15.5 $\pm$ 2.2	7.2 $\pm$ 4.6
6	R	9.1 $\pm$ 3.6	13.0 $\pm$ 10.9	10.6 $\pm$ 4.9	6.8 $\pm$ 4.6	9.4 $\pm$ 2.8
	C	21.8 $\pm$ 3.6	14.2 $\pm$ 16.1	21.9 $\pm$ 3.1	13.3 $\pm$ 11.5	15.4 $\pm$ 4.9
7*	R	9.1 $\pm$ 1.4	9.3 $\pm$ 3.3	16.3 $\pm$ 2.0	9.9 $\pm$ 3.3	9.3 $\pm$ 2.2
	C	17.7 $\pm$ 5.0	15.8 $\pm$ 1.8	15.6 $\pm$ 3.4	9.9 $\pm$ 2.7	14.3 $\pm$ 3.3
Post Avg.	R	14.78 $\pm$ 2.15	15.81 $\pm$ 3.00	15.43 $\pm$ 1.85	14.29 $\pm$ 2.94	11.21 $\pm$ 2.68
	C	11.89 $\pm$ 1.65	13.72 $\pm$ 2.96	18.60 $\pm$ 1.37	11.85 $\pm$ 2.88	9.74 $\pm$ 2.17
Uni. Avg.	R	14.16 $\pm$ 2.86	15.63 $\pm$ 4.03	13.62 $\pm$ 2.26	14.28 $\pm$ 3.70	10.28 $\pm$ 3.39
	C	10.08 $\pm$ 1.32	12.87 $\pm$ 4.58	19.95 $\pm$ 1.39	11.96 $\pm$ 4.03	8.65 $\pm$ 2.71
Bi. Avg.	R	15.62 $\pm$ 1.19	16.04 $\pm$ 1.64	17.84 $\pm$ 1.32	14.30 $\pm$ 1.92	12.45 $\pm$ 1.73
	C	14.32 $\pm$ 2.08	14.85 $\pm$ 0.80	16.79 $\pm$ 1.33	11.71 $\pm$ 1.34	11.19 $\pm$ 1.44

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi: Bilateral, R: Replaced, C: Contralateral

Table 26: Peak Ankle Plantarflexion Moment (Nm, mean $\pm$ std)

Participant	Limb	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	R	-35.36 $\pm$ 2.31	-35.7 $\pm$ 1.69	-36.03 $\pm$ 1.27	-34.59 $\pm$ 2.24	-30.46 $\pm$ 1.38
	C	-26.69 $\pm$ 1.67	-25.2 $\pm$ 0.25	-31.4 $\pm$ 1.72	-26.59 $\pm$ 1.22	-27.85 $\pm$ 1.2
2*	R	-24.34 $\pm$ 1.15	-22.57 $\pm$ 0.44	-30.04 $\pm$ 1.28	-25.75 $\pm$ 0.88	-29.25 $\pm$ 1.9
	C	-40.4 $\pm$ 4.59	-30.63 $\pm$ 2.03	-35.3 $\pm$ 3.53	-40.76 $\pm$ 1.87	-39.91 $\pm$ 2.48
3	R	-20.72 $\pm$ 1.62	-18.5 $\pm$ 1.57	-19.57 $\pm$ 1.66	-15.66 $\pm$ 0.43	-17.34 $\pm$ 0.8
	C	-18.47 $\pm$ 0.76	-18.03 $\pm$ 1.53	-20.35 $\pm$ 0.77	-18.0 $\pm$ 1.53	-19.95 $\pm$ 1.22
4	R	-7.95 $\pm$ 0.82	-11.24 $\pm$ 0.56	-11.07 $\pm$ 0.27	-10.15 $\pm$ 0.52	-8.18 $\pm$ 0.75
	C	-15.81 $\pm$ 1.76	-13.43 $\pm$ 0.75	-7.72 $\pm$ 0.69	-16.08 $\pm$ 1.12	-15.09 $\pm$ 1.04
5	R	-4.0 $\pm$ 0.94	-4.97 $\pm$ 1.69	-2.59 $\pm$ 3.59	-5.1 $\pm$ 0.58	-4.7 $\pm$ 0.44
	C	-8.85 $\pm$ 0.97	-7.19 $\pm$ 1.1	-8.72 $\pm$ 2.09	-10.55 $\pm$ 2.55	-6.33 $\pm$ 0.87
6	R	-9.24 $\pm$ 0.71	-11.27 $\pm$ 4.06	-14.24 $\pm$ 0.6	-10.68 $\pm$ 1.57	-11.22 $\pm$ 0.81
	C	-11.25 $\pm$ 2.83	-11.42 $\pm$ 3.1	-24.61 $\pm$ 2.23	-11.16 $\pm$ 3.99	-12.12 $\pm$ 0.89
7*	R	-19.86 $\pm$ 0.76	-21.26 $\pm$ 1.96	-18.58 $\pm$ 0.58	-20.57 $\pm$ 1.25	-20.14 $\pm$ 1.09
	C	-15.08 $\pm$ 0.54	-15.59 $\pm$ 0.99	-15.93 $\pm$ 1.05	-15.57 $\pm$ 1.01	-15.57 $\pm$ 1.01
Post Avg.	R	-17.35 $\pm$ 10.16	-17.93 $\pm$ 9.3	-18.88 $\pm$ 10.47	-17.5 $\pm$ 9.47	-17.33 $\pm$ 9.28
	C	-19.51 $\pm$ 10.03	-17.36 $\pm$ 7.52	-20.57 $\pm$ 9.84	-19.82 $\pm$ 9.86	-19.55 $\pm$ 10.34
Uni. Avg.	R	-10.48 $\pm$ 7.18	-11.5 $\pm$ 5.53	-11.87 $\pm$ 7.11	-10.4 $\pm$ 4.32	-10.36 $\pm$ 5.36
	C	-13.6 $\pm$ 4.35	-12.52 $\pm$ 4.5	-15.35 $\pm$ 8.42	-13.95 $\pm$ 3.66	-13.37 $\pm$ 5.7
Bi. Avg.	R	-26.52 $\pm$ 7.98	-26.51 $\pm$ 7.99	-28.22 $\pm$ 8.87	-26.97 $\pm$ 7.09	-26.62 $\pm$ 5.64
	C	-27.39 $\pm$ 12.67	-23.81 $\pm$ 7.62	-27.54 $\pm$ 10.24	-27.64 $\pm$ 12.63	-27.78 $\pm$ 12.17

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi: Bilateral, R: Replaced, C: Contralateral

Table 27: Sagittal Hip ROM ($^{\circ}$, mean $\pm$ std)

Participant	Limb	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	R	24.6 $\pm$ 0.3	26.6 $\pm$ 0.2	29.3 $\pm$ 0.3	27.2 $\pm$ 0.7	24.4 $\pm$ 0.3
	C	35.2 $\pm$ 0.2	35.7 $\pm$ 0.4	35.5 $\pm$ 0.2	32.5 $\pm$ 0.3	30.2 $\pm$ 0.3
2*	R	32.8 $\pm$ 0.7	32.4 $\pm$ 0.4	34.8 $\pm$ 0.2	31.2 $\pm$ 0.6	29.8 $\pm$ 0.8
	C	30.3 $\pm$ 0.4	33.8 $\pm$ 0.3	34.1 $\pm$ 0.4	32.8 $\pm$ 0.4	28.8 $\pm$ 0.4
3	R	39.7 $\pm$ 0.9	40.3 $\pm$ 0.7	42.1 $\pm$ 0.5	38.1 $\pm$ 0.3	36.1 $\pm$ 0.4
	C	35.1 $\pm$ 1.2	38.2 $\pm$ 0.8	38.5 $\pm$ 0.8	34.6 $\pm$ 1.0	31.2 $\pm$ 0.7
4	R	43.0 $\pm$ 0.7	45.5 $\pm$ 0.7	48.8 $\pm$ 0.3	42.6 $\pm$ 0.4	40.5 $\pm$ 0.9
	C	48.1 $\pm$ 1.1	47.9 $\pm$ 0.3	49.8 $\pm$ 0.4	46.5 $\pm$ 0.4	43.2 $\pm$ 0.4
5	R	31.8 $\pm$ 0.8	31.8 $\pm$ 0.8	30.6 $\pm$ 0.9	31.1 $\pm$ 0.9	26.3 $\pm$ 0.4
	C	58.3 $\pm$ 0.6	58.9 $\pm$ 0.7	68.5 $\pm$ 0.3	57.9 $\pm$ 0.3	56.5 $\pm$ 0.4
6	R	38.4 $\pm$ 1.4	42.6 $\pm$ 4.3	46.1 $\pm$ 1.6	42.7 $\pm$ 2.3	38.6 $\pm$ 0.7
	C	42.5 $\pm$ 1.1	43.9 $\pm$ 4.9	44.0 $\pm$ 1.3	39.7 $\pm$ 3.2	35.5 $\pm$ 1.2
7*	R	33.8 $\pm$ 0.2	36.2 $\pm$ 0.8	37.4 $\pm$ 0.4	36.4 $\pm$ 1.0	33.9 $\pm$ 0.5
	C	37.3 $\pm$ 1.2	36.0 $\pm$ 0.6	36.9 $\pm$ 0.8	33.8 $\pm$ 0.7	31.7 $\pm$ 0.4
Post Avg.	R	34.87 $\pm$ 0.69	36.49 $\pm$ 1.11	38.44 $\pm$ 0.59	35.60 $\pm$ 0.89	32.79 $\pm$ 0.57
	C	40.97 $\pm$ 0.86	42.05 $\pm$ 1.16	43.91 $\pm$ 0.60	39.67 $\pm$ 0.90	36.73 $\pm$ 0.52
Uni. Avg.	R	38.24 $\pm$ 0.93	40.06 $\pm$ 1.59	41.90 $\pm$ 0.82	38.61 $\pm$ 0.99	35.36 $\pm$ 0.60
	C	46.01 $\pm$ 1.02	47.23 $\pm$ 1.68	50.21 $\pm$ 0.70	44.65 $\pm$ 1.22	41.62 $\pm$ 0.65
Bi. Avg.	R	30.38 $\pm$ 0.37	31.74 $\pm$ 0.47	33.82 $\pm$ 0.29	31.59 $\pm$ 0.76	29.36 $\pm$ 0.53
	C	34.26 $\pm$ 0.64	35.15 $\pm$ 0.46	35.52 $\pm$ 0.47	33.03 $\pm$ 0.48	30.23 $\pm$ 0.35

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi: Bilateral, R: Replaced, C: Contralateral

Table 28: Peak Hip Extension Moment (Nm, mean $\pm$ std)

Participant	Limb	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	R	-24.07 $\pm$ 0.75	-29.87 $\pm$ 1.96	-37.65 $\pm$ 1.07	-24.94 $\pm$ 2.5	-20.52 $\pm$ 2.94
	C	-24.01 $\pm$ 2.0	-30.3 $\pm$ 1.16	-35.58 $\pm$ 0.99	-14.4 $\pm$ 1.42	-16.41 $\pm$ 1.99
2*	R	-30.59 $\pm$ 1.23	-29.92 $\pm$ 1.8	-55.36 $\pm$ 2.71	-37.58 $\pm$ 3.93	-46.44 $\pm$ 1.94
	C	-37.24 $\pm$ 2.96	-31.14 $\pm$ 0.85	-29.63 $\pm$ 1.7	-41.73 $\pm$ 1.6	-37.55 $\pm$ 2.24
3	R	-35.53 $\pm$ 1.53	-23.52 $\pm$ 2.38	-28.49 $\pm$ 1.87	-28.41 $\pm$ 1.68	-21.62 $\pm$ 3.63
	C	-19.54 $\pm$ 1.06	-14.65 $\pm$ 2.46	-15.39 $\pm$ 3.27	-29.96 $\pm$ 1.54	-25.34 $\pm$ 6.58
4	R	-11.2 $\pm$ 1.64	-16.11 $\pm$ 0.87	-12.69 $\pm$ 0.74	-13.26 $\pm$ 1.05	-6.36 $\pm$ 1.58
	C	-19.02 $\pm$ 2.43	-17.37 $\pm$ 0.95	3.27 $\pm$ 0.94	-17.99 $\pm$ 1.87	-17.88 $\pm$ 2.09
5	R	-19.53 $\pm$ 1.1	-22.77 $\pm$ 0.62	-22.27 $\pm$ 2.04	-18.29 $\pm$ 1.55	-14.55 $\pm$ 0.95
	C	-26.31 $\pm$ 1.06	-12.13 $\pm$ 0.6	-17.79 $\pm$ 0.6	-18.79 $\pm$ 1.27	-19.45 $\pm$ 0.61
6	R	-19.9 $\pm$ 3.31	-19.4 $\pm$ 5.26	-29.59 $\pm$ 3.01	-26.36 $\pm$ 7.06	-25.6 $\pm$ 1.74
	C	-14.23 $\pm$ 3.93	-15.26 $\pm$ 5.67	-62.17 $\pm$ 4.96	-16.28 $\pm$ 5.61	-23.14 $\pm$ 1.05
7*	R	-23.0 $\pm$ 1.63	-23.13 $\pm$ 4.2	-16.39 $\pm$ 0.9	-25.42 $\pm$ 2.12	-19.33 $\pm$ 2.12
	C	-15.23 $\pm$ 1.71	-14.11 $\pm$ 0.83	-14.26 $\pm$ 1.45	-15.34 $\pm$ 1.76	-15.34 $\pm$ 1.76
Post Avg.	R	-23.4 $\pm$ 7.32	-23.53 $\pm$ 4.69	-28.92 $\pm$ 13.31	-24.89 $\pm$ 7.12	-22.06 $\pm$ 11.46
	C	-22.22 $\pm$ 7.33	-19.28 $\pm$ 7.38	-24.51 $\pm$ 19.17	-22.07 $\pm$ 9.36	-22.16 $\pm$ 7.1
Uni. Avg.	R	-21.54 $\pm$ 10.15	-20.45 $\pm$ 3.4	-23.26 $\pm$ 7.75	-21.58 $\pm$ 7.06	-17.03 $\pm$ 8.46
	C	-19.78 $\pm$ 4.97	-14.85 $\pm$ 2.16	-23.02 $\pm$ 27.75	-20.76 $\pm$ 6.23	-21.45 $\pm$ 3.4
Bi. Avg.	R	-25.89 $\pm$ 4.11	-27.64 $\pm$ 3.91	-36.47 $\pm$ 19.51	-29.31 $\pm$ 7.16	-28.76 $\pm$ 15.32
	C	-25.49 $\pm$ 11.08	-25.18 $\pm$ 9.6	-26.49 $\pm$ 11.0	-23.82 $\pm$ 15.51	-23.1 $\pm$ 12.53

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi: Bilateral, R: Replaced, C: Contralateral

Table 29: Vertical Pedal Reaction Force (N, mean $\pm$ std) - Difference

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	3.89 $\pm$ 19.25	-15.68 $\pm$ 23.20	58.25 $\pm$ 11.07	44.91 $\pm$ 17.17	77.60 $\pm$ 8.71
2*	68.42 $\pm$ 24.79	38.33 $\pm$ 11.61	30.69 $\pm$ 24.30	63.78 $\pm$ 10.08	59.38 $\pm$ 13.70
3	-40.88 $\pm$ 9.23	-21.31 $\pm$ 15.51	32.88 $\pm$ 10.38	-6.20 $\pm$ 8.60	30.46 $\pm$ 19.32
4	1.24 $\pm$ 10.86	-10.30 $\pm$ 15.15	-24.40 $\pm$ 8.86	12.33 $\pm$ 25.42	27.62 $\pm$ 13.99
5	57.59 $\pm$ 11.02	17.89 $\pm$ 5.83	-0.11 $\pm$ 3.15	45.18 $\pm$ 13.23	52.19 $\pm$ 11.05
6	18.53 $\pm$ 24.44	6.14 $\pm$ 16.68	54.30 $\pm$ 10.02	-4.72 $\pm$ 17.85	18.97 $\pm$ 13.81
7*	-32.48 $\pm$ 6.86	-56.31 $\pm$ 24.18	-6.91 $\pm$ 16.02	-15.71 $\pm$ 6.19	-27.67 $\pm$ 14.12
Post Avg.	10.90 $\pm$ 41.34	-5.89 $\pm$ 30.44	20.67 $\pm$ 31.67	19.94 $\pm$ 31.10	34.08 $\pm$ 34.05
Uni. Avg.	9.12 $\pm$ 40.82	-1.89 $\pm$ 17.36	15.67 $\pm$ 34.85	11.65 $\pm$ 23.88	32.31 $\pm$ 14.13
Bi. Avg.	13.28 $\pm$ 51.10	-11.22 $\pm$ 47.48	27.34 $\pm$ 32.71	30.99 $\pm$ 41.53	36.44 $\pm$ 56.26

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 30: Sagittal Knee ROM ($^{\circ}$, mean $\pm$ std) - Difference

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	8.78 $\pm$ 0.43	6.29 $\pm$ 0.41	2.9 $\pm$ 0.16	1.89 $\pm$ 0.73	0.13 $\pm$ 0.86
2*	-5.74 $\pm$ 0.92	-1.82 $\pm$ 1.95	-1.45 $\pm$ 0.74	-0.61 $\pm$ 1.69	-2.09 $\pm$ 0.95
3	-7.2 $\pm$ 0.49	-3.02 $\pm$ 1.35	-5.32 $\pm$ 1.28	-8.14 $\pm$ 0.73	-5.69 $\pm$ 0.32
4	-4.7 $\pm$ 1.18	-3.88 $\pm$ 0.7	-5.66 $\pm$ 0.48	-8.09 $\pm$ 0.82	-5.96 $\pm$ 1.3
5	18.9 $\pm$ 1.02	12.17 $\pm$ 0.46	22.64 $\pm$ 0.55	11.25 $\pm$ 0.85	21.49 $\pm$ 0.53
6	15.11 $\pm$ 1.7	7.38 $\pm$ 0.94	4.39 $\pm$ 1.01	3.46 $\pm$ 1.8	3.15 $\pm$ 1.03
7*	11.42 $\pm$ 0.68	6.1 $\pm$ 0.62	1.02 $\pm$ 0.43	-1.12 $\pm$ 0.39	2.3 $\pm$ 0.83
Post Avg.	5.22 $\pm$ 10.87	3.32 $\pm$ 6.19	2.65 $\pm$ 9.62	-0.2 $\pm$ 6.77	1.9 $\pm$ 9.34
Uni. Avg.	5.53 $\pm$ 13.38	3.16 $\pm$ 7.89	4.01 $\pm$ 13.26	-0.38 $\pm$ 9.48	3.25 $\pm$ 12.88
Bi. Avg.	4.82 $\pm$ 9.24	3.52 $\pm$ 4.63	0.83 $\pm$ 2.18	0.05 $\pm$ 1.61	0.11 $\pm$ 2.2

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 31: Knee Extension Angle ($^{\circ}$, mean $\pm$ std) - Difference

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	9.04 $\pm$ 0.48	5.93 $\pm$ 0.38	2.37 $\pm$ 0.32	5.00 $\pm$ 0.65	3.97 $\pm$ 0.64
2*	-6.43 $\pm$ 0.95	3.34 $\pm$ 1.63	0.93 $\pm$ 0.68	1.80 $\pm$ 1.50	0.23 $\pm$ 0.60
3	1.95 $\pm$ 0.30	2.47 $\pm$ 1.30	-1.62 $\pm$ 1.30	-3.74 $\pm$ 0.64	-1.96 $\pm$ 0.27
4	5.97 $\pm$ 0.77	3.34 $\pm$ 0.88	0.22 $\pm$ 0.44	-2.09 $\pm$ 0.79	-1.77 $\pm$ 0.46
5	15.61 $\pm$ 0.47	9.63 $\pm$ 0.93	12.26 $\pm$ 0.41	5.47 $\pm$ 0.86	11.24 $\pm$ 0.55
6	7.52 $\pm$ 1.21	2.57 $\pm$ 1.28	0.72 $\pm$ 0.44	-0.13 $\pm$ 0.63	0.08 $\pm$ 1.23
7*	10.45 $\pm$ 0.45	6.62 $\pm$ 0.33	3.00 $\pm$ 0.56	3.99 $\pm$ 0.42	5.37 $\pm$ 0.41
Post Avg.	6.30 $\pm$ 7.00	4.84 $\pm$ 2.66	2.55 $\pm$ 4.53	1.47 $\pm$ 3.59	2.45 $\pm$ 4.76
Uni. Avg.	7.76 $\pm$ 5.73	4.50 $\pm$ 3.44	2.90 $\pm$ 6.32	-0.12 $\pm$ 4.01	1.90 $\pm$ 6.30
Bi. Avg.	4.35 $\pm$ 9.37	5.30 $\pm$ 1.73	2.10 $\pm$ 1.06	3.60 $\pm$ 1.64	3.19 $\pm$ 2.66

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 32: Knee Extension Moment (Nm, mean $\pm$ std) - Difference

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	20.40 $\pm$ 2.03	11.02 $\pm$ 3.35	20.72 $\pm$ 1.05	9.59 $\pm$ 3.24	28.13 $\pm$ 4.80
2*	0.64 $\pm$ 2.74	5.50 $\pm$ 3.31	15.94 $\pm$ 2.98	5.85 $\pm$ 2.23	7.87 $\pm$ 4.14
3	5.50 $\pm$ 1.28	-0.52 $\pm$ 1.37	10.21 $\pm$ 2.85	1.89 $\pm$ 1.49	3.93 $\pm$ 2.01
4	-2.93 $\pm$ 1.93	-1.93 $\pm$ 1.23	2.22 $\pm$ 1.59	0.27 $\pm$ 0.75	-2.04 $\pm$ 1.60
5	6.07 $\pm$ 2.00	14.13 $\pm$ 1.17	10.38 $\pm$ 1.62	9.70 $\pm$ 2.33	8.09 $\pm$ 2.20
6	4.90 $\pm$ 1.93	3.78 $\pm$ 2.99	-7.48 $\pm$ 3.60	1.70 $\pm$ 6.19	5.13 $\pm$ 3.93
7*	7.26 $\pm$ 2.53	6.70 $\pm$ 8.14	3.54 $\pm$ 1.54	16.17 $\pm$ 3.47	10.71 $\pm$ 4.26
Post Avg.	5.98 $\pm$ 7.29	5.53 $\pm$ 5.78	7.93 $\pm$ 9.38	6.45 $\pm$ 5.73	8.83 $\pm$ 9.43
Uni. Avg.	3.39 $\pm$ 4.24	3.86 $\pm$ 7.26	3.83 $\pm$ 8.45	3.39 $\pm$ 4.27	3.78 $\pm$ 4.25
Bi. Avg.	9.43 $\pm$ 10.06	7.74 $\pm$ 2.90	13.40 $\pm$ 8.87	10.54 $\pm$ 5.22	15.57 $\pm$ 10.97

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 33: Knee Abduction Moment (Nm, mean $\pm$ std) - Difference

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	6.15 $\pm$ 0.42	5.45 $\pm$ 1.19	5.93 $\pm$ 0.79	4.29 $\pm$ 1.28	4.25 $\pm$ 1.90
2*	-6.46 $\pm$ 2.84	-8.50 $\pm$ 1.50	-6.70 $\pm$ 1.90	-5.29 $\pm$ 2.43	-9.68 $\pm$ 1.45
3	-4.02 $\pm$ 0.87	-4.41 $\pm$ 0.34	-2.95 $\pm$ 0.52	-2.79 $\pm$ 0.93	-3.89 $\pm$ 0.51
4	-5.16 $\pm$ 0.74	-4.29 $\pm$ 0.33	-1.73 $\pm$ 0.24	-2.31 $\pm$ 0.30	-4.27 $\pm$ 0.41
5	-0.13 $\pm$ 1.39	3.78 $\pm$ 1.28	2.50 $\pm$ 1.62	1.35 $\pm$ 0.70	1.06 $\pm$ 0.71
6	0.19 $\pm$ 3.00	-0.87 $\pm$ 0.93	-5.60 $\pm$ 0.98	-1.12 $\pm$ 0.61	-1.65 $\pm$ 0.59
7*	2.74 $\pm$ 2.35	8.21 $\pm$ 1.01	5.66 $\pm$ 1.39	5.73 $\pm$ 1.21	6.55 $\pm$ 1.06
Post Avg.	-0.96 $\pm$ 4.54	-0.09 $\pm$ 6.09	-0.41 $\pm$ 5.17	-0.02 $\pm$ 3.99	-1.09 $\pm$ 5.53
Uni. Avg.	-2.28 $\pm$ 2.71	-1.45 $\pm$ 3.85	-1.94 $\pm$ 3.38	-1.22 $\pm$ 1.85	-2.19 $\pm$ 2.45
Bi. Avg.	0.81 $\pm$ 6.52	1.72 $\pm$ 8.96	1.63 $\pm$ 7.22	1.58 $\pm$ 5.99	0.37 $\pm$ 8.78

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 34: Knee Power (Nm/s, mean $\pm$ std) - Difference

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	89.50 $\pm$ 6.54	48.97 $\pm$ 13.14	85.89 $\pm$ 3.44	92.41 $\pm$ 11.87	88.02 $\pm$ 14.12
2*	-15.20 $\pm$ 11.44	27.34 $\pm$ 14.07	63.27 $\pm$ 14.84	23.45 $\pm$ 11.87	26.97 $\pm$ 10.77
3	28.45 $\pm$ 3.81	2.29 $\pm$ 5.02	32.26 $\pm$ 10.93	1.55 $\pm$ 5.62	6.52 $\pm$ 6.91
4	2.71 $\pm$ 6.59	0.89 $\pm$ 5.95	16.45 $\pm$ 4.49	6.80 $\pm$ 4.51	-3.42 $\pm$ 5.10
5	25.76 $\pm$ 8.97	62.22 $\pm$ 7.79	46.38 $\pm$ 6.63	32.90 $\pm$ 8.79	21.41 $\pm$ 8.04
6	30.78 $\pm$ 10.61	20.06 $\pm$ 15.47	-28.44 $\pm$ 15.52	7.91 $\pm$ 12.11	21.14 $\pm$ 14.06
7*	19.42 $\pm$ 8.03	12.25 $\pm$ 15.56	5.78 $\pm$ 4.36	39.25 $\pm$ 6.28	22.05 $\pm$ 8.45
Post Avg.	25.92 $\pm$ 32.52	24.86 $\pm$ 23.27	31.66 $\pm$ 38.01	29.18 $\pm$ 31.26	26.10 $\pm$ 29.30
Uni. Avg.	21.93 $\pm$ 12.97	21.37 $\pm$ 28.60	16.66 $\pm$ 32.46	12.29 $\pm$ 14.02	11.41 $\pm$ 12.09
Bi. Avg.	31.24 $\pm$ 53.34	29.52 $\pm$ 18.46	51.65 $\pm$ 41.30	51.70 $\pm$ 36.13	45.68 $\pm$ 36.75

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 35: Sagittal Ankle ROM ($^{\circ}$, mean $\pm$ std) - Difference

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	-6.98 $\pm$ 1.1	-6.75 $\pm$ 2.12	-5.92 $\pm$ 1.94	-5.41 $\pm$ 2.38	-3.58 $\pm$ 1.58
2*	19.5 $\pm$ 2.09	16.86 $\pm$ 3.21	8.36 $\pm$ 2.96	13.26 $\pm$ 1.39	12.37 $\pm$ 2.52
3	17.21 $\pm$ 1.52	11.24 $\pm$ 1.97	0.1 $\pm$ 2.65	14.83 $\pm$ 1.34	6.2 $\pm$ 0.79
4	13.81 $\pm$ 2.01	1.34 $\pm$ 1.89	-3.75 $\pm$ 1.88	8.25 $\pm$ 3.74	4.92 $\pm$ 3.0
5	15.34 $\pm$ 12.3	36.12 $\pm$ 5.48	45.58 $\pm$ 5.06	23.67 $\pm$ 11.47	15.84 $\pm$ 4.85
6	-12.69 $\pm$ 8.12	-1.2 $\pm$ 3.3	-11.23 $\pm$ 4.88	-6.52 $\pm$ 5.17	-5.99 $\pm$ 5.68
7*	-8.63 $\pm$ 4.84	-6.54 $\pm$ 3.51	0.7 $\pm$ 3.91	-0.05 $\pm$ 2.67	-4.99 $\pm$ 3.76
Post Avg.	5.37 $\pm$ 14.06	7.3 $\pm$ 15.46	4.83 $\pm$ 18.97	6.86 $\pm$ 11.3	3.54 $\pm$ 8.68
Uni. Avg.	8.42 $\pm$ 14.14	11.87 $\pm$ 17.03	7.68 $\pm$ 25.7	10.06 $\pm$ 12.73	5.24 $\pm$ 8.93
Bi. Avg.	1.3 $\pm$ 15.79	1.19 $\pm$ 13.57	1.05 $\pm$ 7.15	2.6 $\pm$ 9.61	1.27 $\pm$ 9.64

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 36: Ankle Plantarflexion Moment (Nm, mean $\pm$ std) - Difference

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	8.66 $\pm$ 3.51	10.50 $\pm$ 1.92	4.63 $\pm$ 1.29	8.00 $\pm$ 3.20	2.61 $\pm$ 1.31
2*	-16.07 $\pm$ 4.29	-8.06 $\pm$ 2.64	-5.25 $\pm$ 4.36	-15.02 $\pm$ 2.04	-10.66 $\pm$ 2.14
3	2.25 $\pm$ 1.43	0.47 $\pm$ 1.41	-0.78 $\pm$ 1.53	-2.33 $\pm$ 1.84	-2.61 $\pm$ 1.73
4	-7.86 $\pm$ 1.56	-2.19 $\pm$ 1.28	3.35 $\pm$ 0.87	-5.93 $\pm$ 1.74	-6.91 $\pm$ 1.16
5	-4.85 $\pm$ 1.20	-2.22 $\pm$ 2.76	-6.13 $\pm$ 4.58	-5.45 $\pm$ 2.72	-1.63 $\pm$ 0.73
6	-2.00 $\pm$ 3.25	-0.15 $\pm$ 2.55	-10.37 $\pm$ 2.10	-0.47 $\pm$ 2.94	-0.90 $\pm$ 1.54
7*	4.78 $\pm$ 0.90	5.67 $\pm$ 2.92	2.65 $\pm$ 0.90	5.00 $\pm$ 0.75	4.57 $\pm$ 2.35
Post Avg.	-2.16 $\pm$ 8.35	0.57 $\pm$ 5.99	-1.70 $\pm$ 5.67	-2.31 $\pm$ 7.61	-2.22 $\pm$ 5.24
Uni. Avg.	-3.12 $\pm$ 4.30	-1.02 $\pm$ 1.39	-3.48 $\pm$ 6.01	-3.55 $\pm$ 2.60	-3.01 $\pm$ 2.69
Bi. Avg.	-0.88 $\pm$ 13.30	2.70 $\pm$ 9.63	0.68 $\pm$ 5.23	-0.67 $\pm$ 12.51	-1.16 $\pm$ 8.29

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 37: Sagittal Hip ROM ($^{\circ}$, mean $\pm$ std) - Difference

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	-10.63 $\pm$ 1.02	-9.07 $\pm$ 1.1	-6.21 $\pm$ 0.64	-5.29 $\pm$ 1.31	-5.82 $\pm$ 0.95
2*	2.51 $\pm$ 0.72	-1.37 $\pm$ 1.27	0.66 $\pm$ 0.55	-1.59 $\pm$ 1.34	0.96 $\pm$ 0.32
3	4.58 $\pm$ 0.59	2.1 $\pm$ 0.82	3.54 $\pm$ 0.93	3.54 $\pm$ 0.99	4.95 $\pm$ 0.6
4	-5.12 $\pm$ 1.65	-2.38 $\pm$ 1.68	-0.96 $\pm$ 1.11	-3.91 $\pm$ 0.89	-2.78 $\pm$ 0.99
5	-26.47 $\pm$ 1.55	-27.09 $\pm$ 1.32	-37.88 $\pm$ 0.39	-26.83 $\pm$ 1.45	-30.23 $\pm$ 1.03
6	-4.06 $\pm$ 1.67	-1.31 $\pm$ 2.23	2.07 $\pm$ 1.92	3.0 $\pm$ 1.16	3.03 $\pm$ 1.83
7*	-3.53 $\pm$ 1.38	0.19 $\pm$ 1.1	0.45 $\pm$ 1.16	2.56 $\pm$ 0.86	2.26 $\pm$ 0.78
Post Avg.	-6.1 $\pm$ 10.29	-5.56 $\pm$ 10.11	-5.48 $\pm$ 14.62	-4.07 $\pm$ 10.63	-3.95 $\pm$ 12.15
Uni. Avg.	-7.77 $\pm$ 13.2	-7.17 $\pm$ 13.42	-8.31 $\pm$ 19.81	-6.05 $\pm$ 14.26	-6.26 $\pm$ 16.32
Bi. Avg.	-3.88 $\pm$ 6.58	-3.42 $\pm$ 4.96	-1.7 $\pm$ 3.9	-1.44 $\pm$ 3.93	-0.87 $\pm$ 4.34

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 38: Hip Extension Moment (Nm, mean $\pm$ std) - Difference

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)	C4 (155/155)	C5 (140/140)
1*	21.06 $\pm$ 6.22	9.91 $\pm$ 2.40	13.94 $\pm$ 2.04	19.74 $\pm$ 3.96	3.96 $\pm$ 3.02
2*	1.59 $\pm$ 4.21	-6.33 $\pm$ 3.81	18.99 $\pm$ 4.54	-3.53 $\pm$ 3.78	12.92 $\pm$ 2.13
3	26.62 $\pm$ 2.73	5.09 $\pm$ 3.49	12.30 $\pm$ 2.81	-6.16 $\pm$ 1.65	8.87 $\pm$ 2.52
4	0.05 $\pm$ 0.93	5.26 $\pm$ 1.45	25.81 $\pm$ 1.97	2.09 $\pm$ 0.98	-2.60 $\pm$ 0.99
5	-11.28 $\pm$ 3.45	11.01 $\pm$ 2.17	3.47 $\pm$ 1.73	-3.16 $\pm$ 1.90	-5.31 $\pm$ 1.68
6	8.32 $\pm$ 4.47	-1.04 $\pm$ 4.33	-32.62 $\pm$ 3.27	2.44 $\pm$ 3.43	-2.59 $\pm$ 3.00
7*	5.04 $\pm$ 2.57	9.18 $\pm$ 2.93	1.43 $\pm$ 2.70	11.23 $\pm$ 3.56	2.88 $\pm$ 6.11
Post Avg.	7.34 $\pm$ 12.90	4.73 $\pm$ 6.35	6.19 $\pm$ 19.07	3.24 $\pm$ 9.25	2.59 $\pm$ 6.64
Uni. Avg.	5.93 $\pm$ 15.96	5.08 $\pm$ 4.92	2.24 $\pm$ 24.99	-1.20 $\pm$ 4.18	-0.41 $\pm$ 6.32
Bi. Avg.	9.23 $\pm$ 10.39	4.25 $\pm$ 9.17	11.45 $\pm$ 9.04	9.15 $\pm$ 11.77	6.59 $\pm$ 5.51

*: Indicates bilateral TKA participant, Avg.: Average, Uni.: Unilateral, Bi.: Bilateral

Table 39: Replaced Knee Total Contact Force (N, mean $\pm$ std)

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)
1*	1398.33 $\pm$ 70.33	1309.88 $\pm$ 75.04	1507.76 $\pm$ 22.63
2*	787.27 $\pm$ 75.47	782.03 $\pm$ 51.69	1162.64 $\pm$ 106.29
3	825.59 $\pm$ 47.23	1105.43 $\pm$ 43.67	983.95 $\pm$ 26.41
4	966.77 $\pm$ 167.71	990.11 $\pm$ 99.38	1118.90 $\pm$ 125.75
5	412.98 $\pm$ 11.99	479.76 $\pm$ 39.92	665.76 $\pm$ 9.90
6	772.85 $\pm$ 77.34	724.00 $\pm$ 254.77	929.67 $\pm$ 83.15
7*	661.31 $\pm$ 31.52	661.73 $\pm$ 60.66	885.32 $\pm$ 78.40
Average	830.99 $\pm$ 301.94	857.75 $\pm$ 282.16	1036.28 $\pm$ 264.34

*: Indicates bilateral TKA participant

Table 40: Replaced Knee Medial Compartment Contact Force (N, mean $\pm$ std)

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)
1*	749.55 $\pm$ 68.22	600.77 $\pm$ 32.49	757.98 $\pm$ 8.18
2*	356.84 $\pm$ 55.10	417.26 $\pm$ 37.00	550.92 $\pm$ 78.91
3	311.16 $\pm$ 16.97	464.24 $\pm$ 29.55	397.19 $\pm$ 21.70
4	620.92 $\pm$ 103.39	636.06 $\pm$ 43.63	619.63 $\pm$ 71.43
5	134.66 $\pm$ 4.96	156.00 $\pm$ 11.55	239.18 $\pm$ 18.40
6	372.10 $\pm$ 67.05	394.46 $\pm$ 159.14	415.23 $\pm$ 43.91
7*	499.53 $\pm$ 30.74	481.30 $\pm$ 24.17	537.59 $\pm$ 48.05
Average	432.71 $\pm$ 203.15	447.01 $\pm$ 153.66	502.53 $\pm$ 168.63

*: Indicates bilateral TKA participant

Table 41: Replaced Knee Lateral Compartment Contact Force (N, mean $\pm$ std)

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)
1*	704.45 $\pm$ 45.23	752.74 $\pm$ 61.09	775.22 $\pm$ 39.17
2*	443.13 $\pm$ 25.77	371.68 $\pm$ 21.87	611.72 $\pm$ 31.11
3	523.94 $\pm$ 39.09	642.89 $\pm$ 21.32	589.68 $\pm$ 18.92
4	384.14 $\pm$ 64.93	436.91 $\pm$ 36.15	542.16 $\pm$ 56.76
5	322.00 $\pm$ 13.25	342.28 $\pm$ 33.74	459.81 $\pm$ 13.79
6	418.80 $\pm$ 49.13	355.44 $\pm$ 113.45	523.04 $\pm$ 48.19
7*	243.02 $\pm$ 19.91	252.79 $\pm$ 49.14	360.87 $\pm$ 31.96
Average	434.20 $\pm$ 148.92	446.81 $\pm$ 180.72	551.79 $\pm$ 129.60

*: Indicates bilateral TKA participant

Table 42: Peak Quadriceps Muscle Force (N, mean $\pm$ std)

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)
1*	1586.72 $\pm$ 60.93	1693.96 $\pm$ 34.46	2121.62 $\pm$ 38.23
2*	967.51 $\pm$ 136.28	774.17 $\pm$ 68.21	604.97 $\pm$ 73.74
3	909.76 $\pm$ 49.40	1230.78 $\pm$ 129.08	991.18 $\pm$ 34.71
4	957.72 $\pm$ 167.05	954.92 $\pm$ 109.50	1247.68 $\pm$ 74.54
5	659.42 $\pm$ 72.77	660.68 $\pm$ 20.72	885.15 $\pm$ 56.61
6	833.98 $\pm$ 127.23	687.33 $\pm$ 237.61	965.14 $\pm$ 148.93
7*	610.06 $\pm$ 46.76	597.74 $\pm$ 61.15	897.40 $\pm$ 67.18
Average	932.17 $\pm$ 321.10	942.80 $\pm$ 395.74	1101.88 $\pm$ 487.82

*: Indicates bilateral TKA participant

Table 43: Peak Hamstrings Muscle Force (N, mean $\pm$ std)

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)
1*	278.71 $\pm$ 43.42	315.27 $\pm$ 39.94	261.15 $\pm$ 17.04
2*	380.49 $\pm$ 21.93	421.22 $\pm$ 53.86	766.42 $\pm$ 20.32
3	221.58 $\pm$ 26.01	243.10 $\pm$ 29.54	319.84 $\pm$ 32.24
4	342.64 $\pm$ 48.32	415.54 $\pm$ 36.33	376.08 $\pm$ 31.18
5	190.90 $\pm$ 19.88	306.54 $\pm$ 10.47	366.42 $\pm$ 27.17
6	293.89 $\pm$ 67.16	301.36 $\pm$ 101.91	217.67 $\pm$ 23.39
7*	311.76 $\pm$ 52.56	244.01 $\pm$ 99.23	166.96 $\pm$ 9.16
Average	288.57 $\pm$ 65.91	321.00 $\pm$ 72.58	353.50 $\pm$ 197.52

*: Indicates bilateral TKA participant

Table 44: Peak Gastrocnemius Muscle Force (N, mean $\pm$ std)

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)
1*	343.32 $\pm$ 48.21	335.48 $\pm$ 24.78	323.84 $\pm$ 45.61
2*	474.09 $\pm$ 93.09	415.51 $\pm$ 26.61	570.57 $\pm$ 39.54
3	68.12 $\pm$ 8.76	77.77 $\pm$ 9.86	222.44 $\pm$ 18.84
4	0.00 $\pm$ 0.00	0.00 $\pm$ 0.00	0.00 $\pm$ 0.00
5	12.19 $\pm$ 1.41	14.50 $\pm$ 2.25	17.35 $\pm$ 2.94
6	90.84 $\pm$ 17.36	130.04 $\pm$ 37.55	102.79 $\pm$ 14.68
7*	360.47 $\pm$ 33.22	299.69 $\pm$ 32.00	401.69 $\pm$ 21.88
Average	192.72 $\pm$ 193.92	181.86 $\pm$ 166.67	234.10 $\pm$ 211.55

*: Indicates bilateral TKA participant

Table 45: Peak Rectus Femoris Muscle Force (N, mean $\pm$ std)

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)
1*	243.11 $\pm$ 66.72	286.03 $\pm$ 26.85	207.51 $\pm$ 12.30
2*	255.07 $\pm$ 51.30	174.59 $\pm$ 28.87	48.77 $\pm$ 8.05
3	234.83 $\pm$ 26.56	244.51 $\pm$ 30.14	117.86 $\pm$ 12.79
4	135.25 $\pm$ 24.06	100.22 $\pm$ 15.96	205.62 $\pm$ 31.77
5	328.26 $\pm$ 4.30	328.57 $\pm$ 21.82	387.52 $\pm$ 18.92
6	203.60 $\pm$ 34.13	142.04 $\pm$ 54.77	146.40 $\pm$ 35.30
7*	134.71 $\pm$ 21.66	118.91 $\pm$ 19.86	217.61 $\pm$ 22.61
Average	219.26 $\pm$ 68.83	199.27 $\pm$ 87.98	190.19 $\pm$ 105.97

*: Indicates bilateral TKA participant

Table 46: Peak Semitendinosus Muscle Force (N, mean $\pm$ std)

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)
1*	23.28 $\pm$ 3.47	26.15 $\pm$ 1.94	23.15 $\pm$ 2.70
2*	35.33 $\pm$ 2.69	39.53 $\pm$ 4.79	72.61 $\pm$ 6.04
3	8.46 $\pm$ 2.33	5.52 $\pm$ 0.02	12.88 $\pm$ 5.41
4	20.15 $\pm$ 3.07	18.40 $\pm$ 1.52	27.77 $\pm$ 1.19
5	5.62 $\pm$ 0.01	5.90 $\pm$ 0.39	8.80 $\pm$ 5.68
6	11.66 $\pm$ 4.89	6.99 $\pm$ 1.73	10.57 $\pm$ 2.62
7*	15.94 $\pm$ 4.76	17.58 $\pm$ 7.74	5.90 $\pm$ 0.43
Average	17.21 $\pm$ 10.14	17.15 $\pm$ 12.57	23.10 $\pm$ 23.22

*: Indicates bilateral TKA participant

Table 47: Peak Medial Gastrocnemius Muscle Force (N, mean $\pm$ std)

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)
1*	278.23 $\pm$ 45.59	273.24 $\pm$ 21.15	250.15 $\pm$ 31.40
2*	386.22 $\pm$ 75.67	339.74 $\pm$ 21.75	463.77 $\pm$ 32.55
3	52.43 $\pm$ 7.31	61.71 $\pm$ 8.18	178.28 $\pm$ 15.43
4	0.00 $\pm$ 0.00	0.00 $\pm$ 0.00	0.00 $\pm$ 0.00
5	7.71 $\pm$ 0.99	9.33 $\pm$ 1.60	11.95 $\pm$ 2.66
6	73.85 $\pm$ 16.99	104.53 $\pm$ 31.23	82.73 $\pm$ 12.25
7*	294.93 $\pm$ 27.42	245.01 $\pm$ 26.43	327.72 $\pm$ 18.25
Average	156.20 $\pm$ 158.66	147.65 $\pm$ 136.83	187.80 $\pm$ 171.73

*: Indicates bilateral TKA participant

Table 48: Peak Tibialis Anterior Muscle Force (N, mean $\pm$ std)

Participant	C1 (140/170)	C2 (155/170)	C3 (170/170)
1*	79.03 $\pm$ 6.13	79.85 $\pm$ 4.49	77.71 $\pm$ 7.52
2*	4.59 $\pm$ 0.22	4.93 $\pm$ 0.37	6.22 $\pm$ 0.65
3	5.62 $\pm$ 0.17	4.60 $\pm$ 0.19	8.48 $\pm$ 1.68
4	152.47 $\pm$ 2.30	159.83 $\pm$ 3.34	155.90 $\pm$ 1.74
5	7.48 $\pm$ 0.39	8.30 $\pm$ 0.29	8.00 $\pm$ 0.60
6	84.56 $\pm$ 23.69	123.73 $\pm$ 27.11	62.05 $\pm$ 10.68
7*	94.56 $\pm$ 12.81	65.34 $\pm$ 29.01	60.32 $\pm$ 8.20
Average	61.19 $\pm$ 56.99	63.80 $\pm$ 62.08	54.10 $\pm$ 54.04

*: Indicates bilateral TKA participant

Appendix B: Figures



Figure 1: Data collection setup of ergometer, motion capture, and participant

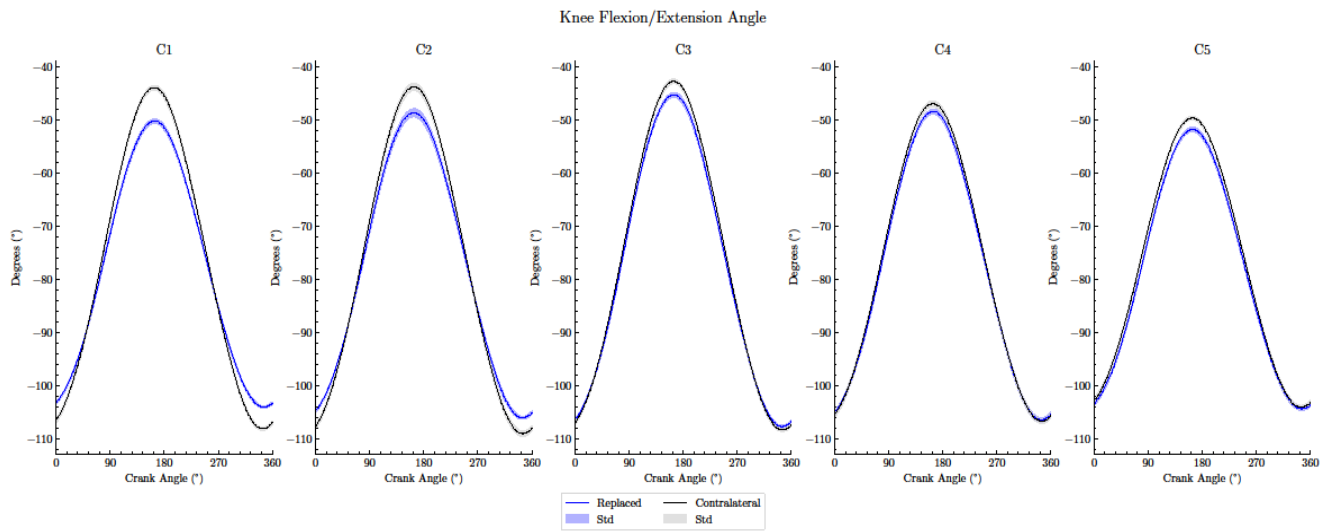


Figure 2: Ensemble curves for sagittal knee joint angle across all conditions. C1: 140/170, C2: 155/170, C3: 170/170, C4: 155/155, C5: 140/140

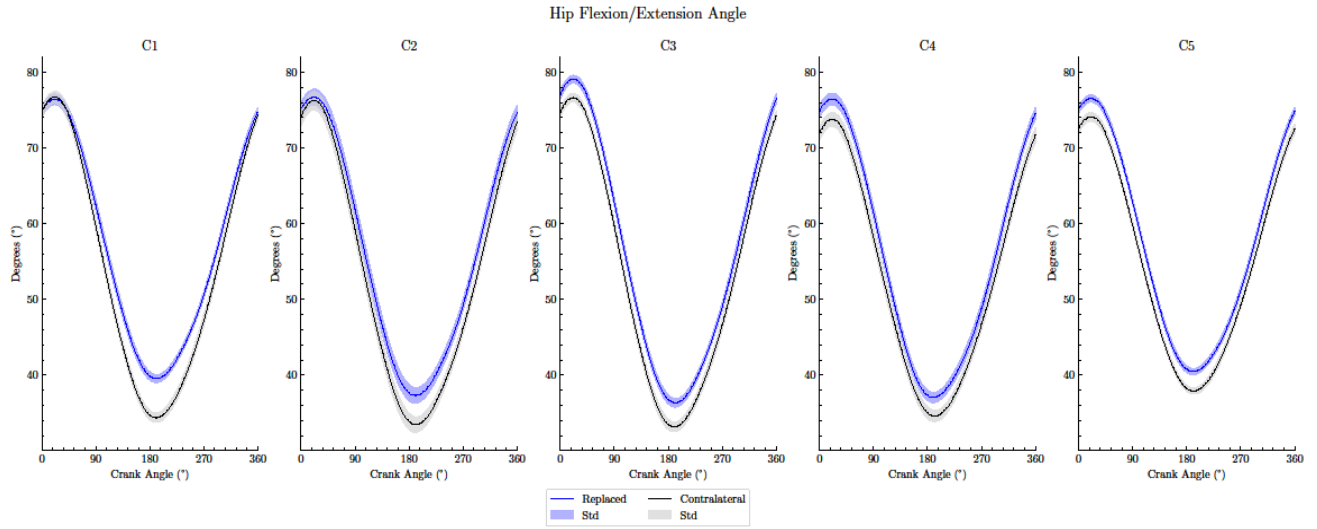


Figure 3: Ensemble curves for sagittal hip joint angle across all conditions. C1: 140/170, C2: 155/170, C3: 170/170, C4: 155/155, C5: 140/140

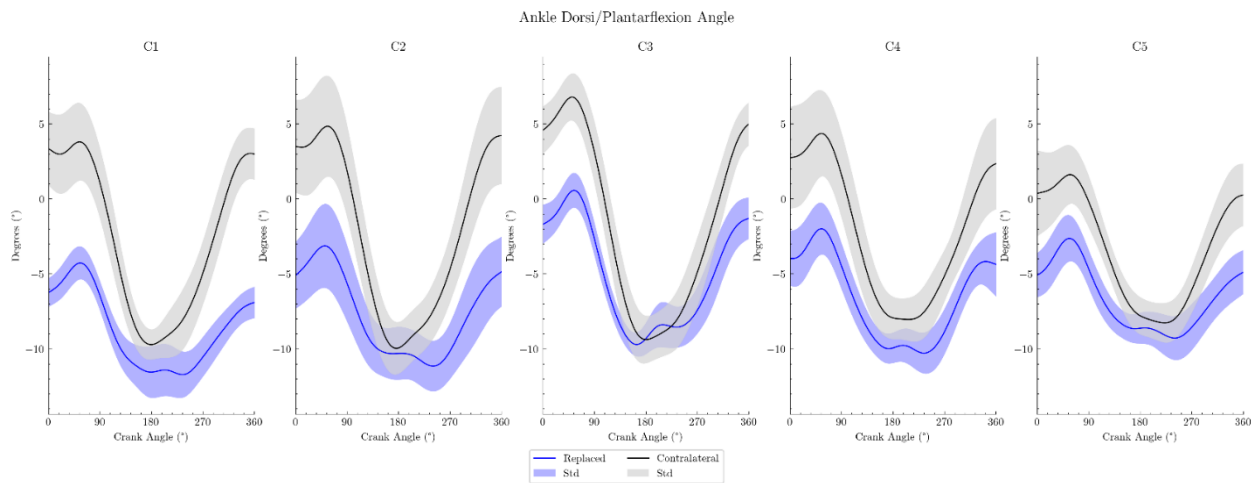


Figure 4: Ensemble curves for sagittal ankle joint angle across all conditions. C1: 140/170, C2: 155/170, C3: 170/170, C4: 155/155, C5: 140/140

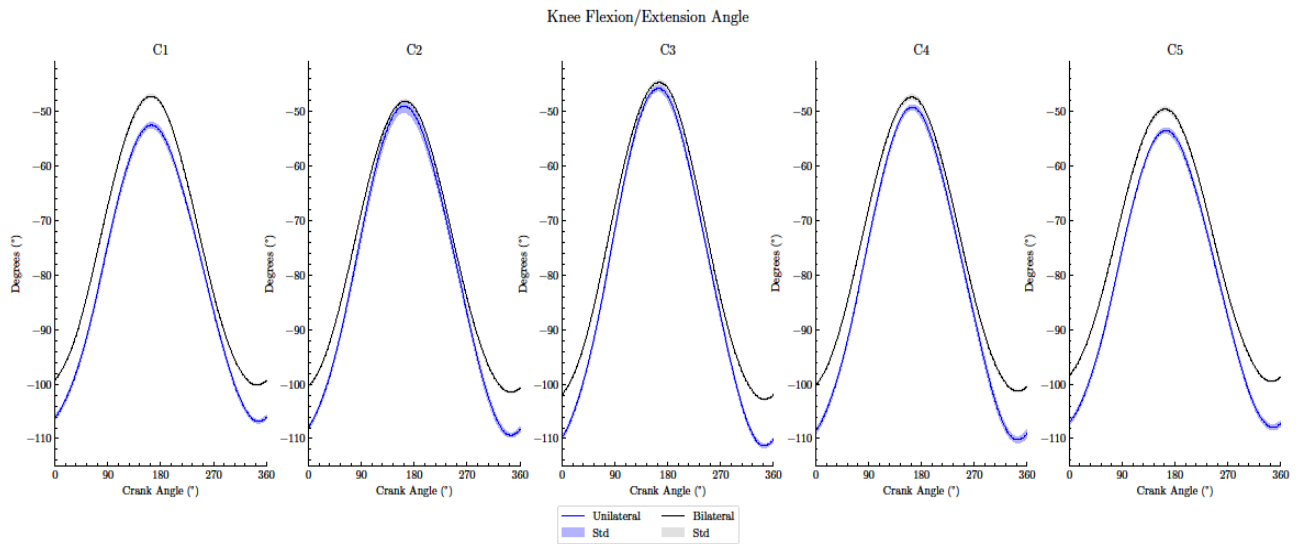


Figure 5: Ensemble curves comparing sagittal knee joint angle between unilateral and bilateral participants for the recently replaced limb's knee joint angle across all conditions. C1: 140/170, C2: 155/170, C3: 170/170, C4: 155/155, C5: 140/140

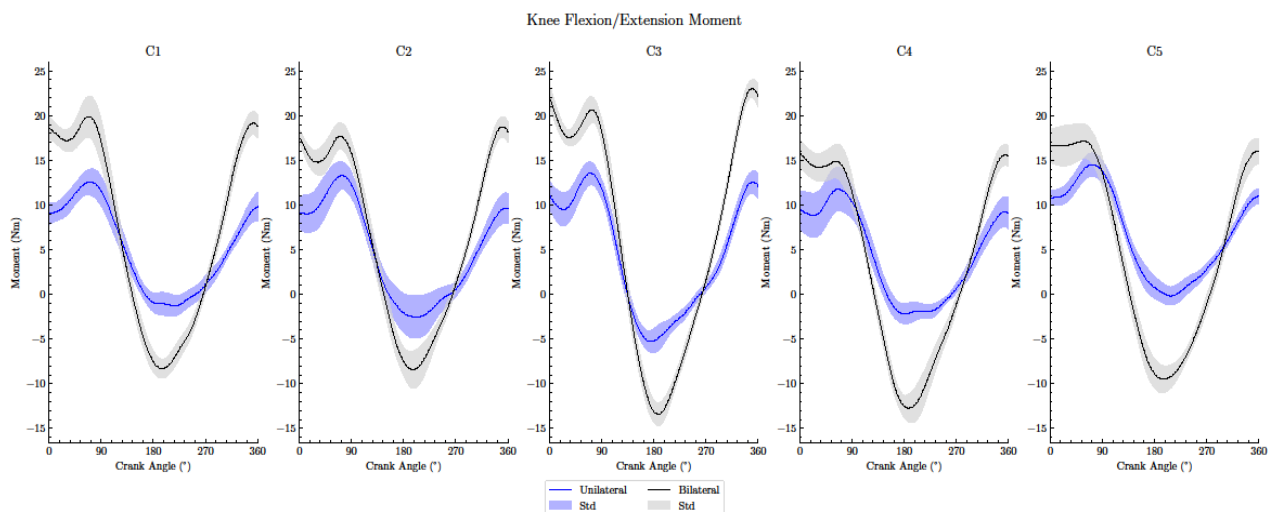


Figure 6: Ensemble curves comparing sagittal knee joint moment between unilateral and bilateral participants for the recently replaced limb's knee joint angle across all conditions. C1: 140/170, C2: 155/170, C3: 170/170, C4: 155/155, C5: 140/140

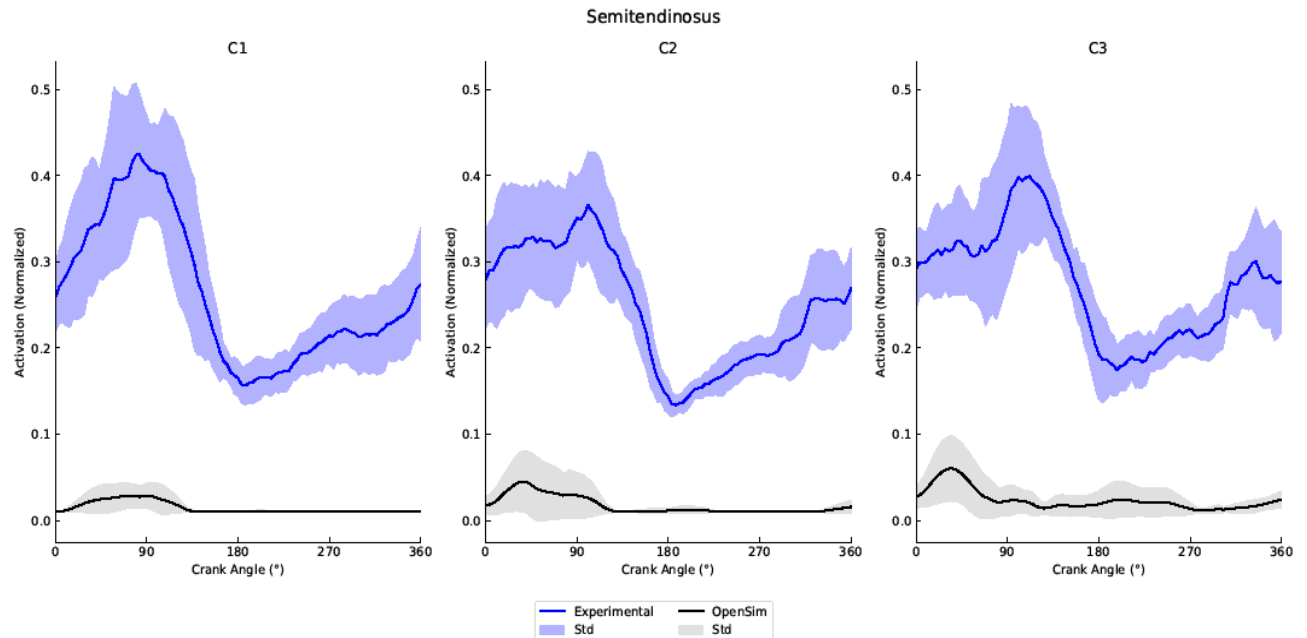


Figure 7: Experimental EMG and OpenSim activation levels for the semitendinosus during cycling. C1: 140/170, C2: 155/170, C3: 170/170

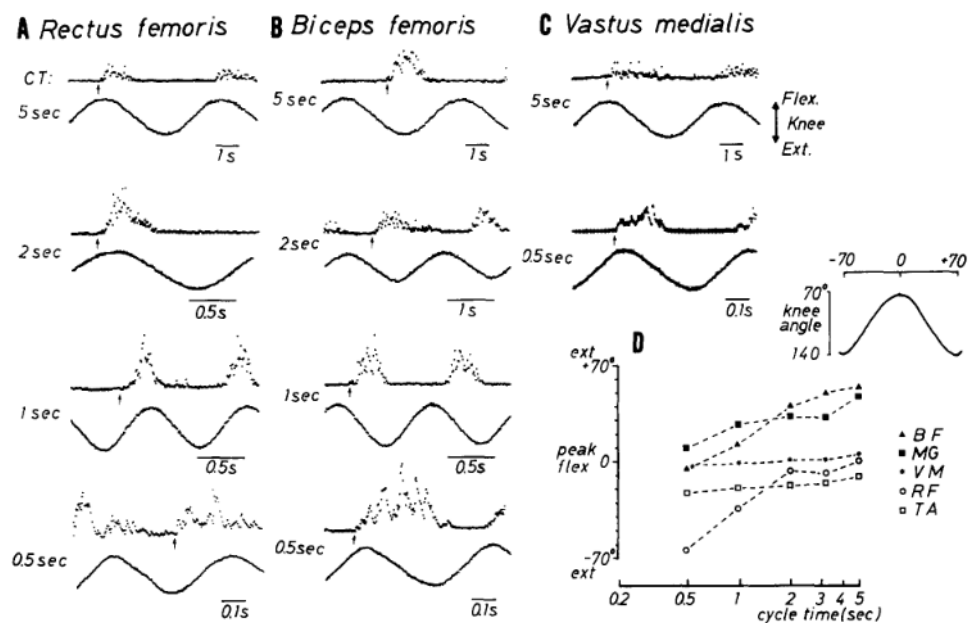


Fig. 1. The forward shift in EMG initiation with increased cycling speed. A: rectified, filtered and averaged EMG of the RF muscle (upper records) triggered by a flexion phase of knee angle (lower curves) during pedalling movements of 5 s, 2 s, 1 s and 0.5 s CT, respectively. B: rectified, filtered and averaged EMG of the BF muscle during pedalling movements of 5 s, 2 s, 1 s and 0.5 s CT, respectively. C: rectified, filtered and averaged EMG of the VM muscle during pedalling movements of 5 s and 0.5 s CT. In all records, an average of 5 cycles were taken from the same subject. For knee angle, upward and downward deflections indicate knee flexion and extension, respectively. D: summary of EMG initiation relative to knee flexion-extension at pedalling rates of different constant velocities. Knee angle data are plotted from peak extension (-70°) to the next peak extension ($+70^\circ$) with peak knee flexion set at 0° . CT, cycle time; BF, biceps femoris; MG, medial gastrocnemius; VM, vastus medialis; RF, rectus femoris; TA, tibialis anterior.

Figure 4: Double burst electromyography pattern of selected musculature, Suzuki [150]

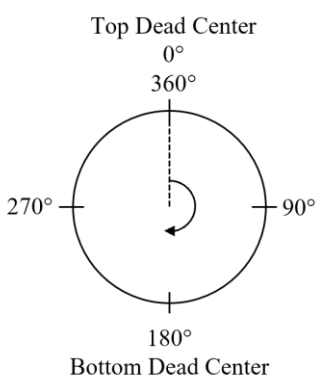


Figure 5: Bicycle crank diagram reference system, adapted from Coyle [334]

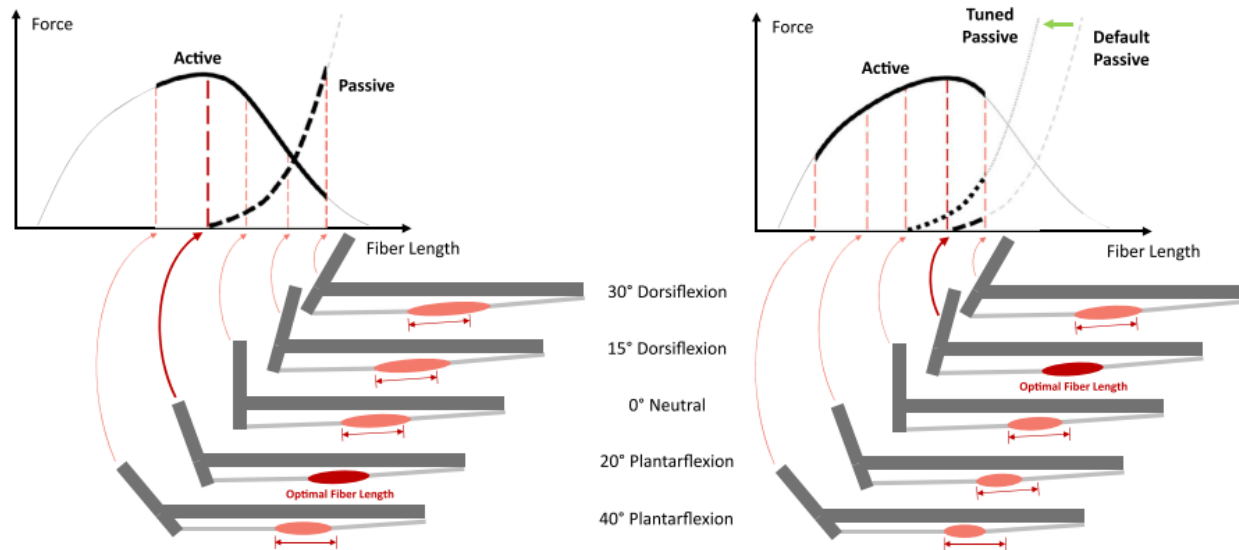


Figure 6: Fiber length relationship at varying joint positions, Chen and Franklin [223]

Appendix C: IRB Approval Letter



THE UNIVERSITY OF
TENNESSEE
KNOXVILLE

February 06, 2024

Songning Zhang, PhD
UTK - Coll of Education, Hlth, & Human - Kinesiology, Recreation & Sports Studies

Re: UTK IRB-23-07760-XP
Study Title: Asymmetric Crank Arm Lengths as an Early Rehabilitation Tool during Cycling in the Lower Limbs of TKA Populations

Dear Songning Zhang:

The UTK Institutional Review Board (IRB) reviewed your application for the above referenced project. It determined that your application is eligible for expedited review under 45 CFR 46.110(b)(1), categories (6) and (7). The IRB has reviewed these materials and determined that they do comply with proper consideration for the rights and welfare of human subjects and the regulatory requirements for the protection of human subjects.

Therefore, this letter constitutes full approval by the IRB of your application (Version 1.3) as submitted, including:

Documents Stamped:

- InformedConsent_2024 (English) - (Version 1.0)
- Patient_Info_Sheet_2024 (Version 1.0)
- AsymmetricCrank_ProtocolScript - (Version 1.0)
- Asymmetric_DataSheet - (Version 1.0)
- Asymmetric Crank PHI Release (Version 1.0)
- TKA Asymmetric Crank Verbal Script - (Version 1.0)
- Asymmetric_CollectionSheet_2024 - (Version 1.0)

Documents Acknowledged:

- KatieStevens_LetterOfSupport - (Version 1.0)
- HaroldCates_LetterOfSupport - (Version 1.0)
- Katie_Stevens_ILA_2024_Signed - (Version 1.0)
- Hal_Cates_ILA_2024_Signed - (Version 1.0)
- SENIAM_MedialGastroc - (Version 1.0)
- SENIAM_TibAnt - (Version 1.0)
- SENIAM_GluteMax - (Version 1.0)
- SENIAM_VastMed - (Version 1.0)

Institutional Review Board / Office of Research & Engagement
1184 White Avenue Knoxville, TN 37926-1150
865-474-7667 865-474-7400 fax 865-474-7668

BIG ORANGE. BIG IDEAS.

Playing Campus of the University of Tennessee System

Consent for Research Participation

Research Study Title: Asymmetric Crank Arm Lengths as an Early Rehabilitation Tool during Cycling in the Lower Limbs of Knee OA and TKA Populations

Researcher(s): Walter Menke, University of Tennessee, Knoxville
Dr. Songning Zhang, University of Tennessee, Knoxville

Why am I being asked to be in this research study?

We are asking you to be in this research study because you have met all the inclusion and exclusion criteria and we believe that you would be a good candidate for the study.

What is this research study about?

The purpose of the study is to identify lower limb kinematics, kinetics, and muscle activity during stationary cycling on cycle ergometry with asymmetric changes of crank arm lengths in patients with TKA during early-stage rehabilitation.

Who is conducting this research study?

The research team at the University of Tennessee, Knoxville Biomechanics Laboratory is conducting this study with the Principal Investigator Songning Zhang, PhD along with co-investigators Walter Menke, MS, and Sean Brown, MS. They are receiving no external funding for this research study.

How long will I be in the research study?

If you agree to participate in this study, you will be involved with 1 study visit. The testing session will last approximately 1.5-2 hours

What will happen if I say “Yes, I want to be in this research study”?

If you agree to be in this study, we will ask you to [do the following](#):

- Participate in 2 visits to our lab. One will occur 1 month before your surgery and the next will occur 1-2 months after your surgery.
- All participation will take place in our lab in the HPER building.
- *During your first test session:* complete a physical readiness questionnaire (PARQ+) to ensure your safety for exercise as well as provide the research team with basic demographic information.
- *During both test sessions:* complete a survey concerning your replaced knee (Knee Injury and Osteoarthritis Outcome Score) as well as the Tampa Scale of Kinesiophobia (TSK) that assesses your mental relationship with movement and your knee.
- *On both testing days,* you will be asked to do the following:
 - Change into appropriate athletic attire (shorts, t-shirt) either owned by you or provided by the lab.

- Complete a brief 3 minute warm up of walking.
- Complete three separate 3-meter time up-and-go tests.
- Complete a passive knee range of motion assessment as well as have the circumference of your knees measured.
- Be fitted with retroreflective markers and electromyography sensors. The sensors are small boxes that will be taped to your skin and pose no risk. Your skin will be shaved and abraded, followed by an alcohol wipe to clean, and prepare the skin for sensors.
- Complete 5 separate two-minute cycling trials on a stationary cycle ergometer with rest provided in between.
- Complete 5 separate walking trials at your own preferred walking pace.
- Complete questionnaires on how hard the activities were and how much pain you had during the trials.

What happens if I say “No, I do not want to be in this research study”?

Being in this study is up to you. You can say no now or leave the study later. If at any time you decide to leave the study, your relationship with UT and the researchers will not be impacted.

What happens if I say “Yes” but change my mind later?

Even if you decide to be in the study now, you can change your mind and stop at any time.

If you decide to stop before the study is completed, no negative consequences will occur to you. Any information we have collected so far will be destroyed and not used in the study.

Are there any possible risks to me?

There is minimal risk in this study. The duration and intensity of exercise are kept at a mild to moderate level. We will ensure you pass the PARQ form, which will help ensure you are safe to exercise. Both walking and cycling pose minimal risk to you during exercise. The researchers in this study are certified in first aid, CPR, and AED. Other people could potentially see you or find out about your participation in the study. We have procedures in place to minimize this risk such as locking all personal information in a locked cabinet. All information will be stored with no identifying information and will be kept using a subject code. All documents will be kept in a locked cabinet and only the investigator will have access to them.

Are there any benefits to being in this research study?

There is a possibility that you may benefit from being in the study, but there is no guarantee that will happen. Possible benefits include physical benefits that come from training. You may also have a reduced asymmetry between your operated and non-operated leg, which could be beneficial in the long-term. Being in a training protocol may lead to increased overall health as well. Even if you don't benefit from being in the study, your participation may help us to learn more about how cycling with feedback can be beneficial for people following a total knee replacement. We hope the knowledge gained from this study will benefit others in the future. You will also be compensated for your time during this study, which will be detailed in “Will I

be paid for being in this research study”. However, parking at the University of Tennessee will be provided at no cost to you.

Who can see or use the information collected for this research study?

We will protect the confidentiality of your information by deidentifying the data such that only a subject number will be assigned to it. If information from this study is published or presented at scientific meetings, your name and other personal information will not be used.

We will make every effort to prevent anyone who is not on the research team from knowing that you gave us information or what information came from you. Although it is unlikely, there are times when others may need to see the information we collect about you. These include:

- Government agencies (such as the Office for Human Research Protections in the U.S. Department of Health and Human Services), and others responsible for watching over the safety, effectiveness, and conduct of the research.
- If a law or court requires us to share the information, we would have to follow that law or final court ruling.
- The Tennessee Orthopedic Clinic (TOC) who will be our site for recruitment.

What will happen to my information after this study is over?

We will not keep your information to use for future research, presentations, and/or publications. Your name and other information that can directly identify you will be deleted from your research data collected as part of the study.

We may share your research data with other researchers without asking for your consent again, but it will not contain information that could directly identify you.

Will I be paid for being in this research study?

You will be compensated with a \$150 gift card upon full completion of the study testing dates. You will not be compensated for partial study completion. Full study completion is defined as completion of both testing dates in their entirety.

Will it cost me anything to be in this research study?

It will not cost you anything to participate in this study. All costs will be taken care of by the researchers.

What else do I need to know?

We use procedures to lower the possibility of these risks happening. Even so, you may still experience problems or injury, even when we are careful to avoid them. Please tell the researcher in charge, Walter Menke (wmenke@vols.utk.edu) about any injuries or other problems that you have during this study.

The University of Tennessee does not automatically pay for medical claims or give other compensation for injuries or other problems.

Who can answer my questions about this research study?

If you have questions or concerns about this study, or have experienced a research related problem or injury, contact the researchers, Walter Menke (wmenke@vols.utk.edu) or over the phone at (865-974-4716). You may also contact my faculty advisor, Dr. Songning Zhang via email at szhang@utk.edu

For questions or concerns about your rights or to speak with someone other than the research team about the study, please contact:

Institutional Review Board
The University of Tennessee, Knoxville
2240 Sutherland Ave, Suite 2
Knoxville, TN 37919
Phone: 865-974-7697
Email: utkirb@utk.edu

STATEMENT OF CONSENT

I have read this form and the research study has been explained to me. I have been given the chance to ask questions and my questions have been answered. If I have more questions, I have been told who to contact. By signing this document, I am agreeing to be in this study. I will receive a copy of this document after I sign it.

Name of Adult Participant

Signature of Adult Participant

Date

Researcher Signature (to be completed at time of informed consent)

I have explained the study to the participant and answered all of his/her questions. I believe that he/she understands the information described in this consent form and freely consents to be in the study.

Name of Research Team Member

Signature of Research Team Member

Date

Appendix E: Patient Info Sheet

Asymmetric Crank Length Study Opportunity

Researchers at the University of Tennessee Knoxville are conducting a study that involves individuals who have recently received a total knee replacement (TKA) from Dr. Harold Cates at Tennessee Orthopaedic Clinics. The goal of the study is to investigate if creating asymmetric crank length conditions during stationary cycling is beneficial for early stage TKA recipients in their rehabilitation procedure.

If you agree to be in this study, we will ask you to [do the following](#):

- Participate in no more than 2 visits to our lab (1.5-2 hours each) in the range of 1 month *before* AND/OR 1-2 months *after* your surgery.
- All participation will take place in our lab in the HPER building on UT Knoxville campus.
- *During a pre-TKA test session*: complete a physical readiness questionnaire (PARQ+) to ensure your safety for exercise as well as provide the research team with basic demographic information.
- *During a post-TKA test session*: complete a survey concerning your replaced knee (Knee Injury and Osteoarthritis Outcome Score) as well as the Tampa Scale of Kinesiophobia (TSK) that assesses your mental relationship with movement and your knee.
- *At either test session*, you will be asked to do the following:
 - Change into appropriate athletic attire (shorts, t-shirt) either owned by you or provided by the lab.
 - Complete a brief 3 minute warm up of walking.
 - Complete three separate 3-meter time up-and-go tests.
 - Complete a passive knee range of motion assessment as well as have the circumference of your knees measured.
 - Be fitted with retroreflective markers and electromyography sensors. The sensors are small boxes that will be taped to your skin and pose no risk. Your skin will be shaved and abraded, followed by an alcohol wipe to clean, and prepare the skin for sensors.
 - Complete 5 separate two-minute cycling trials on a stationary cycle ergometer with rest provided in between.
 - Complete 5 separate walking trials at your own preferred walking pace.
 - Complete questionnaires on how hard the activities were and how much pain you had during the trials.
- You will be compensated with a gift card of \$60 for completing a pre-TKA session, and a \$90 gift card for completing a post-TKA session. If you attend both testing sessions, you will receive a total of \$150

Any questions not answerable by Dr. Harold Cates or Dr. Katie Stevens may be directed to either Walter Menke (wmenke@vols.utk.edu) or Dr. Songning Zhang (szhang@utk.edu).

Inclusion Criteria

- Ages: 50-75
- Willingness to attend either or both testing sessions:
 - One occurring no more than 1 month prior to arthroplasty
 - One occurring 1-2 months after arthroplasty
- *Pre-surgery session*: knee flexion range of motion (ROM) of at least 85°
- *Post-surgery session*: knee flexion range of motion (ROM) of at least 100°
- Male or Female
- Demonstrated asymmetry in their lower limbs as measured by passive knee flexion ROM in clinic with a goniometer
 - And/or flexion/extension contracture of involved side pre-surgery
- OPTIONAL: Use of assistive device for walking/daily activities for operational side pre-surgery

Exclusion Criteria

- Diagnosed osteoarthritis of ankles, hips, or spine
- No contralateral TKA within 4 months of study participation date
- BMI value greater than 40
- Neurological conditions that prevent the patients from cycling or walking without aids
 - Inability to maintain upright posture during cycling (as in, hip hiking or compensation during pedaling)
- Pregnant or Nursing Women
- Cardiovascular Disease
- Presence of partial or total limb amputation

Appendix F: Patient Health Information Release Form



Dear Patient:

Thank you for contacting Tennessee Orthopaedic Alliance East Tennessee Medical Records Department. To better serve you with your request for medical records, Tennessee Orthopaedic Alliance East Tennessee has partnered with MedRecs Management.

MedRecs Management will fulfill your request for records in a safe, secure, and timely manner. The quickest way (typically less than 3 days) for you to receive a copy of your records is to visit our website, todocs.com. Under the Patient Resources tab, you will see Request Medical Records. After clicking this tab, you will be directed to a page with information on obtaining your medical records. Click the "Click Here to Request a Copy of Your Medical Records" button and follow the instructions to complete your request.

If you cannot complete the request via the processing page, you will need to complete and return the attached Authorization form. Please make sure you have specific instructions included as to what records you are requesting and where you are requesting your records be sent. You also have a choice of how you would like to have your records delivered. For records to be delivered directly to you, please choose mail or email. NOTE: If you choose to have the records mailed to you, postage costs will be added and the process can take up to two weeks.

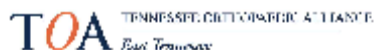
For records to be delivered to another doctor, please choose fax or mail (select only one option). The fax delivery option may only be used for records going to a doctor. Please mail/fax/drop-off the completed Authorization form to the Tennessee Orthopaedic Alliance East Tennessee location you utilize. For Records being sent to Another Health Care Provider, please provide as much contact information for your other Doctor, including the address, phone number or fax.

You can contact a MedRecs Management representative Monday-Friday 9am-6pm Eastern Time by calling: 865-505-7337 or email at: medicalrecords@medrecsmanagement.com

Thank you,
Medical Records Supervisor
Tennessee Orthopaedic Alliance East Tennessee

IRS NUMBER: UTK IRS-23-07760-JP
IRS APPROVAL DATE: 04/29/2025

Authorization to Disclose Protected Health Information
The undersigned authorizes



to release my health information as noted below:

☐ Ft. Sanders Regional - 1419 Clinch Ave, Ste. 100 - Knoxville, TN 37916 - Tx. 865-473-4007
☐ Oak Ridge - 306 Oak Ridge Trpk, Ste 100 - Oak Ridge, TN 37830 - Tx. 865-483-4194
☐ Oak Ridge-Randall Robbins - 941 Oak Ridge Trpk - Oak Ridge, TN 37840 - Tx. 865-483-3407
☐ Padonov/TCC Spine - 9638 Padonov Blvd, Ste. 130 - Knoxville, TN 37923 - Tx. 865-569-4559
☐ Sevierville - 1479 Veterans Boulevard - Sevierville, TN 37863 - Tx. 865-473-4007
☐ Lenoir City - 576 P. Lenoir Med Ctr Dr - Lenoir City, TN 37771 - Tx. 865-568-4437
☐ Seardon - 6484 Kingston Pike - Knoxville, TN 37919 - Tx. 865-473-4007

Patient Information	
Patient Full Name: _____ Other Names? _____	
Patient Address: _____ Date of Birth: _____	
City: _____ State: _____ Zip: _____ Phone #: _____	
Release Information To	
Name/Facility: <u>The University of Tennessee Knoxville</u> Attention: <u>Songning Zhang, Walter Menke</u>	
Address: <u>1914 Andy Holt Avenue</u> Phone: <u>865-974-4716</u>	
City: <u>Knoxville</u> State: <u>TN</u> Zip: <u>37996</u> Fax #: _____	
Email: <u>szhang@utk.edu, wmenke@vois.utk.edu</u> (Please ensure email address is legible!)	
Purpose of Request: ☐ Treatment ☐ Transfer ☒ Other: <u>X-Ray Data</u>	
Please forward Records by: ☐ Mail ☐ Fax (for Dr's Office) ☒ Email ☒ In-Person	
Information to be Released	
☐ Please release a 3 year abstract of my records <i>Abstract includes most notes, labs, procedures & testing</i> ☐ Please release a 5 year abstract of my records _____ ☐ Date Range: _____ ☐ Progress Notes ☐ Radiology Reports ☐ Labs ☐ Operative Reports ☐ Physical Therapy Notes ☐ Other: _____	<i>If you fail to specify, a 3 year abstract will be provided.</i> Pursuant to HIPAA 45 CFR, 164.524, we reserve the right to charge a reasonable cost-based fee for producing and mailing the copies. If you want the entire medical record, the rate will increase proportionally based on the cost. At no time will the cost-based fees exceed Tennessee State law: Statute 63-2-132. Records being sent to another healthcare provider will be sent at no cost.
Authorization to Release Protected Health Information	
I acknowledge and hereby consent to such, that the released information may contain alcohol, drug abuse, psychiatric, HIV testing, HIV results, or AIDS information. _____ (Please Initial)	
I understand that:	
1. I may refuse to sign this authorization and that it is strictly voluntary. 2. My treatment, payment, enrollment or eligibility for benefits may not be conditioned on signing this authorization. 3. I may revoke this authorization at any time in writing, but if I do, it will not have any effect on any actions taken prior to receiving the revocation. Unless otherwise revealed, this authorization will expire on the following date, event or condition: _____. <i>If I do not specify expiration this authorization will expire in 90 days.</i> 4. If the requestor or receiver is not a health plan or health care provider, the released information may no longer be protected by federal privacy regulations and may be disclosed. 5. I understand that I may see and obtain a copy of the information described on this form, for a reasonable copy fee, if I ask for it. I can request a copy of this form after I sign and date it.	
 Please confirm that you have filled out this form in its entirety—If form is incomplete, or if protected information is not released, we may be unable to fulfill this request.	
Signature*: _____ Date: _____	

IRB NUMBER: UTK IRB-23-07780-XP

* For non-emancipated minors under the age of 18, a parent or guardian must sign this form. If patient is unable to sign, a copy of the legal documentation for patient's representative must be supplied with a copy of this form.

Appendix G: Physical Activity Readiness Questionnaire

PAR-Q+

The Physical Activity Readiness Questionnaire for Everyone

The health benefits of regular physical activity are clear; more people should engage in physical activity every day of the week. Participating in physical activity is very safe for **MOST** people. This questionnaire will tell you whether it is necessary for you to seek further advice from your doctor **OR** a qualified exercise professional before becoming more physically active.

GENERAL HEALTH QUESTIONS

Please read the 7 questions below carefully and answer each one honestly: check YES or NO.	YES	NO
1) Has your doctor ever said that you have a heart condition ☐ OR high blood pressure ☐ ?	☐	☐
2) Do you feel pain in your chest at rest, during your daily activities of living, OR when you do physical activity?	☐	☐
3) Do you lose balance because of dizziness OR have you lost consciousness in the last 12 months? Please answer NO if your dizziness was associated with over-breathing (including during vigorous exercise).	☐	☐
4) Have you ever been diagnosed with another chronic medical condition (other than heart disease or high blood pressure)? PLEASE LIST CONDITION(S) HERE: _____	☐	☐
5) Are you currently taking prescribed medications for a chronic medical condition? PLEASE LIST CONDITION(S) AND MEDICATION(S) HERE: _____	☐	☐
6) Do you currently have (or have had within the past 12 months) a bone, joint, or soft tissue (muscle, ligament, or tendon) problem that could be made worse by becoming more physically active? Please answer NO if you had a problem in the past, but it does not limit your current ability to be physically active. PLEASE LIST CONDITION(S) HERE: _____	☐	☐
7) Has your doctor ever said that you should only do medically supervised physical activity?	☐	☐

If you answered **NO** to all of the questions above, you are cleared for physical activity. Please sign the **PARTICIPANT DECLARATION**. You do not need to complete Pages 2 and 3.

- Start becoming much more physically active – start slowly and build up gradually.
- Follow Global Physical Activity Guidelines for your age (<https://www.who.int/publications/item/9789240015128>).
- You may take part in a health and fitness appraisal.
- If you are over the age of 45 yr and NOT accustomed to regular vigorous to maximal effort exercise, consult a qualified exercise professional before engaging in this intensity of exercise.
- If you have any further questions, contact a qualified exercise professional.

PARTICIPANT DECLARATION
If you are less than the legal age required for consent or require the assent of a care provider, your parent, guardian or care provider must also sign this form.

I, the undersigned, have read, understood to my full satisfaction and completed this questionnaire. I acknowledge that this physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if my condition changes. I also acknowledge that the community/fitness center may retain a copy of this form for its records. In these instances, it will maintain the confidentiality of the same, complying with applicable law.

NAME _____ DATE _____

SIGNATURE _____ WITNESS _____

SIGNATURE OF PARENT/GUARDIAN/CARE PROVIDER _____

If you answered **YES** to one or more of the questions above, **COMPLETE PAGES 2 AND 3.**

Delay becoming more active if:

- You are currently experiencing a temporary illness, such as a cold or fever. It is best to wait until you feel better.
- You are pregnant. In this case, talk with your health care practitioner, physician, qualified exercise professional, and/or complete the ePARmed-3r at www.aparmeds.com before becoming more physically active.
- Your health changes. Answer the questions on Pages 2 and 3 of this document and/or talk to your health care practitioner, physician, or qualified exercise professional before proceeding with any physical activity program.

Copyright © 2020 PAR-Q+ Collaborators 1 / 4
01-11-2023

PAR-Q+

FOLLOW-UP QUESTIONS ABOUT YOUR MEDICAL CONDITION(S)

1.	Do you have Arthritis, Osteoporosis, or Back Problems? If the above condition(s) is/are present, answer questions 1a-1c	If NO ☐ go to question 2
1a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments)	YES ☐ NO ☐
1b.	Do you have joint problems causing pain, a recent fracture or fracture caused by osteoporosis or cancer, displaced vertebra (e.g., spondylolisthesis), and/or spondylolysis/pars defect (a crack in the bony ring on the back of the spinal column)?	YES ☐ NO ☐
1c.	Have you had steroid injections or taken steroid tablets regularly for more than 3 months?	YES ☐ NO ☐
2.	Do you currently have Cancer of any kind? If the above condition(s) is/are present, answer questions 2a-2b	If NO ☐ go to question 3
2a.	Does your cancer diagnosis include any of the following types: lung/bronchogenic, multiple myeloma (cancer of plasma cells), head, and/or neck?	YES ☐ NO ☐
2b.	Are you currently receiving cancer therapy (such as chemotherapy or radiotherapy)?	YES ☐ NO ☐
3.	Do you have a Heart or Cardiovascular Condition? This includes Coronary Artery Disease, Heart Failure, Diagnosed Abnormality of Heart Rhythm If the above condition(s) is/are present, answer questions 3a-3d	If NO ☐ go to question 4
3a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments)	YES ☐ NO ☐
3b.	Do you have an irregular heart beat that requires medical management? (e.g., atrial fibrillation, premature ventricular contraction)	YES ☐ NO ☐
3c.	Do you have chronic heart failure?	YES ☐ NO ☐
3d.	Do you have diagnosed coronary artery (cardiovascular) disease and have not participated in regular physical activity in the last 2 months?	YES ☐ NO ☐
4.	Do you currently have High Blood Pressure? If the above condition(s) is/are present, answer questions 4a-4b	If NO ☐ go to question 5
4a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments)	YES ☐ NO ☐
4b.	Do you have a resting blood pressure equal to or greater than 160/90 mmHg with or without medication? (Answer YES if you do not know your resting blood pressure)	YES ☐ NO ☐
5.	Do you have any Metabolic Conditions? This includes Type 1 Diabetes, Type 2 Diabetes, Pre-Diabetes If the above condition(s) is/are present, answer questions 5a-5e	If NO ☐ go to question 6
5a.	Do you often have difficulty controlling your blood sugar levels with foods, medications, or other physician-prescribed therapies?	YES ☐ NO ☐
5b.	Do you often suffer from signs and symptoms of low blood sugar (hypoglycemia) following exercise and/or during activities of daily living? Signs of hypoglycemia may include shakiness, nervousness, unusual irritability, abnormal sweating, dizziness or lightheadedness, mental confusion, difficulty speaking, weakness, or sleepiness.	YES ☐ NO ☐
5c.	Do you have any signs or symptoms of diabetes complications such as heart or vascular disease and/or complications affecting your eyes, kidneys, OR the sensation in your toes and feet?	YES ☐ NO ☐
5d.	Do you have other metabolic conditions (such as current pregnancy-related diabetes, chronic kidney disease, or liver problems)?	YES ☐ NO ☐
5e.	Are you planning to engage in what for you is unusually high (or vigorous) intensity exercise in the near future?	YES ☐ NO ☐

PAR-Q+

6. Do you have any Mental Health Problems or Learning Difficulties? This includes Alzheimer's, Dementia, Depression, Anxiety Disorder, Eating Disorder, Psychotic Disorder, Intellectual Disability, Down Syndrome.
If the above condition(s) is/are present, answer questions 6a-6b. If NO ☐ go to question 7
- 6a. Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments) YES ☐ NO ☐
- 6b. Do you have Down Syndrome AND back problems affecting nerves or muscles? YES ☐ NO ☐
7. Do you have a Respiratory Disease? This includes Chronic Obstructive Pulmonary Disease, Asthma, Pulmonary High Blood Pressure.
If the above condition(s) is/are present, answer questions 7a-7d. If NO ☐ go to question 8
- 7a. Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments) YES ☐ NO ☐
- 7b. Has your doctor ever said your blood oxygen level is low at rest or during exercise and/or that you require supplemental oxygen therapy? YES ☐ NO ☐
- 7c. If asthmatic, do you currently have symptoms of chest tightness, wheezing, laboured breathing, constant cough (more than 2 days/week), or have you used your rescue medication more than twice in the last week? YES ☐ NO ☐
- 7d. Has your doctor ever said you have high blood pressure in the blood vessels of your lungs? YES ☐ NO ☐
8. Do you have a Spinal Cord Injury? This includes Tetraplegia and Paraplegia.
If the above condition(s) is/are present, answer questions 8a-8c. If NO ☐ go to question 9
- 8a. Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments) YES ☐ NO ☐
- 8b. Do you commonly exhibit low resting blood pressure significant enough to cause dizziness, light-headedness, and/or fainting? YES ☐ NO ☐
- 8c. Has your physician indicated that you exhibit sudden bouts of high blood pressure (known as Autonomic Dysreflexia)? YES ☐ NO ☐
9. Have you had a Stroke? This includes Transient Ischemic Attack (TIA) or Cerebrovascular Event.
If the above condition(s) is/are present, answer questions 9a-9c. If NO ☐ go to question 10
- 9a. Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer NO if you are not currently taking medications or other treatments) YES ☐ NO ☐
- 9b. Do you have any impairment in walking or mobility? YES ☐ NO ☐
- 9c. Have you experienced a stroke or impairment in nerves or muscles in the past 6 months? YES ☐ NO ☐
10. Do you have any other medical condition not listed above or do you have two or more medical conditions?
If you have other medical conditions, answer questions 10a-10c. If NO ☐ read the Page 4 recommendations
- 10a. Have you experienced a blackout, fainted, or lost consciousness as a result of a head injury within the last 12 months OR have you had a diagnosed concussion within the last 12 months? YES ☐ NO ☐
- 10b. Do you have a medical condition that is not listed (such as epilepsy, neurological conditions, kidney problems)? YES ☐ NO ☐
- 10c. Do you currently live with two or more medical conditions? YES ☐ NO ☐
- PLEASE LIST YOUR MEDICAL CONDITION(S) AND ANY RELATED MEDICATIONS HERE:

GO to Page 4 for recommendations about your current medical condition(s) and sign the PARTICIPANT DECLARATION.

PAR-Q+

- ✓ If you answered **NO** to all of the **FOLLOW-UP** questions (pgs. 2-3) about your medical condition, you are ready to become more physically active - sign the **PARTICIPANT DECLARATION** below:
It is advised that you consult a qualified exercise professional to help you develop a safe and effective physical activity plan to meet your health needs.
- ▶ You are encouraged to start slowly and build up gradually - 20 to 60 minutes of low to moderate intensity exercise, 3-5 days per week including aerobic and muscle strengthening exercises.
- ▶ As you progress, you should aim to accumulate 150 minutes or more of moderate intensity physical activity per week.
- ▶ If you are over the age of 45 yr and **NOT** accustomed to regular vigorous to maximal effort exercise, consult a qualified exercise professional before engaging in this intensity of exercise.

- ⚠ If you answered **YES** to one or more of the follow-up questions about your medical condition: You should seek further information before becoming more physically active or engaging in a fitness appraisal. You should complete the specially designed online screening and exercise recommendations program - the ePARmed-X+ at www.eparmedx.com and/or visit a qualified exercise professional to work through the ePARmed-X+ and for further information.

- ⚠ **Delay becoming more active if:**
 - ▶ You are currently experiencing a temporary illness, such as a cold or fever. It is best to wait until you feel better.
 - ▶ You are pregnant. In this case, talk to your health care practitioner, physician, qualified exercise professional, and/or complete the ePARmed-X+ at www.eparmedx.com before becoming more physically active.
 - ▶ Your health changes. Talk to your health care practitioner, physician, or qualified exercise professional before continuing with any physical activity program.

- You are encouraged to photocopy the PAR-Q+. You must use the entire questionnaire and NO changes are permitted.
- The authors, the PAR-Q+ Collaboration, partner organizations, and their agents assume no liability for persons who undertake physical activity and/or make use of the PAR-Q+ or ePARmed-X+. If in doubt after completing the questionnaire, consult your doctor prior to physical activity.

PARTICIPANT DECLARATION

- All persons who have completed the PAR-Q+ please read and sign the declaration below.
- If you are less than the legal age required for consent or require the assent of a care provider, your parent, guardian or care provider must also sign this form.

I, the undersigned, have read, understood to my full satisfaction and completed this questionnaire. I acknowledge that this physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if my condition changes. I also acknowledge that the community/fitness center may retain a copy of this form for records. In these instances, it will maintain the confidentiality of the same, complying with applicable law.

NAME _____ DATE _____
SIGNATURE _____ WITNESS _____
SIGNATURE OF PARENT/GUARDIAN/CARE PROVIDER _____

For more information, please contact:
www.eparmedx.com
Email: eparmedx@gmail.com

Creators for PAR-Q+
Wendell G. Goss, David W. Goss, and Dr. David W. Goss in collaboration with the PAR-Q+ Collaboration.
The Physical Activity Readiness Questionnaire for Everyone (PAR-Q+) and ePARmed-X+ are trademarks of the PAR-Q+ Collaboration. ©2014-2023.

By Permission

1. James W. Warburton, D.L. Johnston, J. Goss, and Dr. David W. Goss, in collaboration with the PAR-Q+ Collaboration, are the authors of the PAR-Q+ questionnaire. The PAR-Q+ questionnaire is a trademark of the PAR-Q+ Collaboration. ©2014-2023.

2. Warburton, Goss, Johnston, Goss, and Dr. David W. Goss, in collaboration with the PAR-Q+ Collaboration, are the authors of the ePARmed-X+ questionnaire. The ePARmed-X+ questionnaire is a trademark of the PAR-Q+ Collaboration. ©2014-2023.

3. Goss, Goss, Goss, Goss, and Dr. David W. Goss, in collaboration with the PAR-Q+ Collaboration, are the authors of the PAR-Q+ questionnaire. The PAR-Q+ questionnaire is a trademark of the PAR-Q+ Collaboration. ©2014-2023.

4. Goss, Goss, Goss, Goss, and Dr. David W. Goss, in collaboration with the PAR-Q+ Collaboration, are the authors of the ePARmed-X+ questionnaire. The ePARmed-X+ questionnaire is a trademark of the PAR-Q+ Collaboration. ©2014-2023.

The PAR-Q+ was created using the evidence-based AGREE process (1) by the PAR-Q+ Collaboration created by Dr. David W. Goss, Warburton, Goss, Johnston, Goss, and Dr. David W. Goss. Production of this document has been made possible through financial contributions from the Public Health Agency of Canada and the BC Ministry of Health Services. The views expressed herein do not necessarily represent the views of the Public Health Agency of Canada or the BC Ministry of Health Services.

Copyright © 2014 PAR-Q+ Collaboration 4/ 4
01-11-2023

Appendix H: Knee Injury and Osteoarthritis Outcome Score Survey

Knee Injury and Osteoarthritis Outcome Score (KOOS), English version LK1.0

1

KOOS KNEE SURVEY

Today's date: ____/____/____ Date of birth: ____/____/____

Name: _____

INSTRUCTIONS: This survey asks for your view about your knee. This information will help us keep track of how you feel about your knee and how well you are able to perform your usual activities. Answer every question by ticking the appropriate box, only one box for each question. If you are unsure about how to answer a question, please give the best answer you can.

Symptoms

These questions should be answered thinking of your knee symptoms during the last week.

S1. Do you have swelling in your knee?

Never
☐

Rarely
☐

Sometimes
☐

Often
☐

Always
☐

S2. Do you feel grinding, hear clicking or any other type of noise when your knee moves?

Never
☐

Rarely
☐

Sometimes
☐

Often
☐

Always
☐

S3. Does your knee catch or hang up when moving?

Never
☐

Rarely
☐

Sometimes
☐

Often
☐

Always
☐

S4. Can you straighten your knee fully?

Always
☐

Often
☐

Sometimes
☐

Rarely
☐

Never
☐

S5. Can you bend your knee fully?

Always
☐

Often
☐

Sometimes
☐

Rarely
☐

Never
☐

Stiffness

The following questions concern the amount of joint stiffness you have experienced during the last week in your knee. Stiffness is a sensation of restriction or slowness in the ease with which you move your knee joint.

S6. How severe is your knee joint stiffness after first waking in the morning?

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

S7. How severe is your knee stiffness after sitting, lying or resting later in the day?

None
☐

Mild
☐

Moderate
☐

Severe
☐

Extreme
☐

Pain**P1. How often do you experience knee pain?**Never
☐Mostly
☐Weekly
☐Daily
☐Always
☐**What amount of knee pain have you experienced the last week during the following activities?****P2. Twisting/pivoting on your knee**None
☐Mild
☐Moderate
☐Severe
☐Extreme
☐**P3. Straightening knee fully**None
☐Mild
☐Moderate
☐Severe
☐Extreme
☐**P4. Bending knee fully**None
☐Mild
☐Moderate
☐Severe
☐Extreme
☐**P5. Walking on flat surface**None
☐Mild
☐Moderate
☐Severe
☐Extreme
☐**P6. Going up or down stairs**None
☐Mild
☐Moderate
☐Severe
☐Extreme
☐**P7. At night while in bed**None
☐Mild
☐Moderate
☐Severe
☐Extreme
☐**P8. Sitting or lying**None
☐Mild
☐Moderate
☐Severe
☐Extreme
☐**P9. Standing upright**None
☐Mild
☐Moderate
☐Severe
☐Extreme
☐**Function, daily living****The following questions concern your physical function. By this we mean your ability to move around and to look after yourself. For each of the following activities please indicate the degree of difficulty you have experienced in the last week due to your knee.****A1. Descending stairs**None
☐Mild
☐Moderate
☐Severe
☐Extreme
☐**A2. Ascending stairs**None
☐Mild
☐Moderate
☐Severe
☐Extreme
☐

For each of the following activities please indicate the degree of difficulty you have experienced in the last week due to your knee.

A3. Rising from sitting	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐
A4. Standing	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐
A5. Bending to floor/pick up an object	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐
A6. Walking on flat surface	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐
A7. Getting in/out of car	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐
A8. Going shopping	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐
A9. Putting on socks/stockings	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐
A10. Rising from bed	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐
A11. Taking off socks/stockings	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐
A12. Lying in bed (turning over, maintaining knee position)	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐
A13. Getting in/out of bath	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐
A14. Sitting	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐
A15. Getting on/off toilet	None ☐	Mild ☐	Moderate ☐	Severe ☐	Extreme ☐

For each of the following activities please indicate the degree of difficulty you have experienced in the last week due to your knee.

A16. Heavy domestic duties (moving heavy boxes, scrubbing floors, etc)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A17. Light domestic duties (cooking, dusting, etc)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, sports and recreational activities

The following questions concern your physical function when being active on a higher level. The questions should be answered thinking of what degree of difficulty you have experienced during the last week due to your knee.

SP1. Squatting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP2. Running

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP3. Jumping

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP4. Twisting/pivoting on your injured knee

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP5. Kneeling

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Quality of Life

Q1. How often are you aware of your knee problem?

Never	Monthly	Weekly	Daily	Constantly
☐	☐	☐	☐	☐

Q2. Have you modified your life style to avoid potentially damaging activities to your knee?

Not at all	Mildly	Moderately	Severely	Totally
☐	☐	☐	☐	☐

Q3. How much are you troubled with lack of confidence in your knee?

Not at all	Mildly	Moderately	Severely	Extremely
☐	☐	☐	☐	☐

Q4. In general, how much difficulty do you have with your knee?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Thank you very much for completing all the questions in this questionnaire.

Appendix I: Tampa Scale of Kinesiophobia

TAMPA SCALE

INSTRUCTIONS: Please answer the following questions according to your true feelings, not according to what others think you should believe. Score each statement from strongly disagree to strongly agree by ticking the appropriate box.

Question	Strongly disagree	Disagree	Agree	Strongly agree
1. I'm afraid that I might injure myself if I exercise.				
2. If I were to try to overcome it, my pain would increase.				
3. My body is telling me I have something dangerously wrong.				
4. My pain would probably be relieved if I were to exercise.				
5. People aren't taking my medical condition seriously enough.				
6. My accident has put my body at risk for the rest of my life.				
7. Pain always means I have injured my body.				
8. Just because something aggravates my pain does not mean it is dangerous.				
9. I am afraid that I might injure myself accidentally.				
10. Simply being careful that I do not make any unnecessary movements is the safest thing I can do to prevent my pain from worsening.				
11. I wouldn't have this much pain if there weren't something potentially dangerous going on in my body.				
12. Although my condition is painful, I would be better off if I were physically active.				
13. Pain lets me know when to stop exercising so that I don't injure myself.				
14. It's really not safe for a person with a condition like mine to be physically active.				
15. I can't do all the things normal people do because it's too easy for me to get injured.				
16. Even though something is causing me a lot of pain, I don't think it's actually dangerous.				
17. No one should have to exercise when he/she is in pain.				
SCORING (staff use only)				

(Miller, R.P., Kori, S.H. and Todd, D.D., The Tampa Scale. Unpublished Report, Tampa, FL, 1991.)

Vita

Walt Menke was born in small-town Piner, Kentucky to Henry and Sharon Menke. Walt graduated from Covington Catholic High School in 2015 before starting college adventures at the University of Kentucky. After realizing he wasn't interested in accounting, he shifted his degree to kinesiology with a goal for physical therapy. Upon discovering that he was interested in the theory and development of rehabilitation protocols instead of applying them in the clinic, he decided to stay at UK to complete an MS degree in Biomechanics. At UK, he primarily studied femoroacetabular impingement with some side work on body armor fatigue effects. Given his interest in teaching, he applied for PhD programs in biomechanics and landed himself at the University of Tennessee, Knoxville.

At UT, Walt managed to get entangled in horse, pig, and insole biomechanics as side projects while trying to focus primarily on patients with total knee replacements. After 4 and some change years, Walt has emerged with an MS in Statistics as well as a PhD in Kinesiology & Sport Studies with a Biomechanics specialization. Along with these pieces of flimsy paper, he has gained much more coding and IT experience than he could have bargained for. Despite his love for teaching, he will reserve that for retirement and transition out of academia into the world of data analytics and data science where he can sit (mostly) in silence behind a computer screen.