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HEALTHY PATELLOFEMORAL KINEMATICS AND CONTACT FORCES DURING FUNCTIONAL ACTIVITIES

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*Submitted in total fulfilment of the requirements
of the degree of Doctor of Philosophy*

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Abstract

A better understanding of normal knee function is critical to the treatment of knee disorders. Limited data are available on knee biomechanics during functional activities such as walking, particularly in relation to the articulation of the patella. Three aims were formulated to address this gap in knowledge: 1) analyse the kinematics of the patellofemoral joint during functional activities, 2) determine the region of cartilage contact in the patellofemoral and tibiofemoral joints and their relationship to cartilage thickness, and 3) calculate the distribution of medial-lateral contact loads in the patellofemoral and tibiofemoral joints during level walking.

These aims were achieved by first accurately measuring three-dimensional kinematics of the patellofemoral joint as healthy young people performed six activities: level walking, downhill walking, stair descent, stair ascent, open-chain knee flexion, and standing. These data were examined for notable kinematic characteristics of the patella during ambulatory activities and to determine how the motion of the patella and the tibiofemoral flexion angle are linked (i.e., coupled) together. Cartilage models were created of each participant's knee in order to determine the region of cartilage contact for each of the activities performed, and to identify correlations between cartilage contact and cartilage thickness. Finally, musculoskeletal models with full six degree of freedom patellofemoral and tibiofemoral joints were created, used to calculate the medial-lateral contact loads at the knee during level walking, and finally validated against the measured kinematic data.

These procedures have revealed important findings. Patellar flexion and anterior translation were coupled and linearly related to the tibiofemoral flexion angle. Medial shift and superior translation were likewise coupled to tibiofemoral flexion, and both displayed notable characteristics for all ambulatory activities: the patella shifted laterally at low tibiofemoral flexion angles and underwent rapid superior translation just prior to heel strike. Based on the activities tested here, the patellofemoral joint can effectively be modelled as a one degree of freedom joint. The centroid of cartilage contact for both joints appears to be determined by the tibiofemoral flexion angle, and hence geometry, rather than activity. Patellofemoral contact was concentrated on the lateral side of both the patella and the femur. In each pair of contacting regions within the knee, one side of the pair exhibited a positive relationship between cartilage thickness and contact (i.e., the

medial and lateral tibial plateaus and the patella), while the other exhibited a weak or non-existent relationship (i.e., the medial and lateral femoral condyles in the tibiofemoral joint and the femur in the patellofemoral joint). The patellofemoral joint displayed two peaks in the contact force during level walking, one in early stance and one in swing phase, both at approximately 0.55-times bodyweight. Most of the patellofemoral contact force was transmitted through the lateral facet of the patella. The posterior component of hamstring muscle force contributed to the load transmitted to the patellar facets.

These findings may assist with the diagnosis and treatment of many common knee disorders and will provide a useful source of information for future investigations into the knee.

Declaration

This is to certify that:

- (i) the thesis comprises only my original work towards the PhD except where indicated in the preface,
- (ii) due acknowledgement has been made in the text to all other material used,
- (iii) the thesis is less than 100,000 words in length, exclusive of tables, maps, bibliographies, appendices and footnotes.

Lucas T Thomeer

March 2020

Preface

My supervisor, Professor Marcus G Pandy, advised me on all work presented in this dissertation.

Chapter 3 of this dissertation is based on work that was submitted for publication to Annals of Biomedical Engineering on March 4th, 2020. It describes and presents the results of an experimental study in which patellofemoral kinematics were captured. I worked with Dr. Hans A. Gray and Dr. Shanyuanyue Guan in the collection of this data. I was responsible for processing and analysis of the patella. Tibiofemoral kinematics were acquired in the same experiments; however, they were processed and analyzed by Dr. Gray and Dr. Guan and are not included in this dissertation (though their final results are referenced in Chapter 4 and Chapter 5). The tibiofemoral and patellofemoral kinematics for level walking were published together in the following work:

Gray, H.A., Guan, S., **Thomeer, L.T.**, Schache, A.G., de Steiger, R., Pandy, M.G., 2019. Three-dimensional motion of the knee-joint complex during normal walking revealed by mobile biplane x-ray imaging. J Orthop Res 37, 615-630.

Chapter 4 of this dissertation has not currently been published or submitted for publication. Dr. Guan assisted me in troubleshooting and running the functions written to measure cartilage thickness and calculate cartilage contact described in Chapter 4.

Chapter 5 of this dissertation has not currently been published or submitted for publication. The musculoskeletal model created and optimization problem formulated in it were based on a modelling framework previously developed by Dr Yi-Chung Lin. Dr Lin advised me on modifying and troubleshooting it as I added a patellofemoral joint.

I am responsible for all other work described in this dissertation.

During my candidature I was supported by a Postgraduate Scholarship provided by the University of Melbourne. Work in this dissertation was also, in part, funded by the Australian Research Council (ARC) Discovery Projects Grant Scheme (DP120101973).

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I am grateful to Professor Marcus G Pandy, my primary supervisor, for guiding me through my candidature. He has raised my standards for what high quality biomechanics research can be through both his curiosity to fully understand the cause and implications of any results we find, and desire to only finish with a project when we have made it the best that it can be. I can honestly say that, thanks to Professor Pandy, after almost four years of spending most of my time thinking about the mechanics of the healthy knee, I am even more interested in learning more about the way it works now than when I started.

I am also deeply indebted to Dr Hans A. Gray and Dr Yi-Chung Lin, my co-supervisors, who I relentlessly interrupted and questioned as both possess a wealth of experience for troubleshooting difficult experimental and programing problems and interpreting confusing results. Both were always available to help and gave me kind feedback, even when I just needed a sounding board for my inarticulate theories. In addition, I'm thankful to Dr Kathryn Stok, the chair of my advisory committee, for giving me honest advice at our meetings and encouragement to hurry up and write this dissertation.

A special thanks to all the students and postdocs with whom I shared a lab over the years. Especially Dr. Shanyuanye Guan, my desk mate, who gave me excellent advice, taught me to master MATLAB, showed me the world of modern videogames, and gave me a friend to walk home with after late worknights. I also owe a thanks to Dr Dale Robinson, Dr Wei Gu, and Aaditya Chandramouli, who went on many much-needed coffee and tea breaks with me when I was burning out in the middle of the day.

Thank you to Christine Lim, my fiancée, for putting up with my grouchy moods stemming from optimization problems that wouldn't cooperate, listening to me incoherently ramble about my thoughts on kneecaps, making sure that I did not starve (despite my best efforts), and coming up with an endless supply of knee puns. I'm also grateful to my family, especially my parents Paul and Cynthia Thomeer, for continually coming to visit me even though I moved to the opposite side of the world and for pretending to be interested when I told them what I was working on.

Finally, I would like to thank all the participants who volunteered to take part in this study. Without the time they sacrificed none of the findings presented in this dissertation would have been possible.

Contents

Abstract	i
Declaration	iii
Preface.....	iv
Acknowledgements	vi
Contents	viii
List of Figures.....	xi
List of Tables	xiii
List of Abbreviations	xiv
Chapter 1 Introduction	1
1.1 Motivation for this Dissertation.....	1
1.2 Contributions of this Dissertation.....	3
1.3 Structure of this Dissertation.....	4
Chapter 2 Background	5
2.1 Knee Anatomy.....	5
2.1.1 Knee Joint Complex	5
2.1.2 Bones	7
2.1.3 Knee-spanning Muscles.....	10
2.1.4 Articular Cartilage.....	14
2.1.5 Menisci	14
2.1.6 Tendons and Ligaments.....	15
2.2 Common Disorders of the Knee Joint	17
2.2.1 Osteoarthritis	18
2.2.2 ACL Reconstruction	18
2.2.3 Patellofemoral Pain	18
2.2.4 Etiology of Knee Disorders	19
2.3 Imaging Techniques	19
2.3.1 Static Imaging Techniques.....	19
2.3.2 Dynamic Imaging Techniques.....	20
2.4 Previous Investigations into Healthy Patellofemoral Kinematics.....	22

2.4.1	Walking.....	22
2.4.2	Stair and Ramp Ascent and Descent	23
2.4.3	Simple Activities	24
2.5	Previous Investigations into Healthy Knee Cartilage Contact	25
2.5.1	Cartilage Geometry	25
2.5.2	Cartilage Contact	26
2.6	Previous Investigations into Healthy Knee Contact Forces During Walking.....	27
2.6.1	Contact Models	27
2.6.2	Other Techniques	29
2.6.3	Direct Collocation	30
2.7	Aims of this Dissertation	31
Chapter 3 Healthy Patellofemoral Kinematics during Functional Activities		33
3.1	Introduction	33
3.2	Methods.....	34
3.2.1	Participants.....	34
3.2.2	Data Collection	34
3.2.3	Activities	35
3.2.4	Data Processing	36
3.2.5	Analysis.....	37
3.3	Results.....	38
3.3.1	Activity Velocities	38
3.3.2	Kinematic Features.....	38
3.3.3	Coupling in the Patellofemoral Joint	41
3.4	Discussion	41
Chapter 4 Cartilage Contact in the Knee and its Relationship to Thickness		44
4.1	Introduction	44
4.2	Methods.....	46
4.2.1	Joint Kinematics.....	46
4.2.2	Cartilage Thickness and Contact.....	47
4.2.3	Statistical Analysis	47
4.3	Results.....	48
4.3.1	Cartilage Thickness	48
4.3.2	Centroid of Cartilage Contact.....	49
4.3.3	Cartilage Contact and Thickness.....	50

4.3.4	Cartilage Contact Area.....	52
4.4	Discussion	52
Chapter 5	Load Distribution in the Patellofemoral Joint.....	56
5.1	Introduction	56
5.2	Methods.....	58
5.2.1	Participants and Data Collection	58
5.2.2	Model Development.....	59
5.2.3	Model Evaluation.....	60
5.2.4	Walking Simulation.....	60
5.2.5	Center of Contact	61
5.2.6	Post-Simulation Analysis	62
5.3	Results.....	62
5.3.1	Patellar Facet Size.....	62
5.3.2	Model Evaluation with Cadaver Experiments	62
5.3.3	Model Validation with Gait Experiments	65
5.3.4	Model-Predicted Joint Contact Force.....	67
5.3.5	Contribution of Hamstring Muscles	67
5.3.6	External Knee Adduction Moment.....	68
5.4	Discussion	69
Chapter 6	Conclusion.....	75
6.1	Introduction	75
6.2	Summary of Research Questions and Findings.....	75
6.2.1	Healthy Patellofemoral Kinematics during Functional Activities	76
6.2.2	Cartilage Contact in the Knee and its Relationship to Thickness	76
6.2.3	Load Distribution in the Patellofemoral Joint	77
6.3	Future Work.....	78
6.4	Closing Remarks.....	79
Appendix A		80
Appendix B		97
References		100

List of Figures

Figure 2.1: Knee Joint Complex.....	6
Figure 2.2: Distal Femur.....	7
Figure 2.3: Proximal Tibia	8
Figure 2.4: Proximal Fibula.....	9
Figure 2.5: Patella.....	10
Figure 2.6: Anterior Superficial Muscles of the Thigh	11
Figure 2.7: Posterior Superficial Muscles of the Thigh	12
Figure 2.8: Transverse View of Meniscus.....	15
Figure 2.9: Sagittal View of Knee Ligaments	16
Figure 2.10: Coronal View of Cruciate Ligaments	17
Figure 3.1: Top Down View of Step Locations for Gait Activities	36
Figure 3.2: Anatomical Coordinate System	37
Figure 3.3: Patellofemoral Kinematics and Degree of Coupling	40
Figure 4.1: Cartilage Thickness Map for a Single Subject.....	49
Figure 4.2: Centroid of Cartilage Contact	50
Figure 4.3: Motion of Contact Point on Convex and Concave Surfaces.....	54
Figure 5.1: Passive Ligament Testing	63
Figure 5.2: Evaluation of Modelled Patellofemoral Joint	63
Figure 5.3: Comparison of Model and Experimental Results	64
Figure 5.4: Model-Predicted Active Muscle Forces.....	66

Figure 5.5: Model-Predicted Joint Contact Forces.....	67
Figure 5.6: Posterior Hamstrings Force to Patellofemoral Contact Force	68
Figure 5.7: Correlations to the External Knee Adduction Moment	68
Figure 5.8: Comparison of Model-Predicted to Previously Reported Joint Loading.....	70
Figure 5.9: Forces Acting on the Patella	72
Figure 5.10: EMG Tracking Model Solution	73

List of Tables

Table 3.1: Patellofemoral Kinematics for Dynamic Activities 39

Table 4.1: Two-way ANOVA Results for Cartilage Contact and Activity Performed.. 51

Table 4.2: Depth of Contact to Cartilage Thickness Correlations..... 51

Table 4.3: Area of Cartilage Contact..... 52

Table 5.1: Participant Demographics 58

Table 5.2: Model Ligament Properties 60

Table 5.3: Model-predictions Compared to Motion Capture 65

List of Abbreviations

6-DOF	Six Degree of Freedom
ACL	Anterior Cruciate Ligament
BW	Bodyweight
CMC	Computed Muscle Control
CT	Computed Tomography
EKAM	External Knee Adduction Moment
EMG	Electromyography
LCL	Lateral Collateral Ligament
MCL	Medial Collateral Ligament
MoBiX	Mobile Biplane X-ray
MR	Magnetic Resonance
PCL	Posterior Cruciate Ligament
PF	Patellofemoral
PT	Patellar Tendon
R	Pearson's Correlation Coefficient
r^2	Coefficient of Determination
RMSE	Root Mean Square Error
ROM	Range of Motion
TF	Tibiofemoral

Chapter 1 Introduction

1.1 Motivation for this Dissertation

The knee is the largest joint in the human body (Palastanga et al., 2002). It is both complex, being made up of three bones (Standring, 2016), and weightbearing, experiencing multiple bodyweights of peak force during many common activities (Fregly et al., 2012). The primary function of the knee during ambulation is to shorten the lower limb, through flexion, and lengthen the lower limb, through extension (Palastanga et al., 2002), the importance of which cannot be overstated as it allows for the efficient completion of countless daily tasks, such as, walking, running, jumping, and sitting down.

Unfortunately, the knee is also a common site for injury (Thiem et al., 2013). Osteoarthritis, a leading cause of disability in the elderly in the United States (Lawrence et al., 1998), is a joint disease identified by a gradual loss of articular cartilage resulting in chronic pain and stiffness (Hung and Ahmed, 2017; Kapoor and Mahomed, 2015). Anterior cruciate ligament reconstructions, a knee surgery frequently required for athletes (Mall et al., 2014), are linked to an increased prevalence of knee pain and osteoarthritis in the years following the procedure (Lohmander et al., 2004; Von Porat et al., 2004). Patellofemoral pain, a broad diagnosis for pain in the anterior portion of the knee (Thomee et al., 1999), primarily affects young and active individuals (Baquie and Brukner, 1997; Rothermich et al., 2015), and makes common activities, such as ascending stairs, difficult to perform. While these knee disorders all have different and complex etiologies, improved treatment for each of them begins with a robust understanding of the mechanics and function of the healthy knee.

However, direct measurement of the knee's mechanics presents several difficulties. First, the patella (also called the kneecap, a small sesamoid bone in the anterior knee; see section 2.1.2.4), while vital to the knee's healthy function, is often neglected in studies of the knee because it is small and difficult to image. Second, most imaging techniques (see section 2.3) that are capable of capturing the motion of the knee during daily activities are not able to accurately measure the motion of the joint, and most imaging techniques that are able to accurately capture knee kinematics require the knee to remain stationary, limiting the range of tasks that can be completed. Despite this, many past studies (see

sections 2.4 and 2.5) have attempted to study the motion of the knee joint during activities such as level walking (Lafortune, 1984; Stein et al., 1993), downhill walking (Farrokhi et al., 2015), stair climbing (Lin et al., 2019; Suzuki et al., 2012), knee flexion (Koh et al., 1992; Komistek et al., 2000; Seisler and Sheehan, 2007), lunging (Nha et al., 2008; Sharma et al., 2017), and squatting (Li et al., 2007). Yet these studies have all been limited by factors like restrictive or invasive imaging methodologies or non-healthy participants, either of which compromises the ability of their findings to apply to healthy knees in general.

To overcome the limitations inherent to direct measurements of the knee during functional daily activities, past studies have employed modelling methodologies. The use of a computational model allows for data to be collected and hypotheses to be tested that would otherwise be impossible to obtain *in vivo* in a healthy knee. For the most common daily ambulatory task, level walking, numerous studies in recent years (see section 2.6) have constructed contact models of the knee joint in order to study the motion of the bones in a healthy knee and the contact forces that are regularly transmitted between them (Brandon et al., 2018; Halonen et al., 2016; Hu et al., 2018; Lenhart et al., 2015; van Rossom et al., 2018). However, these studies are all ultimately limited by the same factors as the direct measurement studies. The knee joint is complex and there are many sources of error in model development, making model evaluation critical before a solution can be trusted. As it is not presently possible to directly measure the contact forces that occur *in vivo* within a healthy knee, the next best solution is kinematic evaluation as contact forces and joint kinematics are linked through Newton's Second Law of Motion, but this is only possible with accurate kinematic measurements.

A recent advancement in knee imaging technology has allowed for an improvement over previous studies in this area. Guan et al. (2016a) developed a mobile biplane X-ray imaging system that is capable of noninvasively measuring the motion of a full and healthy knee joint over a large capture volume. The accuracy of these measurements is reported to be less than 1.0 mm for translations and less than 1.5° for rotations (Gray et al., 2017; Guan et al., 2016a), allowing for direct measurement and model evaluation of knee mechanics during the full stride cycle of functional ambulatory activities at an accuracy that was not previously possible.

1.2 Contributions of this Dissertation

The objective of this dissertation was to improve current understanding of the healthy knee joint and its function during everyday dynamic activities. This objective was accomplished by capturing accurate data on the motion of the knee joint complex in ten healthy individuals while they performed six activities: level walking, downhill walking, stair descent, stair ascent, open-chain knee flexion, and standing. These experimental data were then used in conjunction with a modelling framework to address the following three sets of questions (motivations for these individual questions will be provided in sections 3.1, 4.1, and 5.1):

1. What are the full three-dimensional kinematics of the patellofemoral joint during the performed activities? To what extent are each of the 6-degrees-of-freedom of the patellofemoral joint coupled to the flexion-extension angle of the tibiofemoral joint?
2. Where are the locations of cartilage contact in the knee joint complex (i.e., tibiofemoral and patellofemoral joints) during the performed activities? Do thicker regions of cartilage experience more joint contact than thinner regions of cartilage in the knee? Is there a correlation between cartilage thickness and contact?
3. What are the contact loads at the knee during level walking, and how are these loads transmitted through the medial and lateral facets of the patellofemoral and tibiofemoral joints? Do the hamstrings contribute to the load transmitted to the patellar facets? How does the vastus lateralis muscle respond to the external knee adduction moment during walking, and does this response contribute to patellofemoral loading?

1.3 Structure of this Dissertation

This dissertation is structured as follows:

- Chapter 2 presents the background knowledge necessary to understand the subsequent methods and results presented in this dissertation, including a review of the literature relevant to each of the three aims.
- Chapter 3 presents an experimental study on the kinematics of the patellofemoral joint during six activities of daily living: level walking, downhill walking, stair ascent, stair descent, open-chain knee flexion, and static standing.
- Chapter 4 presents an experimental study on the cartilage contact in the patellofemoral and tibiofemoral joints during the aforementioned activities, and its relationship to cartilage thickness.
- Chapter 5 presents a modelling study on the distribution of medial-lateral contact loads in the patellofemoral and tibiofemoral joints during level walking.
- Chapter 6 reviews the findings of this dissertation and gives potential areas of study for future research.

Chapter 2 Background

2.1 Knee Anatomy

The following section gives a brief overview of the human musculoskeletal system in relation to the knee joint.

2.1.1 Knee Joint Complex

The human knee (Figure 2.1) is a synovial joint located in the lower limb which joins the upper leg to the lower leg (Standring, 2016). Often viewed as a simple hinge joint, the primary motion of the knee is in flexion and extension (Palastanga et al., 2002). However, the knee is actually a compound joint that is made up of two individual joints: a) the tibiofemoral (TF) joint, consisting of the femur and the tibia, and b) the patellofemoral (PF) joint, consisting of the femur and the patella (Standring, 2016). The combination of these two individual joints gives the knee 12 possible degrees of freedom in its motion (three rotations and three translations in both the TF and PF joints).

2.1.1.1 Tibiofemoral Joint

The most important function of the TF joint during ambulation is to shorten and lengthen the lower limb through flexion and extension, respectively (Palastanga et al., 2002). This allows the leg to undergo the swing phase of activities such as walking and running. The TF joint is weightbearing and absorbs forces that are several times bodyweight through its two separate contacting surfaces on the medial and lateral sides (van Rossom et al., 2018). Many muscles (2.1.5) and ligaments (2.1.4) cross the TF joint in order to stabilize it against the large forces it is subjected to while allowing for the range of motion necessary for movement (Standring, 2016).

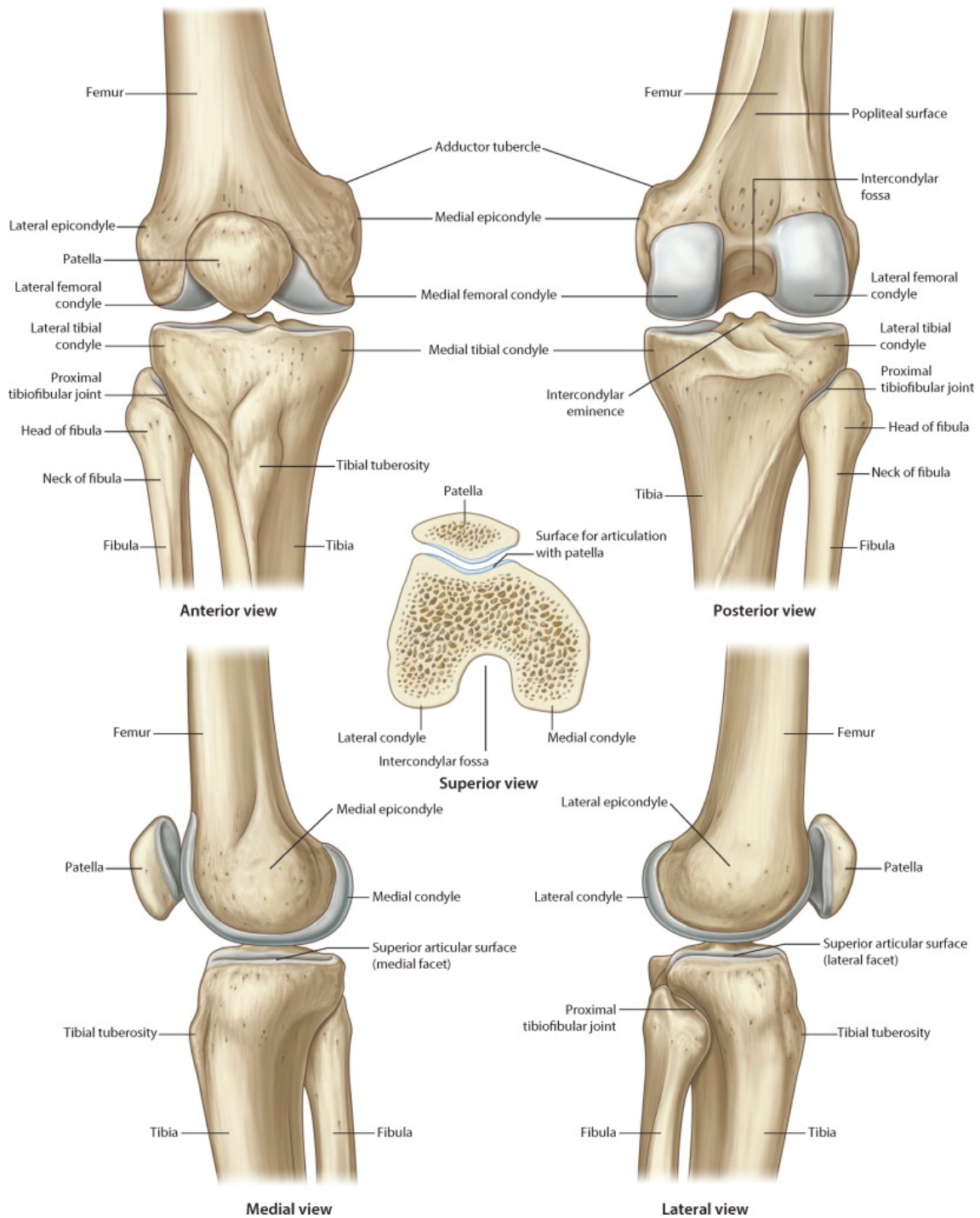


Figure 2.1: Knee Joint Complex. Diagram of the knee joint complex showing the relative positions of the femur, tibia, and patella. Reprinted from Drake (2015) with permission from Elsevier (see page 97 for details).

2.1.1.2 Patellofemoral Joint

The PF joint serves two purposes in the function of the knee joint complex. First, when the TF joint is close to full extension the PF joint provides a mechanical advantage to the quadriceps femoris muscle in generating a knee extension moment (Im et al., 2015). This is critical to proper functioning of the knee as much of the healthy gait cycle takes place

with the knee under 20 degrees of extension (Gray et al., 2019), and the quadriceps femoris muscle's ability to produce force decreases as the TF joint extends (Abernethy et al., 2013). Second, when the TF joint is flexed and the soft tissue of the distal femoral and proximal tibial surfaces would otherwise be exposed to damage from external forces (e.g., kneeling on the floor), the patella slides in front and serves to protect the TF joint (White and Folkens, 2005).

2.1.2 Bones

The human knee includes four total bones: femur, tibia, fibula, and patella. Relative to the soft tissue that surrounds it, bone is rigid and does not deform much when loaded with normal muscle forces. There are two types of macroscopic bone structure: a) compact bone, located in the outer edge of a bone cross-section, it is dense and provides a hard surface for articulation, and b) trabecular bone, located in the middle of a bone cross-section, it is sparse and supports the compact bone (Standring, 2016).

2.1.2.1 Femur

The femur is located in the proximal half of the lower limb and is the largest bone in the human body (Standring, 2016). It can be divided into three sections: a) the proximal end, which contains the femoral head that acts as the ball in the hip's ball and socket joint, b) the femoral shaft, which is cylindrical and connects the two ends of the femur, and c) the distal end (Figure 2.2), which articulates with the patella and tibia to form the PF and TF joints, respectively (Ellis, 2002).

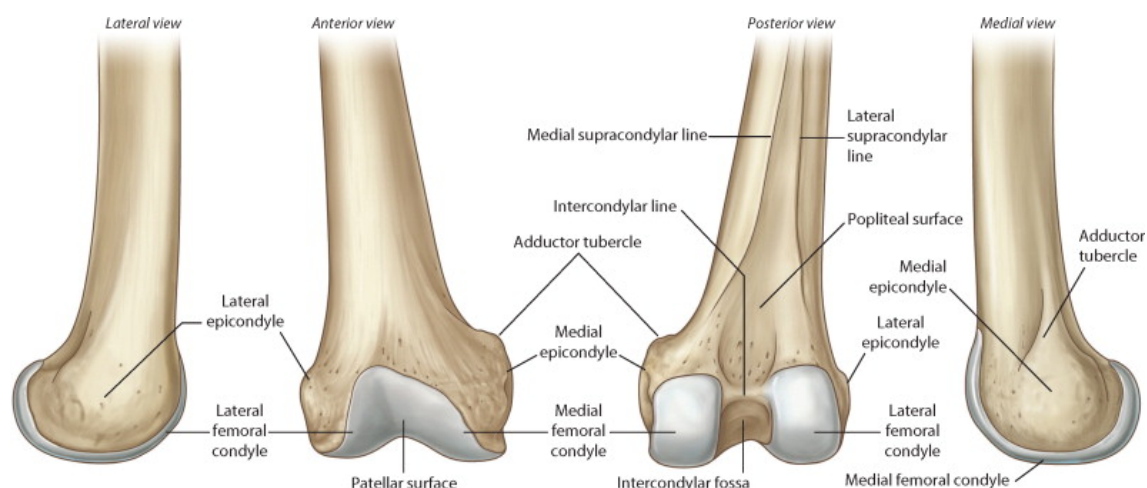


Figure 2.2: Distal Femur. Reprinted from Drake (2015) with permission from Elsevier (see page 97 for details).

The distal end of the femur contains two condyles, one each on the medial and lateral sides, of which the lateral is larger (Palastanga et al., 2002). The inferior and posterior surfaces of the condyles are roughly cylindrical and contact the tibia to form the TF joint. The anterior surfaces of the condyles protrude forward from the femoral shaft to form the trochlear groove (*patellar surface* in Figure 2.2) in the space between them, which the patella rides in as part of the PF joint (Standring, 2016). The gap between the posterior ends of the two condyles is the intercondylar fossa, which serves as an attachment point for the cruciate ligaments (2.1.4.2) (Palastanga et al., 2002).

2.1.2.2 Tibia

The tibia is located below the femur and, together with the fibula (2.1.2.3), makes up the shank portion of the lower leg (Palastanga et al., 2002). The tibia has three sections: a) the proximal end (Figure 2.3), which is roughly flat and serves opposite the femur in the TF joint, b) the tibial shaft, which is roughly triangular in cross-section and connects the two ends of the tibia, and c) the distal end, which is smaller in size than the proximal end and contains surfaces to articulate with the talus and fibula (Ellis, 2002).

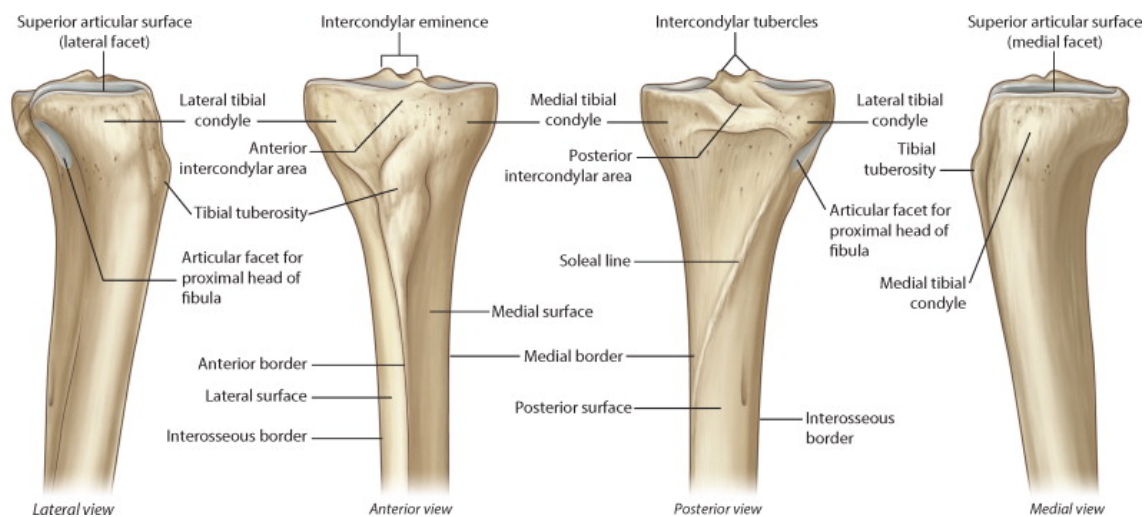


Figure 2.3: Proximal Tibia. Reprinted from Drake (2015) with permission from Elsevier (see page 97 for details).

The proximal end of the tibia contains three areas of interest. First, the superior surface of the tibia contains medial and lateral facets that articulate with the medial and lateral condyles of the femur (Standring, 2016). The medial facet is larger and has a slightly concave shape, while the lateral facet is either flat or slightly convex (Hashemi et al., 2008). Second, the intercondylar eminence protrudes superiorly and is located between the facets. Third, the tibial tuberosity projects out on the anterior edge of the proximal

tibia and serves as the insertion point for the patellar tendon (2.1.4.1) (Palastanga et al., 2002).

2.1.2.3 Fibula

The fibula is located lateral to the tibia in the lower leg. Both the proximal (Figure 2.4) and distal ends of the fibula articulate with the tibia. The proximal end of the fibula serves as an insertion point for ligaments and muscles that cross the lateral side of the TF joint (Ellis, 2002).

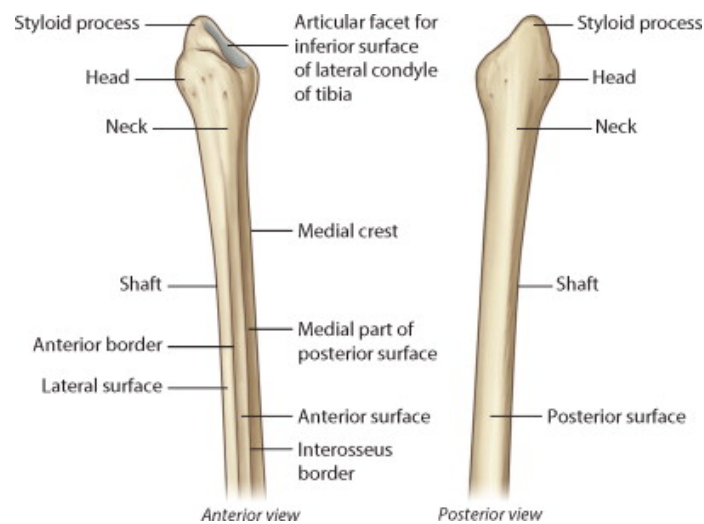


Figure 2.4: Proximal Fibula. Reprinted from Drake (2015) with permission from Elsevier (see page 97 for details).

2.1.2.4 Patella

The patella (Figure 2.5) is a sesamoid bone that is located anterior to the distal end of the femur and articulates in the trochlear groove to form the PF joint (Ellis, 2002). The posterior surface of the patella is divided into medial and lateral facets by a vertically running patellar ridge, which protrudes posteriorly from the facets (Palastanga et al., 2002). The lateral facet is wider and has a larger surface area than the medial facet (Baldwin and House, 2005; Tecklenburg et al., 2006). The quadriceps tendon (2.1.4.1) inserts on the superior surface of the patella, while the patellar tendon (2.1.4.1) originates from the inferior apex of the patella (Standring, 2016). Thus, the patella is responsible for transferring force between the two tendons which allows the quadriceps femoris muscle (2.1.3.1) to extend the TF joint.

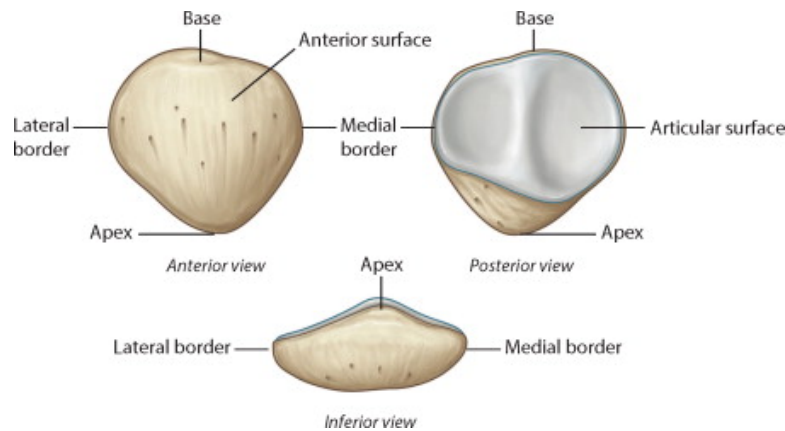


Figure 2.5: Patella. Reprinted from Drake (2015) with permission from Elsevier (see page 97 for details).

2.1.3 Knee-spanning Muscles

There are three different types of muscle in the human body: a) skeletal muscle, b) cardiac muscle, and c) smooth muscle. All three forms of muscle are made of small muscle fibres that contain elements that allow the fibre, and thus the whole muscle, to lengthen or shorten (Standring, 2016). Skeletal muscle is the most common and the only form of muscle that can be voluntarily controlled (Palastanga et al., 2002). For the remainder of this thesis ‘muscle’ will refer to ‘skeletal muscle’. Several notable muscles that cross the knee and contribute to PF and TF joint movement and contact forces are introduced in this section; however, note that this is not a complete detailing of all knee spanning muscles.

2.1.3.1 Anterior Thigh Muscles

The quadriceps femoris (Figure 2.6) is a large muscle located anterior to the femur that is responsible for extending the knee. It has four distinct parts: a) rectus femoris, b) vastus intermedius, c) vastus lateralis, and d) vastus medialis, of which the vastus lateralis is the largest (Standring, 2016).

The rectus femoris originates from two locations on the pelvis, runs approximately down the middle of the thigh in the medial-lateral direction, and inserts into the patella via the patellar tendon (2.1.5.1) (Drake, 2015). This makes the rectus femoris biarticular, flexing the hip and extending the knee when activated. The rectus femoris is in line with and superficial to the vastus intermedius; however, the deep surface of the rectus femoris is smooth to allow it to freely slide over the vastus intermedius and function independently (Palastanga et al., 2002).

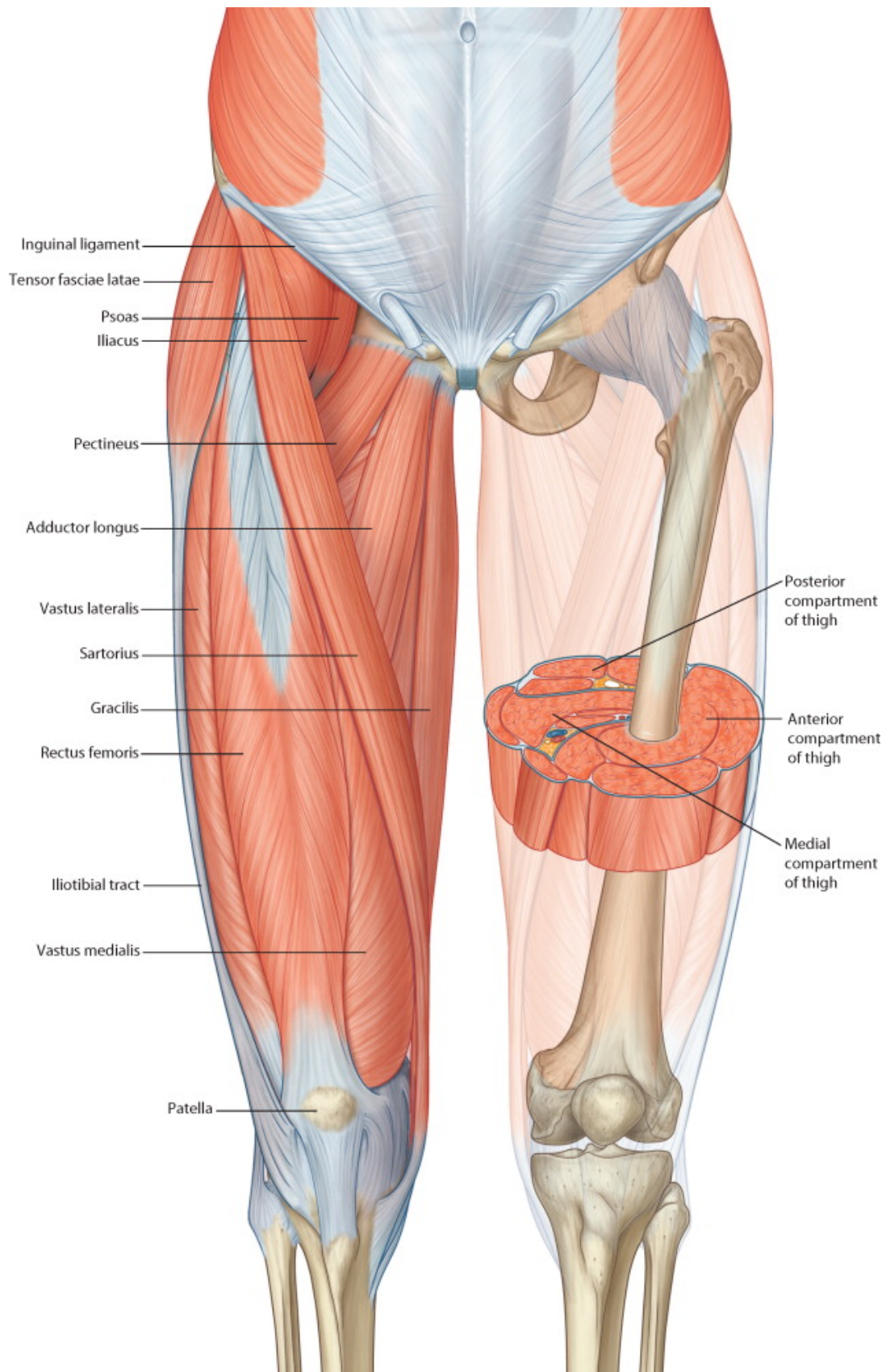


Figure 2.6: Anterior Superficial Muscles of the Thigh. Reprinted from Drake (2015) with permission from Elsevier (see page 97 for details).

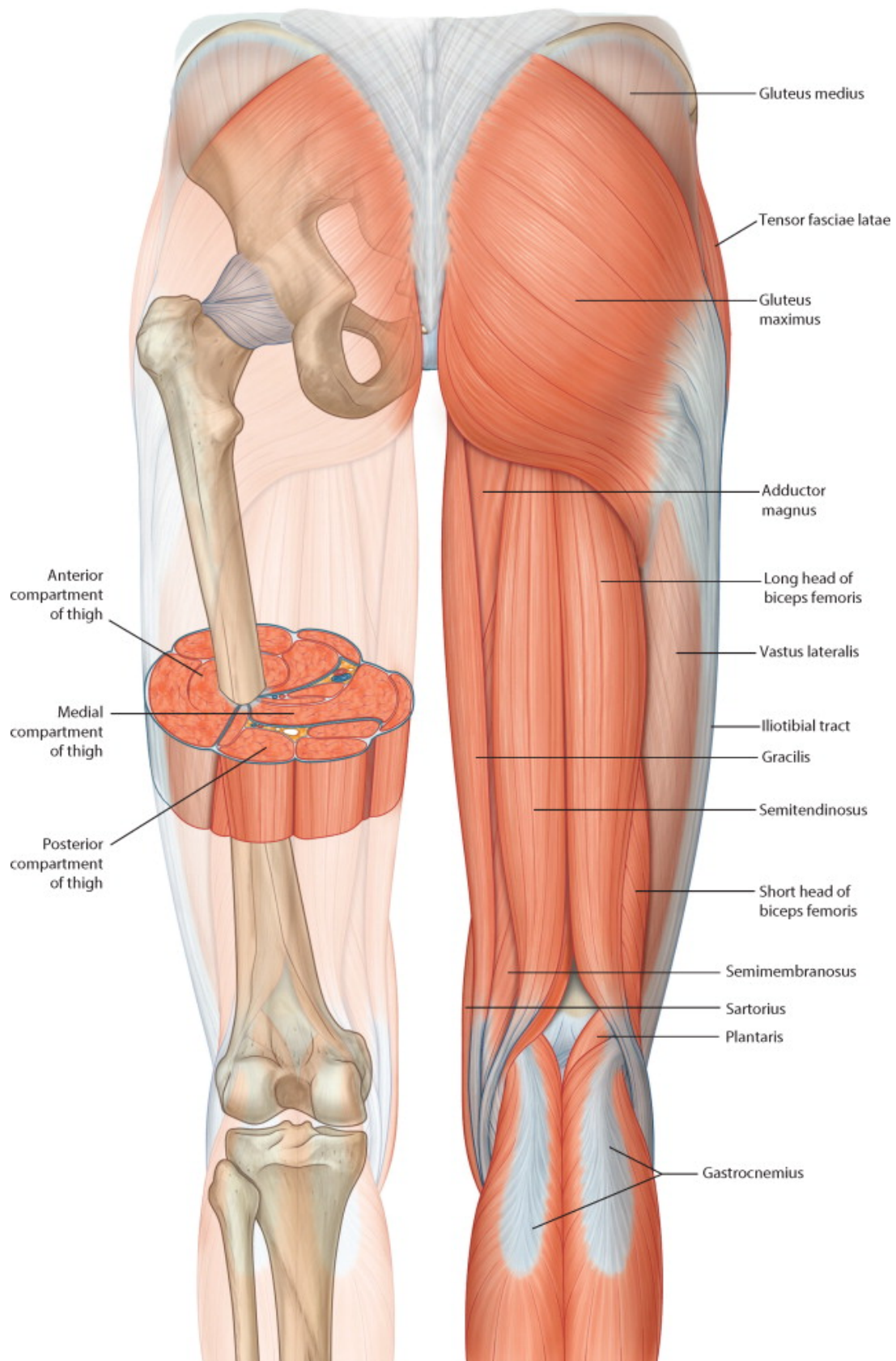


Figure 2.7: Posterior Superficial Muscles of the Thigh. Reprinted from Drake (2015) with permission from Elsevier (see page 97 for details).

The vastus intermedius, vastus lateralis, and vastus medialis are collectively referred to as the vasti. The vasti are all uniarticular, originating in the upper half of the femoral shaft and inserting into the patella via the patellar tendon (Drake, 2015). They extend the knee when activated. The vastus intermedius runs down the center of the thigh and is mostly covered by the rectus femoris (Standring, 2016). The vastus lateralis and vastus medialis run on the lateral and medial sides of the thigh, respectively.

2.1.3.2 Posterior Thigh Muscles

There are three major posterior thigh muscles: a) biceps femoris, b) semitendinosus, and c) semimembranosus (Figure 2.7). These three muscles are collectively referred to as the hamstrings. The hamstrings are a biarticular muscle group, both extending the hip and flexing the knee when activated (Standring, 2016). They oppose the action of the quadriceps.

The biceps femoris originates from two locations that are a considerable distance apart. The long head of the biceps femoris, as well as the semitendinosus and semimembranosus, originates from the posterior surface of the pelvis (Drake, 2015). The short head of the biceps femoris originates from the distal posterior surface of the femoral shaft (Palastanga et al., 2002). This makes the short head of the biceps femoris a uniarticular component of the hamstrings. Both components of the biceps femoris insert into the proximal head of the fibula on the lateral side of the knee (Palastanga et al., 2002). The semitendinosus, which is notable for the long length of its distal tendon, runs down the medial side of the thigh, wraps around the medial condyle of the tibia, and inserts into the proximal medial tibia (Standring, 2016). The semimembranosus runs beneath the semitendinosus down the medial side of the posterior thigh, it inserts in the posterior surface of the medial tibial condyle (Palastanga et al., 2002).

2.1.3.3 Posterior Shank Muscles

The gastrocnemius is a strong biarticular muscle located superficially in the calf (Standring, 2016). The primary purpose of the gastrocnemius is ankle plantar flexion; however, when activated, it also flexes the knee (Palastanga et al., 2002). The muscle originates from two locations, directly superior to the medial and lateral condyles on the posterior surface of the femur, and inserts into the posterior surface of the calcaneus

(Drake, 2015). The medial component of the gastrocnemius is larger than the lateral component (Standring, 2016).

2.1.4 Articular Cartilage

Hyaline cartilage covers the articular surfaces of the femur, tibia, and patella in the knee joint (Palastanga et al., 2002). For the remainder of this thesis ‘cartilage’ will refer to ‘hyaline cartilage’. Its thickness varies from 2 to 5 mm between typical subjects (Coleman et al., 2013; Koo et al., 2005; Li et al., 2005). Articular cartilage reduces friction between the contacting bones and increases the contacting area in the joint which reduces stress (Standring, 2016). While it is an elastic material, cartilage is too thin to act as a shock absorber for the knee (Martin et al., 2015; Standring, 2016). Cartilage is avascular, and nutrients are supplied to it through the synovial fluid contained in the joint capsule surrounding it (Standring, 2016).

2.1.5 Menisci

The menisci of the TF joint (Figure 2.8) are semicircular rings of tough cartilage located superior to the articular cartilage of the medial and lateral tibial facets. The primary purpose of the menisci is to increase the contact area between the round femoral condyles and the flatter tibial facets (Palastanga et al., 2002). The menisci are triangular in cross-section, with the apex of the triangle lying closer to the center of the tibial facet. Both menisci are attached to the tibia and femur through several ligaments, this allows for a small amount of movement relative to the tibia (Standring, 2016).

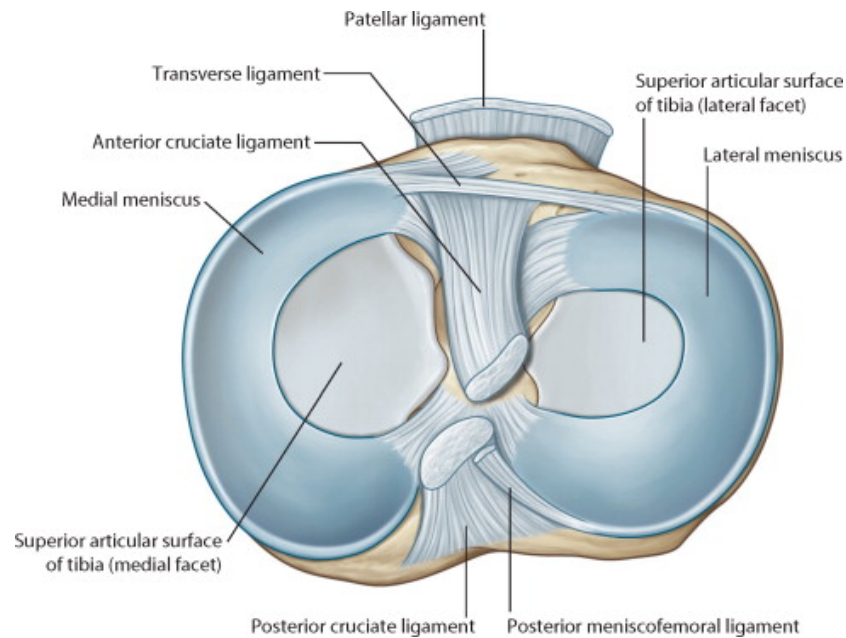


Figure 2.8: Transverse View of Meniscus. Reprinted from Drake (2015) with permission from Elsevier (see page 97 for details).

2.1.6 Tendons and Ligaments

Tendons and ligaments are responsible for transferring force from muscle to bone and passively restricting the motion of bones, respectively (Standring, 2016). They are a form of fibrous tissue called white fibrous tissue that is dense with collagen bundles to give tensile strength (Palastanga et al., 2002). Several notable tendons and ligaments of the knee joint are detailed in this section.

2.1.6.1 Quadriceps and Patellofemoral Tendons

The quadriceps tendon and the patellar tendon, also known as the patellar ligament, (Figure 2.9) are located anterior to the knee and transmit force from the quadriceps to the tibia through the patella. The quadriceps tendon begins individually in each component of the quadriceps femoris muscle, before coming together to form a single thick tendon that attaches to the superior surface of the patella (Standring, 2016). The order and orientation of the components of the quadriceps femoris muscle within the quadriceps tendon varies considerably between individuals (Waligora et al., 2009). The patellar tendon originates from the inferior apex of the patella and inserts on the tibial tuberosity of the tibia. It functions as an extension of the quadriceps tendon by delivering the force generated in the quadriceps to the tibia (Palastanga et al., 2002). The relative angle between the lines-of-action of the quadriceps tendon and patellar tendon in the coronal plane is known as the Q-angle (Smith et al., 2008).

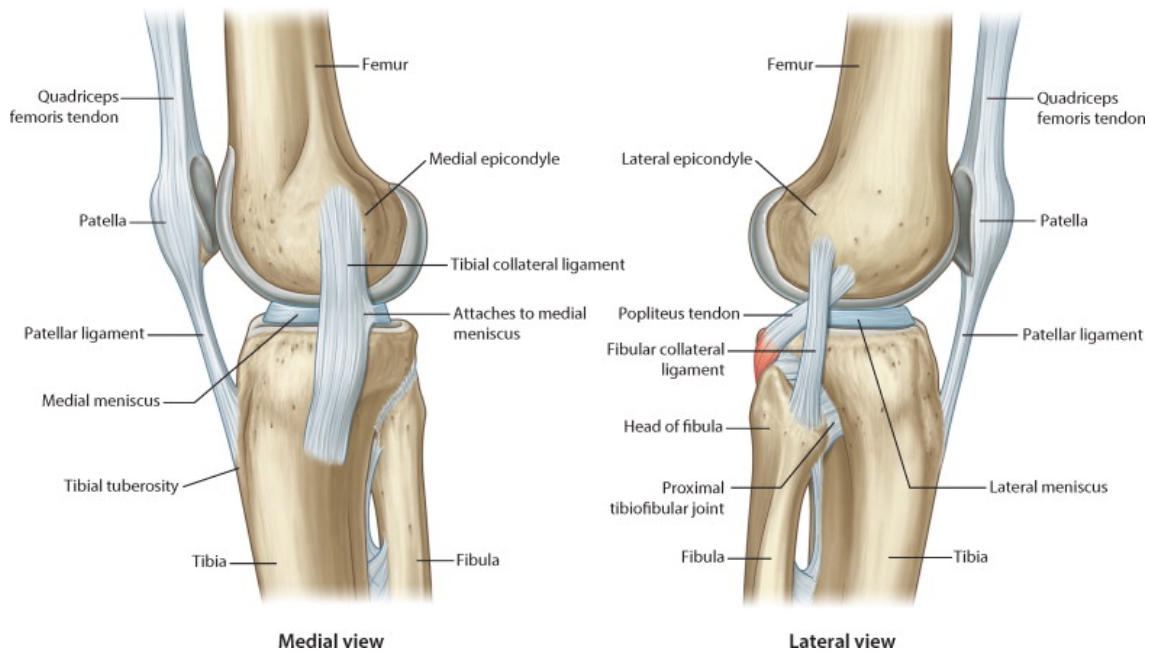


Figure 2.9: Sagittal View of Knee Ligaments. Reprinted from Drake (2015) with permission from Elsevier (see page 97 for details).

2.1.6.2 Collateral Ligaments

The medial and lateral collateral ligaments (Figure 2.9), also known as the tibial and fibular collateral ligaments, respectively, run medial and lateral to the articular surfaces of the TF joint and serve to stabilize it. The medial collateral ligament (MCL) originates on the medial epicondyle of the femur and inserts on both the medial condyle of the tibia and the superior portion of the tibial shaft (Standring, 2016). The MCL resists abduction moments in the TF joint that would cause valgus in the knee. The lateral collateral ligament (LCL) originates on the lateral epicondyle of the femur and inserts on the proximal head of the fibula (Standring, 2016). The LCL resists adduction moments in the TF joint that would cause varus in the knee.

2.1.6.3 Cruciate Ligaments

The anterior and posterior cruciate ligaments (Figure 2.10) are located between the articular surfaces of the TF joint. Both ligaments run approximately along the sagittal plane in opposite directions, crossing in the middle (Standring, 2016). The anterior cruciate ligament (ACL) originates on the medial surface of the lateral femoral condyle and inserts anterior to the intercondylar eminence on the tibia (Palastanga et al., 2002). The ACL stabilizes the TF joint against anterior sliding of the tibia relative to the femur. The posterior cruciate ligament (PCL) originates on the lateral surface of the medial

femoral condyle and inserts posterior to the intercondylar eminence on the tibia (Palastanga et al., 2002). The PCL resists posterior sliding of the tibia relative to the femur in the TF joint. Both ligaments are made of at least two distinct bundles (Standring, 2016).

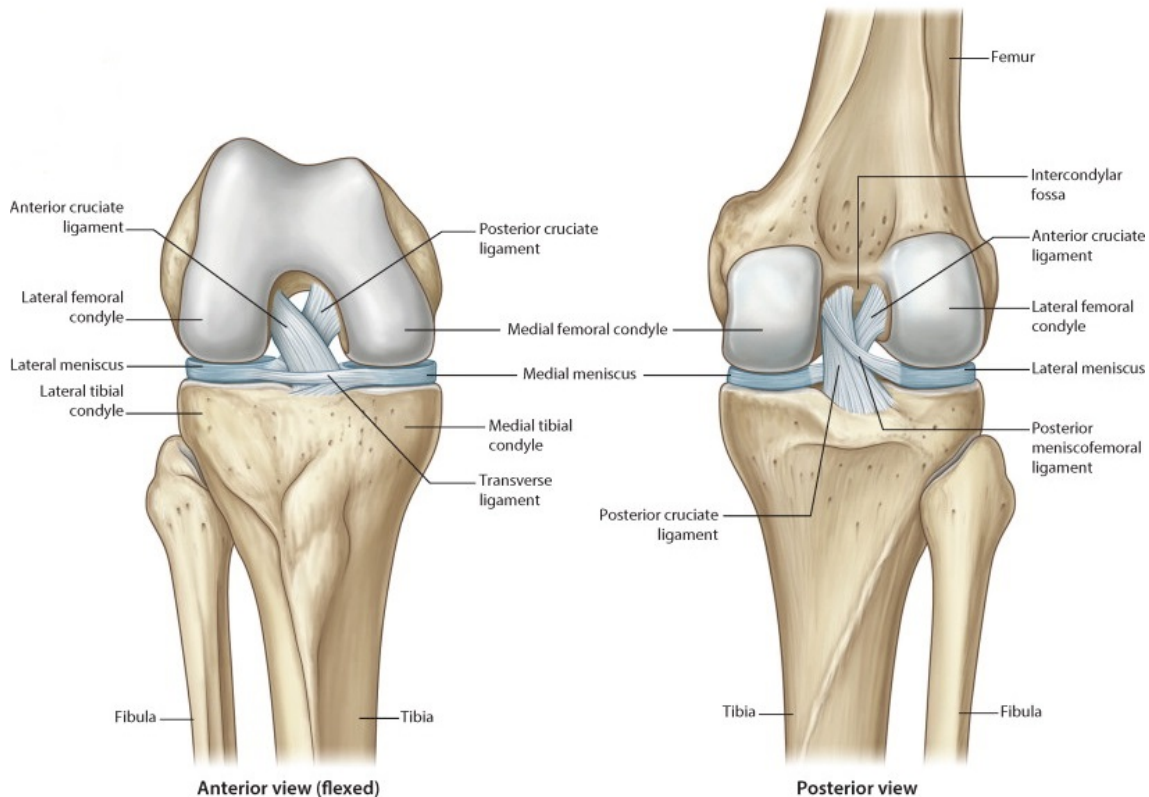


Figure 2.10: Coronal View of Cruciate Ligaments. Reprinted from Drake (2015) with permission from Elsevier (see page 97 for details).

2.2 Common Disorders of the Knee Joint

The following section details a few examples of knee ailments and the commonalities in their etiologies. While this dissertation is focused on healthy knees, the information presented in this section is important for understanding the motivation for the studies that will subsequently be presented.

As a complex and weightbearing joint, the knee is a common site for injury. A survey by Thiem et al. (2013) of 7,828 individuals over 40 years of age in Herne, Germany found that 30.9% of respondents experienced knee pain. This pain can come through many different methods that affect populations of different ages and activity levels; a few examples of possible injury methods are covered below.

2.2.1 Osteoarthritis

Osteoarthritis is a joint disease typified by a gradual loss of articular cartilage (Kapoor and Mahomed, 2015). Its prevalence increases with age, making it a leading cause of disability in the elderly in the United States (Lawrence et al., 1998). Osteoarthritis in the knee and hip is especially prevalent, being responsible for approximately 95% of total knee and hip replacements (Kapoor and Mahomed, 2015). While knee osteoarthritis as a disease is difficult to create strict diagnostic criteria for, individuals who have been diagnosed with it typically report symptoms such as chronic knee pain, difficulty flexing or bearing weight through the knee, and stiffness in the morning (Hung and Ahmed, 2017). Among patients who have knee osteoarthritis it is most common in the PF (22%) and medial TF (44.7%) compartments (Davies et al., 2002). It can exist in any combination of the knee compartments, including all of them or just one.

2.2.2 ACL Reconstruction

ACL injuries are common in sports and often occur in non-contact scenarios when an athlete lands from a jump or rapidly changes direction (Kanosue et al., 2015). ACL injuries come in a range of severities, from being stretched too far but remaining intact to being completely torn. When severe enough, the injury is treated with an ACL reconstruction, a surgical replacement of the ACL (Prodromos, 2018). Mall et al. (2014) found that the incidence of ACL reconstructions has been increasing in recent years, and that people under the age of 20 are most at risk of needing to receive an ACL reconstruction. In addition, ACL reconstructions, even when successful, have also been linked to a higher prevalence of knee pain and osteoarthritis 10 to 15 years later (Lohmander et al., 2004; Von Porat et al., 2004).

2.2.3 Patellofemoral Pain

PF pain is a broad diagnosis that refers to knee pain felt around the patella or more generally in the anterior portion of the knee (Thomee et al., 1999). PF pain often causes difficulty in performing activities in which the knee joint needs to be flexed, such as ascending stairs or rising from a chair. Uncertainty around the exact etiology of PF pain has made it difficult to treat. It has been shown to primarily affect a similar segment of

the population as ACL reconstructions, with young and active individuals being the most likely to be diagnosed (Baquie and Brukner, 1997; Rothermich et al., 2015).

2.2.4 Etiology of Knee Disorders

While all of the knee disorders described above have complex etiologies that are not fully understood, they have all been linked to altered kinematics or loading of the knee joint in some way (Buckwalter et al., 2004; Carter et al., 2004; Maetzel et al., 1997; Sharma et al., 2001; Thomeer et al., 2017; Wilson, 2007; Zabala et al., 2015). However, understanding how and to what extent the mechanics of these knees differs from that of healthy knees is limited by a lack of data on the loading and movement of a healthy knee joint during functional activities (2.4). Further work needs to be done in this area.

2.3 Imaging Techniques

There are many different methodologies for capturing the geometry and kinematics of the bones and soft tissue of the knee joint complex. The following section will define several of these techniques and briefly explain the positives and negatives associated with using them. Only methodologies that can be used on healthy *in vivo* knees are examined here.

2.3.1 Static Imaging Techniques

Static imaging refers to imaging techniques that are used on stationary knees with the goal of accurately capturing the geometry or shape of the bones or soft tissue in the knee joint complex. These techniques all require the subject to remain as still as possible in the capture volume for a variable length of time in order to capture a high-quality image.

2.3.1.1 Computed Tomography Scanning

Computed tomography (CT) is a technique in which a series of x-ray images are gathered from different angles and combined to create cross-sectional images of the scanned object. In other words, as opposed to traditional x-ray imaging, where a three dimensional object is compressed into a two dimensional plane along the direction of the x-ray, CT scanning allows for a single cross-sectional plane to be imaged, increasing the contrast resolution of the image (Hsieh, 2015). CT scans excel in capturing denser objects such as bone or metal joint replacements, as opposed to less dense soft tissue where they are less

effective. As x-rays are used during CT scans, subjects are exposed to radiation (Buzug, 2008).

2.3.1.2 Magnetic Resonance Imaging

Magnetic resonance (MR) imaging is a methodology where three dimensional geometric and material data can be measured from an object by recording the locations and resonant frequencies that atomic nuclei in the object emit when exposed to different electromagnetic waves generated by the MR scanner (Prasad, 2006). In comparison to CT scanning, the primary benefit of MR scanning is in its ability to detect and differentiate soft tissue because of their high water content (Vlaardingerbroek and Boer, 2003). At present the electromagnet waves used in MR scanning have no known negative side effects in subjects (Prasad, 2006).

2.3.2 Dynamic Imaging Techniques

Dynamic imaging refers to imagining techniques that are used on knees that are in motion at the time of imaging in order to capture the kinematics or relative motion of the bones in the knee. These techniques sometimes require a static imaging technique to be done in conjunction with them in order to process the data.

2.3.2.1 Video Motion Capture using Skin Markers

Video motion capture using skin markers is the movement measuring technique where reflective markers are placed over bony landmarks on the skin and the markers' location is recorded over time by multiple cameras, this data is then used to estimate the location of the bony landmarks over time (Andriacchi and Alexander, 2000). The advantages of motion capture are that it is non-invasive, applicable in many different physical environments, and has a large capture volume that can record whole body movements. However, joint kinematics measured using a skin marker system assume there is little to no movement between the skin and the underlying bone, which in many scenarios is not true (Akbarshahi et al., 2010; Cappozzo et al., 1996).

2.3.2.2 Intra-Cortical Bone Pins

Intra-cortical bone pins are stainless steel rods that, on one end, are screwed through the skin into the underlying bone, and on the other end contain a set of three reflective markers (Ramsey et al., 2003). When used in conjunction with motion capture cameras, the bone pins' rigid connection allows for the measurement of skeletal kinematics without the skin motion error of skin markers. However, bone pins are also invasive, cause pain that possibly affects movement, and sometimes deform at large joint angles (Andriacchi and Alexander, 2000; Ramsey et al., 2003).

2.3.2.3 Dynamic Magnetic Resonance Imaging

Dynamic MR imaging describes any MR imaging sequence that has a temporal dimension. While there are multiple methods to do this (Borotikar et al., 2012; Sheehan et al., 1997), they all require a sacrifice in spatial resolution, capturing a few cross-sectional planes over time instead of the whole three dimensional volume of a static MR scan. They are non-invasive and can capture low resolution soft tissue. Conversely, the capture volume of dynamic MR imaging is confined to the bore of the MR scanner and the action to be imaged must be precisely repeated many times during the imaging sequence.

2.3.2.4 Fluoroscopy

Fluoroscopy is an imaging technique where a continuous stream of x-rays are passed through the capture volume in order to record the imaged object in real time (Seeram, 2019). Fluoroscopy machines can capture the real time movement of bones without the use of invasive bone pins; however, they do expose their subjects to ionizing radiation. When compared to MR scanners, fluoroscopy machines are smaller, more portable, and more affordable. Single fluoroscopy machine setups are less accurate than biplane fluoroscopy, as a single fluoroscopy machine has difficulty detecting translations in the direction of capture (Fernandez et al., 2008; Giphart et al., 2012). Stationary fluoroscopy machines are limited by their relatively small capture volume, while fluoroscopy machines that are mounted to gantry systems are able to move to keep the object of interest inside the capture volume, these are known as mobile fluoroscopy machines. However, while multiple mobile single plane fluoroscopy systems have been used in

recent studies (List et al., 2017; Zeller et al., 2017), mobile biplane fluoroscopy is an emerging technology and, as of the writing of this dissertation, data have only been published on a single system located at the University of Melbourne in Parkville, Victoria, Australia (Guan et al., 2016a).

2.4 Previous Investigations into Healthy Patellofemoral Kinematics

Previous studies have investigated the kinematics of the PF joint, motivated by its importance to the knee joint complex and the frequency with which it is injured. This section will give an overview of these previous studies and point out limitations in them that will be improved upon in this dissertation.

There are several important features that will be highlighted when comparing the following studies on PF kinematics to that done in this dissertation. First, PF kinematics should be directly measured *in vivo* using healthy knees, as opposed to using a computational modelling, cadaver, or pathological knee approach, as it is difficult to know how representative these results are of the healthy knee given the assumptions made in modelling and the possible anatomical differences between a pathological and healthy knee. Second, participants should be performing functional activities (e.g., walking, stair climbing), rather than simplified knee flexion activities. While it is possible that simple knee flexion activities will produce PF kinematics that are comparable to more dynamic ambulatory activities, without first measuring the PF kinematics of these functional activities it is not possible to know how important the muscle activations produced in them are to the motion of the patella. Third, PF and TF kinematics should be recorded simultaneously in order to investigate their relationship to each other. Simultaneous recording will ensure that the kinematic measurements are compatible. Finally, accurate imaging techniques should be used that do not interfere with the participant's normal movement patterns as the objective is to measure healthy normative movement (e.g., see section 2.3.2.2 on intra-cortical bone pins).

2.4.1 Walking

In a study of ten individuals with total knee replacements, Guan et al. (2016b) found significant differences in TF kinematics between overground and treadmill walking. For

this reason, preference will be given to overground walking, in contrast to treadmill walking, when examining previous studies on PF kinematics during walking in this section.

Lafortune (1984) conducted the most accurate study of PF kinematics during overground walking to date in his PhD thesis. Five healthy participants were recruited, and joint positions were tracked using bone pins. All six kinematic degrees of freedom were reported for both the PF and TF joints. However, while the TF kinematics were later published (Lafortune et al., 1992), the PF data has never been published. This is possibly because the patella data from two of the five participants had to be discarded because the bone pin shifted. The patella is difficult to track using bone pins because of the extent to which it shifts beneath the skin.

Stein et al. (1993) used single-plane fluoroscopy to measure PF kinematics in eight healthy individuals during treadmill gait. This is the earliest published kinematic data on the PF joint during gait. All six PF kinematic degrees of freedom and the kinematics of the TF joint were not captured due to the limitations of this single-plane fluoroscopy system.

Other studies have predicted walking PF kinematics using methods other than directly measuring *in vivo* healthy knees. Ali et al. (2017) simulated gait for the TF and PF joints using two cadaver knees (age: 50 and 72 years). In separate but similarly designed studies, Lenhart et al. (2015) and Hu et al. (2018) predicted PF and TF kinematics during walking for a single participant using computational contact models driven by estimated muscle forces.

2.4.2 Stair and Ramp Ascent and Descent

The healthy PF kinematics of ascending and descending stairs warrant investigation because of their association with increased PF pain among people with knee disorders (Kujala et al., 1993; Thomee et al., 1999). This section also looks at past studies into ramp descent because of its similarity to stair descent.

No previous study has experimentally measured six degree of freedom (6-DOF) PF kinematics during either stair ascent or descent for a full gait cycle. The best data currently available on stair ascent was acquired by Akbarshahi et al. (2014) for four healthy

individuals using a predictive computational contact model of the PF joint. The subject-specific models were given predicted muscle forces and measured TF joint kinematics for the activity as inputs and output the predicted PF kinematics. The output PF kinematics were compared to the PF flexion angle during stance phase, which was captured using a single-plane fluoroscopy system, to validate the results.

PF kinematics during a step-up activity have been measured. In a step-up activity, the participant begins with one foot on the floor and one foot in front of them on a raised surface, the participant then lifts the foot on the floor and stands with both feet on the raised surface. Both Suzuki et al. (2012) and Lin et al. (2019) experimentally measured the PF kinematics of healthy individuals performing the step-up exercise using biplane and single-plane fluoroscopy, respectively. While not the same as stair ascent, these studies are the best experimentally obtained data sets on the activity currently available.

For ramp descent, Farrokhi et al. (2015) used biplane fluoroscopy to find the PF joint kinematics of 22 healthy individuals as they walked down an incline on a treadmill. Only medial-lateral translation and medial-lateral tilt were reported. As the healthy participants in this study were serving as the controls to a group of participants with TF osteoarthritis, their average age was 68.

2.4.3 Simple Activities

Numerous studies have attempted to quantify the PF kinematics of the knee during simple or non-functional activities. These activities are all similar in that they require the knee to flex and extend without translating, a requirement of dynamic imaging techniques with a small capture volume. Notable examples of these studies are summarized below.

Knee flexion, which can be done while seated or standing, involves the knee starting in a flexed or fully extended position, extending or flexing, and then returning to the starting position. The PF kinematics of knee flexion are well studied because of its simplicity. Koh et al. (1992) reported basic PF and TF kinematics for a single participant during knee flexion using bone pins. Seisler and Sheehan (2007) published a much larger study of the healthy PF and TF kinematics of 34 individuals using dynamic MR imaging. Finally, using single-plane fluoroscopy, Komistek et al. (2000) found the PF kinematics for 14 individuals.

Lunging and squatting are distinct from knee flexion in that they require the subject to support their own bodyweight. This makes them incompatible with some dynamic imaging techniques. Using single-plane and biplane fluoroscopy, respectively, Sharma et al. (2017) and Nha et al. (2008) reported the kinematics of only the PF joint during a lunge. Li et al. (2007) performed a similar study on a static squatting activity, where subjects stopped at regular intervals during the squat so static biplane fluoroscopy images could be captured for their knee, both TF and PF kinematics were reported.

2.5 Previous Investigations into Healthy Knee Cartilage

Contact

Two separate inputs are required to noninvasively quantify the location of articular cartilage contact in healthy PF and TF joints during an activity: the joint kinematics during the activity and a solid model of the joint's cartilage. Combining these two inputs it is possible to determine which sections of articular cartilage overlap or contact the opposing articular surface (e.g, the patellar articular cartilage would contact the femoral articular cartilage) throughout the activity. However, a lack of accurate joint kinematic data during functional activities, as described in the preceding section (Section 2.4), has been a limitation of previous knee cartilage contact studies. Thus, there is a lack of experimental cartilage contact studies involving functional activities and include the PF joint. This section will review the studies done to date.

2.5.1 Cartilage Geometry

MR images are routinely segmented to non-invasively create subject-specific three-dimensional cartilage models of healthy knees (Cohen et al., 1999; Coleman et al., 2013; Koo et al., 2005; Li et al., 2005). The purpose of this is generally to quantify the distribution of cartilage (i.e., a thickness map) covering the articular surface of the bone. Koo et al. (2005) created cartilage models of three porcine knees by both segmenting MR images and laser scanning the dissected knee in order to assess the MR image segmentation method's accuracy. The MR models were accurate with peak errors of less than 1 mm, with the contacting or weightbearing portions of cartilage being the most accurate.

Coleman et al. (2013) will be used as a representative study for quantifying the cartilage thickness of the full healthy knee joint using MR images. Ten healthy knees were scanned and manually segmented. Peak cartilage thickness for the three bones ranged between 4.5 and 5.5 mm. On average, cartilage was thicker in male participants. On the tibia, cartilage was thicker on the lateral side and peaked on the medial side of the lateral facet (~5 mm) and the lateral side of the medial facet (~4 mm) (i.e., towards the intercondylar eminence). On the femur, cartilage thickness peaked in the trochlear groove (~4.7 mm), with a smaller peak on the posterior surface of the medial condyle (~3 mm). On the patella, the thickest portion of cartilage ran in a line from the medial facet to the lateral facet across the patellar ridge (~5.4 mm). Of the three bones the patella had the thickest cartilage, on average.

2.5.2 Cartilage Contact

No study has been published that measures cartilage contact in the TF and PF joints simultaneously. In addition, no study has been published that investigates PF cartilage contact during the full stride cycle of a functional ambulatory activity. Using biplane fluoroscopy for kinematics and MR images for the cartilage model, Suzuki et al. (2012) quantified PF cartilage contact during a step-up activity for 12 healthy individuals and found that cartilage contact was centered on the lateral side of the patella and the femur throughout the activity. Kobayashi et al. (2013) performed a similar analysis during a lunge activity, however it was not a dynamic movement as the participants were instructed to freeze at different levels of knee flexion for the x-ray images to be captured. Using a cadaver knee with an implemented force sensor, Zavatsky et al. (2004) found that the center of contact on the patella was located on the lateral patellar facet during a simulated knee extension activity.

Examining only the TF joint, no study has quantified cartilage contact for an entire gait cycle; however, a few studies have examined the period around heel-strike. MR imaging was used to create the cartilage models in each of the following studies. With a large sample size of 44 healthy participants, Akpinar et al. (2019) reported that level treadmill walking resulted in a greater cartilage contact area at heel-strike compared to downhill treadmill running, as measured using biplane fluoroscopy. Studying the whole stance phase of treadmill gait in eight healthy individuals using biplane fluoroscopy, Liu et al. (2010) concluded that peak cartilage contact occurred in areas with thicker articular

cartilage. Koo et al. (2011) came to the same conclusion one year later when they measured the location of cartilage contact at heel-strike in 17 healthy participants using skin-marker based motion capture to quantify the joint kinematics. This positive relationship between areas of cartilage contact and cartilage thickness was also found by Bingham et al. (2008) and Li et al. (2005) for a forward lunge measured by biplane fluoroscopy.

2.6 Previous Investigations into Healthy Knee Contact Forces During Walking

Due to its important contribution to knee extension and the frequency with which it is injured, previous studies have attempted to calculate the contact force in the PF joint during walking. At present, the most accurate method for doing this is via a three-dimensional contact model, where the contact force is calculated based on a balance between the external ground forces, muscle and ligament forces, inertial forces, and the subsequent deformation of the joint's articular cartilage. Alternative methods for finding PF contact force, such as two-dimensional models or cadaver studies, have also been used for walking and will be covered in this section.

To validate the contact force predictions of a joint contact model the actual joint contact force should be measured and compared. However, direct validation *in vivo* is not possible for practical and ethical reasons. While not as accurate as validating via direct contact force measurement, measuring joint kinematics and comparing them to joint contact model predicted kinematics offers some evaluation of the accuracy of the predictions and is possible for a three-dimensional contact model. To date no previous study into PF joint contact force during walking has been able to validate using either of these methodologies because of a lack of experimental data on the PF joint.

2.6.1 Contact Models

Knee joint contact models are typically used in conjunction with full body or lower-limb musculoskeletal models. In this setup the musculoskeletal model calculates the necessary muscle forces to drive the body to perform the desired action, and the contact model calculates the resulting knee kinematics and contact forces needed to achieve equilibrium with the muscle forces. Ideally, the contact model is built directly into the knee of the

musculoskeletal model so the two can be solved simultaneously; however, it is also possible to generate solutions where the two models are separate, and the contact model is solved after the musculoskeletal model.

While various methods exist, at present only two types of contact models are commonly used for modelling the PF joint: finite element models and elastic foundation models. In the finite element method, the geometry of the joint (e.g., bone, cartilage) is divided into many simple geometric shapes or elements that, when combined, accurately describe the total geometry of the joint. These elements are each given material properties and constraints that define their relationships to each other. While accurate, the finite element method suffers from a large time cost for both creating a model and running it. In the elastic foundation method, articular cartilage is modelled as a series of tightly bunched linear springs where the resting length of the spring is the thickness of the cartilage and the spring constant is the stiffness of the cartilage. Elastic foundation models benefit from being faster to create and run relative to other contact modelling techniques.

For walking, the best knee contact force prediction, to date, was done by Lenhart et al. (2015). An elastic foundation model was created for a single healthy participant using static MR images and built into a generic lower-limb musculoskeletal model. This model was driven to track full body kinematics during walking recorded using skin-marker based motion capture. To validate the model's solution the contact model's resulting kinematics were compared, and matched well with, knee flexion kinematics measured using a dynamic MR scan. TF and PF contact pressures were reported over the gait cycle. However, conclusions on knee contact forces from this study are limited because of the small sample size, one subject, and the limited validation, kinematics were compared to a flexion activity instead of the same activity being studied.

Following on this work, van Rossom et al. (2018) scaled the model created by Lenhart et al. (2015) to predict the peak and average PF and TF contact forces of 15 healthy individuals during a range of activities, including walking. Brandon et al. (2018) conducted a similar study where the model created by Lenhart et al. (2015) was scaled to predict the knee contact forces of 20 healthy children during walking. Both studies have the same major limitations: the contact models were generic and not subject-specific, and they contained no method to validate the contact model results.

Hu et al. (2018) created a single elastic foundation model from publicly available knee MR data and incorporated it into a generic model of the lower-limbs. Motion-capture data of walking was recorded from a separate individual for the musculoskeletal model to track. Resulting PF kinematics were compared to previously published PF kinematics during a knee flexion activity from a different individual in order to validate the results. PF contact force during the gait cycle was published. Despite the quality of the analysis in this study, it is severely limited by using separate subjects for the contact model, motion capture, and results validation.

Halonen et al. (2016) used a two-step modelling approach to predict knee contact forces. Motion capture data was recorded for a single healthy participant and used to drive a generic musculoskeletal model which calculated muscle forces. Then those calculated muscle forces were used to drive a separate model, a finite element model of the TF and PF joints, based on the subject's MR images, which outputted predicted knee contact forces. Validation was lacking, with only TF kinematics for a few degrees of freedom being used to assess the results. Also, the two-step modelling methodology is a limitation as it does not allow for the kinematics of the PF joint in the contact model to affect the musculoskeletal model's choice of which muscles to activate.

2.6.2 Other Techniques

While the studies presented above that make use of contact models represent the best work that has been done to date on knee joint contact during walking, other techniques have been tried as well. These studies are typically older, published before building subject-specific contact models that include the PF joint became more practical. In order to produce knee contact force results without the use of a contact model most of the studies presented below make gross simplifications to the knee joint complex or use non-subject-specific knee joint contact parameters.

Simplified mathematical models were previously commonly used to calculate PF contact forces during walking (Brechtel and Powers, 2002; Chen et al., 2010; Komistek et al., 1998; Moissenet et al., 2017; Shelburne et al., 2005). In these models the motion of the patella is typically governed by constraint equations and the PF contact force is defined as the patella-femur interbody force necessary to satisfy those equations. Moissenet et al. (2017) is an example of this form of study. A musculoskeletal model of a single leg was

constructed in which the TF and PF joints were both hinge joints. Muscle forces were determined by solving a muscle activation optimization problem in which the model tracked the motion of a single individual's gait pattern captured using a motion capture system. The PF constraint forces were reported over the gait cycle. No validation of these contact force results was performed.

2.6.3 Direct Collocation

Direct collocation is an efficient method for solving complex optimal control problems that is commonly used to simulate human movement using musculoskeletal models (Ackermann and Van den Bogert, 2010; Kaplan and Heegaard, 2001; Lin and Pandey, 2017; Lin et al., 2018; Porsa et al., 2016; Stelzer and Von Stryk, 2006). While it has not previously been used in a simulation which includes a PF joint, a brief description will be given here as it will be implemented in Chapter 5.

Trajectory optimization is the process of determining a path (i.e., trajectory) for a system while minimizing some performance criteria (Kelly, 2017). For instance, a person walking from point A to B. The trajectory would include numerous variables as functions of time, such as, the motion of the body, muscle forces, neural excitations, etc. The performance criteria to optimize could be total muscle activation (for efficient walking) or time (for fast walking).

Direct shooting is the traditional method for solving trajectory optimization problems. In direct shooting the control variables (e.g., neural excitations) are discretized and given to the optimizer as variables that can be changed in order to minimize the performance criteria. The states (e.g., muscle activation, muscle force, joint motion) are then determined by forward integrating the dynamical equations that define the system given the controls. While this method benefits from having few design variables, it is computationally expensive (Porsa et al., 2016).

Direction collocation is a more computationally efficient method for solving trajectory optimization problems. Both the controls and states are discretized. This removes the need to forward integrate the computationally expensive dynamical equations in order to solve for the states. Instead, these equations are converted into a series of algebraic constraints

that must be satisfied. These constraints would look as follows when using the trapezoidal rule for integration (Lin and Pandy, 2017):

$$\frac{\underline{x}_{n+1} - \underline{x}_n}{t_{n+1} - t_n} - \frac{\underline{x}'_n + \underline{x}'_{n+1}}{2} = 0$$

where $\underline{x}$ is the vector of states, $\underline{x}'$ is the vector of state derivatives, and t is time. By forcing the optimizer to satisfy this constraint, the correct relationship between the controls and states is maintained while avoiding the time cost of forward integration.

2.7 Aims of this Dissertation

The knee is a complex weightbearing joint in the lower-limb that is critical for functional activities such as walking. The knee is also subject to many different disorders that limit its ability to function properly. With many knee injuries (e.g., ACL reconstructions) eventually leading to PF and TF osteoarthritis, a debilitating disease indicated by a loss of articular cartilage. The etiologies of many of these knee disorders have been tied to altered mechanics of the knee joint. These alterations can manifest in many ways such as the contacting areas of cartilage or the loading pattern of the patella. However, prevention and diagnosis of altered knee mechanics are limited by a lack of knowledge of the mechanics of the healthy knee joint. Capturing the motion of the full knee joint during a functional activity requires an accurate dynamic imaging technique with a large capture volume that, until recently, did not exist. This lack of knowledge of healthy knee joint kinematics has also been a major limitation of any previous study attempting to investigate the cartilage contact patterns of the healthy knee or validate the force prediction of a contact model of the healthy knee.

This present dissertation has been organized into three aims that address the lack of knowledge of healthy knee joint mechanics:

Aim 1 (Chapter 3) focuses on experimentally measuring the PF kinematics of healthy knees during functional activities. This aim makes use of a mobile biplane fluoroscopy machine that has not previously been used to study the kinematics of the healthy PF joint. The motion of the PF joint is described here, and the extent to which each motion is coupled to TF flexion is quantified.

Aim 2 (Chapter 4) focuses on calculating the location of cartilage contact in both the PF and TF joints of healthy knees during functional activities. The kinematic results of Aim 1 will be used to aid in accurately locating the contacting cartilage. The locations of contacting cartilage will be investigated to determine if, in healthy knees, it aligns with areas of greater cartilage thickness.

Aim 3 (Chapter 5) focuses on calculating the medial and lateral contact forces for the PF and TF joints of healthy knees during overground walking. Subject-specific elastic foundation contact models of the PF and TF joints will be created and built into a lower-limb musculoskeletal model to calculate these joint contact forces. The kinematic results of Aim 1 and cartilage contact results of Aim 2 will be used to validate the PF and TF mechanics predicted by the contact model. The simulation's solution will be investigated to search for contributors to PF contact force and its medial-lateral distribution.

Chapter 3 Healthy Patellofemoral Kinematics during Functional Activities

This chapter is based on work that was submitted for publication to *Annals of Biomedical Engineering* on March 4th, 2020.

The level walking kinematic data presented in this chapter were additionally published in the following work:

Gray, H.A., Guan, S., **Thomeer, L.T.**, Schache, A.G., de Steiger, R., Pandey, M.G., 2019. Three-dimensional motion of the knee-joint complex during normal walking revealed by mobile biplane x-ray imaging. *J Orthop Res* 37, 615-630.

3.1 Introduction

The PF joint, a component of the knee joint complex, is critical for efficient knee extension (Im et al., 2015). However, the PF joint is also frequently injured through disorders such as PF pain, mainly affecting young and active individuals, and PF osteoarthritis, mainly affecting older individuals (Davies et al., 2002; Rothermich et al., 2015). The etiologies of these various PF knee disorders have all been linked to altered mechanics of the PF joint (Carter et al., 2004; Thomeer et al., 2017).

This pathway to injury indicates that a robust understanding of the kinematics of the healthy PF joint during everyday activities is essential to the prevention and treatment of PF joint disorders. Previous studies have measured a few healthy PF degrees of freedom during a functional activity (Farrokhi et al., 2015; Stein et al., 1993), or healthy 6-DOF PF kinematics during a non-functional activity such as lunging or squatting (Komistek et al., 2000; Li et al., 2007; Nha et al., 2008; Seisler and Sheehan, 2007; Sharma et al., 2017), but no previous study has reported healthy 6-DOF PF kinematics during a full gait activity. This is because imaging the knee during an activity such as walking requires a dynamic imaging technique that is both accurate and has a large or mobile capture volume.

Assuming that the patella, tibia, and femur are non-deformable bodies, the knee has 12-DOF in total, 3 rotations and 3 translations at the TF joint, and 3 rotations and 3

translations as the PF joint. Despite this, in musculoskeletal modelling simulations the knee is frequently modelled as having only 1-DOF (e.g., it is modelled as a hinge knee) (Correa et al., 2010; Jansen et al., 2014; Klemetti et al., 2014; Lin and Pandy, 2017; Neptune and McGowan, 2016; Pandy et al., 2010). This assumes that the 6-DOF motion of the PF joint is fully determined by (or coupled to) the TF flexion angle. However, the accuracy of this assumption throughout the full gait cycle of various functional activities has not previously been assessed as accurate 3D measurements of patellar motion are needed to perform this calculation.

In this study, a novel mobile biplane X-ray (MoBiX) imaging system (Guan et al., 2016a) was used to capture fluoroscopy images of the knee while six activities were performed: static standing, level walking, downhill walking, stair descent, stair ascent, and open-chain knee flexion. The primary purpose of this study was to measure and report the 6-DOF kinematics of the PF joint during these activities. The secondary purpose of this study was to quantify the extent to which the PF DOFs were coupled to TF flexion angle.

3.2 Methods

3.2.1 Participants

A cohort of ten healthy subjects (6M/4F, 27.1 ± 10.7 years, 68.3 ± 9.0 kg, 168.0 ± 9.3 cm) were recruited based on the following inclusion criteria: a) no knee pain or previous knee surgery, and b) no contraindications to receiving X-ray fluoroscopy, computed tomography (CT) or magnetic resonance scans. Informed consent was acquired prior to testing and all experimental procedures were approved by the Human Research Ethics Committee at the University of Melbourne.

3.2.2 Data Collection

All participants visited the gait lab for two familiarization sessions and two data-collection sessions. Familiarization sessions involved a walkthrough of the experimental procedures and were conducted to introduce participants to the data-collection process. In addition, each participant attended a local imaging center to receive a CT scan ($0.35 \times 0.35 \times 0.50$ mm) of their right knee.

Three sets of data were captured simultaneously during data-collection: biplane fluoroscopic images, full-body motion, and ground reaction forces. Biplane fluoroscopy images were acquired using the MoBiX imaging system (1,024 x 1,024 pixels, 200 frames/s, 1/200 s exposure time, 110 kV, 13.1 mA). Combined with the CT scan, the maximum radiation dosage received was below 0.21 mSv. All participants wore a lead vest. Full-body motion data were captured using a 9-camera motion capture system (VICON, Oxford, UK) operating at 120 Hz. Forty-five reflective skin markers were used. Ground reaction forces were captured using three force plates (AMTI, Watertown, MA) operating at 1,080 Hz.

3.2.3 Activities

Participants performed six activities: standing, open-chain knee flexion, level walking, downhill walking, stair descent, and stair ascent. For the standing activity, participants stood stationary with both feet along opposite edges of a single force plate, arms were straight but abducted away from their torso. For the open-chain knee flexion activity, participants begin with both feet shoulder width apart on a single force plate before beginning a four-step process: 1) the left hip was adducted so that the right foot lifted off the force plate while the right knee remained extended, 2) the right knee was slowly flexed to its fullest extent, 3) the right knee was slowly extended, and 4) the right foot was placed back on the force plate to return to the starting position. The knee flexion and extension cycles were separately processed. Participants were given a pole to aid in maintaining their balance throughout the flexion-extension cycle. Participants were instructed to keep their right thigh stationary throughout Steps 2 and 3. A horizontal board was attached to the pole which was adjusted to lightly touch the thigh of each participant. This board served as a tactile reminder to keep the thigh stationary, it did not restrain motion of the thigh. For the level walking activity (Figure 3.1), each participant took ten steps at a self-selected speed. Average stride length was measured prior to data-collection while each participant walked across the lab several times. This stride length was used to prescribe the starting location and the first three steps of the activity to ensure that steps four, five, and six occurred on a force plate. Data were processed from heel-strikes five through seven, or one complete gait cycle. For the downhill walking, stair descent, and stair ascent activities (Figure 3.1), each participant took six steps at a self-selected speed, starting and ending on a level platform. The ramp used had a 10° decline. The stairs used consisted of

two 17 cm steps. A uniform starting position was prescribed for all participants to ensure that steps two, three, and four occurred on the force plates. Data were processed from heel-strikes three to five. All four gait activities (i.e., level walking, downhill walking, stair descent, and stair ascent) occurred over ground.

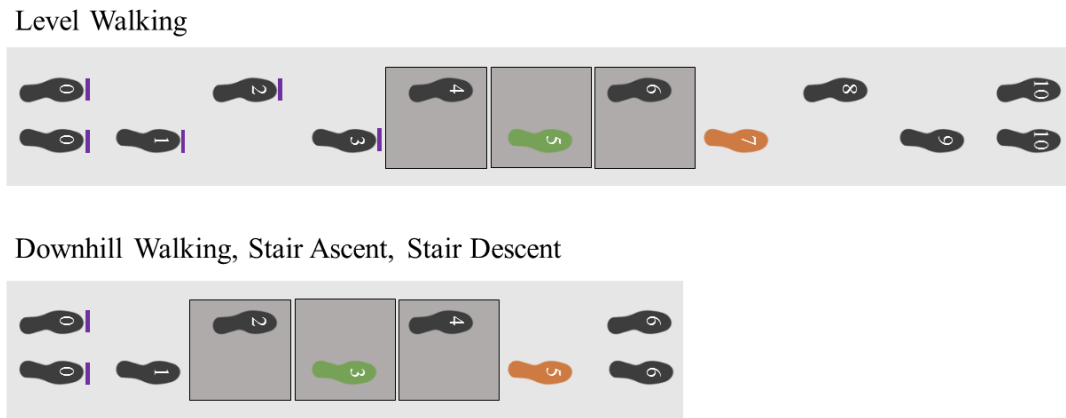


Figure 3.1: Top Down View of Step Locations for Gait Activities. Footprints signify step locations. Black outlined boxes signify force plate locations. Green and orange footprints signify starting and ending heel-strikes of processed gait cycle, respectively. Purple lines signify step locations prescribed by taped marks.

3.2.4 Data Processing

Geometric models of each participant's femur and patella were manually created by segmenting their CT scan using 3D Slicer (Pieper et al., 2006). An anatomical coordinate system (Figure 3.2) based on the work of Grood and Suntay (1983) was defined for each geometric model using Geomagic (3D Systems, Morrisville, NC). For the patella, the origin was defined as the centroid, the anterior axis was defined as the third inertial axis, the lateral axis was defined as being mutually perpendicular to the anterior axis and the patellar ridge, and the superior axis was mutually perpendicular to the anterior and lateral axes. For the femur, the lateral axis was defined as the axis of a cylinder fit to the posterior and distal portions of both condyles, the origin was defined as the point where a perpendicular line dropped from the most distal portion of the intercondylar notch intersected the lateral axis, the superior axis was defined as the axis of a cone fit to the femoral shaft, and the anterior axis was mutually perpendicular to the lateral and superior axes.

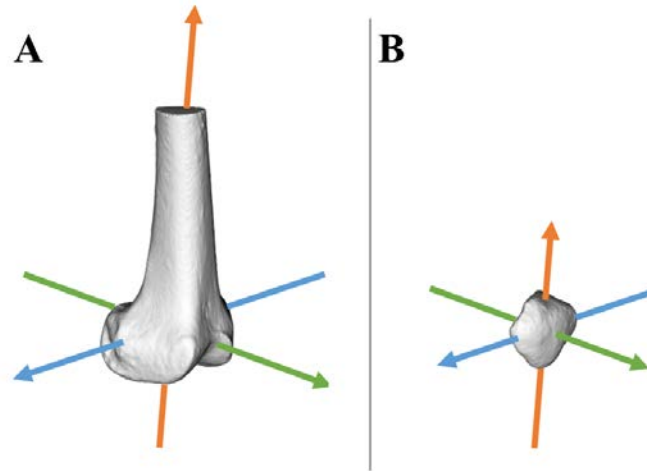


Figure 3.2: Anatomical Coordinate System. The superior-inferior (orange), medial-lateral (blue), and anterior-posterior (green) axes used in this study displayed on the femur (A) and patella (B) of a single subject.

6-DOF PF kinematics were acquired by translating and rotating the geometric models to match the orientation of the bone in the captured fluoroscopy images using custom MATLAB (Mathworks Inc., Natick, MA) software. Maximum root-mean-square errors for these procedures for the intact PF joint have been assessed as 0.37 mm for translations and 1.46° for rotations by comparing to the results of a bead-based method (2 mm diameter zirconium oxide beads) (Gray et al., 2017). Fluoroscopic data were processed at 50 Hz (a decrease from the 200 Hz the images were captured at in order to save time while performing manual pose estimation) and filtered using a fourth-order, low-pass Butterworth filter with a 10 Hz cut-off frequency. The final results were resampled to 201 data points. Full-body motion and ground reaction force data were used to identify gait events.

3.2.5 Analysis

For each dynamic activity, the degree of coupling between each PF motion and TF flexion-extension was quantified. The mean values for each PF motion were plotted against TF flexion-extension, and a spline function was fit to the resulting 201 data points using the MATLAB function ‘fit’ with the value of the ‘smoothing parameter’ set to 0.1. A coefficient of determination (r^2) and root-mean-square error (RMSE) were calculated for each DOF. Perfect coupling would be reflected by an r^2 value of 1.00 and an RMSE value of 0.0. A DOF was considered to be strongly coupled when $r^2 \geq 0.7$, moderately coupled when $0.5 \leq r^2 < 0.7$, and not coupled when $r^2 < 0.5$. This analysis was repeated for each DOF with the data from all the dynamic tasks pooled together. A spline function

was fit to the resulting $5 \times 201 = 1,005$ data points and r^2 and RMSE values were computed as before.

In addition, means and standard deviations were calculated for the 6-DOF PF kinematics of each activity. Student's paired t-tests with a Bonferroni correction were performed to test for differences in the mean, minima, maxima and range of motion between activities. A significance level of 0.05 was used.

3.3 Results

3.3.1 Activity Velocities

Level walking had the highest mean walking speed at 1.31 ± 0.14 m/s. This was followed by downhill walking at 0.84 ± 0.09 m/s, stair ascent at 0.63 ± 0.09 m/s, and stair descent at 0.62 ± 0.05 m/s. The average TF flexion angular velocity of the open-chain knee flexion activity was 125.93 ± 21.03 deg/s (compared to 159.7 ± 26.5 deg/s for level walking).

3.3.2 Kinematic Features

The largest rotational DOF was flexion-extension, with ranges of motion (ROM) from 68.7° to 50.1° (Table 3.1). ROMs for medial-lateral rotation (9.4° to 5.5°) and medial-lateral tilt (9.8° to 7.1°) were much smaller. For the translations, anterior-posterior translation (17.3 to 11.5 mm) and superior-inferior translation (17.5 to 14.5 mm) had similarly large ROMs, with both being larger than medial-lateral shift (7.2 to 3.8 mm).

The patella rapidly translated superiorly and laterally just prior to heel-strike in all dynamic activities (Figure 3.3, first and second column). Relatively little motion was seen throughout the activity for both medial-lateral shift and medial-lateral tilt, except for open-chain knee flexion, which tilted 5 degrees laterally followed by 4 degrees of medial tilt. Compared to standing, the patella was shifted and tilted more medially throughout all dynamic activities. Except for level walking, medial-lateral shift was largely static from contralateral toe-off to ipsilateral toe-off for all gait activities. Stair ascent was more flexed, posteriorly translated, and inferiorly translated in early stance when compared to the other gait activities.

CHAPTER 3: PATELLOFEMORAL KINEMATICS

Table 3.1: Patellofemoral Kinematics for Dynamic Activities. The mean, minima, maxima, and ROM of each DOF and activity. Superscripts indicate significant differences between the corresponding activities. LW, level walk; DW, downhill walk; SD, stair descent; SA, stair ascent; OF, open-chain flexion.

Parameter (Unit)	Meas.	Level Walk	Downhill Walk	Stair Descent	Stair Ascent	Open-chain Flexion
Flexion (deg)	Mean	15.8 ^{DW, SD, SA, OF}	22.2 ^{LW, SD, SA, OF}	29.4 ^{LW, DW, SA, OF}	24.4 ^{LW, DW, SD, OF}	41.4 ^{LW, DW, SD, SA}
	Min	47.3 ^{DW, SD, OF}	50.6 ^{LW, SD, OF}	64.6 ^{LW, DW, SA, OF}	52.0 ^{SD, OF}	73.8 ^{LW, DW, SD, SA}
	Max	-4.8 ^{DW, SA, OF}	0.5 ^{LW}	0.9	1.5 ^{LW}	5.1 ^{LW}
	ROM	52.1 ^{SD, OF}	50.1 ^{SD, OF}	63.7 ^{LW, DW, SA}	50.6 ^{SD, OF}	68.7 ^{LW, DW, SA}
Lateral rotation (deg)	Mean	-3.1	-3.3	-3.5	-4.1	-2.1
	Min	2.0	-0.2	0.4	0.8	0.5
	Max	-7.3	-6.4	-7.5	-8.6 ^{OF}	-5.0 ^{SA}
	ROM	9.3 ^{OF}	6.2 ^{SA}	7.9	9.4 ^{DW, OF}	5.5 ^{LW, SA}
Lateral tilt (deg)	Mean	1.7	1.9	1.1	1.0	2.3
	Min	6.9	6.2	4.9	4.7	5.5
	Max	-2.9	-1.7	-2.4	-2.4	-2.5
	ROM	9.8	8.0	7.3	7.1	8.0
Lateral shift (mm)	Mean	4.4	3.9	3.8	3.7	3.9
	Min	8.0	7.2	7.4	6.6	5.8
	Max	0.8	1.7	1.7	1.4	1.9
	ROM	7.2 ^{OF}	5.6	5.7	5.3	3.8 ^{LW}
Anterior translation (mm)	Mean	46.2 ^{DW, SD, SA, OF}	45.3 ^{LW, SD, SA, OF}	43.4 ^{LW, DW, SA, OF}	44.6 ^{LW, DW, SD, OF}	41.2 ^{LW, DW, SD, SA}
	Min	50.6 ^{OF}	50.0	49.7	49.5	49.1 ^{LW}
	Max	39.1 ^{SD, SA, OF}	38.0 ^{SD, SA, OF}	33.8 ^{LW, DW, SA}	37.0 ^{LW, DW, SD, OF}	31.9 ^{LW, DW, SA}
	ROM	11.5 ^{SD, OF}	12.0 ^{SD, OF}	15.9 ^{LW, DW, SA}	12.6 ^{SD, OF}	17.3 ^{LW, DW, SA}
Superior translation (mm)	Mean	9.9 ^{DW, SD, SA, OF}	6.7 ^{LW, SD}	4.9 ^{LW, DW}	5.9 ^{LW}	4.0 ^{LW}
	Min	17.6 ^{DW, SD, SA, OF}	14.9 ^{LW, OF}	13.5 ^{LW}	14.5 ^{LW}	11.3 ^{LW, DW}
	Max	0.0	-0.9	-1.7	-1.7	-3.2
	ROM	17.5	15.8	15.2	16.2	14.5

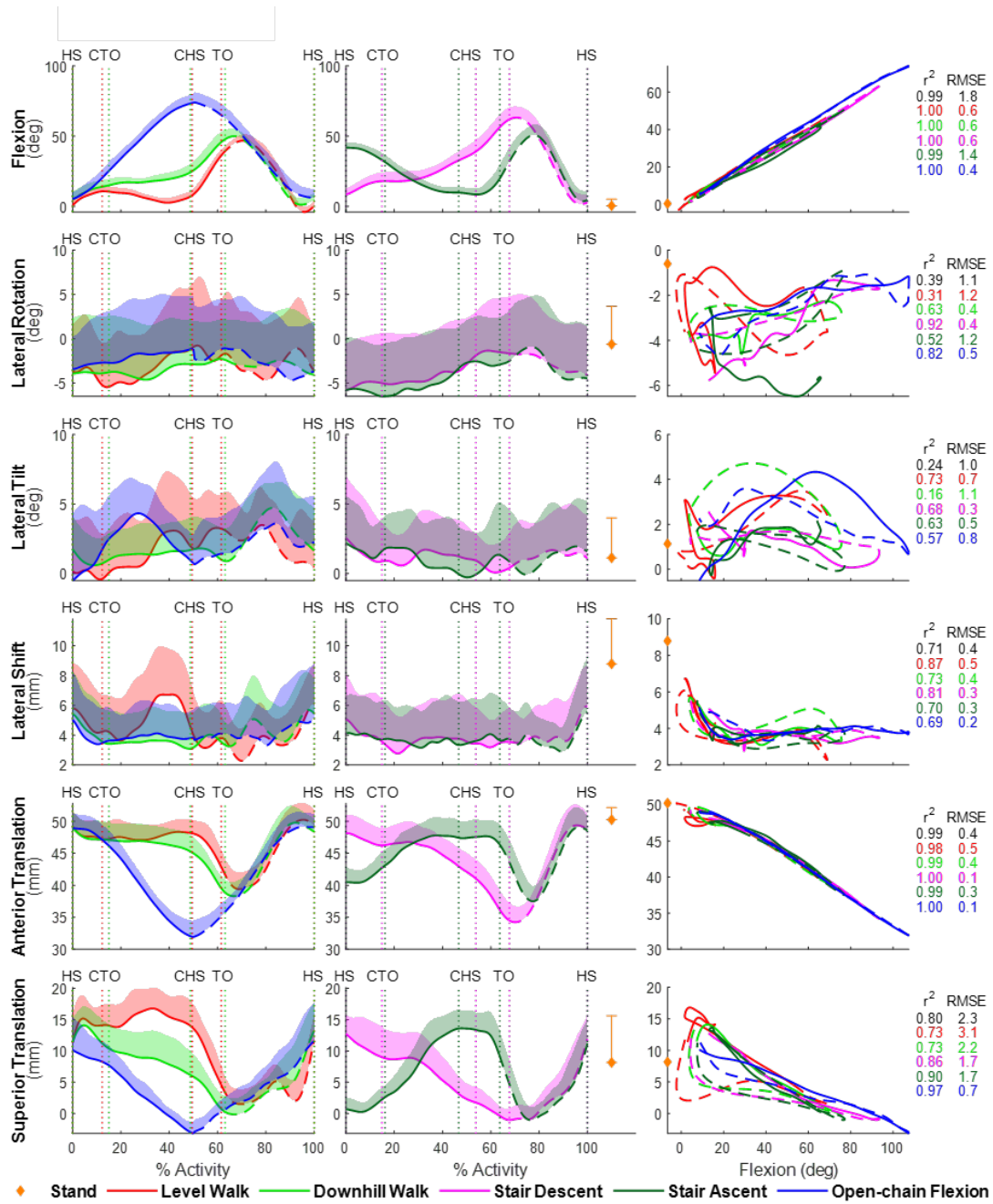


Figure 3.3: Patellofemoral Kinematics and Degree of Coupling. The first and second columns plot the 6-DOF kinematics of the PF joint against the percent of each activity completed. Vertical dotted lines correspond to significant gait events: HS, heel strike; CTO, contralateral toe off; CHS, contralateral heel strike; TO, toe off; HS, heel strike. The third column plots the 6-DOF kinematics of the PF joint against the TF flexion-extension angle. Black numbers represent the degree of coupling of all dynamic activities combined. Colored numbers represent the degree of coupling of each individual activity. The shaded region represents one standard deviation. The solid and dashed lines represent the stance and swing phase, respectively, except for open-chain knee flexion, where they represent knee flexion and knee extension.

3.3.3 Coupling in the Patellofemoral Joint

The degree of coupling can be qualitatively visualized by plotting the PF kinematics against the TF flexion angle (Figure 3.3, third column). Flexion-extension and anterior-posterior translation were nearly perfectly coupled to TF flexion ($r^2 = 0.99$). In addition, both exhibited a linear relationship with TF flexion angle (Pearson's linear correlation coefficients of 0.99 and -0.98, respectively). Medial-lateral shift showed strong coupling ($r^2 = 0.71$). All dynamic activities rapidly shifted laterally at low levels of TF flexion. This trend was qualitatively continued in the medial-lateral shift of the hyperextended knee during standing. Superior-inferior translation was also strongly coupled ($r^2 = 0.80$). However, its RMS error of 2.3 mm was over five times greater than the other translations (0.4 mm for both anterior-posterior translation and medial-lateral shift) due to the divergence of the curves at low TF flexion angles. Medial-lateral tilt ($r^2 = 0.24$) and medial-lateral rotation ($r^2 = 0.39$) were not strongly coupled to TF flexion and neither showed any obvious patterns of movement.

3.4 Discussion

The data presented in this study are the first of their kind to accurately describe the 6-DOF motion of the healthy PF joint during multiple daily ambulatory activities. This was accomplished using a mobile biplane X-ray imaging system. We found that TF flexion was strongly coupled to four of the six motions of the patella: flexion-extension, anterior-posterior translation, medial-lateral shift, and superior-inferior translation. In addition, we quantified the distinct medial-lateral shift of the patella at low TF flexion angles that was characteristic of all gait activities tested. These findings on the basic mechanical function of the PF joint during six common activities will assist in further understanding the function of a healthy knee.

The limitations of this study should be considered when interpreting its results. Due to limits in the vertical range of the MoBiX imaging system the gait cycles for downhill walking, stair ascent and stair descent were not strictly a standard cycle, with the final heel strike for each of these three activities occurring on a level platform. Despite this limitation the stair and ramp activities performed in this study represent an improvement over the limitations imposed by a non-mobile imaging system where only a step-up task (Lin et al., 2019; Suzuki et al., 2012) or the stance phase of downhill walking (Farrokhi

et al., 2015) can be captured. Another limitation is the potential for the MoBiX imaging system to alter a participant's normal gait pattern through its presence. Yamokoski and Banks (2011) quantified the impact of a tracking robot on healthy gait and concluded that individuals did walk faster when being tracked but the differences were small in magnitude.

The results of this study demonstrate the importance of the trochlear groove to patellar motion. Beyond 20° of TF flexion, when the patella is likely firmly seated in the trochlear groove, relatively little medial-lateral shift of the patella was observed. However, at lower TF flexion angles, when the patella leaves the intercondylar groove, the patella translated laterally. This lateral movement has previously been observed by Nha et al. (2008) for a lunge activity, but not for multiple dynamic activities. As the surface of the distal femur is relatively flat in this region, this lateral translation is likely caused by the soft tissue surrounding the patella and not bone geometry. Additionally, this pattern holds for many different instances of quadriceps activation (e.g., both stance and swing phase for the gait activities, open-chain flexion, and the standing activity), ruling out active muscle force as the cause. Instead, this movement can be attributed to passive soft tissue forces. Possibly the pull of the vastus lateralis, as the largest of the vasti muscles (O'Brien et al., 2010), or the combined pull of both the quadriceps tendon and the patellar tendon (which can be quantified using the Q-angle) (Smith et al., 2008).

Superior translation of the patella in the last 20% of the activity was a characteristic of all five dynamic activities. This phenomenon is due to the elasticity of the patellar tendon. During the four gait activities the superior translation occurred at low TF flexion angles. The patella begins in an inferior location caused by a slack patellar tendon, then activation of the quadriceps prior to heel strike stretches the patellar tendon and superiorly translates the patella. The superior translation in open-chain knee flexion is likewise caused by quadriceps activation, however in this case it is the quadriceps co-contracting in response to hamstrings activation.

Four of the six motions of the patella were coupled to TF flexion. Patellar flexion and patellar anterior-posterior translation were nearly perfectly coupled and linearly correlated to tibial flexion in all activities, increased tibial flexion leads to patellar flexion and posterior translation of the patella. However, patellar medial-lateral rotation and tilt are not coupled to tibial flexion. No clear pattern is evident in the motion of either

secondary rotation throughout the four gait activities. The RMSE error associated with the degree of coupling for both is low (lower than the accuracy of the system as reported by Gray et al. (2017)), so the lack of coupling as defined by an r^2 value may be due to the compressed range of motion for both DOFs. Medial-lateral and superior-inferior translation of the patella are both coupled to tibial flexion, but both have weaker couplings than those exhibited by patellar flexion and anterior-posterior translation for reasons discussed above (i.e., lateral translation of the patella at low TF flexion angles and superior-inferior translation of the patella in response to quadriceps activation). Based on these data we can conclude that the patellofemoral joint can effectively be modelled as a 1-DOF joint. However, a 2-DOF joint may often be required to allow for superior-inferior translation of the patella to stretch the patellar tendon.

In summary, the 6-DOF PF kinematics reported here are the first of their kind to be measured in healthy knees performing a range of daily ambulatory tasks. We found strong coupling between TF flexion and four of the six secondary motions of the knee and quantified the kinematic characteristics of the PF joint that persist despite the activity being performed.

Chapter 4 Cartilage Contact in the Knee and its Relationship to Thickness

This chapter is based on work that has not currently been published or submitted for publication.

4.1 Introduction

ACL reconstructions, a surgical procedure performed to replace a ruptured ACL, are common and their incidence is increasing, particularly among individuals younger than 20 years or older than 40 years (Mall et al., 2014). This is a cause for concern because of the well documented link between ACL reconstructions and knee osteoarthritis 10 to 15 years later (Lohmander et al., 2004; Von Porat et al., 2004). One proposed theory for the increased prevalence of osteoarthritis is that the ACL injury and subsequent reconstruction alters the normative kinematics and loading of the joint, leading to degeneration of the cartilage (Andriacchi et al., 2009; Chaudhari et al., 2008; Paschos, 2017). If true, restoration of healthy pre-injury cartilage contact mechanics in the knee would likely lower the incidence of post-surgery knee osteoarthritis.

However, despite its importance in the prevention of osteoarthritis, there is presently a lack of data on the cartilage contact patterns that occur in the TF and PF joints of healthy individuals during everyday functional activities (e.g., stair ascent and descent). Suzuki et al. (2012) determined the centroid of cartilage contact in the PF joint to be on the lateral facet during a step-up exercise using biplane fluoroscopy. Liu et al. (2010) found more motion of the contact centroid on the medial tibial plateau, in comparison to the lateral tibial plateau, of the TF joint during stance phase of level treadmill walking. Akpinar et al. (2019) discovered a larger area of cartilage contact during the first 10% of the level walking cycle, as opposed to downhill running, in the TF joint. Other studies have quantified the patterns of cartilage contact during simple activities such as knee flexion or lunging (Bingham et al., 2008; Li et al., 2005; Moro - oka et al., 2008; Thorhauer and Tashman, 2015; Zavatsky et al., 2004). No study has simultaneously determined the location of cartilage contact in both the healthy PF and TF joints during complete cycles of everyday activities. This is an important gap in the literature to fill as it will provide

insight into the healthy function of articular knee cartilage during daily activities and comparison data to be used when determining the abnormalities in pathological knees.

Gray et al. (2019) estimated the center of contact during one stride cycle of level walking for both the patellofemoral and tibiofemoral joints using the closest-point and lowest-point methods, respectively. The closest-point method, which has also been used by Hoff et al. (1998) and Komistek et al. (2003), defines the center of contact as the point which is the shortest distance between the two surfaces. The lowest-point method defines the center of tibiofemoral contact as the most inferior points on the medial and lateral femoral condyles (DesJardins et al., 2007). Neither of these methodologies requires a geometric cartilage model, making them more cost and time efficient than using cartilage overlap to locate the center of contact. However, DeFrate et al. (2004) found an average error of 4.7 mm in the location of the lateral center of contact when using a bone geometry method of locating the center of cartilage contact.

Previous studies have shown that articular cartilage is responsive to broad changes in overall loading within the knee (Hudelmaier et al., 2003; Jones et al., 2003; Koo and Andriacchi, 2007; Vanwanseele et al., 2003), leading to the prevailing theory of cartilage health, that articular knee cartilage adapts to withstand the cyclical loading patterns of activities of daily living, with cartilage breaking down when loading patterns are changed (Andriacchi et al., 2009; Chaudhari et al., 2008). Extending this theory, previous studies have investigated the relationship between cartilage loading and thickness to see if it holds within different regions of the knee. That is, do regions within the same knee that experience more contact have thicker cartilage? While analysing the stance phase of gait, Liu et al. (2010) found that peak cartilage deformation occurred in regions with thicker cartilage on the tibial plateaus. Scanlan et al. (2013) and Koo et al. (2011) concluded that cartilage contact occurred in thicker regions of cartilage at heel-strike for the medial femoral condyle but not the lateral femoral condyle. Increased cartilage thickness in contacting regions was also found by Bingham et al. (2008) and Li et al. (2005) in a lunge activity for the TF joint. However, at present, no study has simultaneously investigated the strength of the relationship between cartilage contact and thickness in both the tibiofemoral and patellofemoral joints, and no study has investigated this relationship in the same group of participants during the full stride cycle of multiple activities of daily

living. This represents an important avenue for study as at present it is not known whether or not knee loading can be interpreted from the cartilage distribution within a single knee.

The primary goal of this study was to simultaneously measure the locations of cartilage contact in the PF and TF joints during everyday activities. This was done using accurate 12-DOF kinematics of the knee captured using a mobile biplane X-ray imaging system. A secondary goal was to determine if thicker regions of cartilage experience more joint contact than thinner regions of cartilage. It was hypothesized that the centroid of cartilage contact would occur in the lateral facet of the PF joint, and that a positive relationship would be found between cartilage thickness and amount of cartilage contact throughout the knee joint for level walking, downhill walking, stair ascent, stair descent, and open-chain knee flexion, with the strongest correlation in level walking.

4.2 Methods

4.2.1 Joint Kinematics

The PF joint kinematics used in this study are the same as those presented in Chapter 3. While a brief summary is given below, full details on data acquisition may be found there. TF joint kinematics were captured in the same procedure.

Ten healthy individuals (6M/4F, 27.1 ± 10.7 years, 68.3 ± 9.0 kg, 168.0 ± 9.3 cm) participated in this study. All participants performed six activities (standing, open-chain knee flexion, level walking, downhill walking, stair ascent, and stair descent) while biplane fluoroscopic images of their right knee, full-body motion data, and ground reaction forces were simultaneously acquired using the MoBiX imaging system, a 9-camera motion capture system, and three force plates, respectively. Each participant also received a CT scan ($0.35 \times 0.35 \times 0.50$ mm) and a proton-density-weighted turbo spin-echo MR scan ($0.625 \times 0.625 \times 3$ mm) of their extended right knee.

The CT and MR scans were manually segmented using 3D Slicer (Pieper et al., 2006) to create geometric models of the femoral, tibial, and patellar bone surfaces and articular cartilage, respectively. 6-DOF joint kinematics for the PF and TF joints were derived by pose estimating the geometric bone models with the fluoroscopic images.

4.2.2 Cartilage Thickness and Contact

Cartilage thickness was measured for the femur, tibia, and patella of each subject using a four-step method implemented using MATLAB. First, each geometric cartilage model was transformed into a volumetric point cloud, with points placed 0.10 mm apart. Second, the surface of the corresponding geometric bone model was defined as a series of triangles with an average side length of 1.00 mm. Third, a ray was projected from the center of and normal to each triangle in the surface mesh of the bone model. Fourth, the largest three-dimensional distance between intersected points along the ray was assigned as the cartilage thickness for that triangle.

The centroid, area, and amount of cartilage contact were determined in MATLAB using a five-step process. This process was repeated for all 201 frames for each joint, activity, and participant. First, geometric cartilage models were kinematically transformed using the results of pose estimation. Second, both geometric cartilage models were turned into volumetric point clouds with points placed approximately 0.22 mm apart. Third, the contact volume was defined as the intersection of the two cartilage point clouds. Fourth, the centroid of the contact volume was mapped to the closest triangle center on both geometric bone models to determine the location of the contact centroid on both bones in the joint. The centroid of cartilage contact was first determined in the contact volume and then mapped to the bone surface, rather than being calculated initially on the bone surface in order to give greater weight to areas of cartilage that experienced more overlap. Fifth, the individual points in the point cloud defining the contact volume were mapped to the closest triangle center on both geometric bone models to determine the area and amount of cartilage contact on both bones in the joint. In other words, the total area of cartilage contact was determined by summing the individual areas of each triangle that had a point from the point cloud mapped to it, and the number of points from the point cloud that were mapped to a triangle was used as a surrogate for cartilage contact at that location.

4.2.3 Statistical Analysis

A two-way analysis of variance (ANOVA) was used to test for main effects in cartilage thickness for: a) contacting and non-contacting regions of cartilage, and b) activity. If activity was a main effect, then Tukey's post hoc tests were done to determine the activity pairings that differed. A significance level of 0.05 was used.

Spearman's rank correlation coefficients were found between the cumulative amount of contact for an entire activity and cartilage thickness within each of the 6 regions of cartilage contact (i.e., medial tibial cartilage, lateral tibial cartilage, patellar cartilage, femoral cartilage in the PF joint, femoral cartilage in the medial TF joint and femoral cartilage in the lateral TF joint). Cartilage that did not come into contact during an activity were excluded from this analysis. A correlation was considered strong when $r \geq 0.70$ and moderate when $0.50 \leq r < 0.70$.

Paired t-tests for each participant were done analysing the total area of the contacting cartilage between each of the three matching regions of contact: medial tibia to medial femur in the TF joint, lateral tibia to lateral femur in the TF joint, and patella to femur in the PF joint. The medial and lateral facets of the patella were not distinguished in this analysis. A significance level of 0.05 was used with no correction for multiple t-tests.

4.3 Results

4.3.1 Cartilage Thickness

For all participants, the thickest region of femoral cartilage was in the superior portion of the trochlear groove (Figure 4.1). It ranged from 3.5 to 5.5 mm thick. No consistent patterns in thickness were seen in the remaining portions of femoral cartilage. Peak cartilage thickness on the lateral tibial plateau (3.5 to 5.0 mm) was slightly greater and covered a wider area than that on the medial plateau (3.0 to 4.5 mm) (Figure 4.1). Cartilage thickness for both tibial plateaux peaked near the intercondylar eminence. Cartilage thickness in the patella peaked between 4.0 and 6.0 mm for all participants (Figure 4.1). The thickest portion of patellar cartilage occurred in a medial-lateral oriented strip stretching from the patellar ridge to the lateral border of the patella and was located midway up the articular surface of the patella in the superior-inferior direction. Cartilage bordering the central strip of thick cartilage, including the medial facet, ranged in thickness from 1.0 to 3.0 mm.

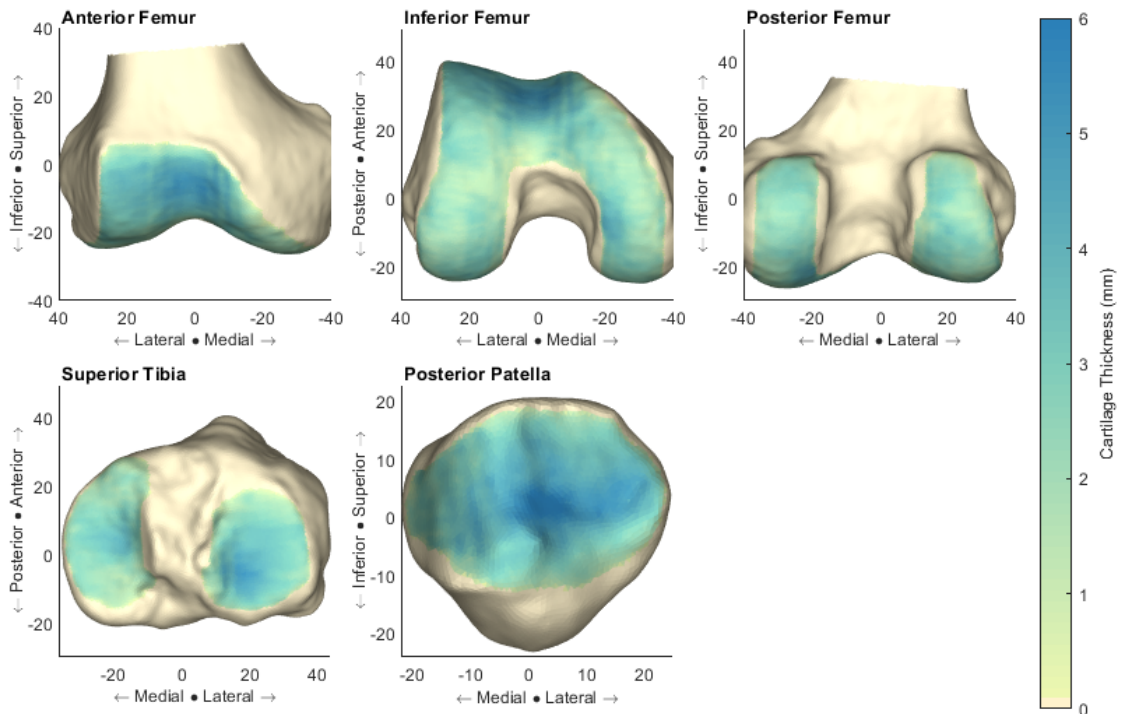


Figure 4.1: Cartilage Thickness Map for a Single Subject. A sample heatmap showing the cartilage thickness for a single subject. All positions are given in mm.

4.3.2 Centroid of Cartilage Contact

There was no noticeable difference in the location of the centroid of cartilage contact for the five dynamic activities (Figure 4.2). Except for the standing activity, contact occurred in the same region. Combining the average data for all six activities, the anterior-posterior range of motion for the TF joint on the medial plateau was larger than on the lateral plateau (13.9 mm compared to 10.3 mm, respectively). The TF contact centroid had little medial-lateral movement on the femoral condyles, with all activities falling into a single line that ran along the middle of the lateral femoral condyle and the lateral side of the medial femoral condyle. Movement of the PF contact centroid mostly occurred in the superior-inferior direction on the patella and the femur. The centroid was located on the lateral patellar facet, just lateral to the patellar ridge. Lateral motion of the contact centroid was seen on both the femur and the patella during deep flexion in the open-chain flexion activity.

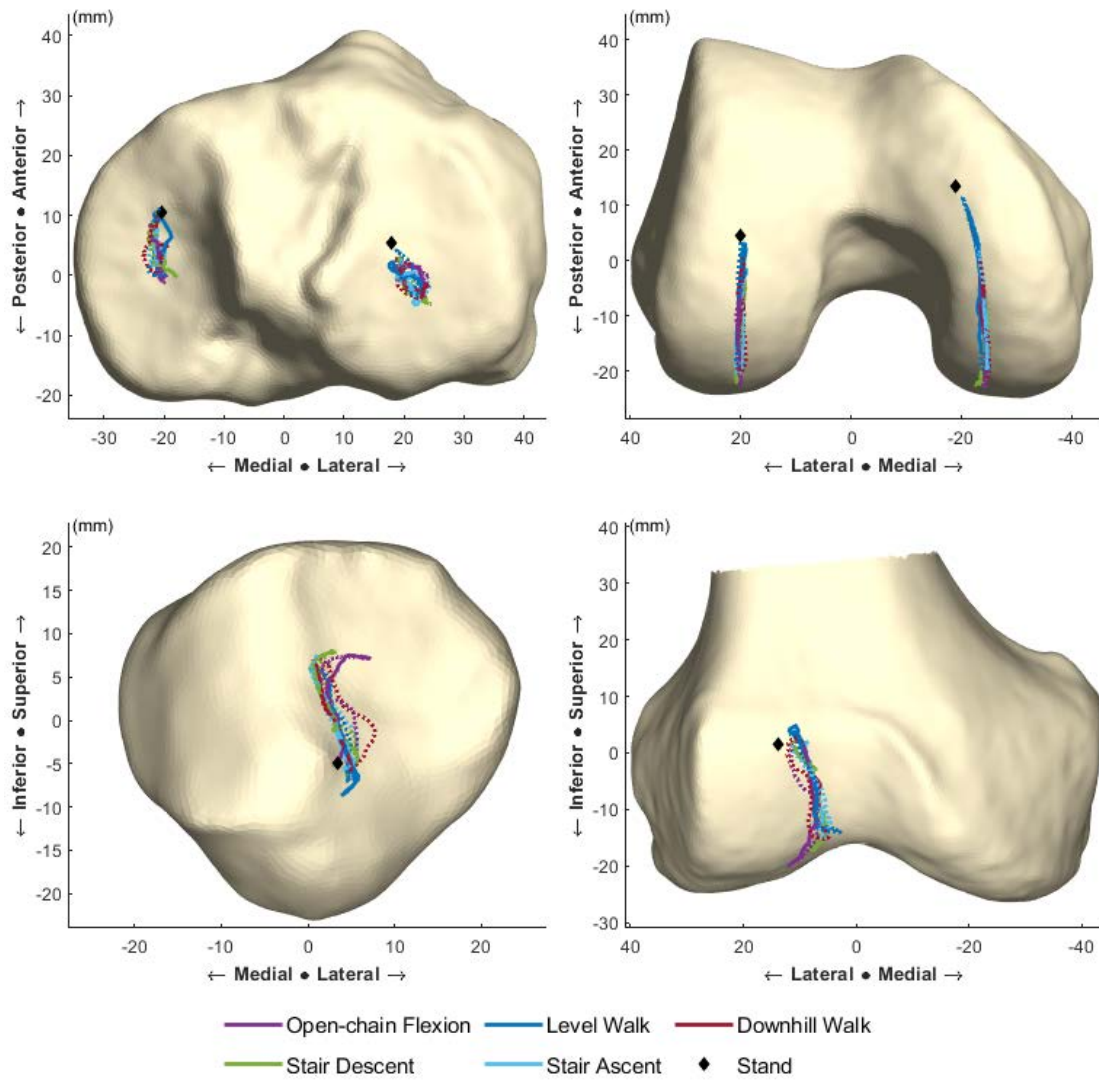


Figure 4.2: Centroid of Cartilage Contact. The tibiofemoral joint is on the top row, separate medial and lateral contact centroids are shown. The patellofemoral joint is on the bottom row, a single contact centroid is shown. For the four gait activities, the solid line indicates stance phase and the dotted line indicates swing phase. For open-chain flexion, the solid line indicates the flexion phase and the dotted line indicates the extension phase. The standing activity is indicated with a black diamond. Centroids are the average of all participants. Bone models are from the participant with closest to the average epicondylar width.

4.3.3 Cartilage Contact and Thickness

In the two-way ANOVA (Table 4.1), contacting regions of cartilage were significantly thicker than non-contacting regions of cartilage in the medial and lateral tibia, the patella, and the patella contacting region of the femur, but not the tibia contacting region of the femur (medial or lateral). The largest difference in thickness between contacting and non-contacting cartilage occurred in the patella, where regions of cartilage that came into contact during an activity were 2.0 mm thicker than regions of cartilage that did not come into contact. This was approximately double the difference in thickness seen in the tibia (0.9 and 1.0 mm for the medial and lateral tibia, respectively), which in turn was

approximately double the difference seen in the patella contacting region of the femur (0.4 mm). Activity performed was only a main effect for the patella. Tukey's post hoc tests revealed two pairs of activities significantly differed at the patella: a) contacting cartilage in open-chain knee flexion was 0.5 mm thicker than contacting cartilage in level walking, and b) contacting cartilage in open-chain knee flexion was 0.5 mm thicker than contacting cartilage in downhill walking. There were no interaction effects.

Table 4.1: Two-way ANOVA Results for Cartilage Contact and Activity Performed. Each column represents a separate ANOVA result. Each row represents an individual variable. The '—' symbol indicates that a main effect was not found (i.e., $p > 0.05$). Where main effects were found for cartilage contact (top row), the direction of the difference is indicated (e.g., "CC > nCC" indicates that contacting cartilage is thicker than non-contacting cartilage), and the magnitude of the difference in mm is indicated in parentheses. Where main effects were found for activity performed (bottom row), the individual activity pairings found to be different using a Tukey's post hoc tests are indicated using the following symbols: LW, level walk; DW, downhill walk; SD, stair descent; SA, stair ascent; OF, open-chain flexion. There were no interaction effects.

	Tibiofemoral				Patellofemoral	
	Medial		Lateral		Patella	Femur
	Tibia	Femur	Tibia	Femur		
Cartilage Contact	CC > nCC (0.9)	—	CC > nCC (1.0)	—	CC > nCC (2.0)	CC > nCC (0.4)
Activity Performed	—	—	—	—	OF > LW (0.5) OF > DW (0.5)	—

Table 4.2: Depth of Contact to Cartilage Thickness Correlations. Spearman's rank correlation coefficients between the measured depth of cartilage contact throughout an entire activity and cartilage thickness. Only segments of cartilage that came into contact were included in the correlation. Done independently for each of the six contact regions (columns) and five dynamic activities (rows). Bottom row shows the average of all activities for that column.

	Tibiofemoral				Patellofemoral	
	Medial		Lateral		Patella	Femur
	Tibia	Femur	Tibia	Femur		
Level Walking	0.46	0.17	0.58	0.11	0.57	0.38
Downhill Walking	0.56	0.32	0.56	0.41	0.65	0.55
Stair Descent	0.63	0.33	0.60	0.53	0.71	0.62
Stair Ascent	0.59	0.38	0.60	0.52	0.70	0.62
Open-chain Flexion	0.52	0.33	0.41	0.48	0.63	0.26
Average	0.55	0.31	0.55	0.41	0.66	0.49

Rank correlation strength varied between the six regions of contact (Table 4.2). Averaged across all activities, cartilage thickness in the patella, medial tibial, and lateral tibial contact regions was correlated to amount of cartilage contact ($r = 0.66, 0.55$, and 0.55 , respectively). Weak or non-existent correlations were found for the three contact regions of the femur ($r < 0.5$). Stair descent had the strongest correlation in five of the six contact

regions (all but the medial femur in the TF joint), and level walking had the weakest correlation in four of the six contact regions (all but the femur in the PF joint and the lateral tibia).

4.3.4 Cartilage Contact Area

The PF region of the femur had the largest area of cartilage contact in all five dynamic activities (16.4 to 19.6 cm²) (Table 4.3). Within each of the three pairs of contacting regions, the area of cartilage contact in the medial femur in the TF joint, lateral femur in the TF joint, and femur in the PF joint were significantly larger than that in the medial tibia, lateral tibia, and patella, respectively. The differences in area of cartilage contact between the five dynamic activities were small; however, the area of cartilage contact during the standing activity was noticeably smaller than that of the dynamic activities.

Table 4.3: Area of Cartilage Contact. The area of cartilage contact for each contact region and activity. Means and standard deviations for the 10 individual participants are shown, as well as the p-values for the paired t-tests for each of the three pairs of contacting regions. Activities are identified as follows: LW, level walk; DW, downhill walk; SD, stair descent; SA, stair ascent; OF, open-chain flexion; ST, standing.

	Tibiofemoral						Patellofemoral			
	Medial			Lateral						
	Contact Area (cm ²)		P	Contact Area (cm ²)		P	Contact Area (cm ²)		P	
	Tibia	Femur		Tibia	Femur		Patella	Femur		
LW	8.4±1.1	12.0±1.5	<0.001	8.1±2.2	10.6±2.2	<0.001	10.1±1.7	16.4±2.1	<0.001	
DW	7.7±1.0	11.2±1.3	<0.001	7.0±1.9	9.5±1.9	<0.001	10.1±1.7	17.3±2.0	<0.001	
SD	7.7±0.8	11.7±1.6	<0.001	7.3±1.5	10.6±1.9	<0.001	10.3±1.7	19.6±1.9	<0.001	
SA	7.9±1.0	10.9±1.1	<0.001	7.1±1.5	9.8±2.1	<0.001	10.2±1.7	17.8±2.1	<0.001	
OF	7.7±1.9	11.7±1.6	<0.001	6.0±1.9	10.0±2.5	<0.001	9.3±1.9	19.5±3.0	<0.001	
ST	6.1±2.2	5.4±2.1	<0.001	5.2±2.0	4.7±1.7	0.001	1.4±0.8	1.5±0.9	0.026	

4.4 Discussion

Making use of accurate 12-DOF knee kinematics captured using a mobile biplane X-ray imaging system, the results presented in this study are the first of their kind to simultaneously measure the locations of PF and TF cartilage contact in healthy individuals during everyday activities. Using these data, we were also able to quantify the relationship between cartilage thickness and cartilage contact in the healthy knee. Based on these results we have accepted our hypothesis that the centroid of cartilage contact will occur on the lateral facet of the PF joint, and rejected our hypothesis that a positive

relationship will be found between cartilage thickness and amount of cartilage contact throughout the knee joint and that it will be strongest in level walking.

There are limitations to this study that should be considered when analysing its results. The meniscus, which serves an important role in force transmission and stabilization within the TF joint (Walker and Erkman, 1975), was not accounted for in this study. Only cartilage-to-cartilage contact was considered in this analysis. In addition, cartilage geometry was assumed to be static, as opposed to an elastic material with geometry that adapts to external forces throughout a day or activity. When measuring changes in TF cartilage that occur over the course of a day, Waterton et al. (2000) found no changes in the total volume of cartilage and the changes in thickness ranged between -0.6 to +0.5 mm from morning to evening. In the current study this would amount to an approximate change of 25% of total cartilage thickness in thinner regions of cartilage and 10% of total cartilage thickness in thicker regions of cartilage.

The patterns of cartilage contact presented in the results above compare favourably to previous studies. In an experimental study of 12 healthy individuals doing a step-up exercise, Suzuki et al. (2012) located the centroid of cartilage contact slightly lateral to the patellar ridge on the patella and slightly lateral to the center of the trochlear groove on the femur. Zavatsky et al. (2004) found the center of force application to be just lateral to the patellar ridge on the patella in a study of 12 cadaver knees performing a simulated knee extension. Likewise, this study located the centroid of cartilage contact in the same region identified by Suzuki et al. (2012) and Zavatsky et al. (2004) for all six activities. This consistency in the centroid's placement on both the patella and the femur, regardless of the activity being performed, leads to the conclusion that the region of cartilage contact is determined by the geometry of the joint and not by active muscle forces that vary with activity. In addition, the centroid's location on the lateral portion of both the patella and femur for the entirety of all six activities suggests that the majority of force transferred through the PF joint is transmitted through the lateral facet.

In the TF joint, Li et al. (2005) measured the center of cartilage-to-cartilage contact for six healthy individuals during weightbearing knee flexion (i.e., a lunge), and found the contact point to lie in the central third of the lateral femoral condyle, the lateral third of the medial femoral condyle, close to the tibial tuberosity on the medial tibial plateau, and to translate in the anterior-posterior direction along the central portion of the lateral tibial

plateau. The TF centers of cartilage contact on the femoral condyles found in this study are in close agreement with those measured by Li et al. (2005), not only for open-chain knee flexion, but for all activities. The point of cartilage contact had little variance between activities on the femoral condyles, indicating that it may be purely a function of TF flexion angle. Conversely, the motion of cartilage contact observed on the tibial plateaus by Li et al. (2005) was not repeated in this study. In these results the center of TF contact on the lateral tibial plateau remained relatively stationary throughout each of the activities, while it translated in the anterior-posterior direction on the medial plateau. These findings are in agreement with those measured by Liu et al. (2010) in eight healthy individuals during the stance phase of level treadmill walking. These differences are possibly due to methodology, where the measurements taken in this study and by Liu et al. (2010) were done on dynamic movements, those taken by Li et al. (2005) were done on individuals performing a quasi-static lunge. Greater translation of the contact point on the medial tibial plateau is likely due to the geometry of the tibial plateau, the medial tibial plateau is concave while the lateral tibial plateau is flat or convex (Figure 4.3) (Hashemi et al., 2008).

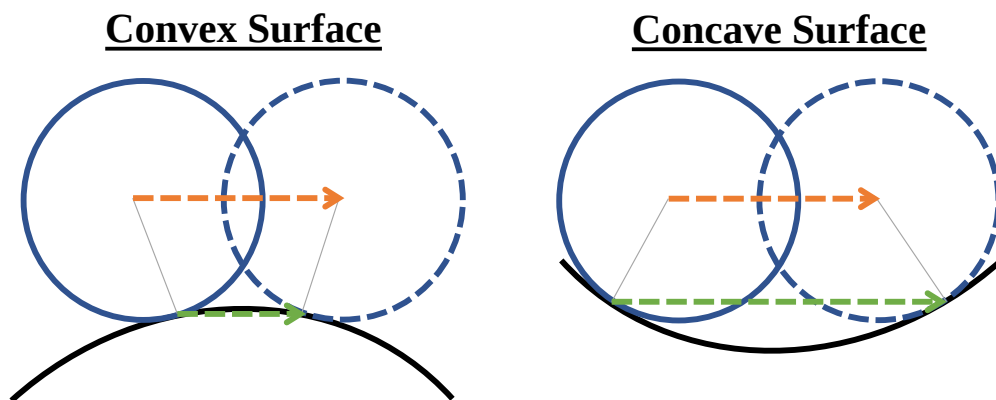


Figure 4.3: Motion of Contact Point on Convex and Concave Surfaces. A visual demonstration of the difference in translation of the contact point between a circle and a convex or concave curve given the same translation in the center of the circle. In the above illustration the radius of the circles (solid and dotted blue lines) are the same. The dotted orange arrow represents the translation of the center of the circle and is the same in both the convex and concave scenario. The green dotted arrow represents the translation of the contact point between the circle and the curve and is greater when the circle is translating on a concave curve.

The relationship between cartilage thickness and cartilage contact found in this study was more complex than hypothesised, but consistent in both methodologies used to assess it (Table 4.1 and Table 4.2). The largest difference in thickness between contacting and non-contacting regions of cartilage and the strongest correlation between amount of cartilage contact over an activity and cartilage thickness were both in the patella, followed

by the tibial plateaus, the femoral cartilage involved in the PF joint, and the femoral cartilage involved in the TF joint. Leading to the conclusion that, in the activities investigated in this study, a clear positive relationship between cartilage thickness and contact exists in the patella and tibia, with no relationship or a more limited relationship in the femur. These results are consistent with those reported by Liu et al. (2010), where greater cartilage deformation occurs in thicker cartilage on the medial and lateral tibial plateaus, but not with Scanlan et al. (2013) and Koo et al. (2011) who concluded that cartilage contact occurred in regions of thicker cartilage for the medial femoral condyle but not the lateral femoral condyle. However, their conclusions were based solely on heel-strike, while the data presented in this study examines the entirety of several activities.

Cartilage developing in areas of greater contact across all regions of the knee joint is too simplistic of a rule to explain the results presented here. If this rule were uniformly true it would hold for the femur, not just the patella and tibia, and the strongest correlations would be for level walking, as this is the most common of the gait activities tested. However, it is notable that in each pair of contacting regions, one side of the pair exhibits a positive relationship between cartilage thickness and contact (i.e., the medial and lateral tibial plateaus and the patella), and one exhibits a weak or non-existent relationship (i.e., the medial and lateral femoral condyles in the TF joint and the femur in the PF joint). In each of these pairs the contacting region that has a positive relationship between cartilage thickness and contact is the smaller region by surface area (Table 4.3). While additional experiments are needed to test this hypothesis, it is a possible explanation for why the relationship between cartilage thickness and contact varies across the knee.

In conclusion, this study is the first of its kind to quantify the cartilage contact patterns of both the PF and TF joints during multiple everyday activities in the healthy knee. Its data shows the locations of the cartilage contact centroids and how they move throughout six activities and quantifies the relationship between cartilage thickness and cartilage contact.

Chapter 5 Load Distribution in the Patellofemoral Joint

This chapter is based on work that has not currently been published or submitted for publication.

5.1 Introduction

Patellar maltracking can increase the compressive stress between the patellar facets and trochlear ridge, causing anterior knee pain and ultimately leading to degenerative changes within the joint (Boden et al., 1997; Elahi et al., 2000). Many studies have investigated the interactions between the internal and external forces acting at the TF joint, for example, how the external knee adduction moment (EKAM) induces a contact force in the medial compartment of the TF joint (Schipplein and Andriacchi, 1991; Zhao et al., 2007). Less is known about the forces acting at the PF joint, like how the knee-crossing muscles act to distribute load across the medial and lateral facets of the patella. This is an important knowledge gap given the frequency of PF pain in the general population, especially those who are active (Rothermich et al., 2015), and the causative link between mechanical loading and progression of knee joint osteoarthritis (Carter et al., 2004).

Experiments on intact cadaver knees have been performed to evaluate the contact pressures present at the PF joint under simulated muscle loading (Hsieh et al., 1998; Huberti and Hayes, 1984; Kumagai et al., 2002; Lee et al., 2003; Li et al., 2004; Mizuno et al., 2001). Computational models have also been used to calculate the forces transmitted by the PF joint during daily activities such as walking (Halonen et al., 2016; Hu et al., 2018; van Rossom et al., 2018). Very few studies have estimated the forces acting in the medial and lateral compartments of the PF joint (Elias et al., 2004; Hirokawa, 1991). Using intact cadaver knees and three-dimensional models of the PF joint in isolation, these authors found the force acting on the lateral patellar facet to be higher than that on the medial facet when a quadriceps force was applied. Using a full-body musculoskeletal model in conjunction with finite element analysis, Akbarshahi et al. (2014) also found higher loads transmitted by the lateral patellar facet during stair ascent.

A higher force acting on the lateral patellar facet can also be anticipated by considering the geometry of the PF joint, as bones grow and adapt to accommodate the external forces applied to them (Robling, 2009). Using the Wiberg patella classification system, Tecklenburg et al. (2006) reported that 90% of all patellae have a larger lateral facet. In an anatomical study of 92 patellae, Baldwin and House (2005) similarly found that the lateral patellar facet is 25% wider than the medial facet. Yoshioka et al. (1987) observed in 32 femurs that the average anterior projection of the lateral femoral condyle is 7 mm larger than that of the medial femoral condyle. A larger lateral patellar facet is presumably the result of a higher lateral PF contact force generated during daily activities like walking.

Loading of the PF joint may be explained by the interactions between the muscles and bones which meet at the knee. Experiments on intact cadavers have shown that hamstring muscle force can increase the pressure between the patella and trochlear groove by increasing posterior translation and external rotation of the tibia relative to the femur (Li et al., 2004). In a series of experiments performed on intact cadaver knees, Li et al. (2004) found that an isolated quadriceps force applied to extend the knee produced anterior translation and internal rotation of the tibia relative to the femur, whereas the addition of an antagonistic hamstring force increased posterior translation and external rotation of the tibia. Posterior tibial translation can increase the angle between the quadriceps tendon and patellar tendon, requiring a higher compressive force to equilibrate the patella, while external tibial rotation can increase lateral patellar tilt and compress the lateral patellar facet into the femur. Thus, the combination of posterior tibial translation and external tibial rotation can increase the force transmitted to the patella during TF flexion.

The medial-lateral distribution of PF contact force may also be influenced by the EKAM. Increased EKAM, which is correlated with increased medial TF loading (Zhao et al., 2007) and TF osteoarthritis (Baliunas et al., 2002), is primarily resisted by the quadriceps during early stance (Shelburne et al., 2006). However, little is known about how the individual quadriceps muscles respond to increased EKAM. Is vastus lateralis force increasing to counteract the EKAM? It is the largest quadriceps muscle by volume, located in the lateral portion of the anterior thigh, and active in early stance phase when EKAM peaks (O'Brien et al., 2010; Pal et al., 2011). The relationship between vastus lateralis and vastus medialis usage has been shown to affect PF kinematics in cadavers

(Goh et al., 1995) and individuals with PF pain (Pal et al., 2011), suggesting that increased use of the vastus lateralis to counteract the EKAM would result in altered knee mechanics.

The primary goal of the present study was to describe and explain the load distribution across the medial and lateral facets of the patella in normal walking. The secondary goals were to determine the extent to which the hamstrings contribute to the load transmitted to the patellar facets, and if vastus lateralis force increases to resist the EKAM, also causing increases in lateral PF contact force. Based on the observed differences in geometry between the medial and lateral patellar facets, we expected the force acting on the lateral facet to be higher. We also hypothesized that the hamstring muscles would contribute to patellar contact force through the posterior force it transmits to the proximal tibia, and that vastus lateralis activation would increase in response to increasing EKAM, drawing the lateral facet of the patella into deeper contact with the femur.

5.2 Methods

5.2.1 Participants and Data Collection

Six young healthy individuals of average build participated in this study (Table 5.1). These individuals were a subset of the cohort described in Chapter 3. Additionally, data collection procedures, while briefly summarized below, are described in detail in Chapter 3. All participants had no history of knee pain and no previous knee surgery.

Table 5.1: Participant Demographics. Height, mass, age, and sex of the individuals modelled in this study.

Height (cm)	Mass (kg)	Age (yrs)	Sex
168	70.4	26.8	M
179	77.8	32.8	M
154	54.2	34.2	F
187	73.5	26.1	M
167	63.3	21.3	F
171	66.8	20.3	F
171 ± 10	67.7 ± 7.6	26.9 ± 5.2	

Each participant walked on level ground at their self-selected speed. Full-body motion, ground reaction forces, and 6-DOF kinematics of the TF and PF joints were measured simultaneously for one walking cycle. Full-body motion was measured using nine motion capture cameras, while ground reaction forces were recorded from three force plates. The

MoBiX imaging system was used to measure 6-DOF movements of the tibia and patella relative to the femur. Each participant also received a CT scan of their right knee for use in the kinematic processing of the biplane X-ray images.

In addition, a fat-suppressed, proton-density-weighted turbo spin-echo ($0.625 \times 0.625 \times 3$ mm) MR scan was taken of each participant's right knee using a 3T scanner (Siemens, Munich, Germany). Articular cartilage lining the femoral condyles, tibial plateaus, and patellar facets were manually segmented with 3D Slicer (Pieper et al., 2006) to create subject-specific geometric models.

5.2.2 Model Development

A three-dimensional musculoskeletal model of the lower limbs was created in OpenSim 3.3 (Delp et al., 2007) and scaled to each participant. The model consisted of 13 rigid segments with 32 degrees-of-freedom. The ipsilateral knee was comprised of 6-DOF TF and 6-DOF PF joints, whereas the contralateral knee was represented as a 1-DOF hinge joint. The pelvis articulated with the ground via a 6-DOF joint. The hip, ankle, and torso were represented as 3-DOF, 2-DOF, and 3-DOF joints, respectively. Eighty muscle-tendon units actuated the skeleton using a muscle-tendon model available in OpenSim (Millard et al., 2013). Twelve Hunt-Crossley spheres (6 under each foot) were used to simulate foot-ground contact, with each foot-ground sphere containing normal, frictional, and damping forces.

Contact forces transmitted by the medial and lateral compartments of the TF and PF joints were simulated using four elastic foundation models. The geometries of the contacting surfaces were based on participant-specific articular cartilage models segmented from MR images. The values of the elastic modulus, Poisson's ratio, and dissipation constant were assumed to be 5 MPa, 0.45, and 8, respectively (Argatov, 2013; Hu et al., 2018; Lenhart et al., 2015). Uniform cartilage thickness was assumed within each elastic foundation model and was based on average thicknesses calculated from the MR scans. Average cartilage thickness was separately calculated for each participant for each of the four elastic foundation models.

The geometries of the knee ligaments and patellar tendon were modelled using 13 nonlinear-elastic bundles (Shelburne et al., 2004). The ACL and PCL were each

represented by an anterior and a posterior bundle. The MCL was represented by a superficial layer comprised of 3 bundles and a deep layer comprised of 2 bundles. The LCL was represented as a single bundle. The patellar tendon (PT) was represented by medial, lateral, and central bundles. Ligament stiffnesses, reference strains, and force-length curves are given in Table 5.2. Attachment sites were manually identified on the geometric bone models segmented from each participant’s CT scan.

Table 5.2: Model Ligament Properties. Properties used to create the 13 nonlinear-elastic bundles that modelled the anterior cruciate ligament, posterior cruciate ligament, medial collateral ligament, lateral collateral ligament, and patellar tendon.

	PCSA Force	Reference Length
aACL	3600	0.040
pACL	4000	0.020
aPCL	3600	-0.150
pPCL	1900	0.030
aMCL	2500	0.020
iMCL	3000	0.040
pMCL	2500	0.020
aDMCL	2000	-0.080
pDMCL	4500	0.030
LCL	3000	0.040
mPT	6000	0.001
cPT	6000	0.008
lPT	6000	0.001

5.2.3 Model Evaluation

Model simulation results when 100 N anterior-posterior shear forces and 5 Nm internal-external rotation torques were applied to the tibia with the femur held fixed were compared to corresponding data obtained from experiments performed on intact cadaveric specimens (Gollehon et al., 1987). Model predictions of PF mechanics during simulated knee extension were also compared to the results of cadaver experiments reported by Zavatsky et al. (2004). In these simulations, a force was applied to the rectus femoris muscle to extend the knee slowly from 90° to 5° over 1 second. Only passive forces were applied by the vasti muscles in the model.

5.2.4 Walking Simulation

A dynamic optimization problem for walking was solved by modifying the data-tracking direct collocation framework described by Lin and Pandy (2017) and Lin et al. (2018) to include a PF joint. In this approach both the states (32 generalized coordinates, 32

generalized velocities, 80 muscle-fibre lengths, 80 muscle-fibre velocities, and 80 muscle activations) and controls (80 muscle excitations) were discretised into 70 nodes, and the dynamic equations of motion were converted into algebraic constraints and included in the cost function as defect errors. The weighting factor for the constraint defining the relationship between PF velocities and accelerations was set to zero, which allowed the quadriceps forces to change without producing large increases in the defect errors associated with the position of the patella. The relationships between the velocities and accelerations of the patella were accounted for through the patella's attachment to the quadriceps tendon. In addition to the defect errors, the cost function included two other terms: a data-tracking constraint and a periodic constraint. For the data-tracking constraint, the optimizer minimized the differences between 13 model-predicted and measured states: the 6-DOF position of the pelvis, ipsilateral knee flexion angle, and magnitudes of the ground reaction forces. For the periodic constraint, all states and controls that were not tracked (377 in total) were required to be identical at the beginning and end of the gait cycle. The optimization problem was solved using the 'fsolve' nonlinear system solver in MATLAB (Mathworks, Natick, MA, USA).

An initial guess for the dynamic optimization solution was developed through a three-step procedure. First, an inverse kinematic problem was solved to minimize the differences between the positions of virtual markers in the model and reflective markers used in the motion capture experiments. Second, with all secondary ipsilateral knee DOFs locked, a computed muscle control (CMC) algorithm was used to calculate the muscle excitations needed to produce the kinematic trajectories calculated in Step 1. Ligament, joint contact, and foot-ground forces were not simulated in this step of the procedure. Third, with the ipsilateral knee flexion angle and hip angles prescribed and the PF joint free to move, a forward dynamics simulation was performed to determine the equilibrium positions of the PF joint consistent with the muscle excitations calculated from CMC. The resulting PF kinematics were then combined with the CMC solution and resampled into 70 nodes to create an initial guess for the dynamic optimization solution.

5.2.5 Center of Contact

For each participant, the centroids of TF and PF cartilage contact was found at each time point during the gait cycle for both the model-predicted and measured kinematics. Each participant's cartilage models were converted into point clouds with a fine density

(average point spacing: 0.1 mm) and transformed into the positions dictated by the model-predicted or measured kinematics. Overlapping points between the two sets of point clouds defined the contact volume. The centroid of the contact volume was mapped to the closest point on the bone models. Further detail on this procedure was given in Chapter 4.

5.2.6 Post-Simulation Analysis

For each participant, RMSE were calculated between the model-predicted and measured TF and PF kinematics, full body kinematics, ground reaction forces, and contact centroid locations. A Pearson's correlation coefficient (R) was computed between the model-predicted PF contact force and the posterior component of the combined hamstrings force in the tibial reference frame. Individual data points from all participants were used for this analysis (70 nodes $\times$ 6 participants = 420 datapoints). Finally, Pearson's correlation coefficients were found between each pair of the following four parameters during stance phase: EKAM, ratio of medial TF contact force to total TF contact force, ratio of lateral PF contact force to total PF contact force, and ratio of vastus lateralis force to the sum of vastus lateralis and medialis forces.

5.3 Results

5.3.1 Patellar Facet Size

The average width and area of the posterior surface of the patella was 41.7 ± 4.4 mm and 1130.1 ± 184.4 mm², respectively. The lateral patellar facet was both wider (26.1 ± 1.3 mm to 15.6 ± 2.5 mm) and had a larger surface area (672.1 ± 111.3 mm² to 458.0 ± 80.2 mm²) than the medial facet.

5.3.2 Model Evaluation with Cadaver Experiments

Model-predicted anterior-posterior translation and internal-external rotation of the tibia relative to the femur for simulated anterior-posterior drawer and axial rotation were within one standard deviation of the in vitro measurements at most TF flexion angles (Figure 5.1). Anterior-posterior laxity near full extension was lower in the model than in the native knee. Model-predicted PF mechanics for knee extension were also within one standard deviation of results obtained from cadaver experiments, except for lateral

patellar rotation and lateral patellar tilt between full extension and 20° of TF flexion, where the patella was more medially rotated and titled in the model compared to experiment (Figure 5.2A). The ratio of PF contact force to input quadriceps force was of a similar shape and visually consistent with direct measurements reported for the full range of TF flexion (Figure 5.2B).

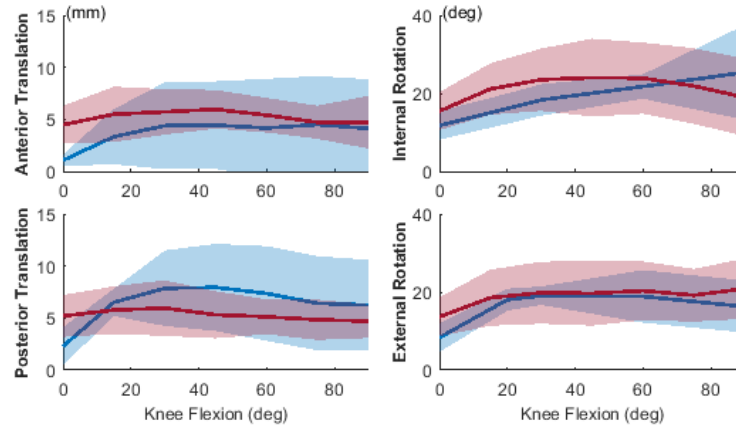


Figure 5.1: Passive Ligament Testing. The average passive knee behavior of all six models (blue) at fixed knee flexion angles. 100 N force applied for anterior-posterior drawer tests. 5 Nm torque applied for internal-external rotation tests. Compared to cadaver data ($n = 16$) reported by Gollehon et al. (1987) (red). Shaded regions indicate one standard deviation.

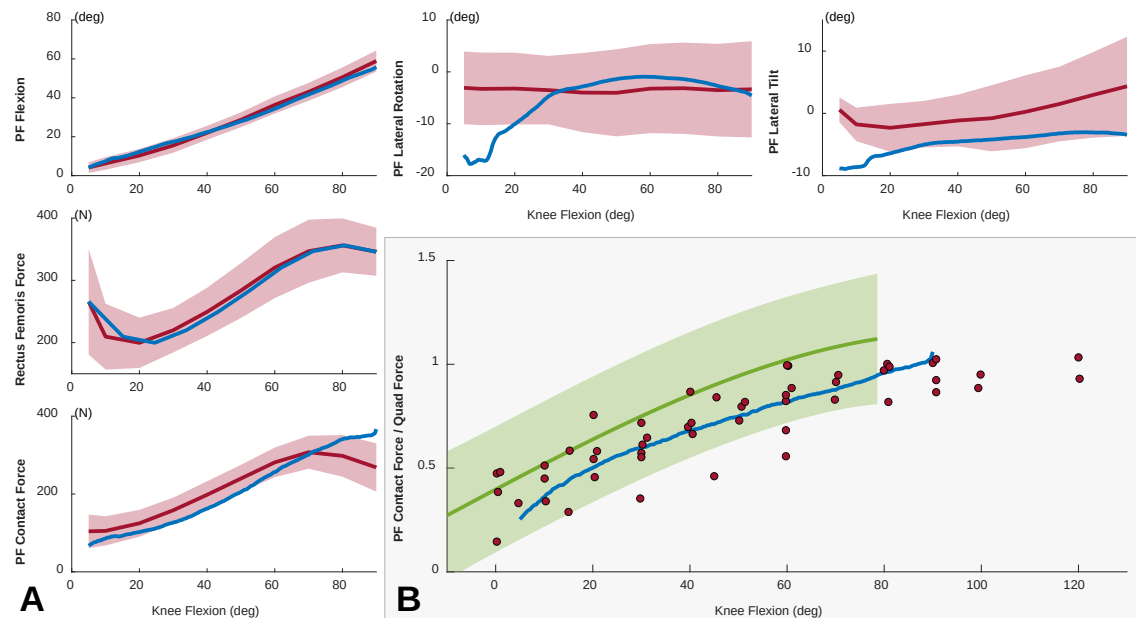


Figure 5.2: Evaluation of Modelled Patellofemoral Joint. Comparisons of model predictions of PF joint kinematics and mechanics to experimental data from cadaver measurements. Panel A shows the PF flexion, lateral rotation, lateral tilt, and contact force of a single model (blue) during a knee extension exercise, given a prescribed rectus femoris force and knee flexion angle. Compared to cadaver data ($n = 12$) reported by Zavatsky et al. (2004) (red). Panel B shows a comparison of the model-predictions and experiment for the ratio of total PF contact force to total quadriceps muscle force. Green line shows a third-order polynomial fit to all model-predicted datapoints for the six walking trails. Blue line shows the result from the knee extension test performed on a single model. Red dots show cadaver measurements summarized by Mason et al. (2008).

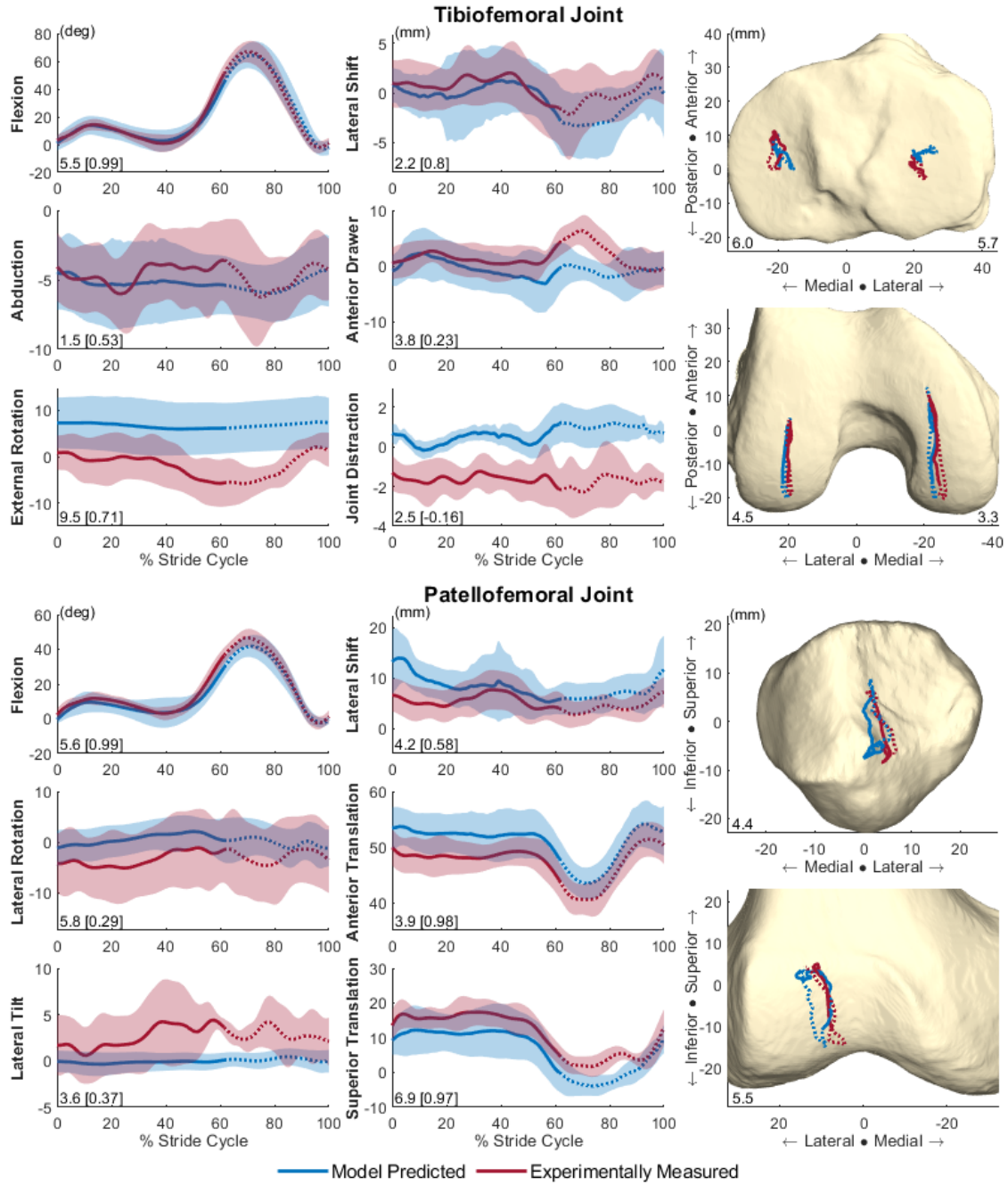


Figure 5.3: Comparison of Model and Experimental Results. Comparison of model-predicted (blue) and experimental (red) kinematics of the patellofemoral and tibiofemoral joints. Solid and dotted lines indicated stance and swing phase, respectively. First and second columns show a comparison of the 12-DOF kinematics. Root Mean Square Errors (RMSE) and Pearson's correlation coefficients (R) are in the bottom left corner of each axes (R in square brackets). Shaded area is one standard deviation. Third column plots the mean centroid of cartilage overlap. One centroid is shown for the patellofemoral joint, separate medial and lateral centroids are shown for the tibiofemoral joint. Bone models shown are for reference only and are from the subject closest to the average epicondylar width.

Table 5.3: Model-predictions Compared to Motion Capture. Average root-mean square errors (RMSE) between model-predicted and experimentally measured full body motion and ground reaction force parameters. Asterisk indicates that a parameter was tracked as part of the cost function. The RMSE for right knee angle reported here is different than that given in Figure 5.3 due to minor differences in the knee flexion angle as measured by the motion capture cameras and the mobile biplane fluoroscopy system. Rotations in deg, translations in mm, forces in N.

		RMSE	
		Right	Left
Pelvis	Tilt	$0.002 \pm 0.001^*$	—
	List	$0.002 \pm 0.001^*$	—
	Rotation	$0.001 \pm 0.001^*$	—
	Anterior	$0.016 \pm 0.005^*$	—
	Superior	$2.200 \pm 0.807^*$	—
	Medial	$0.140 \pm 0.030^*$	—
Lower Limb	Hip Flexion	0.041 ± 0.018	0.400 ± 0.070
	Hip Adduction	0.030 ± 0.015	0.667 ± 0.075
	Hip Rotation	0.018 ± 0.006	0.215 ± 0.076
	Knee Angle	$0.003 \pm 0.002^*$	0.837 ± 0.100
	Ankle Angle	0.051 ± 0.033	0.219 ± 0.061
	Subtalar Angle	0.051 ± 0.031	0.313 ± 0.87
Ground Reaction Force	Anterior	$48.40 \pm 4.67^*$	$41.70 \pm 11.67^*$
	Superior	$66.45 \pm 13.05^*$	$62.82 \pm 12.51^*$
	Medial	$12.71 \pm 4.50^*$	$12.21 \pm 4.19^*$

5.3.3 Model Validation with Gait Experiments

Model-predicted full body kinematics and ground reaction forces were within 0.85° , 2.5 mm, or 70 N of measured values (Table 5.3). Model-predicted knee kinematics for walking were generally within one standard deviation of the experiment data, except for external tibial rotation and joint distraction (Figure 5.3). The tibia was more externally rotated and distracted in the model compared to experiment, with excessive external tibial rotation resulting in increased medial patellar tilt in the model. Model-predicted TF flexion, external tibial rotation, and lateral tibial shift showed strong correlations with measured kinematics ($R > 0.7$), indicating that there was good agreement in the shape of these curves. RMS errors, which measure the absolute difference between the model-predicted and measured kinematics, were < 5 mm for translations and $< 6^\circ$ for flexion and abduction. External rotation was associated with the largest RMS error (9.5°), indicating an offset between the predicted and measured curves. Model-predicted patellar

flexion, anterior patellar translation, and superior patellar translation were each nearly perfectly correlated to the corresponding kinematic measurements ($R \approx 1$), while RMS errors were $< 6^\circ$ for rotations and < 5 mm for lateral shift and anterior translation. The centroids of cartilage contact predicted by the model were in good agreement with measurements obtained at both joints, with all RMS errors ≤ 6 mm (Figure 5.3, third column). In the model and experiment, the centroid of PF joint contact was on the lateral facet throughout the gait cycle, with the calculated centroids lying closer to the central vertical ridge. Overall, the calculated locations of PF and TF joint contact agreed more closely with experiment during stance than swing. Model-predicted locations of joint contact for the medial TF compartment were more consistent with experiment than those calculated for the lateral TF compartment.

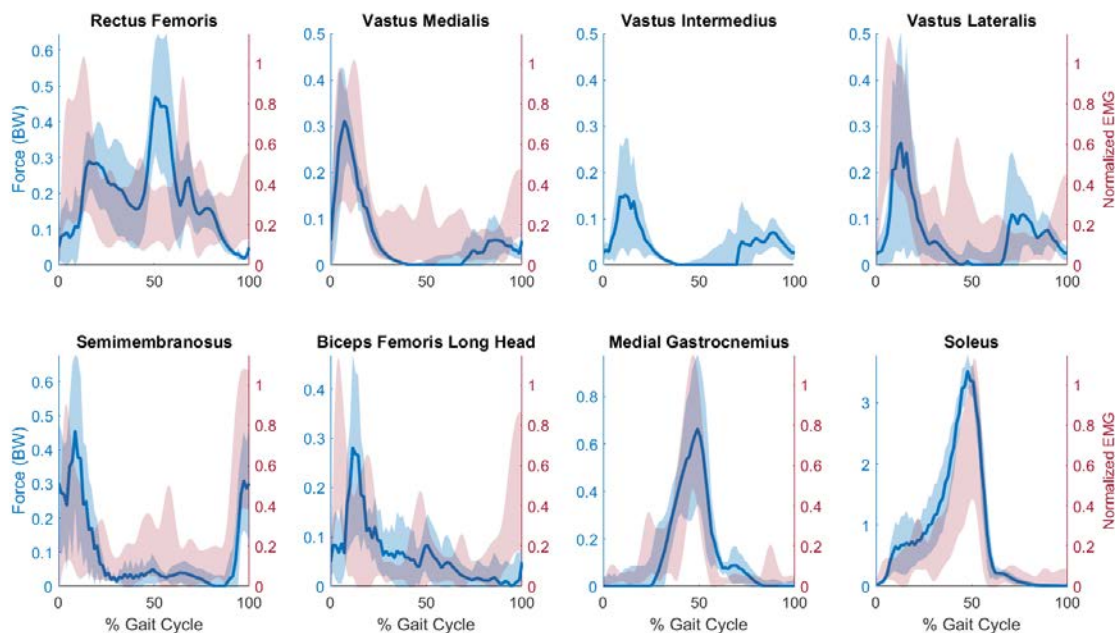


Figure 5.4: Model-Predicted Active Muscle Forces. Model-predicted active muscle forces (blue) plotted over one gait cycle of overground walking. Shaded area shows one standard deviation. EMG data (red) shows one standard deviation of individual muscle activations measured in eight separate subjects during overground walking from the literature (Dorn, 2011). No EMG data for the vastus intermedius.

The sequence and timing of model-predicted active muscle forces were consistent with that measured for young healthy people walking at their self-selected speed (Figure 5.4). In both model and experiment the vastus medialis, vastus lateralis, and hamstrings co-contracted during early stance and the ankle plantarflexors (soleus and gastrocnemius) were activated in late stance. Differences between model and experiment were also evident. Vastus medialis and lateralis were turned off during late stance and early swing in the model whereas moderate levels of EMG activity are observed *in vivo*. In addition,

muscle EMG measurements indicate that the rectus femoris is activated throughout the gait cycle whereas this muscle was turned off during early stance and late swing in the model.

5.3.4 Model-Predicted Joint Contact Force

The total TF contact force peaked twice during stance, with the magnitudes of the first and second peaks calculated to be 3.1- and 2.9-times bodyweight (BW), respectively (Figure 5.5). Total PF contact force also showed two peaks: the first peak occurred during early stance and its magnitude was 0.56 BW while the second peak was slightly higher (0.58 BW) and occurred during swing (Figure 5.5). The standard deviation in the magnitude of the first peak was twice as large as that for the second peak.

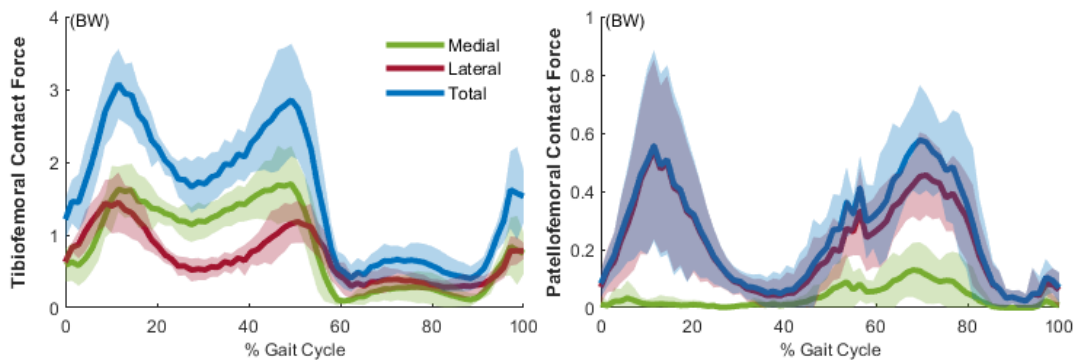


Figure 5.5: Model-Predicted Joint Contact Forces. Tibiofemoral and patellofemoral joint contact forces plotted over one gait cycle of level walking for all six participants in this study. Colours indicate medial (green), lateral (red), and total (blue) joint contact forces. Shaded area shows one standard deviation.

Peak forces in the medial TF compartment were greater than those in the lateral TF compartment, 1.6 BW to 1.4 BW in early stance and 1.7 BW to 1.2 BW in late stance. The force acting on the lateral patellar facet was much higher than that acting on the medial facet. Medial PF contact force was almost zero for much of the stance phase and peaked at 0.15 BW during swing. Lateral PF contact force peaked first during early stance (0.54 BW) and then again in swing (0.51 BW).

5.3.5 Contribution of Hamstring Muscles

The posterior component of the hamstrings force was linearly correlated to total PF joint contact force ($R = 0.61$) (Figure 5.6). The relationship was positive, meaning an increase in posterior hamstrings force was associated with an increase in total PF joint contact force.

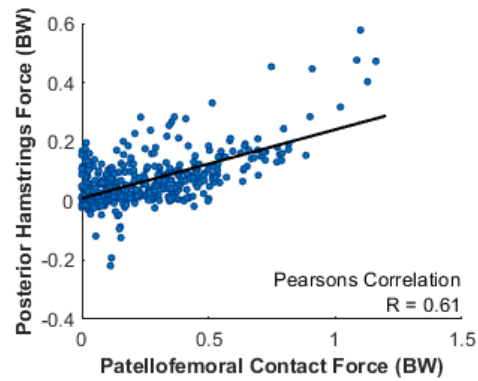


Figure 5.6: Posterior Hamstrings Force to Patellofemoral Contact Force. The posterior component (in the tibial reference frame) of the total hamstrings force plotted against the total patellofemoral contact force. Individual data points plotted for each of the six participants. Linear trendline fit using the ‘polyfit’ function in MATLAB.

5.3.6 External Knee Adduction Moment

EKAM showed a moderate to strong correlation ($R = 0.63$) to the ratio of medial to total TF contact force (Figure 5.7). No other strong correlations were found between EKAM, TF force, PF force, and vastus lateralis force ($R \leq 0.37$ for all).

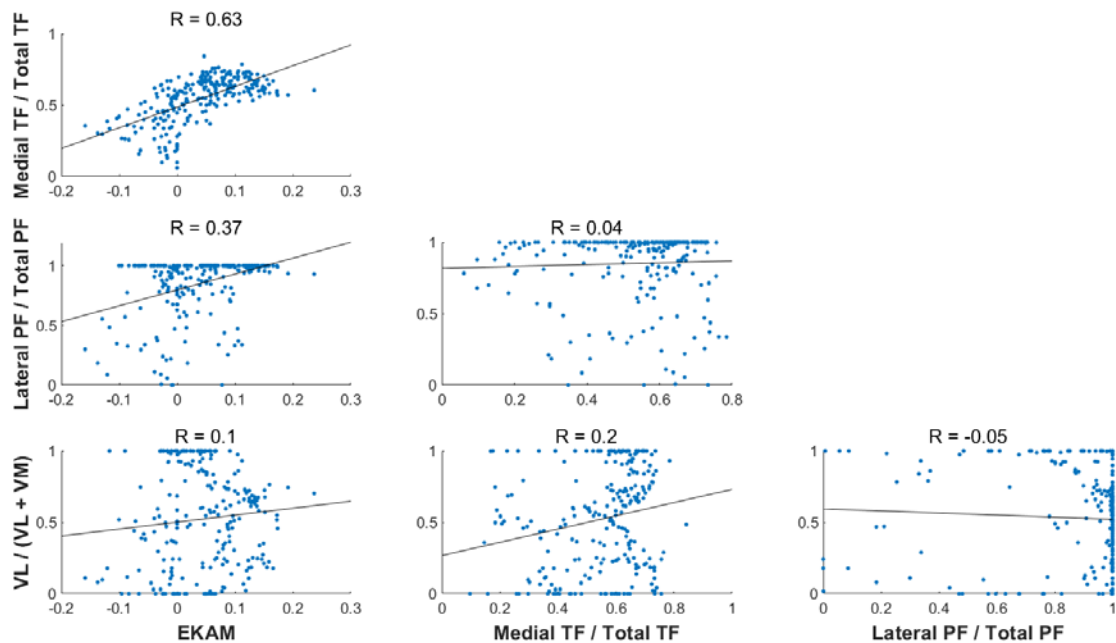


Figure 5.7: Correlations to the External Knee Adduction Moment. Scatterplots and Pearson’s correlation coefficients (R) between the external knee adduction moment (EKAM), ratio of medial tibiofemoral contact force to total tibiofemoral contact force (Medial TF / Total TF), ratio of lateral patellofemoral contact force to total patellofemoral contact force (Lateral PF / Total PF), and ratio of vastus lateralis force to the sum of vastus medialis and vastus lateralis force ($VL / (VL + VM)$). Linear trendline found using MATLAB’s ‘polyfit’ function.

5.4 Discussion

We used a 12-DOF model of the knee-joint complex (6-DOF for each of the TF and PF joints) embedded in a full-body musculoskeletal model to determine load distribution at the PF joint for one cycle of normal walking. We tested three hypotheses. First, that the force transmitted to the lateral patellar facet would be larger than the force transmitted to the medial patellar facet; which we have accepted based on the findings of this study. Second, that the posterior pull of the hamstrings on the proximal tibia will contribute to the contact force produced in the PF joint; which we have accepted based on the findings of this study. Third, that vastus lateralis force will increase in response to increasing EKAM, drawing the lateral facet of the patella into deeper contact with the femur; which we have rejected based on the findings of this study.

Model-predicted contact forces at the TF and PF joints were validated against *in vivo* measurements of 12-DOF TF and PF joint kinematics measured for healthy individuals during walking. Model-predicted TF abduction-adduction, TF medial-lateral shift, PF flexion-extension, PF medial-lateral rotation, PF anterior-posterior translation, and PF superior-inferior translation were within one standard deviation of biplane X-ray imaging measurements (Figure 5.3). The model accurately predicted rapid superior translation of the patella just prior to heel-strike, which Gray et al. (2019) noted as a characteristic kinematic feature of normal walking. However, TF internal-external rotation, TF anterior-posterior drawer, and PF medial-lateral shift did not compare as well, which may be due to inaccuracies in the stiffness characteristics of the knee ligaments assumed in the model and/or structures present in the real knee that were not represented in the model. For example, lateral shift of the patella predicted during early stance may be restrained by the action of the medial patellar retinaculum, which was not modelled in this study. In addition, Lenhart et al. (2015) found that TF internal-external rotation and anterior-posterior drawer were sensitive to the ligament properties assumed in their knee model, which may explain some of the differences between model and experiment observed here. Despite these discrepancies the predicted centroid of cartilage contact was consistent with measurements obtained from biplane X-ray imaging and MRI (Figure 5.3).

Overall, there is general agreement between our calculations of TF and PF joint loading and results reported previously by others. The shape and magnitude of the total TF contact

force were consistent with the findings of previous modelling and experimental studies (Figure 5.8). Sritharan et al. (2012) found that the vasti and the gastrocnemius are the largest contributors to the TF contact force in early and late stance, respectively, which agrees with the muscle timing calculated in the present study (Figure 5.4). Consistent with previous PF models, our calculations show a peak in total PF contact force during early stance (Figure 5.8), when the vasti are active. However, not all previous model predictions show a peak in total PF contact force during swing, as we found here. This peak in the PF contact force during swing is due to the ratio of PF contact force to quadriceps force being greater when the knee is flexed (Thomeer et al., 2017), as it is during swing, meaning a small quadriceps force can cause a large contact force during swing.

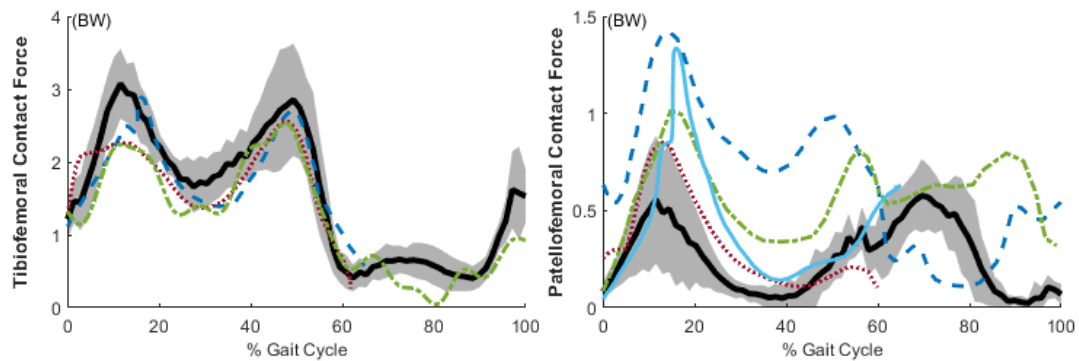


Figure 5.8: Comparison of Model-Predicted to Previously Reported Joint Loading. A comparison of the total joint contact forces predicted in this study (black) to previous studies. Tibiofemoral contact force is compared against: Shelburne et al. (2005), blue dashed line; Sritharan et al. (2012), red dotted line; and Fregly et al. (2012), green dashed line. Patellofemoral contact force is compared against: Hu et al. (2018), dashed blue line; Halonen et al. (2016), dotted red line; Chen et al. (2010), dashed green line; and Shelburne et al. (2005), solid blue line.

However, the magnitude of our predicted total PF joint contact force was lower than that found in previous studies. The cause of this difference may be methodological: the medial and lateral PF retinacula were not simulated in this study, which, in combination, may act to compress the patella into the femur *in vivo*. Alternatively, PF contact force depends on quadriceps muscle force, which in turn depends on the moment arm of the quadriceps and the relative movements of the femur, tibia and patella. The dynamic optimization solution derived in the present study permitted all of these parameters to be determined simultaneously, allowing for a more accurate simulation of knee extensor biomechanics, as opposed to a two-step modelling approach where quadriceps muscle force is solved using a simplified knee model that does not allow for changes in PF kinematics (Halonen et al., 2016). Finally, we modelled the patellar tendon as a nonlinear-elastic element, as

opposed to an inextensible tendon (Hu et al., 2018; Shelburne et al., 2005), allowing for a more realistic simulation of the knee extension process.

Our calculations indicate that most of the compressive load acting at the PF joint during gait is transmitted by the lateral patellar facet. Van Kampen and Huijskes (1990) reported that external rotation of the tibia increases lateral shift of the patella, and our model simulations resulted in a greater amount of external rotation than that found experimentally (Figure 5.3). However, it is not likely that our model's increased tibial external rotation had a significant effect on the medial-lateral load distribution in the PF joint for two reasons: a) the majority of PF contact force was transmitted through the lateral patellar facet in all six of our models, including models where external rotation matched measured values, and b) the predicted and measured centroids of cartilage contact (Figure 5.3) were located in similar positions on the lateral patellar facet. Increased proportional use of the vastus lateralis compared to the vastus medialis was not found to be the cause of a higher force acting in the lateral patellar facet (Figure 5.7). PF contact force may be greater in the lateral compartment for geometric reasons. The lateral anterior femoral condyle and the lateral patellar facet are both larger than their medial counterparts (Baldwin and House, 2005; Tecklenburg et al., 2006; Yoshioka et al., 1987), giving the lateral PF joint more surface area to come into contact. The Q-angle is a measure approximating the angle between the lines-of-action of the quadriceps tendon and patellar tendon. A Q-angle of 0° indicates that the quadriceps tendon and patellar tendon are perfectly aligned with each other whereas a positive/negative Q-angle indicates that the net pull on the patella has a component in the lateral/medial direction (Smith et al., 2008). While Q-angles are not a robust clinical measure for diagnosing patellar disorders, they are consistently found to be positive, generally around 10 to 15° (Smith et al., 2008), as they were in this study ($12.1 \pm 1.2^\circ$). This suggests that the quadriceps muscles and patellar tendon combine to pull the patella laterally into the lateral ridge of the trochlear groove in the healthy knee. Likewise, our model findings show that the quadriceps muscles and the patellar tendon act in unison to pull the patella laterally during all phases of the gait cycle (Figure 5.9). Defrate et al. (2007) came to a similar conclusion in finding that the patellar tendon pulls the patella more laterally at small TF flexion angles, indicating that greater PF joint contact would occur on the lateral side when the knee is extended. This is consistent with our finding that medial PF contact

force is greater during swing, when the knee is more flexed, as opposed to stance, when it is more extended (Figure 5.5).

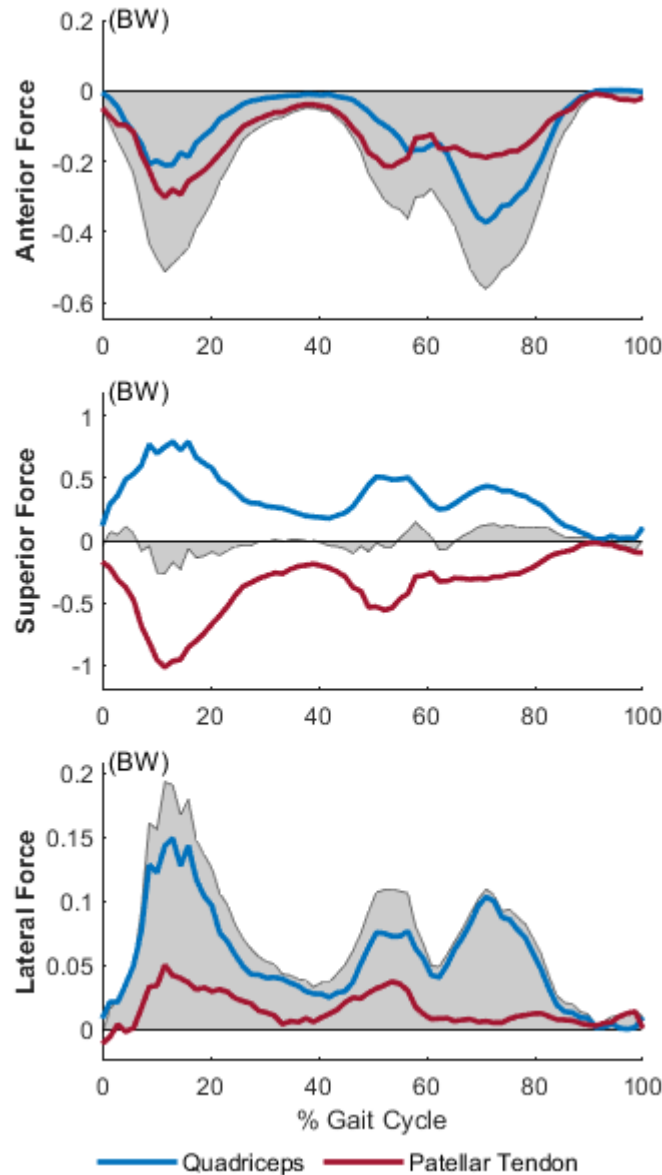


Figure 5.9: Forces Acting on the Patella. Anterior-posterior, superior-inferior, and lateral-medial components of the quadriceps muscles (blue) and patellar tendon (red). Components are taken in the patellar coordinate system. Shaded grey area represents the summation of the quadriceps muscles and patellar tendon forces.

While Sritharan et al. (2012) has shown that, collectively, the vasti resist the EKAM during early stance, our results show that, relative to the vastus medialis, vastus lateralis force does not increase to resist the EKAM. This leaves open the question of what, if any, impact the EKAM has on the relative activation of the individual vasti muscles. In this study the vastus lateralis was assumed to insert on the superior surface of the patella at a single point, whereas anatomically the vastus lateralis inserts on the patella over an area,

both through the quadriceps tendon and directly into the lateral patella. This simplifying assumption may have compromised the vastus lateralis' ability to resist the EKAM in this study. Alternatively, as indicated in the results of this study, it is possible that the relative activation levels of the individual vasti are independent of EKAM. In other words, increasing EKAM does not favour increasing vastus lateralis activation over the other vasti, it causes all the vasti to uniformly increase in activation.

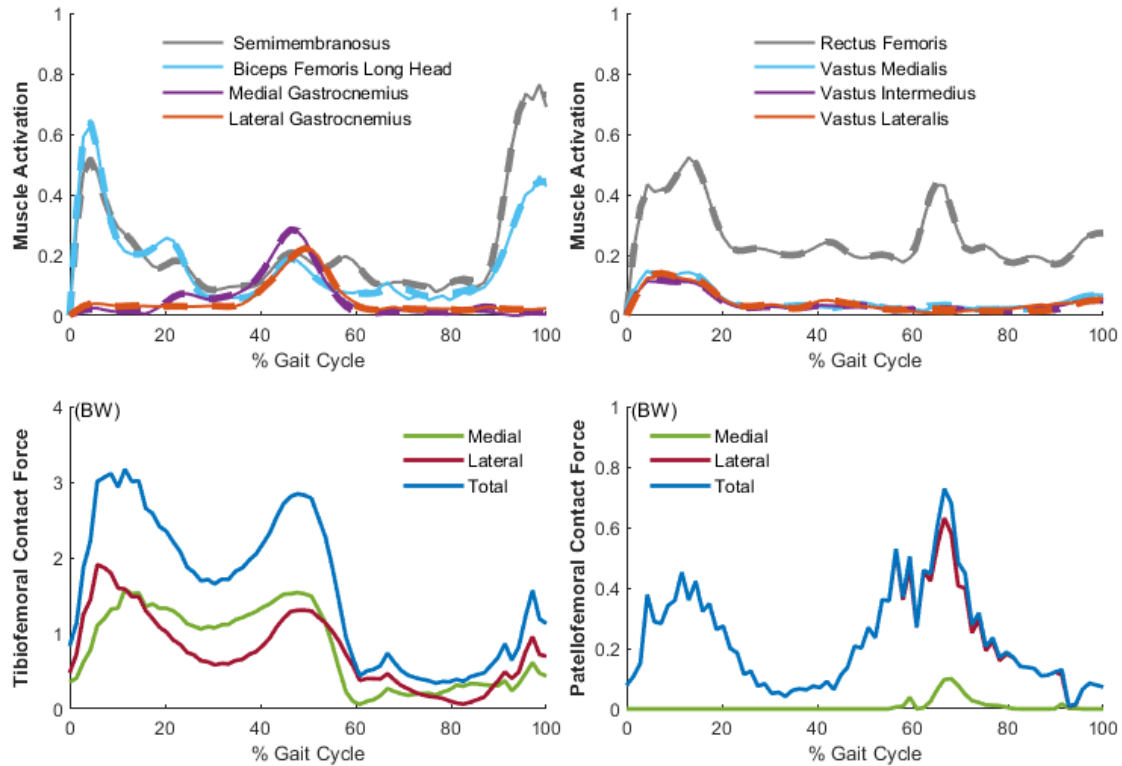


Figure 5.10: EMG Tracking Model Solution. Overground walking solution for a single model with the activations of seven knee-crossing muscles being tracked from EMG data reported by Dorn (2011). The top row shows the simulated activations of the tracked muscles (solid line) and the EMG measured target muscle activations (dotted line). The bottom row shows the tibiofemoral and patellofemoral joint contact forces over the gait cycle.

The quadriceps and hamstring muscles were contributors to the knee joint contact modelled in this study. In order to test the robustness of our findings against errors in model simulated muscle activation, an additional optimization problem was formulated and solved for a single model in which the cost function was modified to include a data-tracking constraint for seven knee-crossing muscle activations: rectus femoris, vastus medialis, vastus intermedius, vastus lateralis, semimembranosus, biceps femoris long head, and gastrocnemius. These muscle activations were constrained to track the mean EMG measurements of eight healthy subjects walking at their self-selected speed (Dorn, 2011). All other components of the cost function were the same as described previously.

The TF and PF joint contact force for this simulation exhibited the same patterns as seen when muscle activations were not tracked (Figure 5.10). Contact force for both joints contained two peaks: one in early stance, and one in late stance and early swing for the TF and PF joints, respectively. Almost all PF joint contact force was transmitted through the lateral patellar facet.

There are limitations of the present analysis that should be considered when interpreting the results. The medial patellofemoral ligament, which is a stabilizer against PF lateral shift when the knee is extended (Amis et al., 2003), was not included in the model. Nonetheless, the model-predicted centroid of PF joint contact compared favourably with experiment (Figure 5.3). A second potential limitation was the way in which the quadriceps muscles were assumed to insert onto the patella. The quadriceps have complicated patellar insertion dynamics, with both the relative anterior-posterior alignment of the muscles forming the quadriceps tendon and the presence or absence of oblique heads of the vastus lateralis and vastus medialis varying considerably between individuals (Waligora et al., 2009). These variations were not accounted for in this study which may alter the effect of individual quadriceps muscles on the mechanics of the patella. Finally, the menisci, which play an important role in both the stability and load transmission of the TF joint, were not modelled in this study.

In summary, this study represents the first solution for 12-DOF knee mechanics during walking to validate its results through direct comparison to experimental kinematics of the activity being performed. Its results show that: a) most PF contact force is transmitted by the lateral facet of the patella, a phenomenon likely driven by the geometry of the joint, b) the hamstring muscles are an important contributor to patellofemoral contact force, and c) increased EKAM does not result in increased vastus lateralis force and lateral PF contact force.

Chapter 6 Conclusion

6.1 Introduction

The main objective of this dissertation was to grow our understanding of the knee's healthy function during daily activities. A specific focus was placed on including the PF joint in each of the studies performed, as this joint is often neglected in studies of the knee because of the complexity that the patella introduces in both experimental imaging and computational modelling of the knee.

The motivation for this objective was twofold. First, healthy functioning of the knee is critical to overall musculoskeletal health. The knee is the largest joint in the human body (Palastanga et al., 2002) and undergoes significant loading during daily activities (Fregly et al., 2012). Its ability to lengthen and shorten the lower limb is important for performing most ambulatory activities (Palastanga et al., 2002). Second, the knee is a common site for musculoskeletal injury (Thiem et al., 2013). Osteoarthritis, ACL reconstructions, and PF pain are three examples of many disorders that affect the knee (Hung and Ahmed, 2017; Mall et al., 2014; Rothermich et al., 2015). These disorders affect a large portion of the general population, both young and old, active and immobile, and each has a different and complex etiology, but improved treatment for each of them begins with an improved understanding of the healthy mechanics of the knee.

To achieve this objective we captured experimental data on the movement of the knee joint as ten healthy participants performed six activities: level walking, downhill walking, stair descent, stair ascent, open-chain knee flexion, and standing. This data capture was made possible through the use of a novel knee imaging methodology, a mobile biplane X-ray imaging system (Guan et al., 2016a), which allowed for noninvasive measurement of the full three dimensional motion of the PF and TF joints while the above activities were performed.

6.2 Summary of Research Questions and Findings

The three sets of questions outlined in section 1.2 and answered in the body of this dissertation are briefly summarized here.

6.2.1 Healthy Patellofemoral Kinematics during Functional Activities

This chapter detailed the experimental procedure that was used to capture the PF kinematic data for six activities, presented the 6-DOF motion of the PF during each activity, and analysed the extent to which each DOF was coupled to the TF flexion angle.

What are the full three-dimensional kinematics of the patellofemoral joint during the performed activities?

Kinematic data were presented in Figure 3.3 and Table 3.1. The most prominent rotational DOF was flexion-extension. The most prominent translational DOFs were anterior-posterior and superior-inferior translation. Medial-lateral patellar shift displayed a similar pattern for all activities, with relatively little motion when the knee was flexed beyond 20°, and a rapid lateral movement when the knee neared full extension. This pattern highlighted the importance of the trochlear groove on the distal femur, which acts to constrain the motion of the patella when the knee is flexed. The patella underwent superior patellar translation in the last 20% of all five dynamic activities tested, emphasizing the impact that the elasticity of the patellar tendon can have on PF kinematics.

To what extent are each of the 6-degrees-of-freedom of the patellofemoral joint linked or coupled to the flexion-extension angle of the tibiofemoral joint?

The degree of coupling between each PF DOF and TF flexion was given in Figure 3.3. Flexion-extension and anterior-posterior translation were almost perfectly coupled to TF flexion. While not as strong, medial-lateral shift and superior-inferior translation were both coupled to TF flexion. Medial-lateral rotation and medial-lateral tilt were not coupled to TF flexion, however, both had very small ranges of motion, indicating that there was not much motion for TF flexion to couple to. It was concluded that, based on the activities tested here, the PF joint can effectively be modelled as having 1-DOF.

6.2.2 Cartilage Contact in the Knee and its Relationship to Thickness

Using the PF kinematics presented in Chapter 3, as well as its associated TF kinematics, the location of cartilage contact during each activity was calculated, and the hypothesis that thicker regions of knee cartilage experience more contact was tested.

Where are the locations of cartilage contact in the full knee joint complex during the performed activities?

The centroids of cartilage contact were presented in Figure 4.2. More anterior-posterior movement of the contact centroid occurred on the medial tibial plateau, compared to the lateral tibial plateau. On the femur TF contact occurred along the middle of the lateral femoral condyle and the lateral side of the medial femoral condyle. PF contact was concentrated on the lateral side of both the patella and the femur. The location of contact appeared to be determined by TF flexion angle, with little variation seen between activities.

Do thicker regions of cartilage experience more joint contact than thinner regions of cartilage in the knee? Is there a correlation between cartilage thickness and contact?

These results were presented in Table 4.1 and Table 4.2. A relationship was found between cartilage thickness and cartilage contact, but it was not consistent for every region of contact. In each pair of contacting regions within the knee, one side of the pair exhibited a positive relationship between cartilage thickness and contact (i.e., the medial and lateral tibial plateaus and the patella), and one exhibited a weak or non-existent relationship (i.e., the medial and lateral femoral condyles in the TF joint and the femur in the PF joint). Activity did not influence this relationship.

6.2.3 Load Distribution in the Patellofemoral Joint

This chapter detailed the creation of a musculoskeletal model with full 6-DOF PF and TF joints. It was used to determine the contact forces and kinematics of the healthy knee during level walking.

What are the contact loads transmitted to the knee during level walking, and how are they transmitted through the medial and lateral facets of the patellofemoral and tibiofemoral joints?

The contact loads during level walking were presented in Figure 5.5. The TF joint underwent two peaks in contact force that both occur during stance phase, both at approximately 3 BW. The majority of its load was transmitted on the medial side. The PF joint also underwent two peaks in contact force, one in early stance, and one in swing

phase, both peaked at approximately 0.55 BW. Almost all the PF contact force went through the lateral facet of the patella.

Do the hamstrings contribute to the load transmitted to the patellar facets?

The relationship between posterior hamstring force and patellar contact force was shown in Figure 5.6. The posterior component of hamstrings force showed a moderate correlation to PF contact force. This indicates that the hamstrings do contribute to the load transmitted to the patellar facets.

How does the vastus lateralis muscle respond to an external knee adduction moment during walking, and does this response contribute to patellofemoral loading?

The correlations done in answering this question were shown in Figure 5.7. While the ratio of medial TF contact force to total TF contact force was positively correlated to the EKAM, as shown in previous studies, no other correlations were found. These results indicate that the vastus lateralis does not increase in activation to oppose the EKAM.

6.3 Future Work

The experimental and modelling results presented in this dissertation can be improved and expanded upon in several ways, some examples of which are given here.

Account for the menisci in the TF joint for both the cartilage contact (Chapter 4) and contact modelling (Chapter 5) studies. The primary purpose of the menisci is to increase the contact area between the round femoral condyles and the flatter tibial facets (Palastanga et al., 2002). While difficult to account for because of its mobility with respect to the knee bones (the menisci are not rigidly attached to the tibial plateaus), its inclusion would have an impact on both the cartilage contact regions that were calculated and the contact forces and kinematics of the knee that were modelled.

Create a mathematic model of PF joint kinematics. Given the close grouping of the PF kinematics when plotted against TF flexion angle presented in Chapter 3, and the conclusion that the PF joint can be modelled as a 1-DOF system, the next step is to create a mathematical model that gives the 6-DOF kinematics of the healthy PF joint as a function of TF flexion angle. This would be a valuable contribution to computational

modelling studies that wish to account for the PF joint without dealing with the complexities that adding an additional 6 DOFs would cause.

Include the medial patellofemoral ligament and a more anatomically accurate representation of the quadriceps muscle in the musculoskeletal model. The medial patellofemoral ligament is a stabilizer against lateral displacement of the patella at low TF flexion angles (Amis et al., 2003; Amis, 2007), one discrepancy in the validation of the model predicted kinematics was excessive lateral displacement of the patella at low TF flexion angles. An anatomically accurate representation of the quadriceps muscle's attachment to the patella should include a vastus medialis obliquus and a vastus lateralis obliquus which can serve to pull the patella medially and laterally, respectively (Amis, 2007; Waligora et al., 2009), the inclusion of these muscles may serve to better test the hypothesis that the vastus lateralis activates to resist the EKAM.

Expand the methodology used to simulate level walking in Chapter 5 to simulate other activities. Increased PF pain is associated with ascending and descending stairs among people with knee disorders (Kujala et al., 1993; Thomee et al., 1999). In addition, van Rossom et al. (2018) found the average and maximum PF contact forces for ascending and descending stairs to be multiple times their magnitude for level walking.

6.4 Closing Remarks

This dissertation has expanded our knowledge of the mechanics of the healthy knee through both experimental and computational methodologies. It has focused on contributing results that include the PF joint and assess the knee during everyday activities, two areas where previous research was lacking. Its conclusions will prove vital in assisting with the diagnosis and treatment of many common knee disorders and will provide a useful source of information for future investigations into the knee

Appendix A

An author-accepted version of the following published work is reproduced here:

Gray, H.A., Guan, S., **Thomeer, L.T.**, Schache, A.G., de Steiger, R., Pandy, M.G., 2019. Three-dimensional motion of the knee-joint complex during normal walking revealed by mobile biplane x-ray imaging. J Orthop Res 37, 615-630.

Three-Dimensional Motion of the Knee-Joint Complex During Normal Walking Revealed by Mobile Biplane X-Ray Imaging

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ABSTRACT: Accurate knowledge of knee kinematics is important for a better understanding of normal joint function and for improving patient outcomes subsequent to joint reconstructive surgery. Limited information is available that accurately describes the relative movements of the bones at the knee in vivo, even for the most common of all activities: walking. We used a mobile X-ray imaging system to measure the three-dimensional motion of the entire knee-joint complex—femur, tibia, and patella—when humans walk over ground at their natural speeds. Data were recorded from 15 healthy individuals (9 males, 6 females; age 30.5 ± 6.2 years). The most pronounced rotational motion of the tibia was flexion-extension followed by internal-external rotation and abduction-adduction (peak-to-peak displacements: 70.7° , 9.2° , and 1.9° , respectively). Maximum anterior translation of the tibia was 6.5 mm and occurred in early swing, coinciding with peak knee flexion and peak internal rotation. The most prominent rotational motion of the patella was flexion-extension (peak-to-peak displacement: 50.5°). The tibia pivoted about the medial compartment of the tibiofemoral joint, conferring greater movements of the contact centers in the lateral compartment than the medial compartment (15.4 and 9.7 mm, respectively). Internal-external rotation, anterior-posterior translation and medial-lateral shift of the tibia as well as flexion-extension and anterior-posterior translation of the patella were each coupled to the knee flexion angle, as were movements of the contact centers at each joint. These fundamental data serve as a valuable resource for evaluating knee joint function in normal and pathological gait. The data are available in Supplementary_Material_Data.xlsx. © 2019 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. *J Orthop Res* 37:615–630, 2019.

Keywords: patellofemoral; tibiofemoral; standing; joint coupling; knee model

The knee is a complex system consisting of three bones and two joints: the femur and tibia articulate at the tibiofemoral joint; and the femur and patella, at the patellofemoral joint. Both joints must withstand high contact forces (well in excess of body weight) during daily activities such as walking.¹ Accurate knowledge of three-dimensional knee kinematics and condylar motion is important for a better understanding of normal and pathological joint function, and ultimately for improving patient outcomes subsequent to total knee arthroplasty, knee-ligament reconstruction and high tibial osteotomy. Knee-joint biomechanics has been studied for well over a century, yet only two studies have recorded the full three-dimensional movements of the femur and tibia for one complete cycle of normal walking (i.e., heel-strike to the next ipsilateral heel-strike),^{2,3} and no data to our knowledge are available to describe the corresponding motion of the patella.

The most common method for measuring joint motion in vivo involves video-based motion capture with reflective markers fixed on the skin. Unfortunately, movement of the skin with respect to bone causes uncertainty in the measured joint kinematics, with root-mean-square (RMS) errors as high as 24° reported for bone motion at the knee.⁴ Errors due to skin movement may be minimized, if not eradicated,

by inserting steel pins directly into bone,^{2,5,6} however, intra-cortical pins cause pain and impingement, potentially altering the subject's natural gait pattern.⁷

Dynamic X-ray imaging (or fluoroscopy) enables accurate, non-invasive measurement of bone motion in vivo. Single and biplane X-ray imaging have been used to measure motion of the healthy knee during both walking and running gaits.^{8–11} In these studies, the imaging equipment remained fixed relative to the ground while subjects ambulated on a treadmill. Thus, even for relatively slow walking speeds, it has been possible to analyze only a portion of the stride cycle because the joint invariably strays beyond the capture volume of the X-ray imaging device.^{8–12}

X-ray fluoroscopy also has been used to measure the locations of the tibiofemoral joint contact centers during weight-bearing activities such as lunging.^{13,14} The results indicate that the contact center moves further in the lateral compartment than in the medial compartment, suggesting that the femur pivots about the medial tibial condyle and that axial rotation of the tibia is coupled to the knee flexion angle.¹⁵ Indeed, the results of cadaver experiments suggest that the healthy knee has two primary axes of rotation fixed in bone: an optimal flexion axis that passes through the centers of the femoral condyles, and a longitudinal axis that is parallel to the mechanical axis of the tibia and passes through the medial joint compartment.^{16,17} These findings have driven recent innovations in total knee replacement design such as the development of a medially constrained knee implant,¹⁸ where the medial tibiofemoral articulation is fashioned as a ball-and-socket joint to restrict anterior-posterior movement of the medial femoral condyle. The

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hypothesis that the healthy knee pivots about the medial tibiofemoral compartment was recently challenged by Koo and Andriacchi¹⁹ who, using skin-surface markers, found the center of rotation in the transverse plane to lie in the lateral compartment during the stance phase of walking.

Here we provide the first non-invasive measurements of the kinematics of the entire knee-joint complex for normal walking. We used a unique Mobile Biplane X-ray (MoBiX) imaging system²⁰ to track and image the three-dimensional motion of the femur, tibia, and patella as healthy young adults walked at their natural speeds over ground. Our specific aims were to (i) obtain simultaneous measurements of 6-degree-of-freedom (6-DOF) tibiofemoral and patellofemoral joint kinematics and condylar motion for one complete stride cycle; (ii) examine the extent to which secondary rotations and translations of the femur, tibia, and patella are coupled to the knee flexion angle; and (iii) determine the location of the center of rotation of the knee in the transverse plane for the stance and swing phases of normal walking.

MATERIALS AND METHODS

Design: Descriptive cross-sectional study.

Level of Evidence: III.

Participants

Fifteen healthy individuals (9 males and 6 females; age 30.5 ± 6.2 years; height, 168.0 ± 9.3 cm; weight, 67.4 ± 8.4 kg) with no knee pain and no history of knee surgery gave informed consent to participate in this study. Ethics approval for the experimental procedures was granted by the Human Research Ethics Committee at the University of Melbourne.

Experimental Protocol

Biplane X-ray images (resolution $1,024 \times 1,024$ pixels) of the right knee were acquired using the MoBiX imaging system (Fig. 1A) sampling at 200 frames/s with an exposure time of 1/200 s. Design specifications and performance of the MoBiX imaging system are described in detail by Guan et al.²⁰ The X-ray units were operated in continuous mode (110 kV and 13.1 mA) with an inter-beam angle of approximately 60° and an X-ray source to image intensifier distance of approximately 1.4 m. Body-segmental motion and ground reaction forces were recorded synchronously with the X-ray images. Full-body 3D motion data were recorded using a 9-camera video-based motion capture system (VICON, Oxford, UK) sampling at 120 Hz, while ground reaction forces were measured using three strain-gauged force plates (AMTI, Watertown, MA) sampling at 1,080 Hz and mounted flush with an 8.4 m long wooden walkway. In addition, a CT scan (voxel size $0.35 \times 0.35 \times 0.50$ mm) of the right knee was obtained with the joint extended. Each participant was exposed to a maximum combined radiation dosage of 0.18 mSv from both biplane X-ray imaging and the CT scan.

All participants completed two sessions. The first session involved a 1-hour familiarization procedure that gave each participant the opportunity to practice walking with the MoBiX system operational. The second session involved gait data collection. Each participant wore a sleeveless top, lead vest, shorts, and sandals. Forty-five retro-reflective skin

markers were placed at predetermined landmarks.²¹ The participant walked back and forth in a straight line several times at a comfortable speed while average stride length was measured and used to mark the locations of the first three steps on the walkway (R1, L1, and R2 in Fig. 1B). Each trial consisted of 10 complete steps, with foot placement prescribed only for the first three steps; the subject was free to choose foot placement, and hence step length, for the remaining seven steps of each trial. This protocol was adopted to ensure the participant used a comfortable step length and stepped onto the three force plates during the 4th, 5th, and 6th steps (L2, R3, L3, respectively, in Fig. 1B). The MoBiX system tracked and imaged the right knee of the participant during the mid-portion of the walking trial, from the 2nd right-heel-strike to the 4th right-heel-strike (i.e., two complete stride cycles from R2 to R4 in Fig. 1B). Kinematic data were processed for one complete stride cycle, from the 3rd right heel-strike to the 4th right heel-strike (R3 to R4 in Fig. 1B). Kinematic and kinetic data were also recorded for a static trial in which the subject stood on one force plate with the arms abducted away from the trunk and the feet set apart at approximately shoulder width.

Knee-Joint Kinematics

Six-DOF tibiofemoral and patellofemoral joint kinematics were derived from the biplane X-ray images by following a pipeline of previously described procedures.²⁰ Image processing, pose estimation, and joint kinematic analysis were performed using a custom program developed in MATLAB (Mathworks Inc., Natick, MA). Maximum RMS errors associated with 6-DOF kinematic measurements for the healthy knee during overground walking were reported to be 0.78 mm and 0.77° for translations and rotations of the tibiofemoral joint²⁰ and 0.37 mm and 1.46° for translations and rotations of the patellofemoral joint.²² Individual geometric models of the femur, tibia and patella, required for pose estimation, were segmented from the CT images using 3D Slicer.²³ An anatomical joint coordinate system²⁴ was developed to describe 6-DOF kinematics of both the tibiofemoral and patellofemoral joints (see Fig. 2). Raw kinematic measurements were filtered using a fourth-order, low-pass Butterworth filter with a cut-off frequency of 10 Hz and then resampled to 201 time points over one stride cycle. Major gait events were identified using the full-body motion data and ground force measurements.²⁵

Condylar Motion and Knee-Joint Center of Rotation

Femoral condylar motion was determined by identifying the approximate geometric center of each femoral condyle and calculating its position with respect to the tibia at each time point during the stride cycle. The centers of the femoral condyles were found by determining the intersection between the medial-lateral axis of the femur (axis X_F in Fig. 2) and the approximate mid-sagittal plane of each condyle.

The kinematic measurements and shapes of the articulating surfaces of the bones were used to estimate the locations of joint contact between the femur and tibia and between the femur and patella. The contact centers in the medial and lateral compartments of the tibiofemoral joint were found by identifying the minimum distances between the medial and lateral condyles of the femur and a transverse plane fitted to the tibial plateau.²⁶ The contact centers for the patellofemoral joint were found by calculating the locations of points

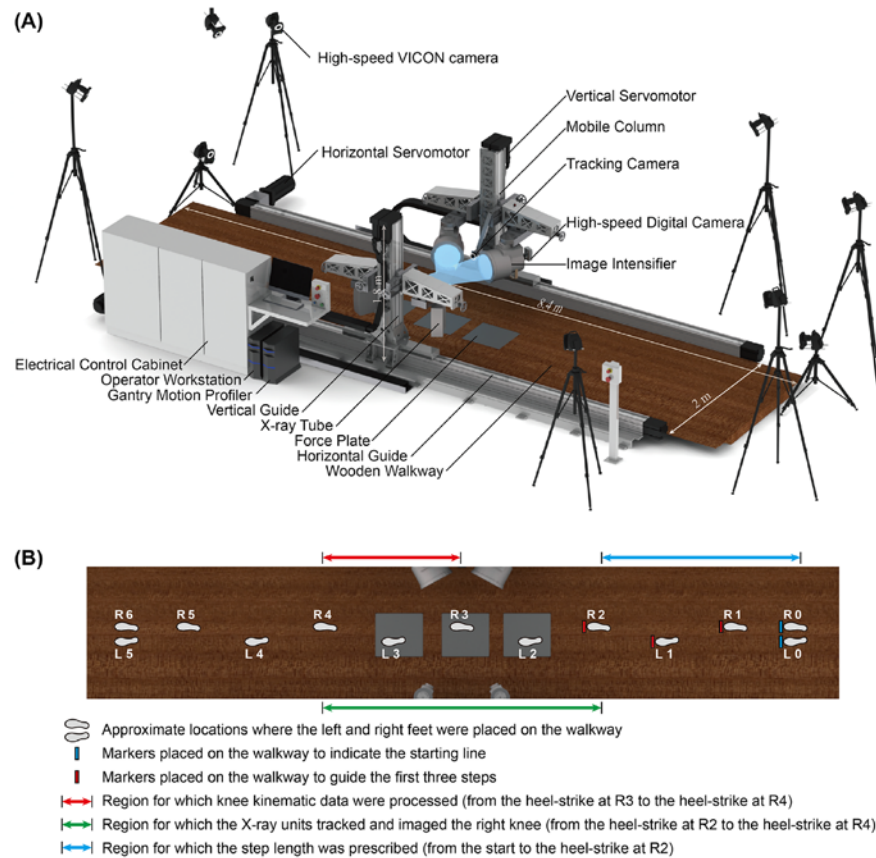


Figure 1. Schematic diagram illustrating the experimental setup for measuring knee-joint kinematics during overground walking. (A) The Mobile Biplane X-ray (MoBiX) imaging system, comprised two X-ray units and a custom robotic gantry mechanism which enabled horizontal and vertical motion along guides. A high-speed camera (Tracking Camera) acquired images of a retro-reflective marker located on the lateral side of the right knee. These images were processed by a real-time computer (Gantry Motion Profiler) to generate velocity commands controlling the motion of the X-ray units via the servomotors. Two high-speed digital cameras mounted on the two image intensifiers were used to capture biplane X-ray images which were then stored on the Operator Workstation. (B) Plan view of the walkway showing locations of the force plates and approximate foot placement used during each gait trial. Kinematic data were processed for one complete stride cycle, from the 3rd heel-strike to the 4th heel-strike of the right leg (R3 to R4).

on the medial and lateral patellar facets that were closest to the femoral surface.

The center of rotation of the knee in the transverse plane was found by projecting the medial-lateral axis of the femur (axis X_F in Fig. 2) onto the transverse plane of the tibia ($X_T O_T Y_T$ plane in Fig. 2) at each time point, and the intersection of these projections was then calculated on a least-squares basis¹⁹ for the stance phase, swing phase, and the entire stride cycle.

Normalization

Joint translations, locations of the femoral condylar centers, locations of joint contact, and the center of rotation of the knee in the transverse plane were each normalized on a

participant-specific basis using a coefficient (C_i) defined as follows:

$$C_i = \frac{\bar{w}}{w_i}$$

where w_i is the femoral bicondylar width measured for the i th participant and $\bar{w}$ is the mean femoral bicondylar width calculated across all participants.²⁷

Statistical Analyses

Means and standard deviations were calculated for tibiofemoral and patellofemoral joint kinematics, motion of the

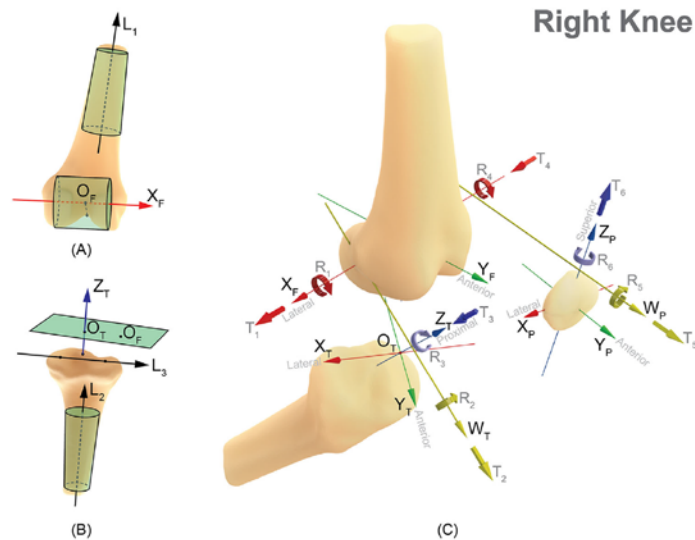


Figure 2. Coordinate systems used to define the relative positions and orientations of the femur, tibia, and patella of a right knee. (A) The femoral coordinate system was constructed by fitting a cylinder to the posterior and distal portions of both femoral condyles. The axis of the cylinder was defined as the femoral X-axis (X_F) pointing to the right. The origin (O_F) was defined as the foot of the perpendicular dropped from the intercondylar notch apex to the X-axis. A cone was fitted to the femoral diaphysis to determine the long axis (L_1) of the femoral shaft. The Y-axis (Y_F) pointed anteriorly and was mutually perpendicular to X_F and L_1 . The Z-axis (Z_F) pointed proximally and was mutually perpendicular to X_F and Y_F . (B) The tibial coordinate system was constructed by fitting a cone to the tibial diaphysis to determine the location of the long axis (L_2) of the tibia. The Z-axis (Z_T) was parallel to L_2 and pointed proximally, passing through the midpoint between the two intercondylar eminences. The relative positions of the femur and tibia, as obtained from the CT scan with the knee unloaded and fully extended, were used to locate the origin (O_T) of the tibial reference frame. The origin (O_T) was located at the foot of the perpendicular dropped from O_F to Z_T . The Y-axis (Y_T) pointed anteriorly and was mutually perpendicular to Z_T and the line (L_3) connecting the approximate center of each tibial plateau. The X-axis (X_T) pointed to the right and was mutually perpendicular to Z_T and Y_T . (C) For the patellar coordinate system, the origin (O_P , not shown) was located at the centroid of the patella. The Y-axis (Y_P) pointed anteriorly while the X-axis (X_P) pointed to the right and was mutually perpendicular to Y_P and the posterior patellar ridge. The Z-axis (Z_P) pointed superiorly and was mutually perpendicular to X_P and Y_P .³⁶ The X_F axis of the femur, Z_T axis of the tibia and Z_P axis of the patella were used as the respective body-fixed axes to define the relative positions and orientations of the bones at the tibiofemoral and patellofemoral joints.³⁴ Two floating axes (W_T and W_P) that were perpendicular to X_F were defined to also be perpendicular to Z_T and Z_P , respectively. For the tibiofemoral joint, flexion (R_1), abduction (R_2), and external rotation (R_3) were defined by the X_F , W_T , and Z_T axes, respectively, while lateral shift (T_1), anterior drawer (T_2) and joint distraction (T_3) were defined along X_P , W_P and Z_P axes, respectively. For the patellofemoral joint, flexion (R_4), lateral rotation (R_5), and lateral tilt (R_6) were defined by the X_P , W_P , and Z_P axes, respectively, while lateral shift (T_4), anterior translation (T_5), and superior translation (T_6) were defined along X_F , W_P , and Z_P , respectively. Arrows in the figure indicate the positive direction of each degree of freedom.

femoral condylar centers and tibiofemoral joint contact centers, and the center of rotation of the knee in the transverse plane. Student's paired *t*-tests were used to compare (i) peak values of 6-DOF knee-joint kinematics and the displacements of the condylar centers and joint contact centers measured for walking against mean values of the corresponding variables measured for standing; (ii) peak-to-peak displacements of the condylar centers on the medial and lateral sides of the knee; and (iii) peak-to-peak displacements of the joint contact centers in the medial and lateral compartments of the knee. Two-sided, two-sample *t*-tests were used to compare parameters between males and females. Differences were considered statistically significant when $p < 0.05$.

We used a curve-fitting technique to assess the degree to which secondary motions of the tibia and patella relative to the femur were coupled to the knee flexion angle. For example, coupling between lateral shift of the tibia and the knee flexion angle was evaluated by fitting a spline function

to the 201 data points in the graph of mean lateral tibial shift versus mean knee flexion; the spline was generated using a MATLAB function called "fit" (Mathworks Inc., Natick, MA), which applied a least-squares approximation with the "smoothing parameter" set to 0.1. The coefficient of determination (r^2) and the Root Mean Square Error (RMSE) associated with this fit were used as indicators of how strongly lateral tibial shift was coupled to the knee flexion angle. An r^2 value of 1.0 and RMSE of 0.0 would indicate perfect coupling between the two variables. Details of these methods are provided in Supplementary_Material_Methods.docx.

RESULTS

Mean walking speed across all participants was 1.28 ± 0.15 m/s (range: 1.03–1.51 m/s). No statistically significant differences were found in either tibiofemoral or patellofemoral joint kinematics measured for the male and female participants, thus all data were pooled.

Tibiofemoral Kinematics

The most pronounced rotational motion of the tibia was flexion-extension (peak-to-peak displacement, 70.7°). The magnitudes of internal-external rotation (peak-to-peak displacement, 9.2°) and abduction-adduction (peak-to-peak displacement, 1.9°) were much smaller, the latter deviating relatively little throughout the stride cycle (Fig. 3A, Table 1). The largest translational motion of the tibia was anterior-posterior translation (peak-to-peak displacement, 6.4 mm) followed by medial-lateral shift (peak-to-peak displacement, 4.0 mm) and compression-distraction (peak-to-peak displacement, 1.0 mm).

The knee was slightly bent at heel-strike and proceeded to flex during early stance to a maximum of 14.1° at contralateral toe-off (Fig. 3A). It then extended to a minimum flexion angle of 1.0° just prior to contralateral heel-strike. Terminal stance was marked by rapid flexion, which continued into swing until the flexion angle reached a maximum of 68.6° immediately after toe-off. The knee then extended for most of the swing phase and became hyperextended just before heel-strike. The tibia was internally rotated for most of the walking cycle (Fig. 3A). Maximum internal rotation (external rotation angle of -5.4°) occurred during swing and coincided with peak knee flexion immediately after toe-off. As the knee extended through swing the tibia rotated externally to its maximum value of 3.8° just before heel-strike.

The tibia translated anteriorly from heel-strike and reached a peak position of 3.7 mm at contralateral toe-off (Fig. 3A). Maximum anterior translation (6.5 mm) occurred in early swing, coinciding with peak knee flexion and peak internal rotation (Fig. 3A). The tibia then translated posteriorly with knee extension. Medial-lateral translation of the tibia was minimal during the first half of the stride cycle (Fig. 3A). During late stance and early swing, the tibia translated medially and then laterally, reaching a peak position of 2.3 mm just before heel-strike. Compression-distraction and abduction-adduction were small with nondescript profiles that appeared to be unrelated to the flexion angle (Fig. 3A and B, Table 1).

Internal-external rotation, anterior-posterior translation and medial-lateral shift were each coupled to the knee flexion angle ($r^2 > 0.76$ for all) (Fig. 3B). An increase in knee flexion was associated with internal rotation, anterior translation, and medial shift of the tibia relative to the femur.

Patellofemoral Kinematics

The most prominent rotational motion of the patella was flexion-extension (peak-to-peak displacement, 50.5°); medial-lateral tilt and medial-lateral rotation were relatively small (peak-to-peak displacements of 4.5° and 3.8°, respectively) and displayed unremarkable profiles (Fig. 4A, Table 1). The largest translational motion was superior-inferior translation

(peak-to-peak displacement, 14.9 mm) followed by anterior-posterior translation (peak-to-peak displacement, 11.3 mm) and medial-lateral shift (peak-to-peak displacement, 4.9 mm).

The time course of patellar flexion was closely similar to that of knee flexion (cf Figs. 3A and 4A). Patellar flexion exhibited two maxima, each coinciding with a peak in the knee flexion angle. The patella flexed through weight acceptance and attained a peak of 10.0° at contralateral toe-off. It then flexed rapidly in late stance and early swing, reaching a peak of 47.4° immediately after toe-off.

The patella underwent rapid superior translation just after heel-strike and continued to translate superiorly during mid-stance, reaching a peak superior position of 17.2 mm (Fig. 4A and B). It then translated inferiorly in late stance before reaching its most inferior position of 2.3 mm shortly after toe-off, when both knee flexion and patellar flexion were maximum. The patella translated superiorly and inferiorly by relatively small amounts (~4 mm) during swing before translating rapidly in the superior direction immediately before heel-strike. Peak-to-peak displacements for patellar rotation, tilt and shift were comparable during stance and swing: the patella rotated, tilted, and shifted by 3.8°, 4.5°, and 4.0 mm, respectively, during stance compared to corresponding movements of 2.5°, 3.1°, and 3.6 mm in swing.

Patellar flexion was tightly coupled ($r^2 = 0.998$) and linearly related (coefficient of determination for linear regression = 0.99) to the knee flexion angle (Fig. 4B). Patellar flexion lagged knee flexion for nearly the entire stride cycle, especially during swing when the knee was highly flexed. Anterior-posterior translation of the patella also was tightly coupled ($r^2 = 0.97$) and linearly related (coefficient of determination for linear regression = 0.94) to the knee flexion angle, with an increase in knee flexion associated with posterior translation of the patella (Fig. 4A and B). Medial-lateral tilt and medial-lateral shift were coupled to knee flexion when the knee was bent beyond 20°.

Condylar Motion and Joint Center of Rotation in the Transverse Plane

The centers of the medial and lateral femoral condyles translated predominantly in the anterior-posterior direction during walking (Fig. 5, Table 2). Both condyles translated posteriorly as the knee flexed during late stance and early swing, with peak-to-peak displacement of the lateral condyle (11.6 mm) being significantly greater than that of the medial condyle (7.6 mm, $p < 0.001$). Peak-to-peak displacements of the two condyles in the medial-lateral direction were similar throughout one stride.

Consistent with the movements of the centers of the femoral condyles, anterior-posterior translation of the tibiofemoral contact center was greater in the lateral compartment than the medial compartment

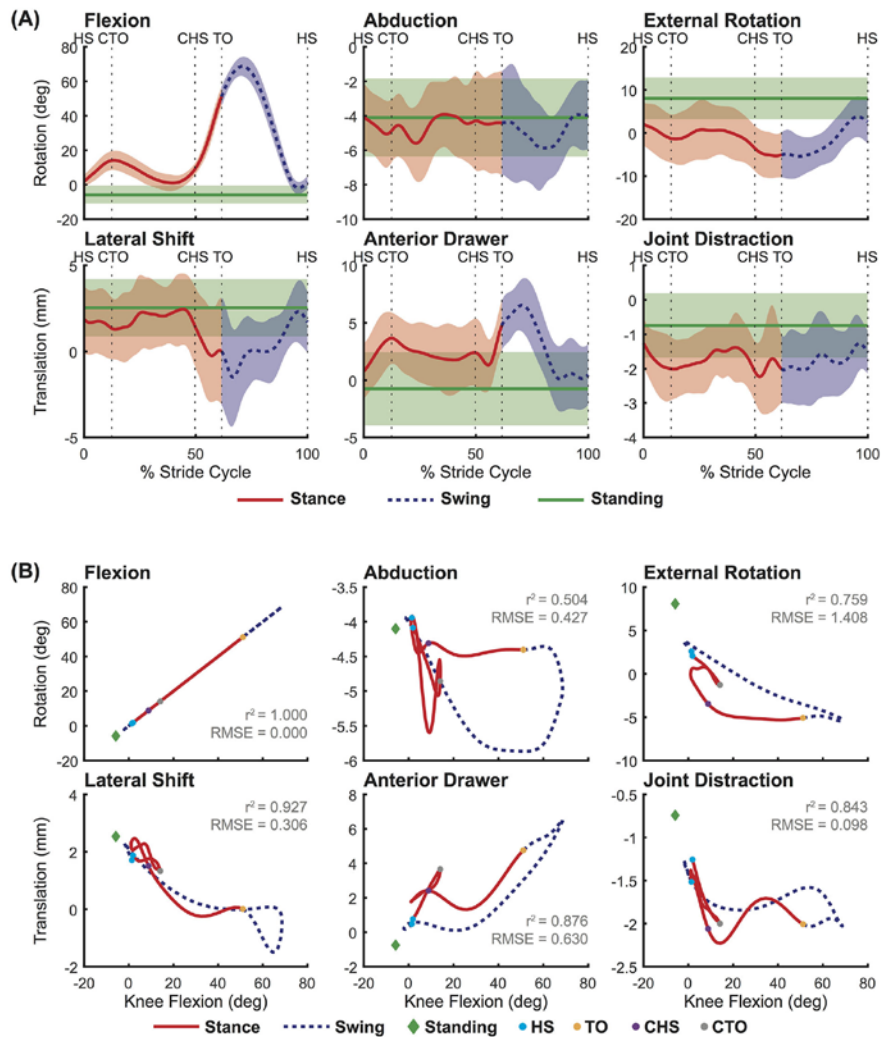


Figure 3. Tibiofemoral joint kinematics describing displacements of the tibia with respect to the femur for one cycle of overground walking (red and blue) and standing (green). (A) 6-DOF tibiofemoral kinematics plotted against percentage of the stride cycle; solid and dotted lines represent mean values calculated for 15 participants and the shaded area represents ± 1 standard deviation from the mean. (B) Mean 6-DOF tibiofemoral displacements plotted against the mean knee flexion angle for all 15 participants. The coefficients of determination (r^2) and root mean square error (RMSE) indicating degree of coupling with knee flexion angle are given. RMSE values are in degrees for data given in the top panel and mm for data in the bottom panel. HS, heel-strike; TO, toe-off; CHS, contralateral heel-strike; CTO, contralateral toe-off.

(Fig. 6A and B). Mean anterior-posterior displacement of the contact center in the lateral compartment was 15.4 mm compared to 9.7 mm in the medial compartment ($p = 0.002$, Table 2). Move-

ments of the contact centers in the medial and lateral compartments were coupled to the knee flexion angle ($r^2 > 0.85$), with posterior translations of the contact centers on the medial and lateral

Table 1. Peak-to-peak Displacement (PTPD) of the Mean 6-DOF Kinematic Parameters Measured for the Tibiofemoral Joint (TFJ) and Patellofemoral Joint (PFJ) for One Stride Cycle of Overground Walking

6-DOF TFJ Kinematics		6-DOF PFJ Kinematics	
Parameter	PTPD	Parameter	PTPD
Rotations (deg)			
Flexion	70.66	Flexion	50.49
Abduction	1.94	Lateral rotation	3.84
External rotation	9.23	Lateral tilt	4.46
Translations (mm)			
Lateral shift	3.96	Lateral shift	4.92
Anterior drawer	6.42	Anterior translation	11.33
Joint distraction	0.97	Superior translation	14.92

tibial plateaux associated with increasing knee flexion (Fig. 6C).

In contrast, the patterns of movement of the contact centers on the medial and lateral facets of the patella were similar (Fig. 7A and B). Both contact centers remained on the inferior half of the patella for most of the stance phase, before translating first superiorly and then inferiorly during swing. Superior-inferior translations of the patellar contact centers were larger than those observed in the medial-lateral direction ($p < 0.02$, Table 2), and were coupled to the knee flexion angle ($r^2 > 0.92$). Superior translations of the contact centers on the medial and lateral facets of the patella were related to increases in knee flexion (Fig. 7C).

The center of rotation of the tibiofemoral joint in the transverse plane was located at an average distance of 17.9 mm medial to the tibial origin over one stride cycle (Fig. 8). During stance the center of rotation was close to the geometric center of the tibial plateau and positioned 5.8 mm medially, whereas in swing it shifted further medially to be located 36.1 mm medial to the tibial origin.

Knee-Joint Configuration in Standing

The knee was more hyperextended and the tibia more externally rotated in standing than at any time during walking. Knee hyperextension in standing was 5.8° compared to a maximum hyperextension of 3.2° attained in walking ($p = 0.02$). The tibia was externally rotated by 8.1° in standing whereas maximum external rotation during walking was 5.0° ($p = 0.002$, Table 2). The tibiofemoral contact center was translated more anteriorly in the lateral compartment (9.3 mm) than in the medial compartment (3.1 mm) during standing ($p = 0.004$) (Fig. 6A and B; Table 2). Anterior translation of the lateral contact center was also greater during standing (9.3 mm) than at any time during walking (6.1 mm peak displacement, $p = 0.04$). The patella was displaced more laterally and anteriorly in standing compared to walking; however,

patellar tilt was not significantly different (Fig. 4A, Table 2).

DISCUSSION

We used a unique mobile biplane X-ray imaging system to obtain simultaneous measurements of 6-DOF kinematics of the healthy tibiofemoral and patellofemoral joints for one complete stride cycle when humans walk over ground at their natural speeds. We found that internal-external rotation, anterior-posterior translation, and medial-lateral shift of the tibia as well as flexion-extension, anterior-posterior translation and, to a lesser extent, medial-lateral tilt and medial-lateral shift of the patella are each coupled to the knee flexion angle. Anterior-posterior translations of the tibiofemoral contact centers were significantly greater in the lateral compartment than the medial compartment whereas no appreciable differences were observed in the movements of the contact centers on the medial and lateral facets of the patella. Posterior translations of the tibiofemoral contact centers and superior translations of the patellofemoral contact centers were associated with increasing knee flexion. The center of rotation of the knee in the transverse plane was located predominantly on the medial side throughout one stride cycle. These fundamental data provide an accurate description of the kinematic behavior of the entire knee-joint complex during normal walking and serve as a valuable resource for evaluating tibiofemoral and patellofemoral function in pathological gait. Accurate knowledge of 6-DOF knee kinematics is also essential for validating computational models of the knee joint. Measured 6-DOF rotations and translations of the tibia and patella relative to the femur are available in Supplementary_Material_Data.xlsx.

One potential limitation of the present study relates to the use of a mobile gantry mechanism which, by translating the imaging units alongside the participants, may have adversely affected their natural gait. To address this concern, we provided ample opportunity for each participant to practice walking with the MoBiX system in operation, first during an initial familiarization session and subsequently on the day of testing immediately prior to data collection. Yamooski and Banks²⁶ quantified the errors associated with the measurement of free-speed gait kinematics when humans are followed by a mobile imaging system. They found that a close-proximity robot motion tracking system increased stride length by just 9 mm and resulted in no statistically significant differences in hip- and knee-joint motion compared to normal gait. A second limitation is that the locations of the joint contact centers were calculated using the surface geometries of the bones reconstructed from a CT scan of each participant's knee. More accurate assessment of the joint contact center locations, particularly near full extension of the knee, may be derived using geometric models of the bones, cartilage, and

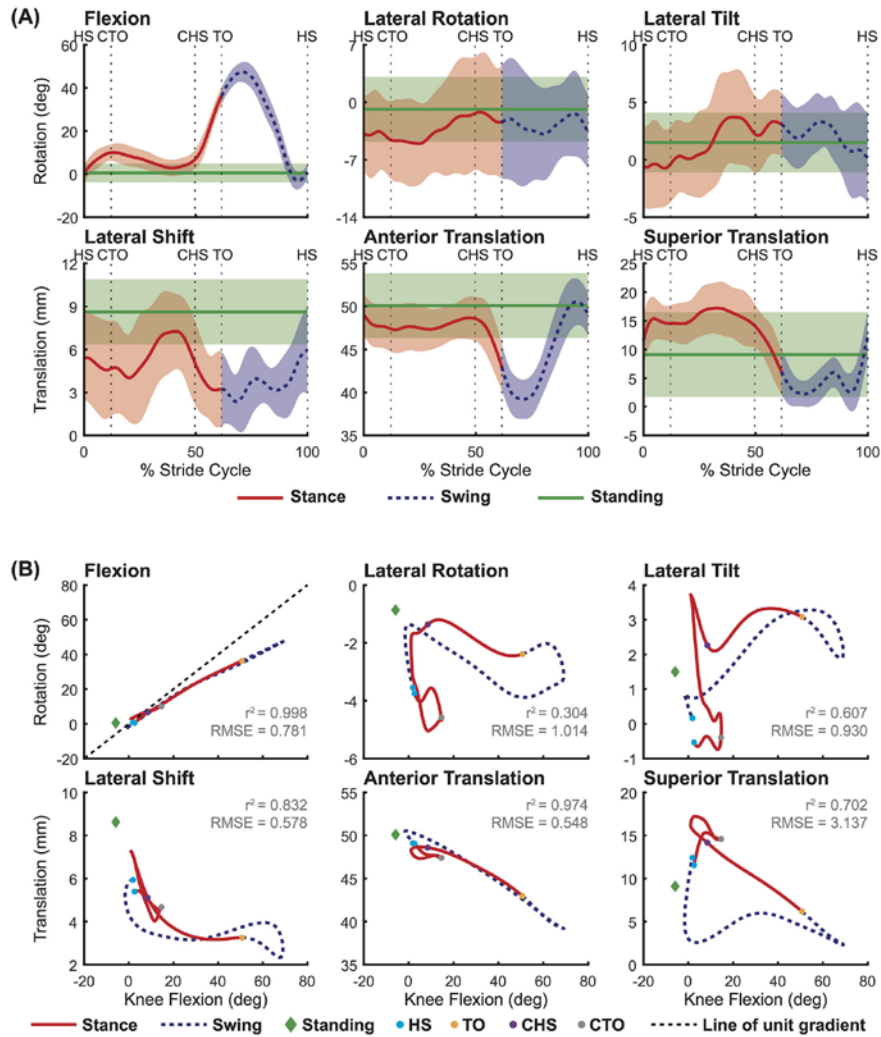


Figure 4. Patellofemoral joint kinematics describing displacements of the patella with respect to the femur for one cycle of overground walking (red and blue) and standing (green). (A) Six-DOF patellofemoral kinematics plotted against percentage of the stride cycle; solid and dotted lines represent mean values calculated for 14 participants and the shaded area represents ± 1 standard deviation from the mean. Patellofemoral motion was measured for only 14 participants because the patella moved outside the X-ray image capture volume for one individual. (B) Mean 6-DOF patellofemoral displacements plotted against the mean knee flexion angle for 14 participants. The black dotted line of unit gradient (top left-hand panel) represents the knee flexion angle. The coefficients of determination (r^2) and root mean square error (RMSE) indicating degree of coupling with knee flexion angle are given. RMSE values are in degrees for data given in the top panel and mm for data in the bottom panel. Positive lateral rotation represents the inferior patella moving laterally; positive lateral tilt represents movement of the lateral side of the patella toward the femur. HS, heel-strike; TO, toe-off; CHS, contralateral heel-strike; CTO, contralateral toe-off.

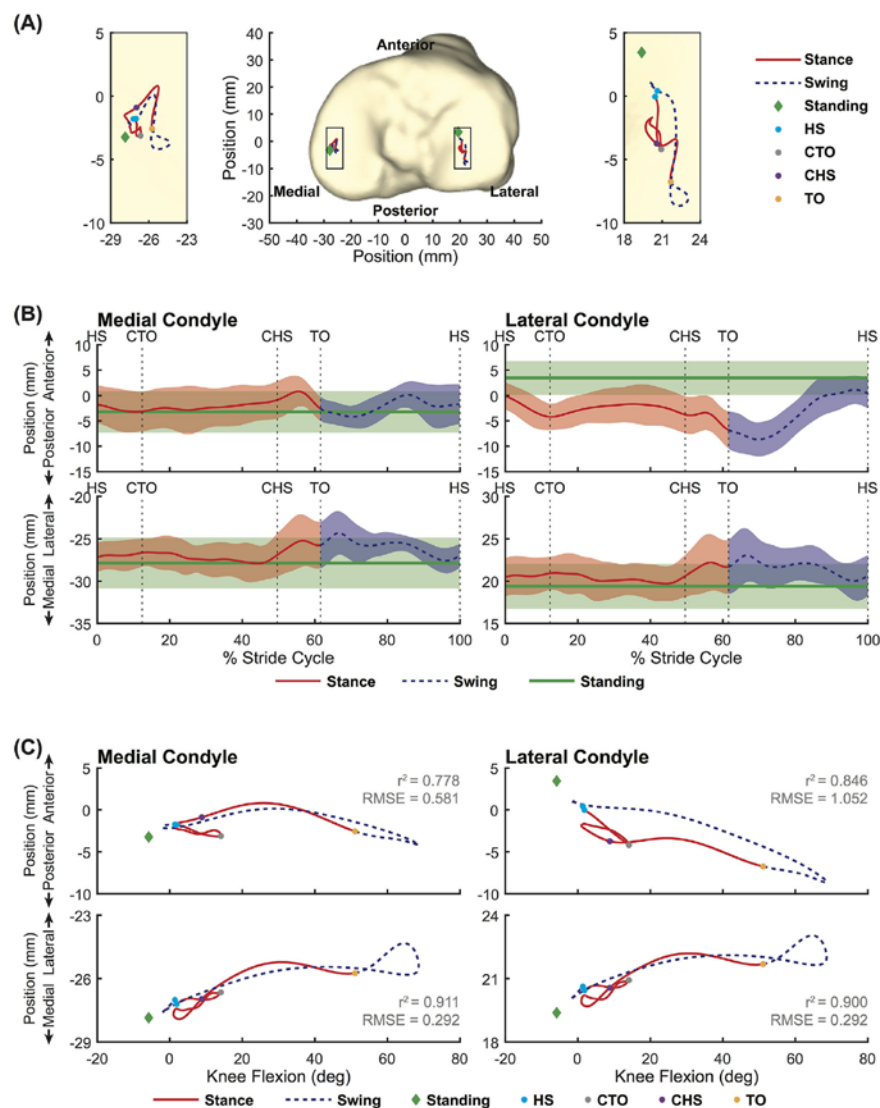


Figure 5. Locations of the centers of the femoral condyles for standing and for one complete cycle of overground walking. (A) Locations of the femoral condylar centers projected onto the plateau of the right tibia as viewed from above. The solid red lines represent the stance phase of walking while the dashed blue lines represent the swing phase. The green diamonds represent standing. (B) Locations of the femoral condylar centers plotted against the percentage of the stride cycle. The solid and dotted lines represent the mean for 15 subjects, and the shaded area represents ± 1 standard deviation from the mean. (C) Locations of the femoral condylar centers plotted against the mean knee flexion angle. The coefficients of determination (r^2) and root mean square error (RMSE in mm) indicating degree of coupling with knee flexion angle are given. HS, heel-strike; CTO, contralateral toe-off; CHS, contralateral heel-strike; TO, toe-off. Results shown in all panels are expressed in the tibial reference frame.

Table 2. Statistical Comparison of 6-DOF Kinematic Parameters, Condylar Motion, and Locations of the Contact Centers for the Tibiofemoral Joint (TFJ) and Patellofemoral Joint (PFJ) Measured for Walking and Standing

Joint	Parameter (Unit)	Dir.	Variable 1				Variable 2				p-Value
			Activity	Con.	Meas.	Value	Activity	Con.	Meas.	Value	
TFJ	Flexion (deg)	-	Stand	-	Mean	-5.81	Walk	-	Mean	21.49	27.30
	Abduction (deg)	-	Stand	-	Mean	-4.10	Walk	-	Mean	-4.68	-0.58
	External rotation (deg)	-	Stand	-	Mean	8.07	Walk	-	Mean	-1.37	-9.44
	Lateral shift (mm)	-	Stand	-	Mean	2.54	Walk	-	Mean	1.07	-1.46
	Anterior drawer (mm)	-	Stand	-	Mean	-0.75	Walk	-	Mean	2.58	3.34
	Joint distraction (mm)	-	Stand	-	Mean	-0.74	Walk	-	Mean	-1.72	-0.97
PFJ	Flexion (deg)	-	Stand	-	Mean	0.54	Walk	-	Mean	15.72	15.18
	Lateral rotation (deg)	-	Stand	-	Mean	-0.86	Walk	-	Mean	-3.06	-2.19
	Lateral tilt (deg)	-	Stand	-	Mean	1.50	Walk	-	Mean	1.71	0.21
	Lateral shift (mm)	-	Stand	-	Mean	8.62	Walk	-	Mean	4.52	-4.10
	Anterior translation (mm)	-	Stand	-	Mean	50.10	Walk	-	Mean	46.15	-3.95
	Superior translation (mm)	-	Stand	-	Mean	9.10	Walk	-	Mean	10.51	1.41
TFJ	Flexion (deg)	-	Stand	-	Mean	-5.81	Walk	-	Min	-3.23	2.58
	External rotation (deg)	-	Stand	-	Mean	8.07	Walk	-	Max	5.04	-3.03
TFJ	Condylar center (mm)	AP	Stand	M	Mean	-3.24	Stand	L	Mean	3.44	6.68
		AP	Stand	L	Mean	3.44	Walk	L	Max	2.04	1.40
		AP	Walk	M	PTPD	7.62	Walk	L	PTPD	11.57	3.95
		ML	Walk	M	PTPD	6.21	Walk	L	PTPD	6.23	0.02
TFJ	Contact center (mm)	SI	Walk	M	PTPD	3.62	Walk	L	PTPD	3.45	0.18
		AP	Stand	M	Mean	3.15	Stand	L	Mean	9.32	6.17
		AP	Stand	L	Mean	9.32	Walk	L	Max	6.19	-3.13
		AP	Walk	M	PTPD	9.69	Walk	L	PTPD	15.38	5.69
PFJ	Contact center (mm)	ML	Walk	M	PTPD	6.87	Walk	L	PTPD	7.67	0.81
		ML	Stand	L	Mean	5.96	Walk	L	Mean	12.08	-6.12
		ML	Walk	M	PTPD	13.21	Walk	L	PTPD	16.21	3.00
		SI	Walk	M	PTPD	21.50	Walk	L	PTPD	20.76	-0.73

Column headers: Dir., Direction; Diff., Difference; Con., Condyle; Meas., Measurement; CI, confidence interval. Direction column: AP, anterior-posterior; ML, medial-lateral; SI, superior-inferior. Condyle column: M, medial; L, lateral. Measurement column: Min, Max, Mean, and PTPD are the Minimum, Maximum, Mean, and Peak-To-Peak Displacements measured over the activity for each participant and averaged across all participants. *Indicates a significant difference ($p < 0.05$).

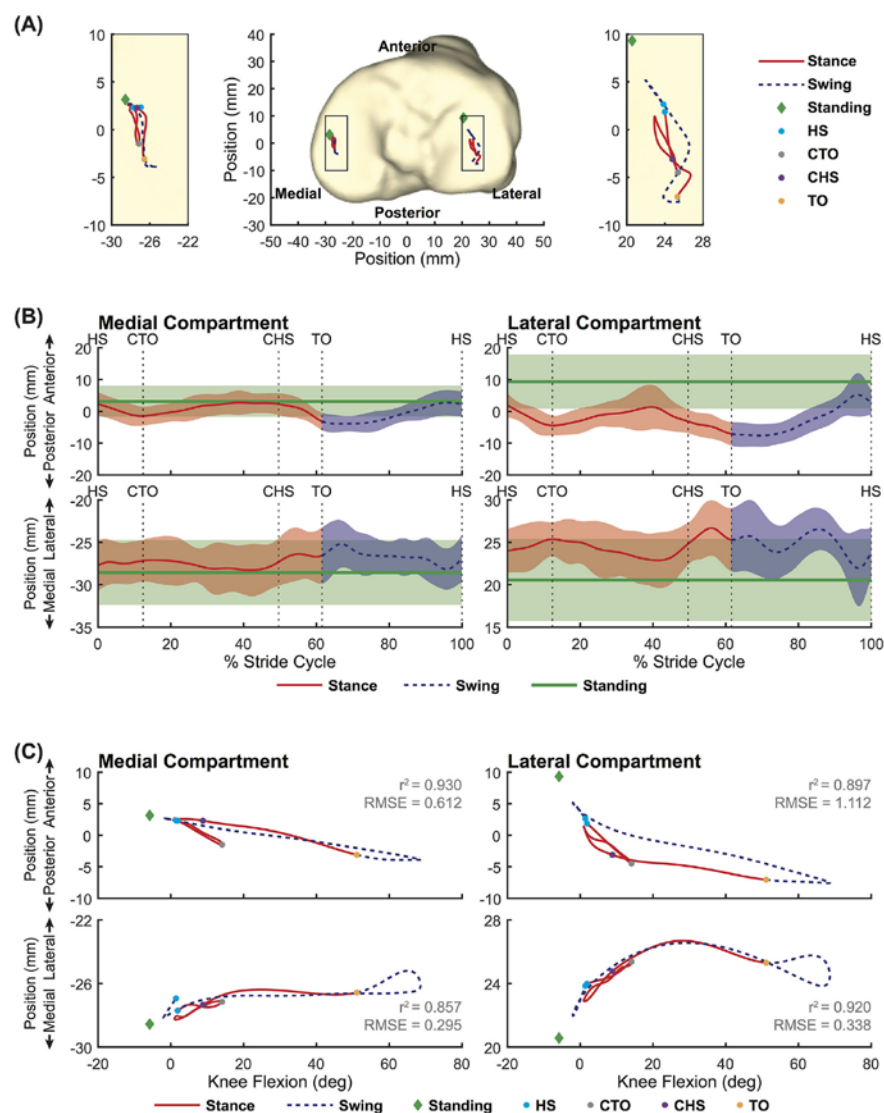


Figure 6. Locations of the tibiofemoral joint contact centers for standing and for one complete cycle of overground walking. (A) Locations of the tibiofemoral contact centers displayed on the right tibia as viewed from above. The solid red lines represent the stance phase of walking while the dashed blue lines represent the swing phase. The green diamonds represent standing. (B) Locations of the tibiofemoral contact centers plotted against the percentage of the stride cycle. The solid and dotted lines represent the mean for 15 subjects, and the shaded area represents ± 1 standard deviation from the mean. (C) Locations of the tibiofemoral contact centers plotted against the mean knee flexion angle. The coefficients of determination (r^2) and root mean square error (RMSE in mm) indicating degree of coupling with knee flexion angle are given. HS, heel-strike; CTO, contralateral toe-off; CHS, contralateral heel-strike; TO, toe-off.

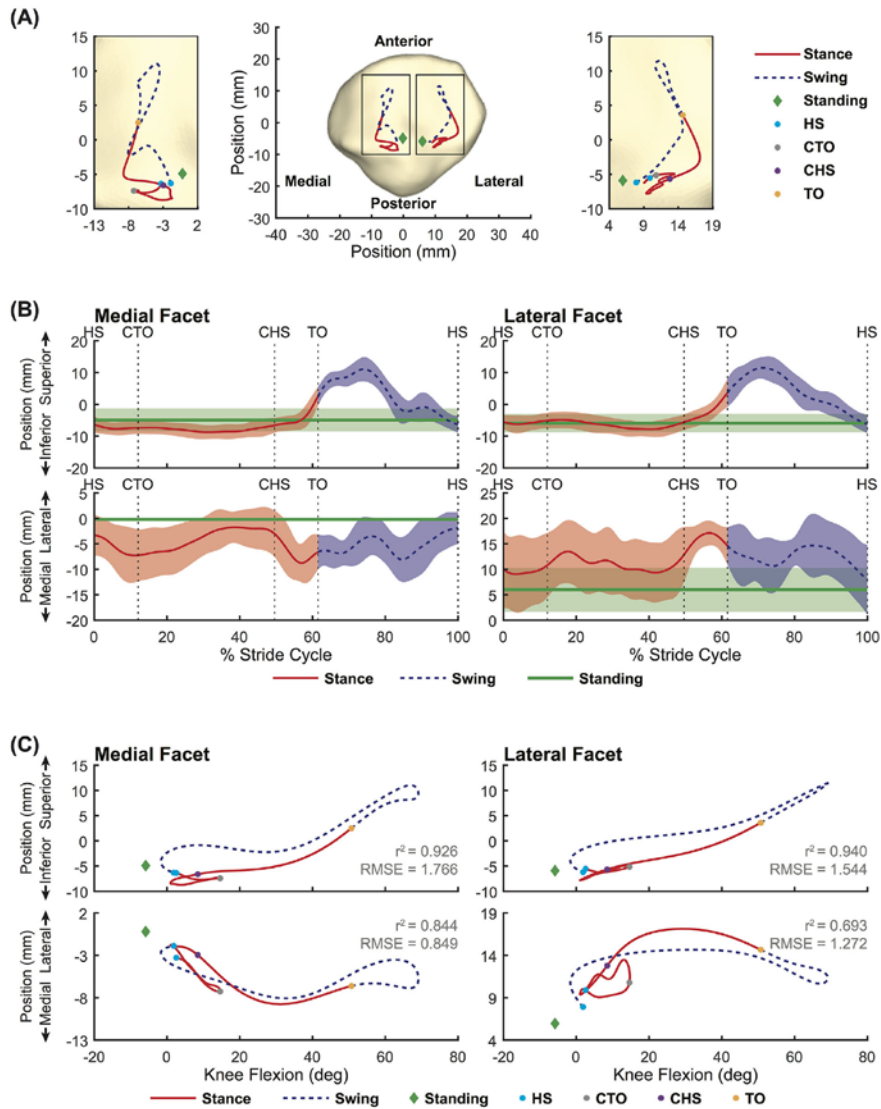


Figure 7. Locations of the patellofemoral joint contact centers for standing and for one complete cycle of overground walking. (A) Locations of the patellofemoral contact centers displayed on the posterior surface of the right patella as viewed from the posterior direction. The solid red lines represent the stance phase of walking while the dashed blue lines represent the swing phase. The green diamonds represent standing. (B) Locations of the patellofemoral contact centers plotted against the percentage of the stride cycle. The solid and dotted lines represent the mean for 14 subjects, and the shaded area represents ± 1 standard deviation from the mean. (C) Locations of the patellofemoral contact centers plotted against the mean knee flexion angle. The coefficients of determination (r^2) and root mean square error (RMSE in mm) indicating degree of coupling with knee flexion angle are given. HS, heel-strike; CTO, contralateral toe-off; CHS, contralateral heel-strike; TO, toe-off.

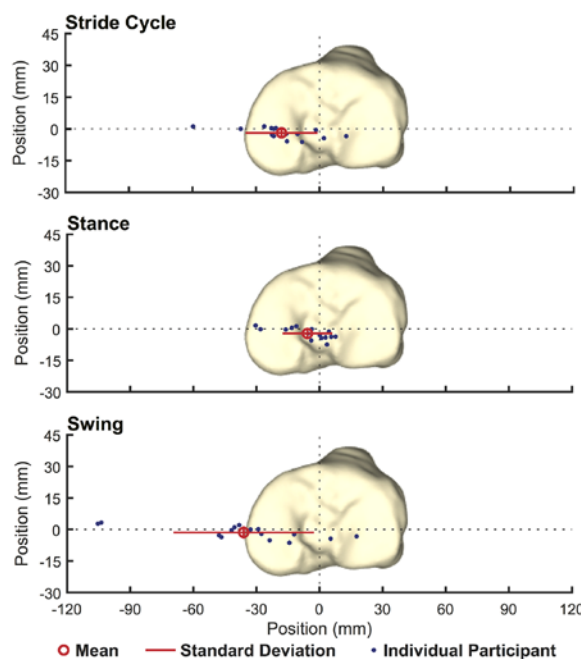


Figure 8. Location of the center of rotation in the transverse plane measured for one stride cycle of overground walking (top panel); for the stance phase only (middle panel); and for the swing phase only (bottom panel). The blue dots signify the center of rotation calculated for each participant while the red circle represents the mean center of rotation calculated for all 15 participants. The red bars represent ± 1 standard deviation from the mean.

menisci obtained from MRI.²⁹ In addition, the locations of the joint contact centers at the patellofemoral joint were calculated by assuming that contact occurred on both facets of the patella. While this is likely to be the case for most of the stride cycle when the patella is firmly entrenched in the trochlear groove, contact between the articular surfaces of the patella and femur may not be guaranteed when the knee is near full extension.

The only comparable datasets describing knee kinematics in healthy overground gait are those of LaFortune et al.² and Andriacchi et al.³ who used bone pins and skin markers, respectively, to measure 6-DOF tibiofemoral joint kinematics during overground walking. Our results for flexion-extension, internal-external rotation and medial-lateral shift are consistent with the corresponding patterns of movement noted by LaFortune et al.,² but there is less agreement in the remaining three degrees of freedom of tibiofemoral joint motion (Fig. 9). The amplitude of anterior-posterior translation measured in the present study is smaller and displays a noticeably different profile for most of the stride cycle. Our measurements of joint compression-distraction are also larger than those of LaFortune, especially during late stance and the first half of swing. These authors assigned 3D coordinate systems to the femur and tibia based on bony landmarks

identified from 2D X-ray images whereas we used more accurate 3D models segmented from CT scans. Intracortical pins also cause pain and impingement, which may have altered the gait patterns of the participants in LaFortune's study, plus this technique is prone to errors caused by vibration, bending, and loosening of the bone pins.⁷ There are notable differences also between our measurements and those of Andriacchi et al.,³ particularly with respect to internal-external rotation, abduction-adduction, anterior-posterior translation and medial-lateral shift (Fig. 9).

Coupled motions have been studied in the unloaded knee by measuring unconstrained passive kinematics in cadaver specimens.³⁰ Internal rotation, adduction, and all three components of tibial translation were found to be coupled to the flexion angle in both passive flexion and extension, implying that the knee behaves as a 1-DOF mechanism with the relative movements of the bones guided solely by ligament geometry and articular contact. Our data show that internal-external rotation, anterior-posterior translation, and medial-lateral shift of the tibia are each coupled to knee flexion-extension during walking, but no clear association was observed between the flexion angle and abduction-adduction (Fig. 3B). We also found that movements of the tibiofemoral contact centers in both the medial and lateral compartments are coupled to

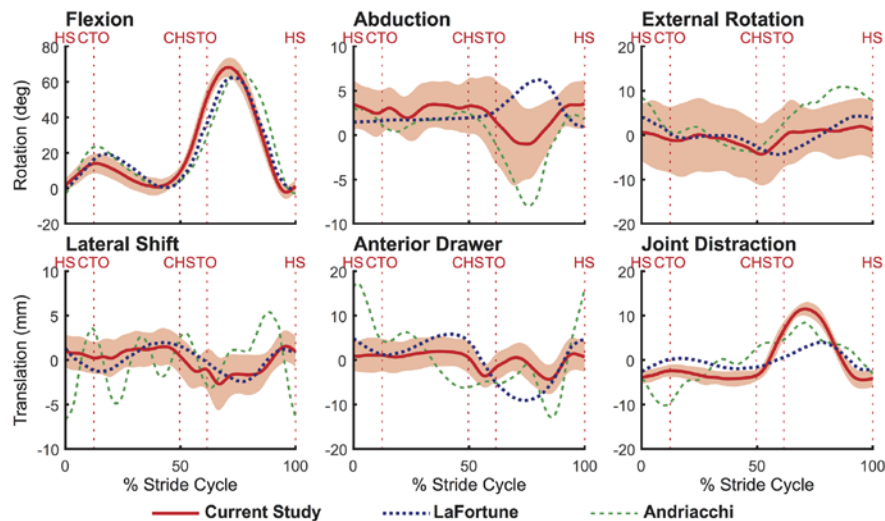


Figure 9. Six-DOF tibiofemoral joint kinematics measured in the present study (red) compared to results reported by LaFortune et al.² (blue) and Andriacchi et al.³ (green). To ensure a meaningful comparison between the three data sets, the results for all three studies are expressed using the bone coordinate systems described by LaFortune et al.² (see also Figs. 4 and 5 in Andriacchi et al.³). The kinematic profiles for lateral shift, anterior drawer, and joint distraction were offset so that the mean value of each trajectory over the stride cycle was zero. The solid and dotted lines represent the mean values for all subjects in each study while the shaded area represents ± 1 standard deviation for data recorded in the present study. HSC, heel-strike; TO, toe-off; CHS, contralateral heel-strike; CTO, contralateral toe-off.

the knee flexion angle, with posterior translations of the contact centers (i.e., rollback of the medial and lateral femoral condyles on the tibial plateau) associated with increasing knee flexion (Fig. 6C). These results are consistent with the hypothesis that the knee may be approximated as a 1-DOF mechanical linkage and suggest that, at least for normal walking, tibial internal-external rotation, anterior-posterior translation and medial-lateral shift depend more heavily upon the soft-tissue structures and bony anatomy of the knee than the external (muscle, gravitational and inertial) forces transmitted at the joint.

Previous investigators have shown that the tibia rotates internally as the knee is flexed, and that this rotation is reversed in the classic “screw-home” pattern during extension.³¹ In the present study, knee extension was associated with external rotation of the tibia, which was particularly evident as the knee approached full extension (Fig. 3B). Indeed, the tibia was externally rotated by as much as 8° while the knee was hyperextended in standing (Fig. 3B), with the tibiofemoral contact center translated 6 mm more anteriorly in the lateral compartment than the medial compartment (green diamonds in Fig. 6A). These results illustrate clearly the existence of a screw-home pattern of movement in the healthy joint.

The tibia was more anteriorly translated at contralateral toe-off compared to its configuration at the same flexion angle during swing (Fig. 3B). Movement of the tibia during early stance, and particularly near contralateral toe-off, may be explained by the action of the quadriceps muscles, which are shown to be active at this time.³² During early stance, when the knee is near extension, the line-of-action of the patellar tendon is such that it applies an anterior shear force to the tibia.³³ Thus, any force transmitted from the quadriceps to the patellar tendon will act to translate the tibia anteriorly at this time.

The quadriceps likely played a major role in stabilizing motion of the patella during terminal swing as the knee approached hyperextension (Fig. 4, Superior Translation). During the latter half of swing the patella descended inferiorly and the patellar tendon was then presumably slack. Activation of the quadriceps muscles prior to heel-strike³⁴ translated the patella rapidly in the superior direction, allowing the patellar tendon to become taut once more. As the knee flexed beyond 20°, anterior-posterior translation, medial-lateral tilt, and medial-lateral shift of the patella were each tightly coupled with the knee flexion angle (Fig. 4B), indicating that the patella was then firmly entrenched within the trochlear groove.

Our measurements suggest that the center of rotation of the knee in the transverse plane lies in the

medial compartment of the tibiofemoral joint during both stance and swing. Using video motion capture with skin markers, Koo and Andriacchi¹⁹ found that the center of rotation was located in the lateral compartment during stance. This discrepancy may be due to the different measurement methods adopted in these two studies. Using biplane X-ray imaging, Liu et al.³⁵ also found the center of rotation in the transverse plane to lie in the lateral compartment during stance. However, the subjects in their study walked on a treadmill at the relatively slow speed of 0.67 m/s. Guan et al.²⁵ found statistically significant differences between treadmill and overground walking with respect to 6-DOF tibiofemoral kinematics and the locations of the joint contact centers when subjects walked at the same speed under these two conditions. In addition, the walking speed of 0.67 m/s reported by Liu et al.³⁵ is approximately one-half of the preferred speed adopted by the subjects in the present study (1.28 m/s). Our findings concur with previous reports which show that the longitudinal axis of rotation passes through the medial tibial plateau, indicating that axial rotation of the knee occurs about the medial compartment of the tibiofemoral joint.^{15–17} Our results also show that anterior-posterior translation of the center of the lateral femoral condyle is significantly greater than that of the medial femoral condyle (Fig. 5A and B), consistent with the concept of a medially constrained total knee replacement design.¹⁸

AUTHORS' CONTRIBUTIONS

HAG, SG, AGS, RS, and MGP designed the study. MGP and RS obtained funding for the research. HAG, SG, and LTT performed the data collection and analysis. HAG, SG, and MGP interpreted the data and drafted the manuscript. All authors edited, revised, and approved the final version. MGP was the chief investigator for the study.

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REFERENCES

- Pandy MG, Andriacchi TP. 2010. Muscle and joint function in human locomotion. *Annu Rev Biomed Eng* 12:401–433.
- LaFortune MA, Cavanagh PR, Sommer HJ, et al. 1992. Three-dimensional kinematics of the human knee during walking. *J Biomech* 25:347–357.
- Andriacchi TP, Alexander EJ, Toney MKK, et al. 1998. A point cluster method for in vivo motion analysis: applied to a study of knee kinematics. *J Biomech Eng* 120:743–749.
- Akbarshahi M, Schache AG, Fernandez JW, et al. 2010. Non-invasive assessment of soft-tissue artifact and its effect on knee joint kinematics during functional activity. *J Biomech* 43:1292–1301.
- Reinschmidt C, Van Den Bogert AJ, Lundberg A, et al. 1997. Tibiofemoral and tibiofemoral motion during walking: external vs. skeletal markers. *Gait Posture* 6:98–109.
- Benoit DL, Ramsey DK, Lamontagne M, et al. 2007. In vivo knee kinematics during gait reveals new rotation profiles and smaller translations. *Clin Orthop Relat Res* 454:81–88.
- Ramsey DK, Wretenberg PF, Benoit DL, et al. 2003. Methodological concerns using intra-cortical pins to measure tibiofemoral kinematics. *Knee Surg Sport Traumatol Arthrosc* 11:344–349.
- Kozanek M, Hosseini A, Liu F, et al. 2009. Tibiofemoral kinematics and condylar motion during the stance phase of gait. *J Biomech* 42:1877–1884.
- Anderst W, Zauel R, Bishop J, et al. 2009. Validation of three-dimensional model-based tibio-femoral tracking during running. *Med Eng Phys* 31:10–16.
- Farrokhi S, Voycheck CA, Klatt BA, et al. 2014. Altered tibiofemoral joint contact mechanics and kinematics in patients with knee osteoarthritis and episodic complaints of joint instability. *Clin Biomech* 29:629–635.
- Kefala V, Cyr AJ, Harris MD, et al. 2017. Assessment of knee kinematics in older adults using high-speed stereo radiography. *Med Sci Sports Exerc* 49:2260–2267.
- Li C, Hosseini A, Tsai T-Y, et al. 2015. Articular contact kinematics of the knee before and after a cruciate retaining total knee arthroplasty. *J Orthop Res* 33:349–358.
- Komistek RD, Dennis DA, Mahfouz M. 2003. In vivo fluoroscopic analysis of the normal human knee. *Clin Orthop Relat Res* 410:69–81.
- Li G, DeFrate LE, Sang EP, et al. 2005. In vivo articular cartilage contact kinematics of the knee: an investigation using dual-orthogonal fluoroscopy and magnetic resonance image-based computer models. *Am J Sports Med* 33:102–107.
- Freeman MAR, Pinskerova V. 2005. The movement of the normal tibio-femoral joint. *J Biomech* 38:197–208.
- Hollister AM, Jatana S, Singh AK, et al. 1993. The axes of rotation of the knee. *Clin Orthop Relat Res* 290:259–268.
- Churchill DL, Incavo SJ, Johnson CC, et al. 1998. The transepicondylar axis approximates the optimal flexion axis of the knee. *Clin Orthop Relat Res* 356:111–118.
- Scott G, Imam MA, Eifert A, et al. 2016. Can a total knee arthroplasty be both rotationally unconstrained and anteroposteriorly stabilised? A pulsed fluoroscopic investigation. *Bone Jt Res* 5:80–86.
- Koo S, Andriacchi TP. 2008. The knee joint center of rotation is predominantly on the lateral side during normal walking. *J Biomech* 41:1269–1273.
- Guan S, Gray HA, Keynejad F, et al. 2016. Mobile biplane X-ray imaging system for measuring 3D dynamic joint motion during overground gait. *Med Imaging IEEE Trans* 35:326–336.
- Lai A, Lichtwark GA, Schache AG, et al. 2015. In-vivo behavior of the human soleus muscle with increasing walking and running speeds. *J Appl Physiol* 118:1266–1275.
- Gray HA, Guan S, Pandy MG. 2017. Accuracy of mobile biplane X-ray imaging in measuring 6-degree-of-freedom patellofemoral kinematics during overground gait. *J Biomech* 57:152–156.
- Pieper S, Lorensen B, Schroeder W, et al. 2006. The N-MIC Kit: ITK, VTK, pipelines, grids and 3D Slicer as an open platform for the medical image computing community. In: 3rd IEEE International Symposium on Biomedical Imaging: Macro to Nano, 2006. Arlington, VA: IEEE. p 698–701.
- Grood ES, Suntay WJ. 1983. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J Biomech Eng* 105:136–144.
- Guan S, Gray HA, Schache AG, et al. 2017. In vivo six-degree-of-freedom knee-joint kinematics in overground and treadmill walking following total knee arthroplasty. *J Orthop Res* 35:1634–1643.

630 GRAY ET AL.

26. Asano T, Akagi M, Tanaka K, et al. 2001. In vivo three-dimensional knee kinematics using a biplanar image-matching technique. *Clin Orthop Relat Res* 388:157–166.
27. Krevolin JL, Pandy MG, Pearce JC. 2004. Moment arm of the patellar tendon in the human knee. *J Biomech* 37:785–788.
28. Yamokoski JD, Banks SA. 2011. Does close proximity robot motion tracking alter gait? *Gait Posture* 34:508–513.
29. Defrate LE, Sun H, Gill TJ, et al. 2004. In vivo tibiofemoral contact analysis using 3D MRI-based knee models. *J Biomech* 37:1499–1504.
30. Wilson DR, Feikes JD, Zavatsky AB, et al. 2000. The components of passive knee movement are coupled to flexion angle. *J Biomech* 33:465–473.
31. Trent PS, Walker PS, Wolf B. 1976. Ligament length patterns, strength, and rotational axes of the knee joint. *Clin Orthop Relat Res* 117:263–270.
32. Sutherland DH. 2001. The evolution of clinical gait analysis part 1: kinesiological EMG. *Gait Posture* 14:61–70.
33. Shelburne KB, Pandy MG, Anderson FC, et al. 2004. Pattern of anterior cruciate ligament force in normal walking. *J Biomech* 37:797–805.
34. Cappellini G, Ivanenko YP, Poppele RE, et al. 2006. Motor patterns in human walking and running. *J Neurophysiol* 95:3426–3437.
35. Liu F, Kozanek M, Hosseini A, et al. 2010. In vivo tibiofemoral cartilage deformation during the stance phase of gait. *J Biomech* 43:658–665.
36. Rainbow MJ, Miranda DL, Cheung RTH, et al. 2013. Automatic determination of an anatomical coordinate system for a three-dimensional model of the human patella. *J Biomech* 46:2093–2096.

SUPPORTING INFORMATION

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References

- Abernethy, B., Kippers, V., Pandy, M.G., Hanrahan, S.J., 2013. Biophysical foundations of human movement. *Human Kinetics*.
- Ackermann, M., Van den Bogert, A.J., 2010. Optimality principles for model-based prediction of human gait. *Journal of biomechanics* 43, 1055-1060.
- Akbarshahi, M., Fernandez, J.W., Schache, A.G., Pandy, M.G., 2014. Subject-specific evaluation of patellofemoral joint biomechanics during functional activity. *Med Eng Phys* 36, 1122-1133.
- Akbarshahi, M., Schache, A.G., Fernandez, J.W., Baker, R., Banks, S., Pandy, M.G., 2010. Non-invasive assessment of soft-tissue artifact and its effect on knee joint kinematics during functional activity. *J Biomech* 43, 1292-1301.
- Akpınar, B., Thorhauer, E., Tashman, S., Irrgang, J.J., Fu, F.H., Anderst, W.J., 2019. Tibiofemoral Cartilage Contact Differences Between Level Walking and Downhill Running. *Orthop J Sports Med* 7, 2325967119836164.
- Ali, A.A., Harris, M.D., Shalhoub, S., Maletsky, L.P., Rullkoetter, P.J., Shelburne, K.B., 2017. Combined measurement and modeling of specimen-specific knee mechanics for healthy and ACL-deficient conditions. *J Biomech* 57, 117-124.
- Amis, A., Firer, P., Mountney, J., Senavongse, W., Thomas, N., 2003. Anatomy and biomechanics of the medial patellofemoral ligament. *The Knee* 10, 215-220.
- Amis, A.A., 2007. Current concepts on anatomy and biomechanics of patellar stability. *Sports Med Arthrosc Rev* 15, 48-56.
- Andriacchi, T.P., Alexander, E.J., 2000. Studies of human locomotion: past, present and future. *J Biomech* 33, 1217-1224.
- Andriacchi, T.P., Koo, S., Scanlan, S.F., 2009. Gait mechanics influence healthy cartilage morphology and osteoarthritis of the knee. *The Journal of Bone and Joint Surgery. American volume*. 91, 95.

- Argatov, I., 2013. Mathematical modeling of linear viscoelastic impact: Application to drop impact testing of articular cartilage. *Tribology International* 63, 213-225.
- Baldwin, J.L., House, C.K., 2005. Anatomic dimensions of the patella measured during total knee arthroplasty. *The Journal of arthroplasty* 20, 250-257.
- Baliunas, A., Hurwitz, D., Ryals, A., Karrar, A., Case, J., Block, J., Andriacchi, T., 2002. Increased knee joint loads during walking are present in subjects with knee osteoarthritis. *Osteoarthritis and cartilage* 10, 573-579.
- Baquinie, P., Brukner, P., 1997. Injuries presenting to an Australian sports medicine centre: a 12-month study. *Clin J Sport Med* 7, 28-31.
- Bingham, J., Papannagari, R., Van de Velde, S., Gross, C., Gill, T., Felson, D., Rubash, H., Li, G., 2008. In vivo cartilage contact deformation in the healthy human tibiofemoral joint. *Rheumatology* 47, 1622-1627.
- Boden, B.P., Pearsall, A.W., Garrett, W.E., Feagin, J.A., 1997. Patellofemoral instability: evaluation and management. *JAAOS-Journal of the American Academy of Orthopaedic Surgeons* 5, 47-57.
- Borotikar, B.S., Sipprell III, W.H., Wible, E.E., Sheehan, F.T., 2012. A methodology to accurately quantify patellofemoral cartilage contact kinematics by combining 3D image shape registration and cine-PC MRI velocity data. *Journal of biomechanics* 45, 1117-1122.
- Brandon, S.C.E., Thelen, D.G., Smith, C.R., Novacheck, T.F., Schwartz, M.H., Lenhart, R.L., 2018. The coupled effects of crouch gait and patella alta on tibiofemoral and patellofemoral cartilage loading in children. *Gait Posture* 60, 181-187.
- Brechter, J.H., Powers, C.M., 2002. Patellofemoral stress during walking in persons with and without patellofemoral pain. *Med Sci Sports Exerc* 34, 1582-1593.
- Buckwalter, J.A., Saltzman, C., Brown, T., 2004. The impact of osteoarthritis: implications for research. *Clinical orthopaedics and related research* 427, S6-S15.
- Buzug, T.M., 2008. Computed tomography. [electronic resource] : from photon statistics to modern cone-beam CT. Springer.

Cappozzo, A., Catani, F., Leardini, A., Benedetti, M.G., Croce, U.D., 1996. Position and orientation in space of bones during movement: experimental artefacts. *Clin Biomech (Bristol, Avon)* 11, 90-100.

Carter, D.R., Beaupre, G.S., Wong, M., Smith, R.L., Andriacchi, T.P., Schurman, D.J., 2004. The mechanobiology of articular cartilage development and degeneration. *Clin Orthop Relat Res*, S69-77.

Chaudhari, A.M., Briant, P.L., Bevell, S.L., Koo, S., Andriacchi, T.P., 2008. Knee kinematics, cartilage morphology, and osteoarthritis after ACL injury. *Med Sci Sports Exerc* 40, 215-222.

Chen, Y.-J., Scher, I., Powers, C.M., 2010. Quantification of Patellofemoral Joint Reaction Forces During Functional Activities Using a Subject-Specific Three-Dimensional Model. *Journal of Applied Biomechanics* 26, 415-423.

Cohen, Z.A., McCarthy, D.M., Kwak, S.D., Legrand, P., Fogarasi, F., Ciaccio, E.J., Ateshian, G.A., 1999. Knee cartilage topography, thickness, and contact areas from MRI: in-vitro calibration and in-vivo measurements. *Osteoarthritis Cartilage* 7, 95-109.

Coleman, J.L., Widmyer, M.R., Leddy, H.A., Utturkar, G.M., Spritzer, C.E., Moorman, C.T., 3rd, Guilak, F., DeFrate, L.E., 2013. Diurnal variations in articular cartilage thickness and strain in the human knee. *J Biomech* 46, 541-547.

Correa, T.A., Crossley, K.M., Kim, H.J., Pandy, M.G., 2010. Contributions of individual muscles to hip joint contact force in normal walking. *Journal of biomechanics* 43, 1618-1622.

Davies, A.P., Vince, A.S., Shepstone, L., Donell, S.T., Glasgow, M.M., 2002. The radiologic prevalence of patellofemoral osteoarthritis. *Clin Orthop Relat Res*, 206-212.

Defrate, L.E., Nha, K.W., Papannagari, R., Moses, J.M., Gill, T.J., Li, G., 2007. The biomechanical function of the patellar tendon during in-vivo weight-bearing flexion. *J Biomech* 40, 1716-1722.

DeFrate, L.E., Sun, H., Gill, T.J., Rubash, H.E., Li, G., 2004. In vivo tibiofemoral contact analysis using 3D MRI-based knee models. *Journal of biomechanics* 37, 1499-1504.

- Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E., Thelen, D.G., 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE Trans Biomed Eng* 54, 1940-1950.
- DesJardins, J.D., Banks, S.A., Benson, L.C., Pace, T., LaBerge, M., 2007. A direct comparison of patient and force-controlled simulator total knee replacement kinematics. *Journal of biomechanics* 40, 3458-3466.
- Dorn, T.W., 2011. Computational modelling of lower-limb muscle function in human running. University of Melbourne.
- Drake, R.L., 2015. Gray's atlas of anatomy, Second edition. ed. Churchill Livingstone.
- Elahi, S., Cahue, S., Felson, D.T., Engelman, L., Sharma, L., 2000. The association between varus–valgus alignment and patellofemoral osteoarthritis. *Arthritis & Rheumatism: Official Journal of the American College of Rheumatology* 43, 1874-1880.
- Elias, J.J., Wilson, D.R., Adamson, R., Cosgarea, A.J., 2004. Evaluation of a computational model used to predict the patellofemoral contact pressure distribution. *Journal of biomechanics* 37, 295-302.
- Ellis, H., 2002. *Clinical Anatomy : A Revision and Applied Anatomy for Clinical Students*. John Wiley and Sons, Inc, Oxford, U.K.
- Farrokhi, S., Meholic, B., Chuang, W.N., Gustafson, J.A., Fitzgerald, G.K., Tashman, S., 2015. Altered frontal and transverse plane tibiofemoral kinematics and patellofemoral malalignments during downhill gait in patients with mixed knee osteoarthritis. *J Biomech* 48, 1707-1712.
- Fernandez, J.W., Akbarshahi, M., Kim, H.J., Pandy, M.G., 2008. Integrating modelling, motion capture and x-ray fluoroscopy to investigate patellofemoral function during dynamic activity. *Comput Methods Biomech Biomed Engin* 11, 41-53.
- Fregly, B.J., Besier, T.F., Lloyd, D.G., Delp, S.L., Banks, S.A., Pandy, M.G., D'Lima, D.D., 2012. Grand challenge competition to predict in vivo knee loads. *J Orthop Res* 30, 503-513.

Giphart, J.E., Zirker, C.A., Myers, C.A., Pennington, W.W., LaPrade, R.F., 2012. Accuracy of a contour-based biplane fluoroscopy technique for tracking knee joint kinematics of different speeds. *Journal of biomechanics* 45, 2935-2938.

Goh, J., Lee, P., Bose, K., 1995. A cadaver study of the function of the oblique part of vastus medialis. *The Journal of bone and joint surgery. British volume* 77, 225-231.

Gollehon, D.L., Torzilli, P., Warren, R., 1987. The role of the posterolateral and cruciate ligaments in the stability of the human knee. A biomechanical study. *The Journal of bone and joint surgery. American volume* 69, 233-242.

Gray, H.A., Guan, S., Pandy, M.G., 2017. Accuracy of mobile biplane X-ray imaging in measuring 6-degree-of-freedom patellofemoral kinematics during overground gait. *J Biomech* 57, 152-156.

Gray, H.A., Guan, S., Thomeer, L.T., Schache, A.G., de Steiger, R., Pandy, M.G., 2019. Three-dimensional motion of the knee-joint complex during normal walking revealed by mobile biplane x-ray imaging. *J Orthop Res* 37, 615-630.

Grood, E.S., Suntay, W.J., 1983. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J Biomech Eng* 105, 136-144.

Guan, S., Gray, H.A., Keynejad, F., Pandy, M.G., 2016a. Mobile Biplane X-Ray Imaging System for Measuring 3D Dynamic Joint Motion During Overground Gait. *IEEE Trans Med Imaging* 35, 326-336.

Guan, S., Gray, H.A., Schache, A.G., Feller, J., de Steiger, R., Pandy, M.G., 2016b. In vivo six - degree - of - freedom knee - joint kinematics in overground and treadmill walking following total knee arthroplasty. *Journal of Orthopaedic Research*.

Halonen, K.S., Mononen, M.E., Jurvelin, J.S., Toyras, J., Klodowski, A., Kulmala, J.P., Korhonen, R.K., 2016. Importance of Patella, Quadriceps Forces, and Depthwise Cartilage Structure on Knee Joint Motion and Cartilage Response During Gait. *J Biomech Eng* 138.

Hashemi, J., Chandrashekar, N., Gill, B., Beynnon, B.D., Slauterbeck, J.R., Schutt Jr, R.C., Mansouri, H., Dabezies, E., 2008. The geometry of the tibial plateau and its

influence on the biomechanics of the tibiofemoral joint. *The Journal of Bone and Joint Surgery. American volume.* 90, 2724.

Hirokawa, S., 1991. Three-dimensional mathematical model analysis of the patellofemoral joint. *Journal of biomechanics* 24, 659-671.

Hoff, W.A., Komistek, R.D., Dennis, D.A., Gabriel, S.M., Walker, S.A., 1998. Three-dimensional determination of femoral-tibial contact positions under in vivo conditions using fluoroscopy. *Clinical Biomechanics* 13, 455-472.

Hsieh, J., 2015. *Computed tomography : principles, design, artifacts, and recent advances*, Third edition. ed. SPIE.

Hsieh, Y.F., Draganich, L.F., Ho, S.H., Reider, B., 1998. The effects of removal and reconstruction of the anterior cruciate ligament on patellofemoral kinematics. *Am J Sports Med* 26, 201-209.

Hu, J., Chen, Z., Xin, H., Zhang, Q., Jin, Z., 2018. Musculoskeletal multibody dynamics simulation of the contact mechanics and kinematics of a natural knee joint during a walking cycle. *Proc Inst Mech Eng H* 232, 508-519.

Huberti, H., Hayes, W., 1984. Patellofemoral contact pressures. The influence of q-angle and tendofemoral contact. *The Journal of bone and joint surgery. American volume* 66, 715-724.

Hudelmaier, M., Glaser, C., Englmeier, K.H., Reiser, M., Putz, R., Eckstein, F., 2003. Correlation of knee - joint cartilage morphology with muscle cross - sectional areas vs. anthropometric variables. *The Anatomical Record Part A: Discoveries in Molecular, Cellular, and Evolutionary Biology: An Official Publication of the American Association of Anatomists* 270, 175-184.

Hung, C., Ahmed, M.S., 2017. Knee Osteoarthritis, in: Kahn, S.B., Xu, R.Y. (Eds.), *Musculoskeletal Sports and Spine Disorders: A Comprehensive Guide*. Springer International Publishing, Cham, pp. 251-255.

Im, H.S., Goltzer, O., Sheehan, F.T., 2015. The effective quadriceps and patellar tendon moment arms relative to the tibiofemoral finite helical axis. *Journal of biomechanics* 48, 3737-3742.

Jansen, K., De Groote, F., Duysens, J., Jonkers, I., 2014. How gravity and muscle action control mediolateral center of mass excursion during slow walking: A simulation study. *Gait & Posture* 39, 91-97.

Jones, G., Ding, C., Glisson, M., Hynes, K., Ma, D., Cicuttini, F., 2003. Knee articular cartilage development in children: a longitudinal study of the effect of sex, growth, body composition, and physical activity. *Pediatric research* 54, 230-236.

Kanosue, K., Ogawa, T., Fukano, M., Fukubayashi, T., 2015. *Sports injuries and prevention*. Springer.

Kaplan, M.L., Heegaard, J.H., 2001. Predictive algorithms for neuromuscular control of human locomotion. *Journal of Biomechanics* 34, 1077-1083.

Kapoor, M., Mahomed, N.N., 2015. *Osteoarthritis : pathogenesis, diagnosis, available treatments, drug safety, regenerative and precision medicine*. Adis.

Kelly, M., 2017. *An Introduction to Trajectory Optimization: How to Do Your Own Direct Collocation*.

Klemetti, R., Steele, K.M., Moilanen, P., Avela, J., Timonen, J., 2014. Contributions of individual muscles to the sagittal- and frontal-plane angular accelerations of the trunk in walking. *Journal of Biomechanics* 47, 2263-2268.

Kobayashi, K., Hosseini, A., Sakamoto, M., Qi, W., Rubash, H.E., Li, G., 2013. In vivo kinematics of the extensor mechanism of the knee during deep flexion. *Journal of biomechanical engineering* 135.

Koh, T.J., Grabiner, M.D., De Swart, R.J., 1992. In vivo tracking of the human patella. *J Biomech* 25, 637-643.

Komistek, R.D., Dennis, D.A., Mabe, J.A., Walker, S.A., 2000. An in vivo determination of patellofemoral contact positions. *Clin Biomech (Bristol, Avon)* 15, 29-36.

- Komistek, R.D., Dennis, D.A., Mahfouz, M., 2003. In vivo fluoroscopic analysis of the normal human knee. *Clin Orthop Relat Res*, 69-81.
- Komistek, R.D., Stiehl, J.B., Dennis, D.A., Paxson, R.D., Soutas-Little, R.W., 1998. Mathematical model of the lower extremity joint reaction forces using Kane's method of dynamics. *Journal of Biomechanics* 31, 185-189.
- Koo, S., Andriacchi, T.P., 2007. A comparison of the influence of global functional loads vs. local contact anatomy on articular cartilage thickness at the knee. *J Biomech* 40, 2961-2966.
- Koo, S., Gold, G.E., Andriacchi, T.P., 2005. Considerations in measuring cartilage thickness using MRI: factors influencing reproducibility and accuracy. *Osteoarthritis Cartilage* 13, 782-789.
- Koo, S., Rylander, J.H., Andriacchi, T.P., 2011. Knee joint kinematics during walking influences the spatial cartilage thickness distribution in the knee. *Journal of biomechanics* 44, 1405-1409.
- Kujala, U.M., Jaakkola, L.H., Koskinen, S.K., Taimela, S., Hurme, M., Nelimarkka, O., 1993. Scoring of patellofemoral disorders. *Arthroscopy* 9, 159-163.
- Kumagai, M., Mizuno, Y., Mattessich, S.M., Elias, J.J., Cosgarea, A.J., Chao, E.Y., 2002. Posterior cruciate ligament rupture alters in vitro knee kinematics. *Clin Orthop Relat Res*, 241-248.
- Lafortune, M.A., 1984. The use of intra-cortical pins to measure the motion of the knee joint during walking. Ph. D. thesis, Pennsylvania State University.
- Lafortune, M.A., Cavanagh, P.R., Sommer, H.J., 3rd, Kalenak, A., 1992. Three-dimensional kinematics of the human knee during walking. *J Biomech* 25, 347-357.
- Lawrence, R.C., Helmick, C.G., Arnett, F.C., Deyo, R.A., Felson, D.T., Giannini, E.H., Heyse, S.P., Hirsch, R., Hochberg, M.C., Hunder, G.G., Liang, M.H., Pillemer, S.R., Steen, V.D., Wolfe, F., 1998. Estimates of the prevalence of arthritis and selected musculoskeletal disorders in the United States. *Arthritis Rheum* 41, 778-799.

Lee, T.Q., Morris, G., Csintalan, R.P., 2003. The influence of tibial and femoral rotation on patellofemoral contact area and pressure. *Journal of Orthopaedic & Sports Physical Therapy* 33, 686-693.

Lenhart, R.L., Kaiser, J., Smith, C.R., Thelen, D.G., 2015. Prediction and Validation of Load-Dependent Behavior of the Tibiofemoral and Patellofemoral Joints During Movement. *Ann Biomed Eng* 43, 2675-2685.

Li, G., DeFrate, L.E., Zayontz, S., Park, S.E., Gill, T.J., 2004. The effect of tibiofemoral joint kinematics on patellofemoral contact pressures under simulated muscle loads. *J Orthop Res* 22, 801-806.

Li, G., Papannagari, R., Nha, K.W., DeFrate, L.E., Gill, T.J., Rubash, H.E., 2007. The coupled motion of the femur and patella during in vivo weightbearing knee flexion. *Journal of Biomechanical Engineering-Transactions of the Asme* 129, 937-943.

Li, G., Park, S.E., DeFrate, L.E., Schutzer, M.E., Ji, L., Gill, T.J., Rubash, H.E., 2005. The cartilage thickness distribution in the tibiofemoral joint and its correlation with cartilage-to-cartilage contact. *Clinical biomechanics* 20, 736-744.

Lin, Y.C., Pandy, M.G., 2017. Three-dimensional data-tracking dynamic optimization simulations of human locomotion generated by direct collocation. *J Biomech* 59, 1-8.

Lin, Y.C., Walter, J.P., Pandy, M.G., 2018. Predictive Simulations of Neuromuscular Coordination and Joint-Contact Loading in Human Gait. *Ann Biomed Eng*.

Lin, Z., Tang, Y., Tan, H., Cai, D., 2019. Patellofemoral kinematic characteristics in anterior cruciate ligament deficiency and reconstruction. *BMC Musculoskelet Disord* 20, 82.

List, R., Postolka, B., Schütz, P., Hitz, M., Schwilch, P., Gerber, H., Ferguson, S.J., Taylor, W.R., 2017. A moving fluoroscope to capture tibiofemoral kinematics during complete cycles of free level and downhill walking as well as stair descent. *PloS one* 12, e0185952.

- Liu, F., Kozanek, M., Hosseini, A., Van de Velde, S.K., Gill, T.J., Rubash, H.E., Li, G., 2010. In vivo tibiofemoral cartilage deformation during the stance phase of gait. *Journal of biomechanics* 43, 658-665.
- Lohmander, L., Östenberg, A., Englund, M., Roos, H., 2004. High prevalence of knee osteoarthritis, pain, and functional limitations in female soccer players twelve years after anterior cruciate ligament injury. *Arthritis & Rheumatism: Official Journal of the American College of Rheumatology* 50, 3145-3152.
- Maetzel, A., Makela, M., Hawker, G., Bombardier, C., 1997. Osteoarthritis of the hip and knee and mechanical occupational exposure--a systematic overview of the evidence. *J Rheumatol* 24, 1599-1607.
- Mall, N.A., Chalmers, P.N., Moric, M., Tanaka, M.J., Cole, B.J., Bach Jr, B.R., Paletta Jr, G.A., 2014. Incidence and trends of anterior cruciate ligament reconstruction in the United States. *The American journal of sports medicine* 42, 2363-2370.
- Martin, R.B., Burr, D.B., Sharkey, N.A., Fyhrie, D.P., 2015. *Skeletal tissue mechanics*, Second edition. ed. Springer.
- Mason, J.J., Leszko, F., Johnson, T., Komistek, R.D., 2008. Patellofemoral joint forces. *J Biomech* 41, 2337-2348.
- Millard, M., Uchida, T., Seth, A., Delp, S.L., 2013. Flexing computational muscle: modeling and simulation of musculotendon dynamics. *J Biomech Eng* 135, 021005.
- Mizuno, Y., Kumagai, M., Mattessich, S., Elias, J., Ramrattan, N., Cosgarea, A., 2001. Q-angle influences tibiofemoral and patellofemoral kinematics. *J Orthop Res* 19.
- Moissenet, F., Chèze, L., Dumas, R., 2017. Individual muscle contributions to ground reaction and to joint contact, ligament and bone forces during normal gait. *Multibody System Dynamics* 40, 193-211.
- Moro - oka, T.a., Hamai, S., Miura, H., Shimoto, T., Higaki, H., Fregly, B.J., Iwamoto, Y., Banks, S.A., 2008. Dynamic activity dependence of in vivo normal knee kinematics. *Journal of Orthopaedic Research* 26, 428-434.

Neptune, R.R., McGowan, C.P., 2016. Muscle contributions to frontal plane angular momentum during walking. *Journal of Biomechanics* 49, 2975-2981.

Nha, K.W., Papannagari, R., Gill, T.J., Van de Velde, S.K., Freiberg, A.A., Rubash, H.E., Li, G., 2008. In vivo patellar tracking: clinical motions and patellofemoral indices. *J Orthop Res* 26, 1067-1074.

O'Brien, T.D., Reeves, N.D., Baltzopoulos, V., Jones, D.A., Maganaris, C.N., 2010. Muscle-tendon structure and dimensions in adults and children. *Journal of anatomy* 216, 631-642.

Pal, S., Draper, C.E., Fredericson, M., Gold, G.E., Delp, S.L., Beaupre, G.S., Besier, T.F., 2011. Patellar maltracking correlates with vastus medialis activation delay in patellofemoral pain patients. *Am J Sports Med* 39, 590-598.

Palastanga, N., Field, D., Soames, R., 2002. *Anatomy and Human Movement : Structure and Function*. Elsevier Health Sciences, Oxford.

Pandy, M.G., Lin, Y.-C., Kim, H.J., 2010. Muscle coordination of mediolateral balance in normal walking. *Journal of biomechanics* 43, 2055-2064.

Paschos, N.K., 2017. Anterior cruciate ligament reconstruction and knee osteoarthritis. *World journal of orthopedics* 8, 212.

Pieper, S., Lorensen, B., Schroeder, W., Kikinis, R., Year The NA-MIC Kit: ITK, VTK, pipelines, grids and 3D slicer as an open platform for the medical image computing community. In *Biomedical Imaging: Nano to Macro*, 2006. 3rd IEEE International Symposium on.

Porsa, S., Lin, Y.C., Pandy, M.G., 2016. Direct Methods for Predicting Movement Biomechanics Based Upon Optimal Control Theory with Implementation in OpenSim. *Ann Biomed Eng* 44, 2542-2557.

Prasad, P.V., 2006. *Magnetic resonance imaging. [electronic resource] : methods and biologic applications*. Humana Press.

Prodromos, C.C., 2018. *The anterior cruciate ligament : reconstruction and basic science*, Second edition. ed. Elsevier.

- Ramsey, D., Wretenberg, P., Benoit, D., Lamontagne, M., Nemeth, G., 2003. Methodological concerns using intra-cortical pins to measure tibiofemoral kinematics. *Knee Surgery, Sports Traumatology, Arthroscopy* 11, 344-349.
- Robling, A.G., 2009. Is bone's response to mechanical signals dominated by muscle forces? *Medicine and science in sports and exercise* 41, 2044.
- Rothermich, M.A., Glaviano, N.R., Li, J., Hart, J.M., 2015. Patellofemoral pain: epidemiology, pathophysiology, and treatment options. *Clin Sports Med* 34, 313-327.
- Scanlan, S.F., Favre, J., Andriacchi, T.P., 2013. The relationship between peak knee extension at heel-strike of walking and the location of thickest femoral cartilage in ACL reconstructed and healthy contralateral knees. *Journal of biomechanics* 46, 849-854.
- Schipplein, O., Andriacchi, T., 1991. Interaction between active and passive knee stabilizers during level walking. *Journal of Orthopaedic Research* 9, 113-119.
- Seeram, E., 2019. *Digital Radiography : physical principles and quality control*, Second edition. ed. Springer.
- Seisler, A.R., Sheehan, F.T., 2007. Normative three-dimensional patellofemoral and tibiofemoral kinematics: a dynamic, in vivo study. *IEEE Trans Biomed Eng* 54, 1333-1341.
- Sharma, A., Grieco, T.F., Zingde, S.M., Dennis, D.A., Anderle, M.R., Komistek, R.D., 2017. In Vivo Three-Dimensional Patellar Mechanics: Normal Knees Compared with Domed and Anatomic Patellar Components. *J Bone Joint Surg Am* 99, e18.
- Sharma, L., Song, J., Felson, D.T., Cahue, S., Shamiyeh, E., Dunlop, D.D., 2001. The role of knee alignment in disease progression and functional decline in knee osteoarthritis. *JAMA* 286, 188-195.
- Sheehan, F.T., Zajac, F.E., Drace, J.E., 1997. Using cine phase contrast magnetic resonance imaging to non-invasively study in vivo knee dynamics. *Journal of biomechanics* 31, 21-26.
- Shelburne, K.B., Pandy, M.G., Anderson, F.C., Torry, M.R., 2004. Pattern of anterior cruciate ligament force in normal walking. *J Biomech* 37, 797-805.

Shelburne, K.B., Torry, M.R., Pandy, M.G., 2005. Muscle, ligament, and joint-contact forces at the knee during walking. *Med Sci Sports Exerc* 37, 1948-1956.

Shelburne, K.B., Torry, M.R., Pandy, M.G., 2006. Contributions of muscles, ligaments, and the ground-reaction force to tibiofemoral joint loading during normal gait. *Journal of Orthopaedic Research* 24, 1983-1990.

Smith, T.O., Hunt, N.J., Donell, S.T., 2008. The reliability and validity of the Q-angle: a systematic review. *Knee Surgery, Sports Traumatology, Arthroscopy* 16, 1068-1079.

Sritharan, P., Lin, Y.C., Pandy, M.G., 2012. Muscles that do not cross the knee contribute to the knee adduction moment and tibiofemoral compartment loading during gait. *Journal of Orthopaedic Research* 30, 1586-1595.

Standring, S., 2016. *Gray's anatomy : the anatomical basis of clinical practice*, Forty-first edition. ed. Elsevier Limited.

Stein, L., Endicott, A., Sampalis, J., Kaplow, M., Patel, M., Mitchell, N., 1993. Motion of the patella during walking: a video digital-fluoroscopic study in healthy volunteers. *AJR. American journal of roentgenology* 161, 617-620.

Stelzer, M., Von Stryk, O., 2006. Efficient forward dynamics simulation and optimization of human body dynamics. *ZAMM - Journal of Applied Mathematics and Mechanics/Zeitschrift für Angewandte Mathematik und Mechanik: Applied Mathematics and Mechanics* 86, 828-840.

Suzuki, T., Hosseini, A., Li, J.S., Gill, T.J.t., Li, G., 2012. In vivo patellar tracking and patellofemoral cartilage contacts during dynamic stair ascending. *J Biomech* 45, 2432-2437.

Tecklenburg, K., Dejour, D., Hoser, C., Fink, C., 2006. Bony and cartilaginous anatomy of the patellofemoral joint. *Knee Surgery, Sports Traumatology, Arthroscopy* 14, 235-240.

Thiem, U., Lamsfuß, R., Günther, S., Schumacher, J., Bäker, C., Endres, H.G., Zacher, J., Burmester, G.R., Pientka, L., 2013. Prevalence of self-reported pain, joint complaints

and knee or hip complaints in adults aged ≥ 40 years: a cross-sectional survey in Herne, Germany. PLoS One 8, e60753.

Thomee, R., Augustsson, J., Karlsson, J., 1999. Patellofemoral pain syndrome: a review of current issues. Sports Med 28, 245-262.

Thomeer, L.T., Sheehan, F.T., Jackson, J.N., 2017. Normalized patellofemoral joint reaction force is greater in individuals with patellofemoral pain. Journal of biomechanics 60, 238-242.

Thorhauer, E., Tashman, S., 2015. Validation of a method for combining biplanar radiography and magnetic resonance imaging to estimate knee cartilage contact. Med Eng Phys 37, 937-947.

Van Kampen, A., Huiskes, R., 1990. The three - dimensional tracking pattern of the human patella. Journal of Orthopaedic Research 8, 372-382.

van Rossom, S., Smith, C.R., Thelen, D.G., Vanwanseele, B., Van Assche, D., Jonkers, I., 2018. Knee Joint Loading in Healthy Adults During Functional Exercises: Implications for Rehabilitation Guidelines. J Orthop Sports Phys Ther 48, 162-173.

Vanwanseele, B., Eckstein, F., Knecht, H., Spaepen, A., Stüssi, E., 2003. Longitudinal analysis of cartilage atrophy in the knees of patients with spinal cord injury. Arthritis & Rheumatism: Official Journal of the American College of Rheumatology 48, 3377-3381.

Vlaardingerbroek, M.T., Boer, J.A., 2003. Magnetic Resonance Imaging. [electronic resource] : Theory and Practice, Third edition. ed. Springer Berlin Heidelberg.

Von Porat, A., Roos, E.M., Roos, H., 2004. High prevalence of osteoarthritis 14 years after an anterior cruciate ligament tear in male soccer players: a study of radiographic and patient relevant outcomes. Annals of the rheumatic diseases 63, 269-273.

Waligora, A.C., Johanson, N.A., Hirsch, B.E., 2009. Clinical anatomy of the quadriceps femoris and extensor apparatus of the knee. Clin Orthop Relat Res 467, 3297-3306.

Walker, P.S., Erkman, M.J., 1975. The role of the menisci in force transmission across the knee. Clinical orthopaedics and related research, 184-192.

Waterton, J.C., Solloway, S., Foster, J.E., Keen, M.C., Gandy, S., Middleton, B.J., Maciewicz, R.A., Watt, I., Dieppe, P.A., Taylor, C.J., 2000. Diurnal variation in the femoral articular cartilage of the knee in young adult humans. *Magnetic Resonance in Medicine: An Official Journal of the International Society for Magnetic Resonance in Medicine* 43, 126-132.

White, T.D., Folkens, P.A., 2005. *The human bone manual*. Elsevier.

Wilson, T., 2007. The measurement of patellar alignment in patellofemoral pain syndrome: are we confusing assumptions with evidence? *J Orthop Sports Phys Ther* 37, 330-341.

Yamokoski, J.D., Banks, S.A., 2011. Does close proximity robot motion tracking alter gait? *Gait & posture* 34, 508-513.

Yoshioka, Y., Siu, D., Cooke, T., 1987. The anatomy and functional axes of the femur. *J Bone Joint Surg Am* 69, 873-880.

Zabala, M.E., Favre, J., Andriacchi, T.P., 2015. Relationship between knee mechanics and time since injury in ACL-deficient knees without signs of osteoarthritis. *The American journal of sports medicine* 43, 1189-1196.

Zavatsky, A.B., Oppold, P.T., Price, A.J., 2004. Simultaneous in vitro measurement of patellofemoral kinematics and forces. *Journal of biomechanical engineering* 126, 351-356.

Zeller, I.M., Sharma, A., Kurtz, W.B., Anderle, M.R., Komistek, R.D., 2017. Customized versus patient-sized cruciate-retaining total knee arthroplasty: an in vivo kinematics study using mobile fluoroscopy. *The Journal of arthroplasty* 32, 1344-1350.

Zhao, D., Banks, S.A., Mitchell, K.H., D'Lima, D.D., Colwell, C.W., Jr., Fregly, B.J., 2007. Correlation between the knee adduction torque and medial contact force for a variety of gait patterns. *J Orthop Res* 25, 789-797.