Femoroacetabular Impingement Syndrome and Total Hip Arthroplasty: Joint Biomechanics Before and After Surgery

Danilo Santos Catelli

Thesis submitted to the University of Ottawa
in partial fulfillment of the requirements for the degree of

DOCTOR OF PHILOSOPHY

in Human Kinetics

School of Human Kinetics
Faculty of Health Sciences
University of Ottawa

Thesis supervisor
Prof. Mario Lamontagne, Ph.D.

to the memory of my mom

o sonho que um dia juntos sonhamos,

hoje se torna realidade...

Acknowledgements

"If it were easy, everyone would have a Ph.D." this is not only a quote that my supervisor would say to me every time I would complain about something in the past five years, but this is, in fact, the title of a Ph.D. thesis presented at Washington State University a decade ago. Moreover, all I can say is that this journey was indeed difficult; however, if it was not for the people listed below, this would have been impossible!

Firstly, my supervisor and friend, Prof. Mario Lamontagne, who gave me unconditional support since I started my course application until this moment. Opening doors that I would never expect to be opened, and who brings the most joyful research stories during lunch time! *Merci* Mario!

I would like to also express my sincerest gratitude to Dr. Paul Beaulé, for his continual support, very insightful clinical aspects, and encouragement every time we would meet. "Keep up with the good work" he would say. *Merci à toi aussi* Paul, your hard work and tireless initiatives are incredibly inspiring for any researcher.

I would like to thank my Thesis Advisory Committee members, Prof. Jing-Xian Li and Prof. Daniel Benoit, for the guidance and support along these past years. Your inputs were crucial for this thesis to be presented today. I hope that our research paths will cross again. I would like to also thank Prof. Ryan Graham, for his valuable mentoring and support.

I would like to give a special thank to Prof. Ilse Jonkers and Dr. Mariska Wesseling, for their exceptional support during the past couple years; always willing to help me, especially during the frustrating times with the simulations. Thank you also for hosting me last summer in Leuven and making me feel like part of your lab. It meant a lot to me!

To my lab colleagues, a more than special thanks! My day-to-day life would have been a nightmare if it wasn't for you! First of all, I have to mention Geoff – although you are "gone", you have still been a daily presence, and have helped, mentored and guided me with the wisdom of a nobleman, but yet with the commitment of a loyal friend – thank you! Also to Erik, whom I met on the first day of my classes, and for the last three years has become a loyal, honest colleague! Nevertheless, all the ones that one day were (or that still are) at my side, your collaboration was essential for this achievement: Sarah, Giulia, Mohammad, Teresa, Brent, Luzia, Nok Yeung, Mike, Matt, Andy, Romain and Will (my French curling mate!). Thank you all! =)

To all my Brazilian-born friends that nowadays are, literally, my Canadian family: Firma, Fran, Fer, Celso, Nat, Arthur, Sarah & Val. You made my weekends a blast, and my relaxing times more exciting – thanks for the support and encouragement!

To my dad and family that managed to be supportive and loving even with the distance! I am here today because one day my mom and dad decided to unconditionally support my education. Thank you to my sis, cousins, aunts, uncles, and in-laws for their support.

A better country cannot be built without investing in education; a better future cannot be reached without science. Thank you to the President Dilma Rousseff and to the Minister of Education Fernando Haddad for believing in a better future for Brazil.

I also would like to thank all of the funding agencies that helped with the development of this thesis, providing scholarship and travel awards that allowed me to present this research at many conferences around the world: CAPES (Brazil), The Hans K. Uhthoff Fellowship, The Nicole Bégin-Heick Scholarship, FGPS, GSAÉD, CUPE 2626, School of Human Kinetics, Force & Motion Foundation, ISB, ISBS, and OpenSim.

Last, but definitely not least, thank you to my loving and beautiful wife, Rayane! You are definitely my other half - my best half! Thank you for all the support, caring, love, and for making all my days beautiful! Today we are three, and our Bella is the most precious gift I could have ever dreamed about… *Amo vocês!*

Abstract

Surgical interventions on the hip joint have greatly increased over the past decade, with the cumulative cost total hip arthroplasties (THA) alone exceeding $400B/year by 2020. Although positive patient-reported outcomes and satisfaction after THA and hip preservation for cam femoroacetabular impingement (FAI) are among the highest in orthopaedics, a limited number of research has investigated the biomechanics of dynamic activities following-up the surgery. This doctoral thesis examined the kinematics, muscle force component, and hip contact loading in pre- and postoperative patients during the deep squat motion. Specifically, this research: 1) examined muscle strength and pelvic kinematics in asymptomatic FAI, 2) examined lower-limb kinematics and muscle activity in postoperative patients who underwent either THA or FAI correction during a deep squat task, and 3) examined muscle force contributions and hip contact forces (HCF) during dynamic motion in postoperative FAI patients.

First, clinical and medical imaging evaluations classified the participants into three groups: symptomatic FAI, asymptomatic FAI (FAD – participants had the cam deformity, but no pain), and healthy controls. The FAD participants had significantly greater hip extensor strength compared to the FAI and CTRL groups, which allowed them to achieve greater pelvic mobility and squat as deep as the CTRL group.

Second, at the follow-up for the FAI surgery the patients showed increased pelvic ROM during the squat, and weakness associated with hip flexion and hip flexion-with-abduction were associated with postoperative alterations. For the THA follow-up analyses, the patients using a dual-mobility (DM) prosthesis reached an anterior pelvic tilt similarly to the CTRL during the dynamic parts of the squat; however, without returning its neutral tilt at the bottom of the squat, while the single-bearing (SB) prosthesis was associated with excessive hip abduction during the squat.

Third, a generic full-body musculoskeletal model (MSKM) was optimized to allow for the analysis of tasks with a high range of motion (ROM; e.g. deep squat task), which controlled muscle moment arms during the high joint flexions to avoid the model's motor tendon units (MTU) to penetrate the bony structures and respect the anatomical via points. Simulation performed during gait demonstrated that FAI patients enhance medial-lateral hip stability postoperatively, allowing reduced dynamic forces of the muscles associated with the sagittal aspect of the gait due to a less compensatory strategy to stabilize the hip joint. Furthermore, simulations performed during deep squat showed a higher anterior pelvic tilt in postoperative FAI patients as a 'restore to native' mechanism once the cam-deformity was no longer present. Increased semimembranosus force was linked to higher vertical HCF and total magnitude.

The outcomes of this research include findings for gait and squat analyses that provide a better understanding of the pelvic mobility and hip muscle forces in hip diseases. *In silico* models can improve biomechanical assessment of postoperative patients in order to quantify surgical effectiveness and support clinicians in making subject-specific case decisions. The contributions also lay on the assertion of helping us to formulate future research directions in biomechanics applied to the orthopaedics field.

Preface

The contents of this dissertation are organized into five sections, with each section arranged into multiple chapters, as follows:

I. *Opening*, which introduces the research statement, background, literature review, and the study design,

II. *Asymptomatic Individuals*, which examines strength and motion parameters that differentiate asymptomatic individuals from preoperative ones,

III. *Squatting Motion Analysis after Hip Surgery*, which examines postoperative patients who underwent both hip preservation surgery and total hip arthroplasty during a high range of motion task: the deep squat,

IV. *Musculoskeletal Modelling: from Gait to Squat*, which examines the dynamic muscle forces and the hip contact forces in postoperative patients who underwent hip preservation surgery for femoroacetabular impingement during the two task conditions,

V. *Closing*, which includes the points of discussion, limitations, and concluding remarks.

The dissertation is comprised of six original research articles, on which I was lead author. I was fully involved with: 1) the conception of the studies; 2) data acquisition, analysis and interpretation; 3) drafting, revising, and submission of the articles for peer review in scholarly journals. At the time of thesis submission, the six articles were either published, under revision, or prepared for peer review in journals specializing in areas of clinical orthopaedics and biomechanics. Each of the six articles, within this doctoral thesis, was formatted the journal's and publisher's requirements. Lastly, I have no conflicts of interest to report and certify that the research ethics of both institutions (University of Ottawa and The Ottawa Hospital) approved the investigation protocols (Appendix A – Ethics). All investigations were conducted in conformity with ethical research principles and informed consent for participation in the study was obtained.

Table of Contents

List of Abbreviations

ADL: Activities of Daily Living
AM: Adductor Magnus
ANOVA: Analysis of Variance
ASIS: Anterior-Superior Iliac Crest
avg: Average
BF: Biceps Femoris
BMI: Body Mass Index
BW: Body Weight
CI: Confidence Interval
CT: Computerized Tomography
CTRL: Healthy Control
deg: Degrees
DL: Daily Living
DM: Dual Mobility
DOF: Degrees of Freedom
EMG: Electromyography
F: Female
FABER: Hip Flexion with Abduction and External Rotation Test
FADIR: Hip Flexion with Adduction and Internal Rotation Test
FAI: Femoroacetabular Impingement
FAD: Femoroacetabular Deformity (asymptomatic)
Gmax: Gluteus Maximus
Gmed: Gluteus Medius
GRF: Ground Reaction Force
JCF: Joint Contact Forces
HCF: Hip Contact Forces
HHD: Hand-Held Dynamometer
HOOS: Hip Disability and Osteoarthritis Outcome Score
ID: Inverse Dynamics
IMAT: Intermuscular Adipose Tissue
IK: Inverse Kinematics
M: Male
MA: Moment Arm
mo: Months
MRI: Magnetic Resonance Imaging
MSK: Musculoskeletal
MSKM: Musculoskeletal Model
MTU: Musculotendon Unit
MVIC: Maximum Voluntary Isometric Contraction
N/A: Not Applicable
OA: Osteoarthritis
PE: Polyethylene
PeakLE: Peak Linear Envelope
PL: Peroneus Longus
Post-Op: Postoperative
Pre-Op: Preoperative
PROMs: Patient Reported Outcomes Measures
PSIS: Posterior-Superior Iliac Crest
QOD: Quality of Life
RF: Rectus Femoris
ROM: Range of Motion
RRA: Residual Reduction Algorithm
Sart: Sartorius
SB: Single Bearing
SD: Standard Deviation
SO: Static Optimization
SRA: Sports and Recreation Activities
ST: Semitendinosus
SPM: Statistical Parametric Mapping
TA: Tibialis Anterior
TFL: Tensor Fasciae Latea
THA: Total Hip Arthroplasty
UOMAM: University of Ottawa Motion Analysis Model
VI: Vastus Intermedius
VL: Vastus Lateralis
VM: Vastus Medialis
WS: Wrapping Surface
y: Years

List of Figures

List of Tables

I Opening

1 Introduction

Statement | Rationale | Background

1.1. Statement

Hip surgery interventions have become more frequent over the past decade (Beaulé et al., 2007a; Bozic et al., 2013) with more than 360,000 total hip arthroplasty (THA) performed annually in North America (CDC, 2010; CJRR, 2015). This amount will cost approximately $405B annually by the year 2020 (Bombardier et al., 2011). This also does not consider hip arthroscopy procedures, that have been used as both diagnostics and treatments of femoroacetabular impingement (FAI), chondral injuries and osteoarthritis (OA), and synovitis (Byrd, 2006; Diulus et al., 2006).

The positive patient-reported outcome measures (PROMs) and surgery satisfaction are among the highest in orthopaedics for both THA (Learmonth et al., 2007) and FAI (Philippon et al., 2013) interventions. For FAI, however, limited studies were able to define biomechanical changes during dynamic motions (Rylander et al., 2013, 2011), and controversial results (Bedi et al., 2011a; Lamontagne et al., 2011b; Malagelada et al., 2015) make them even more difficult to generate clinical relevance. This invokes the controversy if more conservative management of the pathology (Casartelli et al., 2017; Loudon and Reiman, 2014) can be an option to reduce the overall cost towards Medicare.

Quantitative description of the kinematics and kinetics of body-segmental movement are commonly used in gait-analysis experiments operated with high-frequency infrared cameras, along with ground-reaction force platforms and surface electromyography systems (Pandy, 2001). The data provided by these tools cannot explain how muscles work in concert to produce a coordinated motion, or how muscle forces will impact joint contact loading. Computer simulation of human motion has provided new insights to solve this deficiency (Pandy, 2001; Vaughan, 1984), and studies using this approach have increased a lot in the past decade. The quantitative explanations of how the components of the musculoskeletal (MSK) system interact demonstrates great promise for enhancing treatment of pathologies and diseases that may limit human mobility (Hicks et al., 2015). With current literature on postoperative FAI focusing primarily on a quantitative description of kinematics and kinetics of the lower limb during level walking, to our knowledge, no studies have investigated MSK interactions or reported the progression of muscle forces and joint contact loading during extreme dynamic motion (e.g. squat) from pre- to post-surgery.

With the ultimate goal to better understand how muscle and hip contact forces (HCF) change after hip surgery in a medium-term follow-up during an extreme hip flexion task, the purpose of this research is to handle the question: *what effect will hip surgeries have with respect to the MSK system performance in a deep squat task?* This research proposes to:

- Evaluate how asymptomatic individuals with FAI cam-type deformity differ from the preoperative symptomatic ones (Chapter 4);
- Examine the postoperative kinematics and muscle contraction patterns during a squat performance in both FAI and THA patients (Chapters 5 and 6);
- Examine the muscle forces and HCF during a level walking to deep squat tasks after surgical correction of the cam-type deformity (Chapters 7, 8 and 9).

1.2. Rationale

Although the FAI etiology remains unclear, the clinical and biomechanical interest associated with femoroacetabular impingement has greatly increased in the past years (Leunig et al., 2009), especially since it has been considered as a leading factor in the development of hip osteoarthritis (OA) (Rintje Agricola et al., 2013; Ganz et al., 2008). In cases with a more severe and larger cam deformity, anterosuperior cartilage and chondrolabral damage are expected (Beaulé et al., 2005; Beck et al., 2005; Leunig et al., 2004). The stress generated with the increase of the contact area between the femoral head and the acetabulum is likely, by Wolff's law (Wolff, 1892), increases the subchondral bone density developing a stiffer bone, that over time the cartilage tissue is more prone to wear and leading to the development of OA (Beck et al., 2005; Ganz et al., 2003; Hart et al., 2009; Murphy et al., 1995; Speirs et al., 2013a). Once the OA is well established, and conservative medical therapy has failed to reduce stiffness, swelling, and pain, total hip arthroplasty becomes a treatment option (Bottai et al., 2015) with very good long-term results (Learmonth et al., 2007).

In order to avoid the advancement of OA, individuals with early clinical and biomarker (e.g. high-resolution magnetic resonance imaging) signs of FAI are commonly referred to a hip-preserving surgery (Beaulé et al., 2009; Freeman et al., 2014). Femoral osteochondroplasty restores the femoral head-neck offset to re-establish a proper spherical femoral head (Aoki et al., 2016; Bedi and Kelly, 2013), and although it indicates high rates of success in term of positive outcomes scores and reduced pain (Beck et al., 2004; Clohisy et al., 2010; Freeman et al., 2014; Stevens et al., 2010), inconclusive kinematic outputs from previous researches (Bedi et al., 2011a; Brisson et al., 2013; Lamontagne et al., 2011b; Malagelada et al., 2015; Rylander et al., 2013, 2011) may bring the questioning if a conservative care may be a management option for symptomatic individuals with FAI (Casartelli et al., 2016).

When considering that a high percentage of the general population have the FAI deformity - but will not notice any FAI symptoms and will go through life without any consequences of its presence (Chakraverty et al., 2013) – it is important to characterize the differences of the asymptomatic individuals with the preoperative patients, in order to evaluate if there are differences in their hip biomechanics. Assessment of dynamic muscle forces and hip contact loading in postoperative patients may support the surgical choice and provide key information for pre- and postoperative rehabilitation protocols.

1.3. Background

1.3.1. From FAI to OA

Although the etiology of FAI is still unclear (Ganz et al., 2008; Leunig et al., 2009), it is believed to be multifactorial (Hart et al., 2009) with a combination of determinants as favorable genetics, soft tissue damage, cartilage defect, labral lesions, hip anatomical geometry, joint stiffness and/or neuromuscular imbalances (Lamontagne et al., 2015). Excessive or repetitive hip motion activities, post-surgical changes and malunion after a fracture have also been pointed as major risk factors for developing FAI (Chaudhry and Ayeni, 2014; Dimmick et al., 2013). Nonetheless, the link between the activity level and the femur malformation has been suggested long ago (Murray and Duncan, 1971), and more recent studies are establishing the link between the high level of physical activity training during late childhood and late adolescence and FAI formation (Agricola et al., 2014; Carsen et al., 2014; Siebenrock et al., 2011; Siebenrock and Schwab, 2013).

High active adolescent indicates a reduction in the hip internal rotation as they are getting older while being involved in sports (Siebenrock et al., 2011). However the cam morphology does not occur before pre-epiphyseal closure (Carsen et al., 2014). Physical activity, when done during childhood direct, affects not only bone density but also bone geometry (cross-sectional area and periosteal circumference) even on those subjects that ceased sports activity for up to 6.5 years (Nilsson et al., 2009). Moreover, the subjects who previously performed sports exercises with high impact loading have a greater cortical bone size, associating the sport-specificity with the bone transformation.

On one hand, while physical activities can generate positive effects during childhood and early adolescence on bone geometry (periosteal augmentation) and reduce the risk of fracture (or develop osteoporosis) later in life, even when physical activities are ceased (the benefits remain once the stimuli were done while the body was still in growth – open epiphyseal cartilage) (Vainionpää et al., 2005). The same stimuli could also mean an onset for the bone malformation, as the beginning of a cam abnormality (Agricola et al., 2014; Carsen et al., 2014; Siebenrock et al., 2013). Thus, it is speculated that sports involving jumping and explosive actions when practiced in a high intensity/frequency per week during bone maturation may be associated with the risk of developing FAI.

As femoroacetabular impingement has been associated as a cause of early OA of the hip (Beck et al., 2005; Ganz et al., 2008, 2003; Hart et al., 2009; Siebenrock et al., 2011), surgical treatments have been reported and recommended as a mechanism that provides not only symptoms relief, but also may decelerate the joint degenerative process (Ganz et al., 2003). To analyze the damage caused by FAI over time, it is important to understand the progress of impingement in the hip joint better.

It has been speculated that the cam deformation induces a migration of the femoral head to a more anterosuperior position (Eijer and Hogervorst, 2017), causing a reduction of the contact area between the femoral head and the acetabular cartilage, and therefore, higher compressive and shear forces at the joint (Eijer and Hogervorst, 2017). The higher stress at the joint which will lead to an increase of the subchondral bone density (Speirs et al., 2013b, 2013a; Wolff, 1892). A stiffer bone at the impingement site causes damage to the anterosuperior acetabular cartilage (Figure 1.1), leading the patient to a more severe degenerative condition (Ganz et al., 2003; Hart et al., 2009).

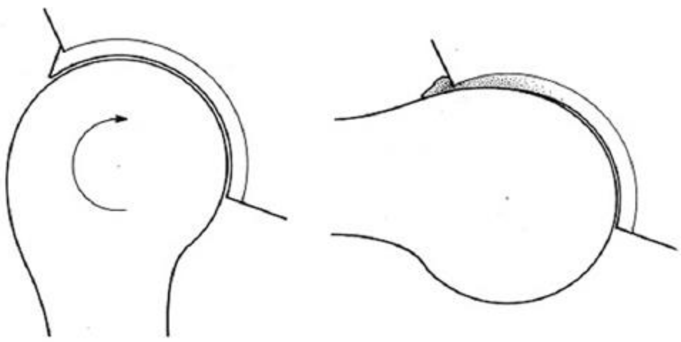


Figure 1.1. Mechanism of damage in cam impingement on a lateral hip view. During flexion the non-spherical femoral head impacts the acetabulum rim, compressing the cartilage and pushing it centrally until it dislocates off the subchondral bone. Reproduced with permission of PLSclear (Beck et al., 2005).

Once OA is established, treatment options will depend on symptoms severity, which in chronic cases options include a total hip arthroplasty (THA) (Taruc-Uy and Lynch, 2013).

1.3.2. *Asymptomatic Population*

The cam-type deformity limits hip range of motion (ROM) for internal rotation associated with flexion during clinical examinations (Carsen et al., 2014; Siebenrock et al., 2011) and dynamic movements (Kennedy et al., 2009; Lamontagne et al., 2009). Although leading to clinical symptoms, the cam pathomechanism is not completely understood (Lamontagne et al., 2015). Heretofore, no associations were found between clinical symptoms and α-angles, as many asymptomatic subjects have their α-angle measured higher than 55º, which suggests that cam presence and severity cannot be determined by pain during mechanical impingement (Lamontagne et al., 2015).

Agricola et al. (2013a) highlighted the difference between cam abnormality and cam impingement, affirming that association of both is not truly straightforward and factors such as femoral or acetabular orientation may also be involved as a risk to develop symptoms. The authors suggested that although cam impingement is a risk factor to OA; most patients with a positive radiograph indication will not develop OA; as other risk factors such as repetitive impinging movements (e.g. hip flexion and hip internal rotation), injury or acetabular morphology can contribute to the development of OA. Perhaps an individual with cam impingement will only become symptomatic if exposed to specific activities (i.e. practicing high impact sports).

Alternatively, other anatomical joint measurements could be more reliable as a mechanism to not only recognize cam morphology but also associate it with clinical symptoms and a pathomechanism diagnosis. Ng et al. (2015, 2016b) found an indication that reduced femoral neck-shaft angle and reduced pelvic ROM are valuable measurements that can identify individuals with cam symptomatic over the asymptomatic ones. This type of analysis contributes to distinguishing individuals better, in addition to providing extra data that could allow us to better understand the pathomechanics that lead to arthritic changes. Although there were recent efforts to link the anatomical parameters associated with the FAI pathomechanism, there is still a gap in the literature that does not delineate how hip muscle strength combined with dynamic motion may affect FAI symptomatology.

1.3.3. Motion Analysis: The Squat Option

The plasticity of the human body is extraordinary when considering the endless types of movements that someone can achieve. The human body can sprint faster than 12 ms^{-1} (Hernández Gómez et al., 2013), resist to over 9x bodyweight during drop-landings in gymnastics (Mcnitt-Gray et al., 1993), perform elaborate movements with coordinated muscular skill as when performing *parkour* (Grosprêtre and Lepers, 2016), or even adapt to singular situations such as in the act of giving birth (Desseauve et al., 2017). However, one of the first skills that humans learn during infancy that represents one of the most important milestones in the development of the motor control is walking (Sutherland, 1997); and this is probably the skill that a healthy person performs most often during one's life.

It is not a coincidence that gait is the task most performed in motion analysis, especially when concerning clinical application. A humble search in PubMed[1] using the keyword ‘gait analysis’ and ‘gait biomechanics’ identified 19,496 and 2,916 items, respectively; while the search for ‘squat analysis’ and ‘squat biomechanics’ identified 1,391 and 234 items, respectively. Gait analysis is used to assess individuals with different pathological conditions (Hainisch et al., 2012; Morris et al., 2001), making it a valuable tool to quantify joint kinematics and kinetics in order to provide insights to improve rehabilitation protocols and orthopaedic surgeries. However, level walking cannot be considered a challenging task for a healthy adult population. It provides limited ROM, with only 20-30º of hip flexion range, which in the case of FAI for example, is not considered enough to cause impingement (Chegini et al., 2009).

Other types of activities of daily living (ADL) are also commonly performed to evaluate clinical patients: stairs (Hammond et al., 2017; Lamontagne et al., 2011a), sitting and standing (Lamontagne et al., 2012), inclined and declined walking (Stansfield and Nicol, 2002), squat (Lamontagne et al., 2011b), among others. The deep squat, although a popular exercise among athletes

---

[1] The search was performed in July 2018.

and in fitness centres, is not a common task for mostly of the ordinary people of the modern western culture (Rane and Corstiaans, 2008). However, when compared with the other ADLs listed above, this is the one that provides the largest range of hip and knee flexions, bringing both joints to its limit, while supporting the body weight.

There are several variations of the squat exercise, but in order to be reliable and valid, is important to be specific and consistent throughout the movement analyzed in clinical and laboratorial settings. The deep squat task (Figure 1.2), when done in controlled settings, offers the opportunity of evaluating the performance of the lower limb muscle and joints in a closed kinetic chain task with very high joint ranges.

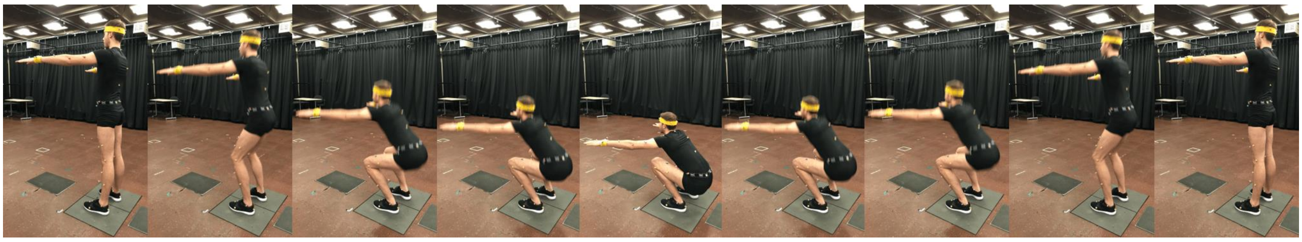


Figure 1.2. Deep squat task, the participant is looking straight forward, with the arms held out straight at shoulder level. Feet are positioned hip-width apart and pointing straight ahead, and the heels remain down during the duration of the task. The participant was instructed to squat as deep as possible without leaning the trunk forward at a controlled speed.

When thinking about a successful surgical intervention in the hip joint, associated with the elimination of the pain and improvement of the clinical symptoms, is the improvement of the overall joint performance. Moreover, we consider the squat as a key task to evaluate the hip joint performance postoperatively.

1.3.4. In silico Analysis: Neuromusculoskeletal Modelling

Research profiles in clinical biomechanics often report kinematics, kinetics, and neuromuscular control associated with an injury or diseases, as a mechanism for describing one’s movement without performing an invasive assessment. Herewith, many research questions still cannot be answered as the direct measurement of muscle forces and joint contact forces are not ethically feasible. Alternatively, and as a result of the advancements in imaging and computing technology, musculoskeletal modelling has been able to explore further the data collected in the laboratory setting to provide individual muscle and hip contact forces. Also, individual variables (e.g. muscle parameters) can be manipulated in order to simulate a cause-effect environment (Smale, 2018).

Although some researches have described strength profiles of preoperative (Casartelli et al., 2011; Diamond et al., 2016b) and postoperative (Casartelli et al., 2014) FAI patients, these have been limited to isometric and isokinetic assessments. Heretofore, only one *in silico* analysis of patients with FAI (Ng et al., 2018b) has been done during gait, and it was able to describe the actions of hip muscles and the HCF along the stance phase. This study opens the doors to advancement in this field, while it was able to identify differences in the hip flexors acting forces that impacted the HCF in preoperative FAI patients compared to healthy participants.

Still, questions regarding this matter only build up: how would the muscle forces during a dynamic task in these patients be affected post-surgery? Would that affect their HCF as well? What would be their response in a task with higher lower-limb joint ROM (such as the deep squatting)?

With most of the recent generic models are designed for walking and running tasks (Arnold et al., 2010; Rajagopal et al., 2016), the musculotendon units (MTU) within the model do not respond well to tasks with high ROM (Figure 1.3). The MTUs cross the pelvis and the femur since these are not represented as solid structures in the model. To control the MTU paths and avoid them to penetrate the bones, solid structures in various shapes near the joint are implemented in the model as so-called wrapping surfaces (WS). Although five WS are present on each side of the recently published Rajagopal's generic model (Rajagopal et al., 2016), MTUs start overlapping bones with just over 85º of hip flexion (not considering hip abduction and external rotations that also occur during the deep squatting task).

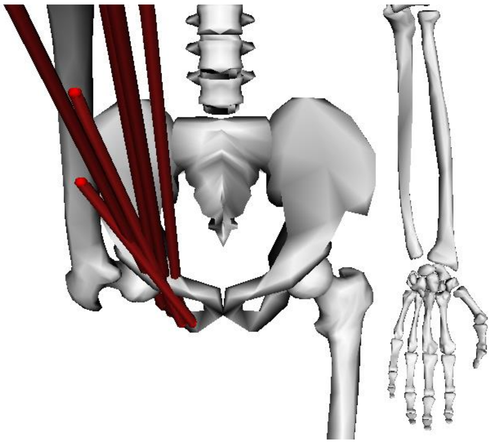


Figure 1.3. Frontal view of the generic model (Rajagopal et al., 2016) in the OpenSim simulation software (Delp et al., 2007), that shows the posterior MTU crossing the pelvis and the head of the femur when the hip joint is placed at 130º of flexion.

Thus, while the simulation software does offer possibilities to investigate neuromusculoskeletal changes in a symptomatic population further, the challenge regarding the line of action of the MTU while performing high ROM tasks must be priory solved in order to simulate deep squat trials.

2 Literature Review

Hip Surgical Interventions | Musculoskeletal Modelling

2.1. Hip Surgical Interventions

For both FAI and hip OA, corrective surgery is recommended by clinicians as a method to restore a normal femoral head-neck offset. Surgical approach for FAI correction can be performed using an open (Ganz et al., 2001; Lavigne et al., 2004), arthroscopic (Guanche and Bare, 2006; Philippon et al., 2008; Weiland and Philippon, 2005) or combined technique (Clohisy and McClure, 2005; Lincoln et al., 2009). The treatment options for a hip with OA correction vary from osteotomy (Ninomiya and Tagawa, 1984; Teratani et al., 2010), hip resurfacing (Caplan et al., 2014; Girard et al., 2006; Heintzbergen et al., 2013) and total hip replacement or THA (Ethgen et al., 2004; Huo et al., 2009). In the following subchapters, we will describe some concepts regarding these two hip conditions, further explore the surgical possibilities and point out what is the state of art of the biomechanical research on this field.

2.1.1. Femoroacetabular Impingement

Almost 20 years ago, in 1999, the FAI concept, also know as hip impingement syndrome, was first described as an impingement produced by the femoral head-neck junction on the anterior rim of the acetabulum (Myers et al., 1999), due to an osseous overgrowth at the femoral head-neck junction (cam impingement), at the acetabular rim (pincer impingement), or at both of them (mixed impingement) (Ganz et al., 2003; Tannast et al., 2007).

While isolated pincer deformity is rare, with an overall prevalence of 7% in the FAI patients, and its link with the development of OA has not been confirmed (Agricola et al., 2013b); the cam impingement affects more than 17% of men and 4% of women (Gosvig et al., 2008). It leads not only to pain but also to limited ROM during ADLs (Brisson et al., 2013; Kennedy et al., 2009; Lamontagne et al., 2009; Ng et al., 2015). Cam impingement is characterized by increased radius of the femoral head at the junction of its neck. The abnormal deformity is defined by the alpha angle (Figure 2.1) which exceeds the value of 50.5 º (Barton et al., 2011; Hack et al., 2010; Lamontagne et al., 2011b) for less conservative to 55º (Kassarjian et al., 2007; Nötzli et al., 2002; Nouh et al., 2008) in the oblique-axial plane.

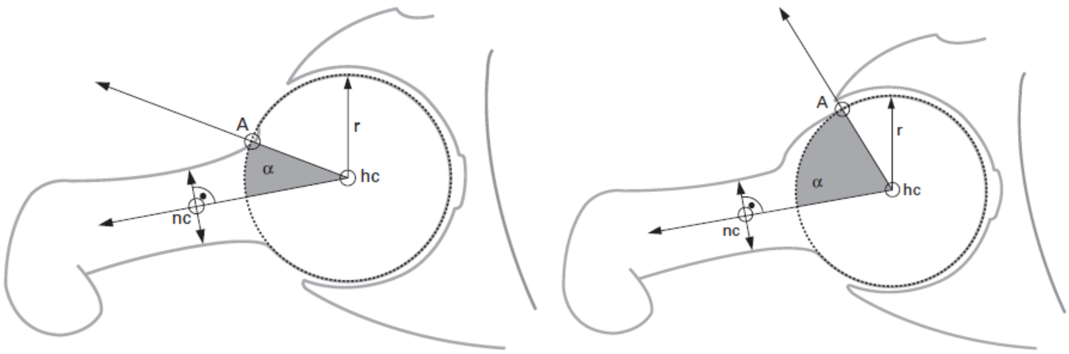


Figure 2.1. α-angle – angle formed by a line crossing the centre of the femoral head (hc) and the femoral neck (nc), and the line connecting the centre of the femoral head with the first point that exceeds the radius of the femoral epiphysis (A), the figure on the left represents a healthy subject, while the one on the right represents a cam FAI. Reproduced with permission of PLSclear (Nötzli et al., 2002).

Hip and groin pain are the primary symptoms of FAI, and a typical gesture used by the patients to report the pain location is opposing the index finger and the thumb along with the lower line of the iliac crest – forming a "C" sign (Byrd, 2014; Lamontagne et al., 2015; Leunig et al., 2009) (Figure 2.2, A). Clinical examinations such as the FABER (hip flexion with abduction and external rotation, Figure 2.2, B) and FADIR (hip flexion with adduction and internal rotation, Figure 2.2, C) test, have shown a sensitivity greater than 96% and 88% to diagnose FAI, respectively (Wilson and Furukawa, 2014). Moreover, studies suggest that FAI is a leading factor in the development of hip OA, particularly in younger adults (Agricola et al., 2013a; Ganz et al., 2008).

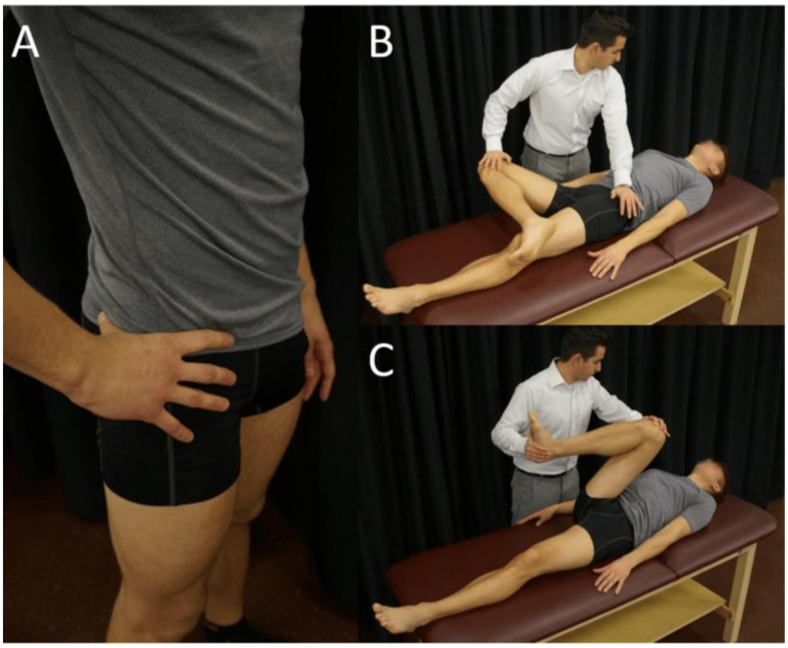


Figure 2.2. A) The "C" sign, the typical gesture used by the FAI patients to describe the deep interior hip pain. B) FABER test (hip flexion, abduction, external rotation): after flexing the hip at around 45 degrees, the examiner externally rotates and abducts the patient's leg. C) FADIR test (hip flexion, adduction, internal rotation): the examiner moves the leg passively into full hip flexion and then into adduction with internal rotation.

As cam impingement affects mostly young and active adults (Beck et al., 2005; Ganz et al., 2008; Nötzli et al., 2002), early recovery from surgical intervention is of utmost importance to allow the patients to return to their usual physical practices, improving quality of life, prevent joint degeneration, and reduce the rate of OA development as a result of FAI (Beaulé et al., 2007b; Eijer et al., 2001; Leunig et al., 2009; Pollard, 2011).

2.1.1.1. FAI correction

The hip osteochondroplasty, also referred as a hip preservation surgery, attempts to restore the femoral head sphericity and avoid the impingement (Freeman et al., 2014; Lavigne et al., 2004). The early stage intervention is highly recommended before major cartilage damage is present (Beck et al., 2005; Fayad et al., 2013; Ganz et al., 2008). Controversially, some surgeons might want to proceed with an intervention even in (pain-free) asymptomatic cases that also present limited hip ROM and initial cartilage damage (Ganz et al., 2008; Pollard et al., 2010), as the surgery is most successful in the absence of secondary degenerative changes. As for the osteochondroplasty procedure, some of the most common surgical techniques are the surgical dislocation of the hip, and the hip arthroscopy (Freeman et al., 2014).

Surgical dislocation of the hip is rarely undertaken for reasons other than arthroplasty, and it can be accomplished through an anterior, lateral, or posterior approach (Ganz et al., 2001). The posterior technique requires an incision which releases most of the posterior fibres of gluteus medius, allowing an almost 360º visualization of the femoral head and complete access to the acetabulum for inspection (Ganz et al., 2001). This approach also allows both acetabular and resection osteoplasties following surgical dislocation, allowing for impingement-free movement between the acetabulum and femoral head (Lavigne et al., 2004). A mini-open anterior approach with arthroscopic assistance technique has also been frequently used (Beaulé et al., 2017), and it allows direct visualization of the anterior femoral head-neck junction, with an incision that avoids muscle resection (i.e. the approach passes between the rectus femoris and the tensor fascia latae muscles), and has a low complication rate (Laude et al., 2009).

On the other hand, arthroscopic surgery provides an interior viewing of the femoroacetabular joint followed by the removal of excessive non-spherical femoral head tissue, which causes the impingement at larger ROM (Guanche and Bare, 2006). This minimally invasive treatment may better restore hip mechanics. Also, hip arthroscopy for FAI has been reported to produce excellent improvement in clinical outcomes and high level of patient satisfaction in the short-term in the adolescent population (Philippon et al., 2008), and allow most of the athletes to return to the same competitive level of sports participation (Reiman et al., 2018).

Aside from the benefits and disadvantages of each procedure, the FAI correction surgery is expected to delay the onset and progression to end-stage arthritis in young and active patients, preserving their joint and possibly avoiding a THA intervention (Pun et al., 2015). Still, it is important to remember that not all patients with the FAI morphology need surgical intervention as identification of the varying spectrum of hip pathologies that present as FAI (i.e. comprehensive history, physical examination and critical analysis of radiological indicators) is a crucial prerequisite for successful treatment (Fayad et al., 2013).

2.1.1.2. Motion Analysis: Preoperative and Postoperative FAI correction

Several cross-sectional studies compared motion analysis of preoperative FAI with healthy control participants (CTRL) (King et al., 2018). These studies focused primary the following tasks: level walking (Diamond et al., 2016a; Farkas et al., 2015; Hetsroni et al., 2015; Hunt et al., 2013; Ng et al., 2018b; Rutherford et al., 2018; Samaan et al., 2017) and squat (Bagwell et al., 2016; Diamond et al., 2017; Kumar et al., 2014; Lamontagne et al., 2009), but others such as stairs (Hammond et al., 2017) and sit-to-stand (Samaan et al., 2017) were also investigated. In sum, when compared to healthy individuals, patients with FAI exhibit alterations in hip and pelvis movement strategies during walking (Diamond et al., 2016a; Hetsroni et al., 2015; Hunt et al., 2013; Kennedy et al., 2009; Ng et al., 2018b) and deep squatting (Lamontagne et al., 2009). However, several limitations involve these studies, with primarily not controlling the type of deformity of the FAI participants and also not controlling if the healthy participants were asymptomatic FAI, which may account for over 20% of the general population (Frank et al., 2015; Hack et al., 2010).

To our knowledge, only six studies have investigated motion analysis following the surgical intervention (Table 2.1), which four of them analyzed level walking (Brisson et al., 2013; Malagelada et al., 2015; Rylander et al., 2011) or level walking and stair climbing (Rylander et al., 2013), one analyzed the squat task (Lamontagne et al., 2011b) and one simulated the hip ROM (Bedi et al., 2011a).

Table 2.1. Summary of motion analysis studies that investigated postoperative FAI patients

Author (year)	FAI morphology	Surgical Approach	Follow-Up (months)	Age Average	Sample Size	Female (%)	Task	Key Findings
Malagelada et al. (2015)	cam and mixed	mini-open	12	40.8	14	35.7	gait	increased support time
Brisson et al. (2013)	cam	open (4) or combined arthroscopic with mini-open (6)	21.1	29.9	10	30	gait	no pre/post differences *vs* controls: reduced hip frontal and sagittal ROM; smaller peak hip abduction and internal rotation moments; decreased peak hip power generation
Rylander et al. (2013)	pincer and mixed	arthroscopy	12	35.4	17	29.4	gait and stair climbing	gait: restored gait patterns; stairs: reduced hip transversal and sagittal ROM remained
Lamontagne et al. (2011b)	cam	open (4) or combined arthroscopic with mini-open (6)	28.8	29.9	10	30	squat	improved squat depth increased knee and ankle flexions
Rylander et al. (2011)	pincer and mixed	arthroscopy	12	33.1	11	27.3	gait	increased hip sagittal ROM
Bedi et al. (2011a)	cam, pincer and mixed	arthroscopy	3	25.9	10		computer-assisted 3D modelling ROM	improved hip flexion and hip internal rotation (with hip flexed at 90º)

From the gait performed studies, Rylander et al. (2011) revealed that postoperative FAI (pincer and mixed) patients improved on the operative side in sagittal plane hip ROM, in maximum hip flexion. Although not controlling analgesic intake, this study also revealed a disruption in the normally smooth motion of the hip passing from flexion into extension in 5 of 11 patients preoperatively (3 cam, 1 pincer, 1 mixed), that was reduced to 2 patients postoperatively (1 pincer, 1 mixed). The postoperative hip flexion angle reversals in the patient with a pincer lesion reduced in magnitude and prevalence. No significant differences were seen in frontal plane hip ROM (Rylander et al., 2011). Brisson et al. (2013) although not revealing pre/post differences, showed that when compared with the CTRL, cam-type (only) FAI patients had decreased hip frontal and sagittal plane ROM preoperatively that remained reduced postoperatively. Kinetic testing showed smaller peak hip abduction moments and internal rotation moments in the postoperative patients also when compared with the CTRL (Brisson et al., 2013). Malagelada et al. (2015) showed postoperative hip ROM improvement in all three planes of the affected hip for the FAI (cam and mixed) patients, however without showing a statistical significance. The only gait parameters that presented significance postoperatively in their study were a higher support time of the affected side and lowered contralateral braking force (Malagelada et al., 2015).

Rylander et al. (2013) analyzed both gait and stair climbing tasks. The findings revealed that the preoperative FAI, when compared to CTRL, showed a reduced hip sagittal plane ROM and hip internal rotation during both tasks. Moreover, although these kinematic variables were almost restored to normal level postoperatively during gait, they remained significantly reduced during stair climbing. Also, pelvic transverse plane ROM and maximum pelvic anterior tilt remained increased compared with CTRL. When divided into subgroups, the cam patients displayed more internal rotation preoperatively in both walking and stair climbing compared with the pincer patients, which showed the largest postoperative gains in hip internal rotation during the gait (Rylander et al., 2013).

Lamontagne et al. (2011b) reported on the pre- and postoperative cam-type FAI patients during deep squatting task and showed a deeper squat during post-operative measures as compared with their preoperative measure. The squat depth improvement was mainly due to greater knee flexion and ankle dorsiflexion peak angles, since no significant differences were detected concerning the kinematics of the pelvis, nor the affected hip (Lamontagne et al., 2011b).

Lastly, Bedi et al. (2011a) used a three-dimensional (3D) computational tomography (CT) images of affected hip pre- and postoperative to perform a dynamic simulation in which the model was stimulated to define the hip ROM up until the impingement to occur, with the pelvis fixed in space. The simulations output improvement in hip flexion and hip internal rotation (when the joint was flexed at 90º) (Bedi et al., 2011a). The pitfall of this technique is that it does not take into account the role of the soft tissue while limiting or guiding the motion; neither capsulolabral, nor musculotendinous (not to mention neurovascular) structures were considered in the simulations (Sampson and Safran, 2015).

Although the use of arthroscopic surgery for FAI has risen recently (Reiman et al., 2016), studies evaluating the effect of this surgical correction on lower limb biomechanics are still limited. Half of the reported studies showed significant improvements in hip kinematics, during gait (Rylander et al., 2013, 2011) and in the overall ROM in 3D simulation (Bedi et al., 2011a). Other two studies did not reveal hip kinematic differences between pre- and postoperative patients (Brisson et al., 2013; Malagelada et al., 2015). Also, the squat depth improvement was associated with the increased motion on the knee and ankle joints rather than the hip (Lamontagne et al., 2011b), while restricted motion was still observed during stair climbing in patients who had improved their hip kinematics during level walking (Rylander et al., 2013). While one could argue that the incongruencies between these studies may be a consequence of the motion capture marker set choice or the CTRL group comparison criteria (Sampson and Safran, 2015), the conflicting results for the surgical intervention during gait may be due to the applied surgical technique – open/mini-open/combined (Brisson et al., 2013; Malagelada et al., 2015) or arthroscopy (Rylander et al., 2013, 2011) – the FAI morphology type – cam/pincer/mixed (Malagelada et al., 2015; Rylander et al., 2013, 2011) or cam only (Brisson et al., 2013) – the follow-up period – 12 months (Malagelada et al., 2015; Rylander et al., 2013, 2011) or more (Brisson et al., 2013) – or even the sex criteria.

The results review that although in all studies whose surgical approach was the arthroscopy resulted in some hip kinematics improvements (Bedi et al., 2011a; Rylander et al., 2013, 2011); however, the concern still remains regarding the overall effect of the surgical interventions on hip kinematics during tasks with a larger ROM such as the stair climbing (Rylander et al., 2013) or the squat (Lamontagne et al., 2011b), which did not seem to result in hip function improvements during the follow-up. This opens the gap in the literature to better understand how the surgical correction for FAI would affect the neuromuscular system in a dynamic situation. With the cam-type morphology being more associated with OA development (R. Agricola et al., 2013a, 2013b; Rintje Agricola et al., 2013), would the hip osteochondroplasty surgery indeed result in preservation or merely a delay of progression of hip OA?

2.1.2. Total Hip Arthroplasty

Total hip arthroplasty (THA), is one of the most frequently performed, cost-effective and successful reconstructive procedures in orthopaedic surgery, with over 1 million surgeries being performed every year worldwide (Ethgen et al., 2004; Lavernia et al., 2015; Pivec et al., 2012). The OA is the most prevalent condition requiring this surgical approach, and as its major risk factor – ageing and obesity – are currently increasing in our population (Taruc-Uy and Lynch, 2013), the number of THAs is also expected to increase to a large extent in the future decades (Kurtz et al., 2007). Other than that, THA is indicated in the acute hip fracture, for osteonecrosis, and for childhood hip problems (Canadian Institute for Health Information, 2013).

A traditional hip prosthesis (i.e. single-bearing – SB) replaces the femoral head-neck junction and the acetabular shell by a single ball-and-socket joint. However, an emerging concept in THA prostheses is the dual mobility system that has an additional polyethylene liner between the manufactured femoroacetabular prosthesis.

2.1.2.1. Dual mobility Concept

In order to address issues of high dislocation rates and ROM limitations, a dual mobility (DM) design was developed as a possibility to improve the ordinary single joint prostheses.

The DM design consists of two articulations (Figure 2.3): the first between the manufactured femoral head and the polyethylene liner, and the second at the interface between the convex surface of the polyethylene liner and the acetabular shell (De Martino et al., 2014). Experimental settings have proved that DM design demonstrated greater ROM when compared to SB with similar head sizes, and also the dislocation rates had no occurrences in patients after a follow-up period of 3 to 6 years (Guyen et al., 2007; Langlais et al., 2008; Leclercq et al., 2008).

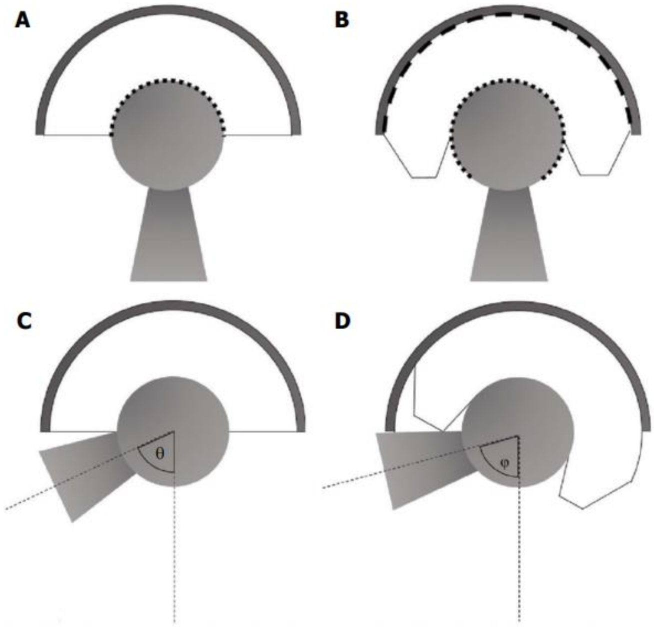


Figure 2.3. Common metal-on-polyethylene (SB) implants (A) include one articulation between the femoral head and the acetabular liner (dashed line). A DM cup (B) consists of two distinct articulations, which allows for greater ROM before impingement of femoral neck occurs – C and D, angle ϕ > angle θ. Reproduced with permission of WJGnet (De Martino et al. 2014).

The DM acetabular cups have been shown to reduce the risk of dislocation (Langlais et al., 2008) and have been typically recommended in cases where the patients are at high risk of dislocation (i.e. patients with previous spinal fusion) (Mudrick et al., 2015) and also in revision surgeries (Philippot et al., 2009) or primary cases for femoral neck fractures (Tarasevicius et al., 2010). The DM cup showed its advantage for increasing ROM *in vitro* (Guyen et al., 2007). However, the benefit of this increased ROM during activities of daily living (ADLs) is unclear. Also, its potential advantages during primary THA have not been examined and concerns regarding its potential for increased wear, intraprosthetic dislocation, and risk of groin pain have risen (De Martino et al., 2017; Hamadouche et al., 2012; Swiontkowski, 2013; Waddell et al., 2016). Recent wear study has found no difference in implant wear between DM and the traditional 22-mm SB metal-polyethylene bearings (Adam et al., 2014). Postoperative improvement in gait and kinematic parameters in THA patients with DM or SB implants have been shown, although they still did not reach the level of the controls (CTRLs) (Kolk et al., 2014; Martz et al., 2016). Therefore, testing a dynamic task that requires a large ROM, such as the deep squat, would be the utmost to determine the benefit of the DM (during primary THA) over the traditional SB implants.

2.2. Musculoskeletal Modelling

MSKMs provide a non-invasive manner to study human movement and predict the effects of interventions, allowing one to study neuromuscular coordination, estimate muscle forces and joint contact loads (Arnold et al., 2001; Damsgaard et al., 2006; Delp et al., 2007, 1990; Hamner et al., 2010). Bones and soft tissue properties can be modelled from experimental data (e.g. cadaveric studies (Brand et al., 1982; Friederich and Brand, 1990; Wickiewicz et al., 1983), medical imaging (Handsfield et al., 2014; Valente et al., 2014)) and combined with marker trajectories and ground reaction forces from functional evaluations, to estimate joint angles, joint moments, muscle forces or joint contact forces (JCF) (Buchanan et al., 2004; Delp et al., 2007; Mantovani, 2016).

OpenSim (Stanford University, USA; Delp et al., 2007; Seth et al., 2018), AnyBody (AnyBody Technology, Denmark; Damsgaard et al., 2006) and SIMM (Motion Analysis, USA; Delp et al., 1990) are some software applications that enable computational modelling of musculoskeletal anatomy and dynamic simulation of movements. These software allow graphic visualization of bones, muscles and ligaments, providing visual feedback of simulations. Many factors of the model can be modified and customized to fit the characteristics of the subjects, and also add-on tools can be used to transfer patient-specific 3D images into the simulator (e.g. NMSBuilder; Valente et al., 2017). For being a free open-source base and offering a repository of free models that can be modified according to the user's specific necessities, OpenSim is the chosen software to be used in this study.

Multiple modelling steps are required to go from the markers trajectories and ground reaction force data experimentally-measured in the research laboratory to the desired simulated muscle forces and JCF outputs, and these will be described next.

2.2.1. Scaling

In an ideal situation, subject-specific bone structure should be used as an input of the model coming from CT exams (Martelli et al., 2015; Valente et al., 2014). However, generic musculoskeletal model is still used most often in simulations, since it does not require a more complex examination, it becomes cheaper and faster, being more feasible to use in clinical applications. The process of matching the measured distances of the reflective markers (obtained in a motion capture recording) with a set of virtual markers of a generic model, which are located in the same anatomical landmarks is called scaling (Delp et al., 2007; Hicks et al., 2015).

Accurate scaling provides more authentic and reliable results for the dynamic movements and moments (Hicks et al., 2015). Although scaling aims to match the subjects' mass properties and the dimensions of the body segments (Delp et al., 2007), the musculotendon properties (i.e. maximum muscle force, tendon slack length, and optimal fiber length) are considered a complex scaling process and are usually not scaled to each specific patient. Although some studies have given attention to this matter (Garner and Pandy, 2003; Kainz et al., 2018; Manal and Buchanan, 2004; Winby et al., 2008), due to the pratical application of this approach, the scaling of these components is commonly done using a length-dependent scale factor (Delp et al., 2007; Hicks et al., 2015).

2.2.2. Inverse Kinematics

To track the participant's motion and calculate joints angles, an inverse kinematics approach is applied with surface marker trajectories as input (Lu and O'Connor, 1999). The experimental markers are tracked and replaced by their respective virtual marker in the pre-existing model, and then decomposition of the orientations of two adjacent body segments into Cardan angles will allow for the calculation of joint angles (Cappozzo et al., 1995; Wu et al., 2002). During this process, an equation (1) tries to minimize the weighted squared error between the experimental and the virtual markers while respecting joint constraints (Delp et al., 2007). Tracking error is computed, and one can choose a weight factor that defines the sensitivity for the tracking accuracy; markers that are considered more reliable (e.g. predefined using a CT scan) can have a higher weight and influence the final result to a greater extent (Mantovani, 2016).

$$(1)\ Squared\ Error = \sum_{i=1}^{markers} w_i(\vec{x_i}^{experimental} - \vec{x_i}^{virtual})^2 + \sum_{j=1}^{joint\ angles} w_j(\theta_j^{experimental} - \theta_j^{virtual})^2$$

$\vec{x_i}^{experimental}$ and $\vec{x_i}^{virtual}$ are the three-dimensional positions of the i^{th} marker, $\theta_j^{experimental}$ and $\theta_j^{virtual}$ are the values of the j^{th} joint angle w_i and w_j are factors that allow markers and joint angles to be weighted differently (Delp et al., 2007).

Inverse kinematics analysis has become widespread in clinical biomechanics, and since computation modelling requires it for both inverse and forward dynamic processing (Hicks et al., 2015), prevention of nonphysiological motions makes inverse kinematics more robust to noise and help to maintain reliable joint kinematics during simulations.

2.2.3. Inverse Dynamics and Residual Reduction Algorithm

Using body kinematics and ground reaction forces (GRF) as input, the net forces and torques at each joint are calculated using equations of motion (i.e. the sum of mass, segmental accelerations and the gravitational force are equal to the generalized forces; Bresler and Frankel, 1950; Pandy and Andriacchi, 2010; Robertson et al., 2004). This process is known as inverse dynamics.

Furthermore, some inconsistencies on kinetic output due to noise in the marker trajectories and joint angle data, or inaccuracies in the model geometry and mass distribution, that can be nominated as residuals, must be reduced in order to provide a better representation of the joint moments (Delp et al., 2007; Hamner et al., 2010). A Residual Reduction Algorithm (RRA) has been introduced to reduce these inconsistencies (Delp et al., 2007). Since the torso is frequently modelled as one rigid body together with the head and is the largest part of the body, its centre of mass location becomes more difficult to be estimated, thus more likely to introduce error (Delp et al., 2007). The RRA will adjust mass distribution and the torso centre of mass location to reduce residuals that may create inconsistencies in the joint forces and moments output (Delp et al., 2007). Still, kinematic and GRF cut-off frequencies and optimization algorithm choices may affect the RRA outcomes (Samaan et al., 2016).

2.2.4. Static Optimization

Invasive *in vivo* measurements of muscle force besides being technically challenging is ethically questionable (Komi et al., 1996; Pourcelot et al., 2005). Approaches such as static optimization (SO) (Ng et al., 2018b; Wesseling et al., 2018) and electromyography-driven (EMG-driven; Sartori et al., 2013, 2012) have been good clinical options in MSK to estimate muscle forces during a movement. The EMG-driven models combine inverse dynamics with surface electromyography (EMG) data as input into a forward muscular model (Sharif Shourijeh et al., 2016), potentially propagating some mistrust of the EMG such as crosstalk (Farina et al., 2004; Winter et al., 1994), movement artifact (De Luca et al., 2010; Reaz et al., 2006) or even processing (De Luca, 1997; Fridlund and Cacioppo, 1986) and normalization (Benoit et al., 2003; Mirka, 1991).

SO uses musculoskeletal geometry and joint moments calculated from inverse dynamics to estimate individual muscle forces at each instant in time (Edwards et al., 2016; Penrod et al., 1974). The forces are calculated using a given cost-function (e.g. the sum of the muscle stresses squared) that imposes the equality between the external and internal torques (Brand et al., 1982; Edwards et al., 2016). Additionally, other constrains (e.g. maximal muscular force, compressive joint load, or passive elements constrains) may be used to improve the muscle forces outputs. SO has a low computational cost (Morrow et al., 2014; Raikova and Aladjov, 2002), which makes it interesting and popular for muscle forces prediction.

Validation of the SO muscle activations is commonly done with EMG linear envelopes (Hicks et al., 2015; Lund et al., 2012). Since EMG data are not inputted in the simulation, it seems to provide a credible approach for this task (Crowninshield et al., 1978; Glitsch and Baumann, 1997).

However, it is important to highlight that EMG excitations and the muscle activation predicted through SO are not the same phenomenon. While there is a time lag (electromechanical delay) between the onset of the neural drive (EMG) and the beginning of the force production (Hug et al., 2015), the SO considers that activation corresponds to instantaneous force production and also does not acknowledge muscle and kinematic states. In this manner, shifting the SO activations over the EMG excitations is an indispensable step during the validation procedure (Hicks et al., 2015).

2.2.5. Joint contact forces

The only way to perform a direct *in vivo* JCF measurement is using an instrumented prosthesis that will replace the joint itself, and in a hip scenario will take measurements at the femoral head (Brand et al., 1994; Davy et al., 1988; Stansfield et al., 2003). However, its major disadvantage is that it can only be performed in a surgical population that need a joint replacement. Also its loading outputs are based in an artificial structure with a completely different material condition and mechanical behaviour when compared to a human bone/joint. Also, the inverse dynamics only considers external and inertial forces and denies the contribution of muscles, motors and other actuators when calculating the generalized forces at the joints.

Computational simulations of MSK systems on the other hand, use the estimated muscle forces as an input (Modenese et al., 2011), which have been speculated as a major contributor of the total HCF (Lu et al., 1998; Ng et al., 2012; Steele et al., 2010), and a suitable approach to this cause (Stansfield et al., 2003). The JCF is performed in OpenSim as *JointReaction* in the Analysis tool and calculates the reaction load acting at the joint centre of both child and parent bodies (Steele et al., 2012). In the HCF case, the simulation output will report the vectors (anterior-posterior, superior-inferior and medial-lateral) of compressive forces that the hip carries to prevent translation of the head of the femur (ball) through the acetabular shell (socket).

2.2.6. *Verification and Validation of MSK Simulations*

Modelling the MSK systems and simulating its motion may involve methods that represents i) musculoskeletal geometry, includes properties of bodies, its mass and inertial properties; which include the joints, the connection between anatomical segments and the muscle geometry; ii) muscle-tendon dynamics, which includes the parameter for the Hill-type muscle model; iii) multibody dynamics, which provides the equations of motion to allow to calculate accelerations, velocities and position over time; iv) muscle control, which coordinates the muscle forces that drive motion; and v) bone contact forces, characterized by the internal joint loading (Hicks et al., 2015). The modelling choices of each of these components may affect the simulations outputs and therefore, verification and validation by comparing experimental data and testing robustness are wanted (Hicks et al., 2015). Considering that the crucial components of simulations that affect this thesis are the musculoskeletal geometry, the neural control and the contact forces, these will be more detailed in the sequence.

Anatomical based models of musculoskeletal geometry represent physiological joint kinematics and muscle path geometry to calculate internal joint loads and accelerations produced by muscles during movements (Hicks et al., 2015). Specifically about the muscle geometry, MTUs can include detailed path information, and muscle moment arms (MA) can be measured in cadavers directly using tendon excursion or load experiments (An et al., 1983; An et al., 1984; Spoor and Leeuwen, 1992). MA can also be estimated experimentally using imaging, such as Magnetic Resonance Imaging (MRI), Computadorized Tomography (CT) scans, or digitalization of cadavers to determine the line of action of a muscle (An et al., 1984; Spoor and Leeuwen, 1992; Arnold et al., 2000). Muscle origin and insertions points are combined with via points and wrapping surfaces to represent constrains from the retinacula and to prevent muscles from penetrating bones or failing its line of action (Hicks et al., 2015). And although more recent models (Rajagopal et al., 2016; Lai et al., 2017) are based in a both cadaver and MRI muscle architecture and skeletal geometry data set, making them suitable for sagittal plane activities such as walking, running and cycling; experimentally measured moment arms of the lower limbs (Németh and Ohlsén, 1985) still offer a limited output range, which makes it challenging to design wrapping surfaces or via points that truly represents physiological moment arms for tasks with large ROM, such as deep squatting.

In some cases, an experimental input from EMG is used to estimate the neural command and drive motion or muscle dynamics (Sartori et al., 2013, 2012). In tracking simulations, however, a model of neural control is not required since they depend only on joint angles and net moments as input to predict muscle activations (e.g.: static optimization; Hicks et al., 2015). In this case, comparing the predicted muscle activity to measured EMG from the same experiment as a mechanism of validating the neural control. To do that so, one should determine the agreement of onset/offset timing between the experiment and the simulations (Hicks et al, 2015), after accounting for electromechanical delay between EMG and simulated activation (Corcos et al., 1992). Additionally, some tracking simulations also apply extra moments or forces (i.e. reserve actuators being used) in order to supply for insufficient muscle force-generating capacity or ignored passive structures (Delp et al., 2007); in case of reserve actuators being active in a model, it is recommended to verify if their moments are smaller than 5% of the total net joint moments (Hicks et al., 2015).

In inverse simulations, the quality of the joint reaction forces is limited by the quality of the input measurements (i.e. kinematics; kinetics; muscle forces), as well as the joint definitions (i.e. joint type and restrictions), segment inertial properties and the muscle geometry. Therefore, avoiding eventual errors in the inputs can also avoid the propagation of errors to the outputs. The best practices of validation of contact forces requires that the output forces stay within 2 standard deviation (SD) of experimental joint forces from instrumented implants for similar motion (Hicks et al., 2015).

3 Study Design

Gaps in the Literature | Framework | General Methods

3.1. Gaps in the Literature

From the review of literature on hip surgical interventions due to either FAI or OA (Chapter 2), it was apparent that not many gait analysis studies have been performed in postoperative patients, and these are mostly done in a level walking condition, overlooking tasks with higher ROM. Moreover, the previous two chapters of this thesis highlighted three main areas for development in this field:

1. Examine the muscle strength and lower limb mobility in asymptomatic individuals with a cam-type deformity that may differentiate them from the symptomatic patients;
2. Enhance the understanding of the postoperative effect of both hip surgical procedures, hip preservation and THA on hip biomechanics during a high ROM task such as the deep squat;
3. Simulate the dynamic muscle forces and HCF in postoperative FAI in different ADLs.

Anatomical, physiological and functional alterations occur in both patients FAI and THA (Huo et al., 2009; Mario Lamontagne et al., 2015; Ng et al., 2016b), and disregarding the surgery procedures choice, a change on these alterations are expected in a post-operatory (post-op) scenario. Still, a follow up usually finds a significant reduction in pain, but besides patients reporting an improvement in general locomotion, it is difficult to determine important changes in gait analysis or functional capacity evaluations using statistical analysis of single analyzed variables (Lamontagne et al., 2011b; Malagelada et al., 2015; Ng et al., 2014; Rylander et al., 2011). Therefore, a more robust biomechanical evaluation providing the assessment of hip muscle forces from pre- and post-op conditions becomes necessary to obtain HCF and verify if there are alterations that may provide some indication about the most appropriate treatment (surgical or conservative) in FAI patients that are still borderline in clinical signs and diagnostic imaging.

3.2. Framework

With the objective to better understand the consequences of hip surgery on cam FAI and hip OA patients, the purpose of this research was to address the question at large: *what are the changes in muscle performance and HCF after hip surgical procedure during a deep squat task?* With a cohort that includes patients who underwent hip preservation surgery (for cam FAI) and THA (for hip OA), asymptomatic cam FAI and healthy CTRL participants, this research was unfolded into six main studies, written as manuscripts:

1. **Changes in Neuromuscular Control in Asymptomatic Individuals during Squat**: which examined isometric muscle strength and lower-limb kinematics in order to test if they would differentiate them from preoperative symptomatic patients (Chapter 4);
2. **Cam FAI Patients after Hip Preservation Surgery – Squat**: which evaluated isometric muscle strength, squat kinematics and muscle activity in a 2-year follow-up of the corrective surgery (Chapter 5);
3. **OA Patients after THA – Effect of DM during a Squat Task**: which compared lower-limb kinematics of patients with either DM or SB implants during a squat task (Chapter 6);
4. **Gait Modelling in Postoperative Cam FAI**: which examined hip muscle forces and HCF using MSK modelling during level walking task (Chapter 7);
5. **Adaptation of a Generic Model to Perform a Squat Task**: which tested the effect of including wrapping surfaces at the knee and hip joints in a generic MSKM in order to control the muscles moment arm to perform a squat task (Chapter 8);
6. **Squat Modelling in Postoperative Cam FAI**: which examined hip muscle forces and HCF using MSK modelling during a deep squat task (Chapter 9).

While the general methods will be described in the next section, an illustrative framework of the methodology of the six studies that involve this thesis is presented in figure 3.1.

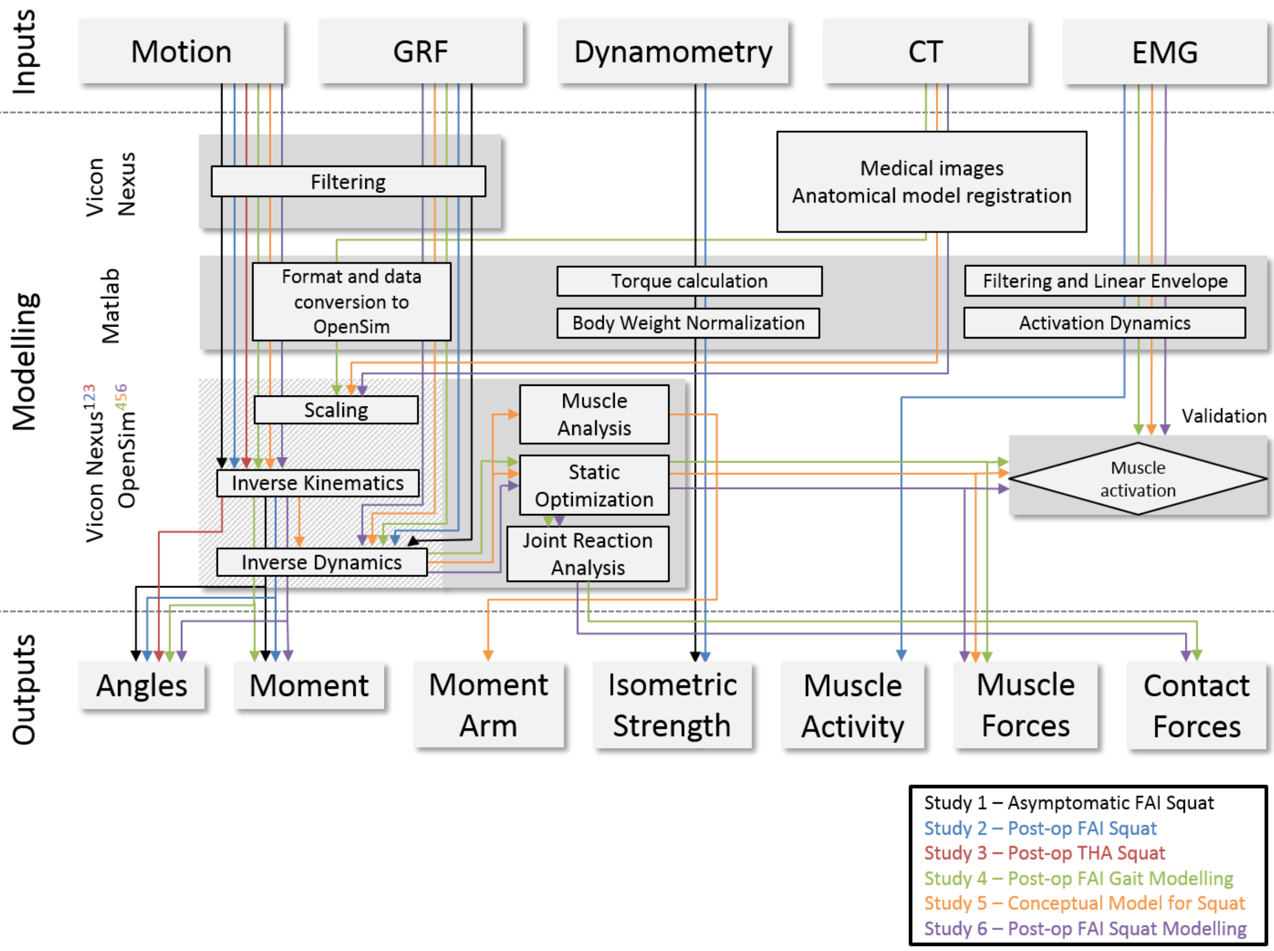


Figure 3.1. Thesis scheme: a methodological framework. The inputs of the simulation were: motion data (marker trajectories), ground reaction forces (GRF), isometric dynamometry, pelvic and knee Computed Tomography (CT) and hip muscles electromyography (EMG). Trajectories and GRF were filtered and converted into OpenSim compatible format files. From the CT a subject-specific knee and hip bone dimensions were measured, and their distances were imported into OpenSim. EMG data were filtered to estimate the muscle activation. The musculoskeletal model was scaled, and inverse kinematics, dynamics, static optimization and joint reaction analysis were run to estimate angles, torques, and muscle and contact forces. The muscle activations estimated from static optimization was compared to those obtained by EMG data for indirect validation. The torque of the measured isometric dynamometric data was calculated and was body-weight normalized in order to define the isometric strength. The coloured lines highlight the parts of the methodological framework involved in the different studies.

3.3. General Methods

3.3.1. Participants

This thesis was part of two larger research programs: one focused in FAI patients, which involved 68 participants (52 took were included in the study); and another focused in patients who underwent THA surgery, which involved 43 participants (36 were included in the study). A summary of the demographic information of the patients that took part in one of the studies of this thesis can be found below on table 3.1.

Table 3.1. Summary of the participants' demographics used in both surgical studies, hip preservation for FAI, and THA.

Demographics		Preoperative			Postoperative			
		Number (M/F)	Age (y)	BMI (kg/m^2)	Number (M/F)	Age (y)	BMI (kg/m^2)	Follow-up (mo)
Hip Preservation Surgery	FAI	14/2	38.5 ± 8.0	26.8 ± 5.0	11/0	36.2 ± 7.4	25.6 ± 3.6	25.1 ± 1.1
	FAD	15/3	32.5 ± 7.1	25.7 ± 1.9	N/A			
	CTRL	16/2	32.8 ± 7.0	25.5 ± 3.3	N/A			
THA Surgery	DM	8/4	63.1 ± 5.6	28.1 ± 2.7	8/4	63.6 ± 5.6	28.3 ± 0.9	6.8 ± 0.8
	SB	10/2	62.7 ± 4.9	29.6 ± 4.8	10/2	63.3 ± 4.9	29.4 ± 3.9	6.8 ± 1.3
	CTRL	6/6	62.5 ± 9.5	25.8 ± 3.8	N/A			

Data are reported as mean ± SD unless otherwise indicated. BMI, body mass index; CTRL, healthy control; DM, dual-mobility; F, female; FAD, asymptomatic with femoroacetabular deformity; FAI, symptomatic with femoroacetabular impingement; M, male; mo, months; N/A, not applicable; SB, single bearing; THA, total hip arthroplasty; y, years.

All the recruitment was done by the Clinical Research staff of the Division of Orthopaedics at The Ottawa Hospital, and it was based on the following selection criteria:

Hip Preservation Surgery:

Symptomatic cam-type impingement (FAI): Only patients diagnosed with cam-type FAI were included, identified by an alpha angle larger than 50.5º in the axial or 60º in the radial views on CT data (Beaulé et al., 2012; Khanna et al., 2014; Nötzli et al., 2002) and presented a positive FADIR impingement test (Byrd, 2014; Lamontagne et al., 2015). The participants must have had experienced hip pain longer than six months near the groin/lateral aspect of the hip, and were assigned for FAI corrective surgery. Patients were excluded if they had any musculoskeletal or neurological disorders, degenerative diseases, previous major lower limb injuries, exhibited signs of hip osteoarthritis (Tönnis grade > 1; Tönnis and Heinecke, 1999), or a BMI greater than 35 kg/m^2. Four of the FAI patients underwent corrective surgery, an osteochondroplasty of the femoral head-neck junction, with an open approach with surgical dislocation and 7 had surgery with an arthroscopic approach, all performed by the same surgeon. A recommended 6-week physiotherapy program followed surgeries. Their follow-up motion caption analysis happened with a minimum of 2 years after the surgery.

Asymptomatic with femoroacetabular deformity (FAD): Participants recruited from the general population who had been selected to participate of the study as healthy controls, since they have never experienced hip pain, however their CT medical images had revealed signs of femoroacetabular deformity consistent with cam-type FAI (i.e. alpha angles larger than 50.5º and 60º in the axial or radial views, respectively). They have not performed a follow-up motion analysis.

Healthy control (CTRL): Participants recruited from the general population, and matched for sex, age, and body mass index (BMI) to the FAI group. Participants were excluded from the CTRL group in the presence of dysplasia, hip osteoarthritis, cartilage narrowing, previous major lower limb injuries, and pain. They have not performed a follow-up motion analysis. All CTRL went through medical imaging.

Total Hip Arthroplasty Surgery:

Degenerative osteoarthritis (OA): Participants who were diagnosed with non-inflammatory degenerative hip OA and were assigned for a unilateral primary THA. The patients were prospectively randomized to either a DM or an SB implants using the Medacta system (Medacta International, Switzerland). The patients suffering from any other lower limb joint disorders other than the one for THA, with any other joint arthroplasty at the ipsilateral or contralateral limb, with evidence of active infection, with neurologic or musculoskeletal disease that may adversely affect gait or weight-bearing, with neuropathic joints, and with requiring structural bone grafts were excluded from participating in the study. Participants who were unwilling to perform the squat task because of either fear or pain, or that were unable to reach the minimum squat depth of 90% of their leg length or because the patients were also excluded of the study. Their follow-up motion caption analysis happened with a minimum of 6 months after the surgery.

Healthy control (CTRL): Participants recruited from the general population, and matched for sex, age, and body mass index (BMI) to the THA groups. Participants were excluded from the CTRL group in the presence of dysplasia, hip osteoarthritis, cartilage narrowing, previous major lower limb injuries, and pain. They have not performed a follow-up motion analysis.

All the participants were properly instructed before the procedure and signed an informed consent form. The study was approved by the Ottawa Hospital Research Ethics Board and the University of Ottawa Health Sciences and Science Research Ethics Board (appendix).

For the study presented in Chapter 4 from the initial 68 participants, 16 were excluded of the study for various reasons (CT malperformance: 1 FAI; recent surgical treatment: 1 FAI; data collection complications: 1 FAI and 1 FAD; EMG malfunction or poor signals: 2 FAI, 4 FAD, and 1 CTRL; did not perform a minimum of 90º of knee flexion during squat: 1 FAI, 2 FAD, and 2 CTRL), and a total of 52 participants were included in the analysis. The studies presented in Chapters 5 & 7 included the same participants and 11 of the 12 FAI participants who underwent hip preservation surgery for cam FAI (4 open and 7 arthroscopy) and who came back for postoperative motion analysis, the exclusion of the solo participant was due to postoperative obesity. The study reported in the Chapter 6 measured seven surgical participants excluded from the initial cohort (had surgery elsewhere: 1 SB; lost follow-up: 1 DM; unable to perform minimum squat requirement: 2 SB; refuse to squat preoperatively due to pain: 1 SB and 1 DM; did not adhere to squat protocol: 1 DM). The study presented in Chapter 8 that customized a model to perform a deep squat task involved only one participant. Lastly, the clinical study reported in Chapter 9 included 10 of the 12 FAI participants who underwent hip preservation surgery for cam FAI (3 open and 7 arthroscopy) and who came back for postoperative motion analysis, the exclusion of the participants was due to postoperative obesity and inappropriate feet position during the squat.

3.3.2. Equipment

The motion capture took place at the Human Movement Biomechanical Laboratory (University of Ottawa), and its apparatus included: ten infrared Vicon MX-13 cameras (VICON, Oxford, UK) sampled at 200 Hz, two fixed Bertec force plates (models FP4060-08, Bertec Corporation, Columbus, US) and two portable Kistler force plates (models 9286BA, Kistler Instruments Corp, Winterthur, CH) always sampled at 1000 Hz. Ground reaction forces and marker trajectories were recorded and synchronized through Vicon Nexus software (versions 1.8 and 2.5, VICON, Oxford, UK). Electromyography (EMG) signals were acquired at 1000 Hz with FreeEMG300 (BTS Bioengineering, Milan, Italy) with 16 channels and Bagnoli Desktop EMG System (Delsys, Boston, US) with eight channels. Participants wore a skin-tight (i.e. spandex) black shorts and short sleeve shirts for participants' comfort. Reflective markers with 12mm of diameter were placed according to the University of Ottawa Motion Analysis Model (UOMAM; Mantovani and Lamontagne, 2016) and were placed preferably on the skin. EMG probes were placed according to the SENIAM guidelines (Hermens et al., 1999) and under the spandex suit to avoid wobbling.

The CT scans were performed at The Ottawa Hospital, and the images were acquired with either the Toshiba Acquilion (Toshiba Medical Systems Corporation, Otawara, Japan) or the Discover CT750 (GE Healthcare, Mississauga, Canada).

3.3.3. Protocol

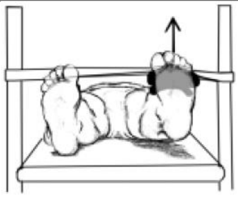
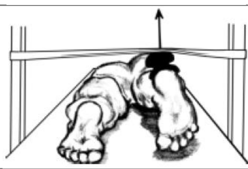
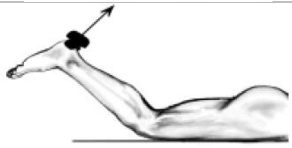
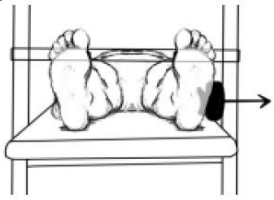
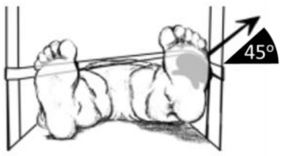
Although this thesis incorporates the results of two larger research programs (i.e. THA and FAI), all participants underwent the same motion capture protocol. The main exception was that only the participants included in the FAI study underwent the CT scan analysis. Whereas the details of the protocol can be directly found in the specific manuscripts of this thesis (chapters 4 to 9), a brief description will follow this section.

Data collections for the FAI study were initiated at The Ottawa Hospital where the CT scans were performed. Prior to the CT imaging test, participants were instrumented with eight radiopaque surface markers (markers with a metallic core that reflect during the scan), placed on the pelvis and on the knees, in correspondence with the following bony anatomical landmarks: left and right anterior and posterior superior iliac spines (ASIS and PSIS, respectively), and left and right medial and lateral epicondyles. With the option of repositioning the radiopaque markers during the scan in order to ensure the best marker placement with the bony landmarks, the radiopaque markers were further replaced by the reflective markers used in motion capture analysis. The analysis of this placement has shown a high inter-rater agreement for the pelvis bony landmarks (i.e. median error of 1.1 mm on ASIS and 1.3 mm on PSIS; Mantovani et al., 2016) and we consider that increased the placement reliability within our motion capture protocol. After finishing the CT scan at the hospital, the participants were transferred to the Human Movement Biomechanics Laboratory (University of Ottawa) to progress to the motion capture protocol.

Once in the laboratory, participants from both studies initiated the preparation for the data collection. They read and signed the written consent form, and were asked to wear the black spandex suit in order to have a tight fit. Anthropometric measurements such as weight, height, leg length, and knee and ankle widths were recorded. After, the participants were asked to perform a 5 minutes warm-up in a cycle ergometer and were also invited to perform 5 minutes of uninstructed stretching.

EMG probes were then placed bilaterally on *rectus femoris, gluteus maximus, biceps femoris, semitendinosus, gluteus medius* and *tensor fasciae latae* muscles according to the SENIAM guidelines (Hermens et al., 1999). A protocol of a maximum voluntary isometric contraction (MVIC) was then performed in order to both normalize the signals collected during the dynamic contractions (Benoit et al., 2003), and to record the maximum isometric strength of the hip movements of the participants. A hand-held dynamometer (HHD, Manual Muscle Testing System Model 01163, Lafayette Instrument, Lafayette, US) was used during the MVIC recording according to the postures detailed on Table 3.2 (Mantovani, 2016). Each MVIC task last 5 seconds and was performed twice with a 30 seconds rest period in between trials.

Table 3.2. Posture description for Maximum Voluntary Isometric Contraction protocol

Muscle	**Task**	**Posture**	**Illustration**
Rectus Femoris	Hip Flexion	Supine with feet shoulder-width apart. HHD is placed between the ankle and the strap.	
Gluteus Maximus	Hip Extension	Prone with feet shoulder-width apart. HHD is placed between the heel and the strap.	
Biceps Femoris and *Semitendinosus*	Knee Flexion	Prone with knee bent at 45º. HHD is placed between the heel and the strap.	
Gluteus Medius	Hip Abduction	Supine with feet shoulder-width apart. HHD is placed between the ankle and the lateral beam.	
Tensor Fasciae Latae	Hip Flexion with Abduction	Supine with feet shoulder-width apart. HHD is placed at 45º between the ankle and the corner between the lateral beam and the strap.	

Following the MVIC protocol, the participants were outfitted with the reflective markers according to the UOMAM marker set (Mantovani and Lamontagne, 2016) – illustrated in the appendix (Figure A.1). The radiopaque markers on the pelvis and knees placed during the CT scan were then replaced by the reflective markers and tapped over to guarantee their placement location during the entire data collection.

The motion capture then proceeded to a static trial, where the participants had their feet parallel, pointing forward and one on each force plate and hip-width apart, and with their arms flexed forward at the shoulder height with the palms facing downwards. The participants were also instructed to maintain their pelvis in a natural/comfortable position during the static trial. The static trial was further used for labelling and scaling purposes. Following the static trial, the participants performed specific ADL tasks that included gait, inclined walking (up and down), deep squat, sit-to-stand, stand-to-sit, stairs (up and down) and dynamic hip ROM[2]. While the tasks were performed in a random sequence, for this thesis the level walking and the deep squat were the only two tasks observed. Although no external load was added to any of the tasks, we acknowledge that the amount of consecutive tasks, including the practice trials that the participants were allowed to perform to get comfortable with the motion, may have induced the participant to fatigue and/or generated a training effect, which may cause bias in these studies. However, the time in between different tasks, necessary to setup the equipment, was considered long enough to allow the participants to rest in between tasks.

The level walking task was executed at a self-selected pace, and the practice trials allowed the researcher to evaluate the participants' stride length in order to determine the starting point of the trial with the goal of hitting three consecutive force plates without aiming them. The initial position of the deep squat task was the same as the static position, and the participants were instructed to squat the deepest as possible while keeping their heels on the ground, and maintaining their balance. The participants were also instructed to keep their arms straight up during the whole trial. Also, although the squat trial was also performed in a self-selected pace, the participants were instructed to perform the movement at a slow pace, with a controlled rhythm (2s descending, 2s at the bottom, 2s ascending and 2s at the top), thus avoiding abrupt differences among the performances. Both tasks the level walking and the deep squatting had five valid trials recorded.

[2] Due to a Research Ethics Board suggestion, participants of the THA study were instructed to bring and wear their own comfortable sports shoes during the motion analysis, while the participants of the FAI study remained barefoot during the entire data collection protocol.

3.3.4. Data Processing

Motion capture data was initially processed in the Vicon Nexus software (versions 1.8 and 2.5, VICON, Oxford, UK) where the marker trajectories were labelled and filtered (Woltring, mean squared error = 15 mm^2). The ground reaction forces were also filtered (zero-lag, fourth order Butterworth, cut-off 6Hz), and the trials were cropped according to the events. For the studies 1, 2 and 3 (chapters 4, 5 and 6), the data was still modelled in the Vicon Nexus software, according to the UOMAM (Mantovani and Lamontagne, 2016) to obtain joint angles and moments; whereas for the studies 4, 5 and 6 (chapters 7, 8 and 9) the filtered data were prepared to OpenSim file format (i.e. converted from the .c3d to .trc and .mot files format; Mantoan et al., 2015) for further processing. All data were cropped according to the events (walking: stance phase, based on the second first and second last frames of the GRF on the involved force plate; squatting: based on the maximum hip extension point – standing – and lowest depth point – squatted, and separated into descending and ascending phases) and time normalized for interpretation.

All simulations were performed in OpenSim 3.3 (Delp et al., 2007) using the model described in chapter 8, which was elaborated based on the full-body MSKM designed by Rajagopal (Rajagopal et al., 2016) and updated by Lai (Lai et al., 2017). By default within the full-body bony geometry, the model includes 37 degrees of freedom to define joint kinematics, Hill-type models (Millard et al., 2013; Zajac, 1989) of 80 muscle-tendon units actuating the lower limbs, and 17 ideal torque actuators driving the upper body. Also, the model's musculotendon parameters are derived from previous anatomical measurements of 21 cadaver specimens (Ward et al., 2009) and magnetic resonance images (MRI) of 24 young healthy subjects (Handsfield et al., 2014). Consistently, for every participant included in the modelling study (chapters 7 and 9), the generic MSKM was scaled with the markers that were previously placed during the CT scan given a weight 100 times higher than the other ones. Inverse kinematics and inverse dynamics tools were used to compute joint angles and net joint moment for each degree of freedom, while the static optimization tool was used to compute muscle forces, which minimized the sum of squared muscle activation (Buchanan and Shreeve, 1996; Todorov and Jordan, 2002; van Bolhuis and Gielen, 1999). A peak force of 1N (gait) or 10N (squat) was defined for the reserve actuators for the three hip coordinates in order to satisfy the optimization outputs. The *JointReaction* analysis tool calculated HCF as three-dimensional vectors acting on the acetabulum and expressed either in the pelvic (gait study – chapter 7) or the femoral (squat study – chapter 9) coordinate systems. Both muscle forces and HCF were normalized to body weight.

The EMG data were processed using custom software designed in MATLAB (MathWorks, Natick, US), where all signals were band-pass filtered (zero-lag, 4th order Butterworth, cut-off 20-400Hz) to remove bias and skin motion artifacts, full-wave rectified, and low-pass filtered at 6 Hz (zero-lag fourth-order Butterworth filter) in order to obtain the linear envelope. The linear envelopes of each muscle were then amplitude-normalized by their respective MVIC values, obtained an average activation of the 2-central-seconds' window of the MVIC tasks.

The variable of interest constituted of isometric strength measures, joint angles, joint moments, muscle activity, muscle moment arm (study 5 – chapter 8), muscle forces and HCF. Also, the statistical analysis done is described in each one of the studies.

II Asymptomatic Individuals

4 Asymptomatic Participants with Femoroacetabular Deformity Demonstrate Stronger Hip Extensors and Greater Pelvis Mobility during the Deep Squat Task

Danilo S. Catelli [1,2] | Erik Kowalski [1] | Paul E. Beaulé [3] | Kevin Smit [3] | Mario Lamontagne [1,3,4]

[1]School of Human Kinetics, University of Ottawa, Ottawa, Ontario, Canada.
[2]CAPES Foundation, Ministry of Education of Brazil, Brasilia, Brazil.
[3]Division of Orthopaedic Surgery, University of Ottawa, Ottawa, Ontario, Canada.
[4]Department of Mechanical Engineering, University of Ottawa, Ottawa, Ontario, Canada.

Contents of this chapter published as an article in *The Orthopaedic Journal of Sports Medicine®* | July 2018

Catelli DS, Kowalski E, Beaulé PB, Smit K, Lamontagne M, Asymptomatic Participants with Femoroacetabular Deformity Demonstrate Stronger Hip Extensors and Greater Pelvis Mobility during the Deep Squat Task. *Orthop J Sports Med*, 2018; 6(7):2325967118782484.

DOI: 10.1177/2325967118782484

4.0. Abstract

Background: Cam-type femoroacetabular impingement (FAI) is a femoral head-neck deformity that causes abnormal contact between the femoral head and acetabular rim, leading to pain. However, some individuals with the deformity do not experience pain and are referred to as having a femoroacetabular deformity (FAD). To date, only a few studies have examined muscle activity in patients with FAI, which were limited to gait, isometric and isokinetic hip flexion, and extension tasks.

Purpose: To compare (1) hip muscle strength during isometric contraction and (2) lower limb kinematics and muscle activity of patients with FAI and FAD participants with body mass index-matched healthy controls during a deep squat task.

Study Design: Controlled laboratory study.

Methods: Three groups of participants were recruited: 16 patients with FAI (14 male, 2 female; mean age, 38.5 ± 8.0 years), 18 participants with FAD (15 male, 3 female; mean age, 32.5 ± 7.1 years), and 18 control participants (16 male, 2 female; mean age, 32.8 ± 7.0 years). Participants were outfitted with electromyography electrodes on 6 muscles and reflective markers for motion capture. The participants completed maximal strength tests and performed 5 deep squat trials. Muscle activity and biomechanical variables were extrapolated and compared between the 3 groups using 1-way analysis of variance.

Results: The FAD group was significantly stronger than the FAI and control groups during hip extension, and the FAD group had a greater sagittal pelvic range of motion and could squat to a greater depth than the FAI group. The FAI group activated their hip extensors to a greater extent and for a longer period of time compared with the FAD group to achieve the squat task.

Conclusions: The stronger hip extensors of the FAD group are associated with a greater pelvic range of motion, allowing for greater posterior pelvic tilt, possibly reducing the risk of impingement while performing the squat, and resulting in a greater squat depth compared with those with symptomatic FAI.

Clinical Relevance: The increased strength of the hip extensors in the FAD group allowed these participants to achieve greater pelvic mobility and a greater squat depth by preventing the painful impingement position. Improving hip extensor strength and pelvic mobility may affect symptoms for patients with FAI.

4.1. Introduction

Cam-type hip morphology, which in some individuals is associated with femoroacetabular impingement (FAI), affects the articular surfaces of the anterosuperior portion at the femoral head-neck junction[4,14] and has been implicated as a cause of labral-chondral damage as well as an early cause of hip osteoarthritis.[14,24,33,49] Several studies have suggested that cam-type FAI, defined by an aspherical femoral head and/or insufficient femoral head-neck offset[25,50] with larger alpha angles,[41] could result in anterior hip or groin pain, labral tears, and damage to the acetabular articular cartilage.[7,15,25,51] Symptomatic patients demonstrate a positive flexion, adduction, and internal rotation (FADIR) test finding,[7,35,42] greater superoposterior femoral coverage and a higher pelvic incidence,[16,40] a decreased neck-shaft angle,[38,39] and decreased range of motion (ROM),[27,29,30] which may result in greater mechanical stress at the anterosuperior portion of the acetabulum.[38] However, some individuals with a cam deformity may not experience symptoms or clinical signs,[1,17,18,26,28,37,46] and to our knowledge, no study has compared muscular activity in symptomatic patients with FAI to asymptomatic individuals with a femoroacetabular deformity (FAD) during a deep squat task.

When compared with healthy population, patients with FAI were shown to have muscle weakness in all hip muscle groups except for the hip extensors and internal rotators.[9] During maximal isometric hip flexion, electromyography (EMG) activity was lower in patients with FAI compared with healthy participants[9] for the tensor fasciae latae (TFL) but not the rectus femoris (RF) muscle. Another study that examined hip flexion strength under isometric and isokinetic conditions showed hip flexor weakness in patients with FAI under both conditions compared with healthy control participants, but no differences existed in muscle activity.[8] To date, only a few studies have reported on muscle activity in patients with FAI during daily activities.[8,9,11] However, these studies have only compared presurgical symptomatic patients with FAI with healthy control participants; they did not track or include asymptomatic individuals with cam morphology (FAD) in motion analysis. It remains unclear if this muscle weakness is part of the pathological process of FAI or something that could be modified by conservative treatment. Also, muscle weakness and muscle imbalance can be determinants for joint stability.[53]

Usually reported as pain-free and not involving the range of hip impingement, subtle gait alterations in FAI patients have been reported.[6,23,27,48] Patients with FAI have also reported pain triggered by sitting in a low chair.[32] A functional task that requires large sagittal hip and pelvic ROM and that may lead to impingement, such as squatting, may be a more challenging task that better reproduces the motion of sitting; squatting may also be demanding enough to evaluate lower limb function.

The objectives of this study were (1) to compare hip muscle strength during maximum voluntary isometric contraction (MVIC) and (2) to compare lower limb kinematics and muscle activity of symptomatic patients with FAI and asymptomatic participants with FAD during a deep squat task and compare the results with those of healthy body mass index–matched controls (CTRL group). It was hypothesized that the FAI group may show less hip flexion and hip abduction during squatting when compared with the FAD and CTRL groups. It was also hypothesized that patients with FAI would have weaker hip flexor muscles and consequently higher normalized muscle activation when performing the squat trials. This was expected as a way for patients with FAI to compensate for their weakness while performing the same task.

4.2. Methods

4.2.1. Participants

After approval from the hospital's and university's ethics committees, 16 patients with FAI were initially recruited by clinical research staff from the senior orthopaedic surgeon's (P.E.B.) clinical practice during a 2-year recruitment period at the local hospital. Thirty-six participants (31 male, 5 female) were recruited from the community to serve as controls. Initial radiographs of all the CTRL participants were taken to screen for the presence of a cam-type deformity. Several CTRL participants showed the presence of a cam deformity but did not experience any clinical symptoms. After this finding, the decision was made to have all participants (with and without the cam-type deformity) undergo full radiographic screening using low-dose computed tomography (CT). CT from the pelvis to knee was performed using a clinical CT scanner (Aquilion CT Scanner [Toshiba] or Discovery CT750 HD [GE Healthcare]). The CT scans of all participants were read by a musculoskeletal radiologist to confirm the presence of a cam deformity. Alpha angles greater than 50.5° (anteriorly at the 3:00 clock-face position about the femoral neck) or 60° (anterosuperiorly at 1:30) were considered positive for cam morphology.[3,17,28,45,52] Participants with neurological or musculoskeletal disorders, degenerative diseases, or any previous major lower limb injuries or surgeries were excluded from the study.

Based on the CT scans, we divided our cohort of participants into 3 groups: those with symptomatic cam-type FAI, those with a cam deformity but no symptoms (FAD), and the CTRL group (Table 4.1). Patients with FAI had experienced hip pain for longer than 6 months near the groin/lateral aspect of the hip and had produced a positive impingement test (FADIR) result.[7,31] The CT scans of participants with FAD indicated the presence of a cam deformity, but these participants did not experience any hip pain or produce a positive impingement test finding. CTRL participants did not have the presence of a cam deformity as indicated by CT, nor did they experience any hip pain or produce a positive FADIR test finding. As a matter of comparison, the affected hip in patients with FAI with a bilateral cam deformity was the one with greater clinical signs; in the FAD group, the side of interest was the one with the larger alpha angle, and the selected hip for the CTRL participants was based on their dominant leg.

The initial cohort was composed of 68 participants; however, 16 were excluded for various reasons (CT malperformance: 1 FAI; recent surgical treatment: 1 FAI; data collection complications: 1 FAI and 1 FAD; EMG malfunction or poor signals: 2 FAI, 4 FAD, and 1 CTRL; did not perform a minimum of 90° of knee flexion during squat: 1 FAI, 2 FAD, and 2 CTRL). A total of 52 participants were included in the analysis (Table 4.1).

Table 4.1. Demographics and patient-reported outcome measures for FAI, FAD and CTRL participants[a].

		FAI (n = 16)	FAD (n = 18)	CTRL (n = 18)
Presence of cam deformity		Yes	Yes	No
Positive impingement test		Yes	No	No
Sex (male/female)		14/2	15/3	16/2
Height (m)		1.74 ± 0.07	1.77 ± 0.09	1.74 ± 0.09
Age (years) [b]		38.5 ± 8.0	32.5 ± 7.1	32.8 ± 7.0
BMI (kg/m^2)		26.8 ± 5.0	25.7 ± 1.9	25.5 ± 3.3
Axial alpha angle (°)		57 ± 6	58 ± 7	43 ± 4
Radial alpha angle (°)		67 ± 5	70 ± 7	52 ± 5
Femoral neck-shaft angle (°)[b]		123±3	127±3	127±2
HOOS	Symptoms[b]	65.6 ± 14.4	95.3 ± 7.2	97.8 ± 5.7
	Pain[b]	66.3 ± 16.3	98.2 ± 4.6	98.8 ± 4.2
	Activities of daily living[b]	75.7 ± 17.9	99.6 ± 1.2	100 ± 0
	Sport/Recreation[b]	56.3 ± 22.0	97.9 ± 6.4	99.7 ± 1.5
	Quality of life[b]	39.5 ± 19.3	95.8 ± 8.3	98.6 ± 4.6

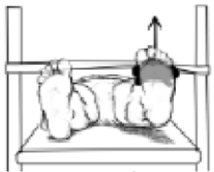
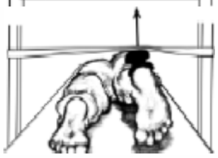
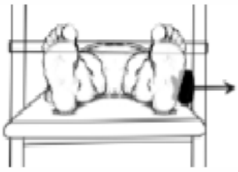
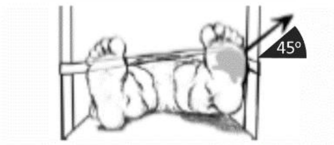
[a] Data are reported as mean ± SD unless otherwise indicated. CTRL, control; FAD, femoroacetabular deformity; FAI, femoroacetabular impingement; HOOS, Hip disability and Osteoarthritis Outcome Score.
[b] The FAI group differed significantly from the FAD and CTRL groups ($P < .001$).

4.2.2. Protocol

Following the CT scans, participants were transferred to the motion analysis laboratory. After warming up for 5 minutes on a cycle ergometer and performing uninstructed stretching, participants completed 2 trials of a sit-and-reach flexibility test while barefoot.[54]

After the flexibility test, participants were instrumented with wireless EMG probes (FREEEMG 300; BTS Bioengineering) placed on the RF, biceps femoris (BF), semitendinosus (ST), TFL, gluteus medius (GMed), and gluteus maximus (GMax) muscles according to the SENIAM guidelines.[19,20] Muscle strength activity was recorded using a handheld dynamometer (Manual Muscle Testing System Model 01163; Lafayette Instrument) and the EMG system during MVIC in the following movements: hip flexion, hip extension, hip abduction, hip flexion with hip abduction, and knee flexion (Table 4.2). Participants were verbally encouraged to complete 2 MVIC trials of 5 seconds for each selected motion as forcefully as possible without causing pain. A rest period of 30 seconds was provided between the 2 MVIC trials. Participants were then outfitted with 45 reflective markers according to the University of Ottawa Motion Analysis Model marker set.[34] To improve accuracy, the markers at the anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), and lateral and medial epicondyles were placed according to identification through the CT scans. All participants remained barefoot for motion capture testing to standardize the movement because shoes with different heel-stack heights could have affected the squat.

Table 4.2. Hip muscle strength produced during the MVIC for the FAI, FAD and CTRL groups normalized by body weight[a].

Movement	Illustration	Normalized torque (Nm/kg)		
		FAI	FAD	CTRL
Hip flexion[bc]		1.56±0.62	2.12±0.74	2.11±0.63
Hip extension[bd]		1.62±0.82	2.13±0.80	1.69±0.67
Hip abduction		1.39±0.45	1.53±0.50	1.60±0.51
Hip flexion with hip abduction		1.36±0.41	1.50±0.51	1.60±0.50
Knee flexion		0.88±0.38	0.99±0.35	0.92±0.35

[a]Data are reported as mean ± SD. CTRL, control; FAD, femoroacetabular deformity; FAI, femoroacetabular impingement; MVIC, maximum voluntary isometric contraction.
[b]The FAI group differs significantly from the FAD group ($P < 0.05$)
[c]The FAI group differs significantly from the CTRL group ($P < 0.05$)
[d]The FAD group differs significantly from the CTRL group ($P < 0.05$)

Motion capture was performed using 10 infrared cameras (MX13; Vicon Motion Systems) sampled at 200 Hz and 2 force plates (Force Plate FP4060-08; Bertec) measuring ground-reaction forces at 1000 Hz. Data were recorded, synchronized, and labeled using Nexus software (version 1.8.5; Vicon Motion Systems).

The participants performed 5 squat trials in a controlled position with their feet pointing forward and hip-width apart with each foot on a force plate. They were instructed to keep their arms elevated in front of the torso at shoulder width during the task. Participants were instructed to squat as deeply as possible without lifting their heels off the floor. The task was performed at the participants' self-selected pace, with a brief pause at the bottom of the squat. An adjustable bench was set to one-third the height of the participants' tibia for safety to prevent them from falling.

4.2.3. Statistical Analysis

Three-dimensional kinematic data were processed and filtered. The ground-reaction force data were filtered and used to calculate joint kinetics. Joint kinematics and kinetics were taken as the average of the 5 squat trials.

Maximal squat depth (percentage of leg length) was defined as the lowest point attained by the origin of the pelvis (calculated as the midpoint between the left and right ASIS and PSIS markers) during the squat, divided by the participant's leg length, which corresponded to the averaged linear distance between the participant's medial malleoli and ASIS. A lower value indicated a deeper squat.

EMG data were processed using custom software designed in MATLAB (MathWorks). All EMG signals were high-pass filtered, bias removed, and rectified, and a low-pass filter defined their linear envelope. The peak level of activation of the MVIC linear envelope was then taken as the amplitude normalization value. Signals were time-normalized for the squat descent and squat ascent phases separately. The selected variables were linear envelope peak (PeakLE), time to reach the PeakLE, and total muscle activity (iEMG), which was the integral of the linear envelope. All variables were averaged across the respective groups for each muscle and phase of the squat.

Data were assessed for normality using the Shapiro-Wilk test. One-way analysis of variance was used to examine differences between the 3 groups, with a Bonferroni post hoc test conducted to determine between-group differences ($P < .05$).

4.3. Results

No significant differences existed between the 3 groups for the sit-and-reach flexibility test findings. Differences in strength values during MVIC trials occurred for the hip flexion and hip extension movements (Table 4.2). The FAI group had lower hip flexion strength compared with both the FAD ($P = .003$) and CTRL ($P = .003$) groups. The FAD group had greater hip extension strength compared with the FAI ($P = .026$) and the CTRL ($P = .047$) groups. No significant differences in strength existed between the groups for knee flexion, neutral hip abduction, or hip abduction with hip flexion.

The maximal squat depths achieved during the squat cycle were the following: FAI, 39.4% ± 12.3%; FAD, 30.0% ± 12.2%; and CTRL, 27.1% ± 8.8% (Figure 4.1). The FAI group was unable to achieve as deep a squat as the FAD ($P < .001$) and CTRL ($P < .001$) groups, while no differences were found between the FAD and CTRL groups ($P = .252$).

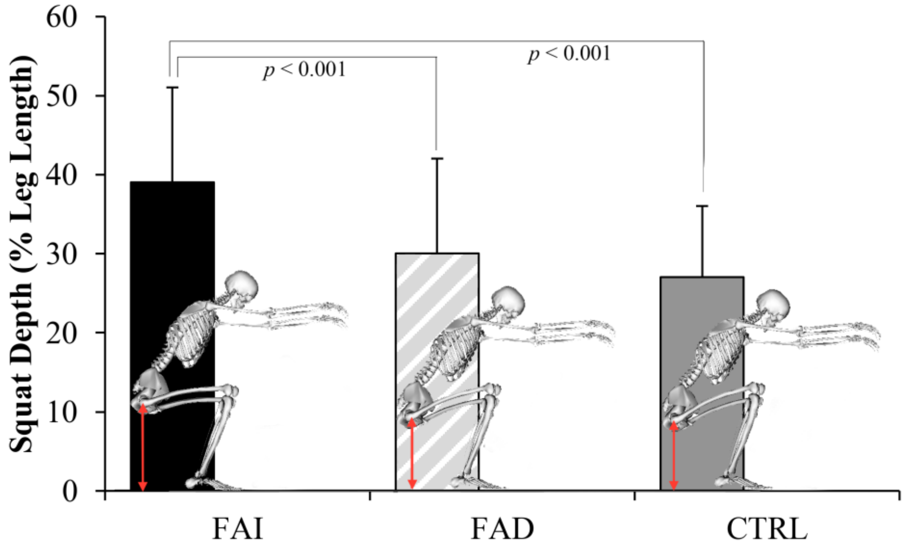


Figure 4.1. Maximum squat depth achieved for the femoroacetabular impingement (FAI), femoroacetabular deformity (FAD), and control (CTRL) groups.

Pelvic sagittal ROM is illustrated in Figure 4.2. During the first half of the descent phase of the squat, pelvic ROM was significantly lower (P = .015) for the FAI group (12.0° ± 5.0°) compared with the FAD group (18.4° ± 6.0°) but not compared with the CTRL group (16.9° ± 3.8°) (Figure 4.2B). During the second half of the descent phase of the squat, pelvic ROM was significantly lower for the FAI group (7.2° ± 4.1°) compared with the FAD (14.2° ± 7.2°) and CTRL (12.7° ± 6.6°) groups (P = .006 and .037, respectively) (Figure 4.2C). During the first half of the ascent phase of the squat, pelvic ROM was significantly lower (P = .039) in the FAI group (7.2° ± 3.7°) than in the CTRL group (12.5° ± 7.5°) (Figure 4.2E).

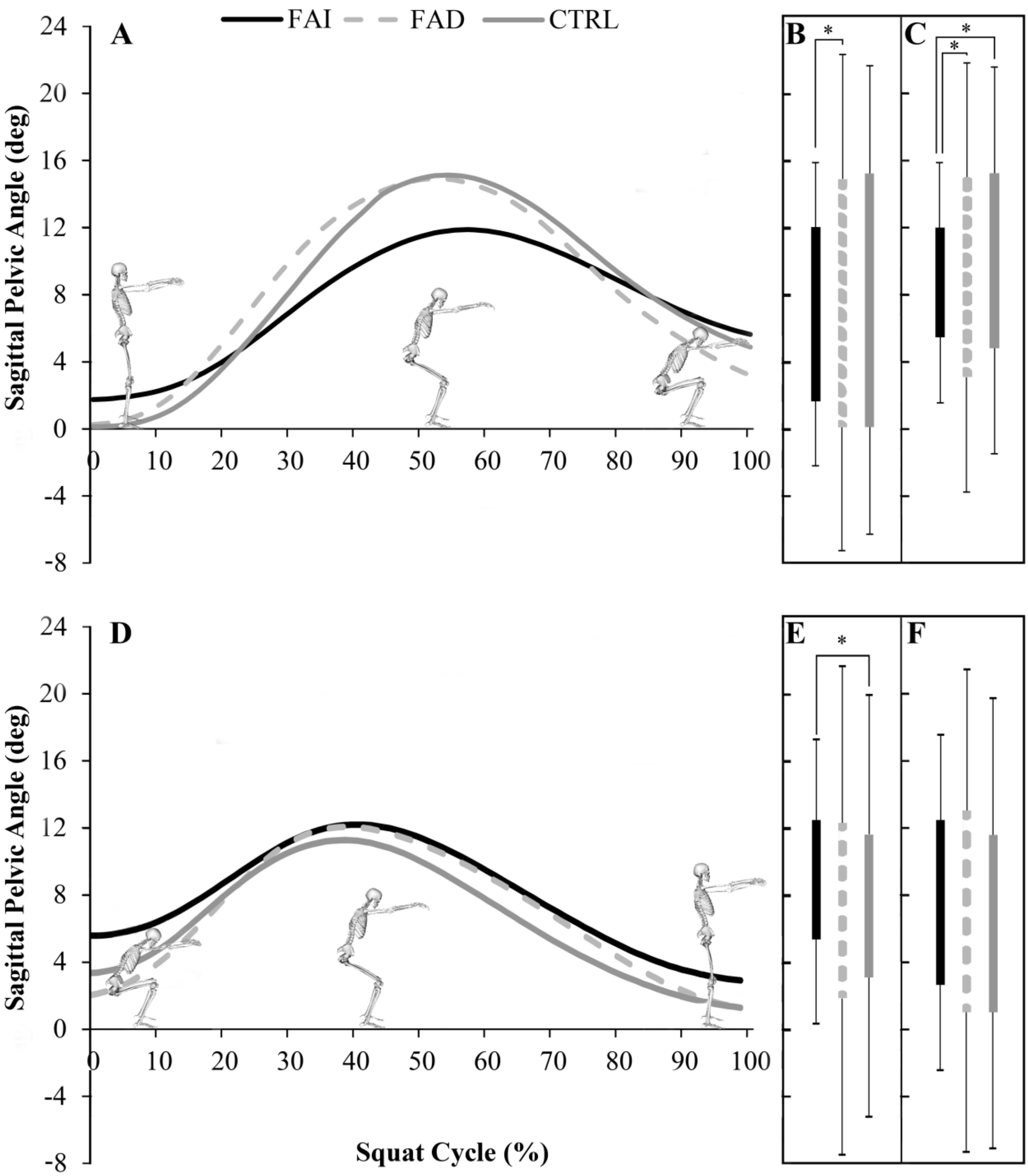


Figure 4.2. Sagittal pelvic tilt and range of motion (ROM) during a squat task in the femoroacetabular impingement (FAI), femoroacetabular deformity (FAD), and control (CTRL) groups. (A) Sagittal pelvic tilt during the descent phase, (B) trough-to-peak ROM during the descent phase, (C) peak-to-trough ROM during the descent phase, (D) sagittal pelvic tilt during the ascent phase, (E) trough-to-peak ROM during the ascent phase, and (F) peak-to-trough ROM during the ascent phase. *Significant difference in ROM between groups ($P < .05$).

Peak hip flexion was lower for the FAI group compared with the CTRL group for both the descent phase (P= .038) and the ascent phase (P = .028) of the squat (Figure 4.3). During the descent phase of the squat, sagittal hip ROM was significantly (P = .025) lower in the FAI group (88.6° ±

23.5°) compared with the CTRL group (103.8° ± 10.6°) but not compared with the FAD group (97.5° ± 11.8°) (Figure 4.3B). During the ascent phase of the squat, sagittal hip ROM was significantly (*P* = .037) lower in the FAI group (90.7° ± 20.5°) compared with the CTRL group (103.5° ± 11.0°) but not compared with the FAD group (97.2° ± 11.2°) (Figure 4.3D). No significant differences in hip abduction or hip joint kinetics (frontal and sagittal) existed between any of the groups.

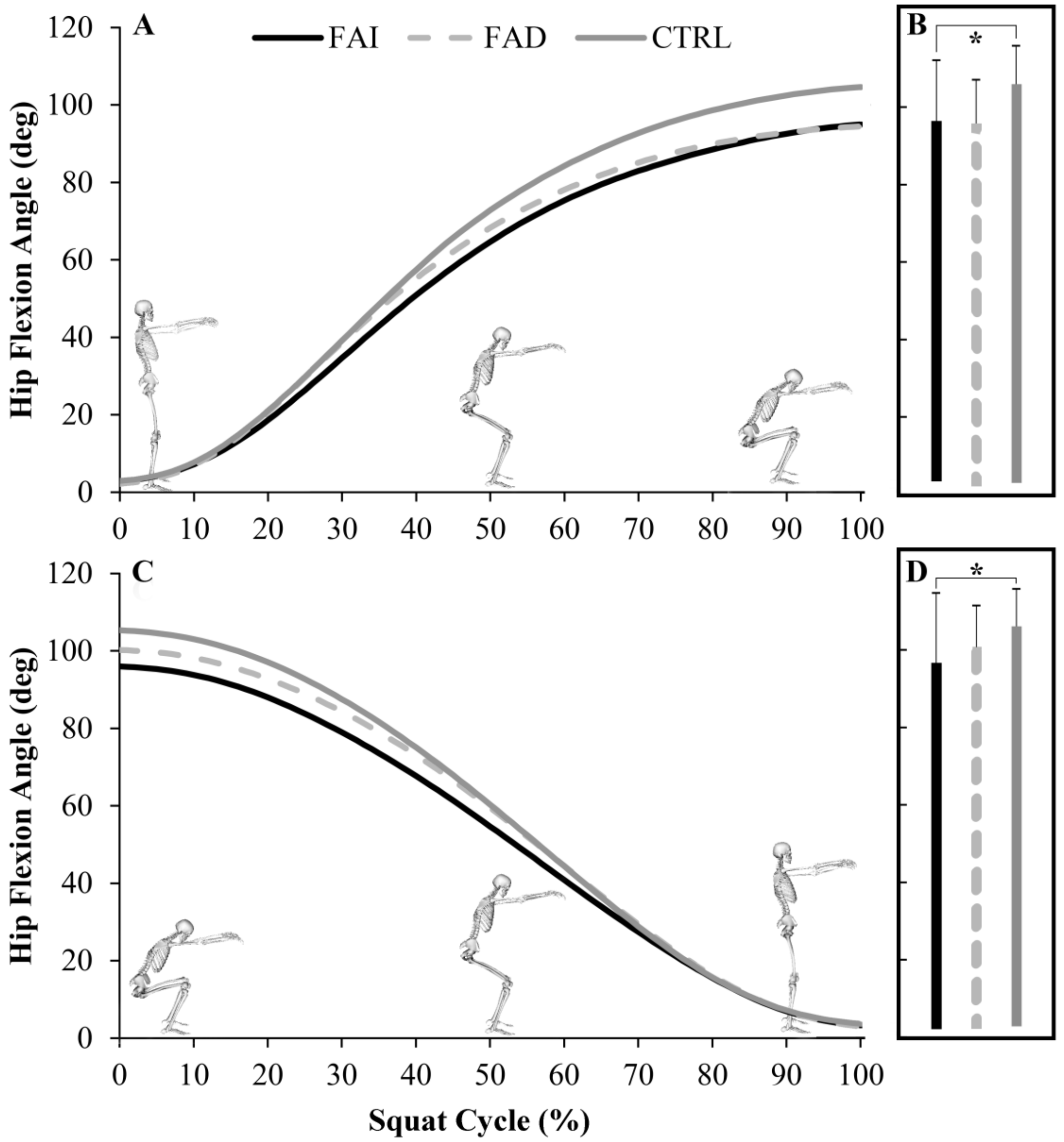


Figure 4.3. Sagittal hip movement during a squat task in the femoroacetabular impingement (FAI), femoroacetabular deformity (FAD), and control (CTRL) groups. (A) Hip flexion during the descent phase, (B) range of motion (ROM) during the descent phase, (C) hip flexion during the ascent phase, and (D) ROM during the ascent phase. *Significant difference in ROM between groups ($P < .05$).

For the EMG analyses, because the signals were normalized by their maximum, the muscle activity results were inversely proportional to the muscle's ability to produce force, as a weaker muscle will need higher muscle activity to perform the same task as its normal-strength counterpart. The FAI group had a significantly greater PeakLE for the BF and ST muscles compared with the FAD group for both squat descent and squat ascent (Figure 4.4). During squat descent, the PeakLE of the RF muscle was significantly lower for the FAD group compared with both the FAI and CTRL groups (Figure 4.4A). During squat ascent, the PeakLE of the GMax muscle was significantly lower (*P* = .005) for the FAD group than for the FAI group (Figure 4.4B). No significant differences in the PeakLE were observed for the TFL and GMed muscles or in the time to reach the PeakLE in any of the groups for any of the muscles.

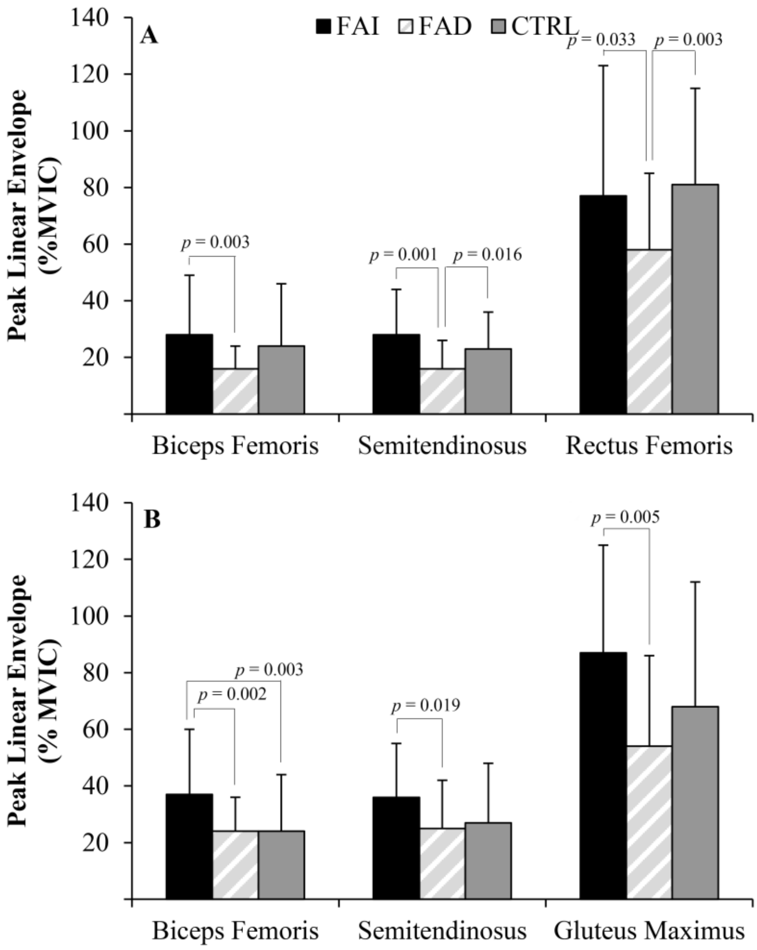


Figure 4.4. Linear envelope peak for the femoroacetabular impingement (FAI), femoroacetabular deformity (FAD), and control (CTRL) groups during the (A) squat descent and (B) squat ascent tasks. MVIC, maximum voluntary isometric contraction.

The FAI group had significantly greater iEMG for the BF and ST muscles compared with the FAD group for both squat descent and squat ascent (Figure 4.5). During squat ascent, the FAI group had significantly greater iEMG for the GMax muscle compared with the FAD group (P = .045) and the CTRL group (P = .046) (Figure 4.5B). No significant differences in iEMG were observed for the RF, TFL, and GMed muscles.

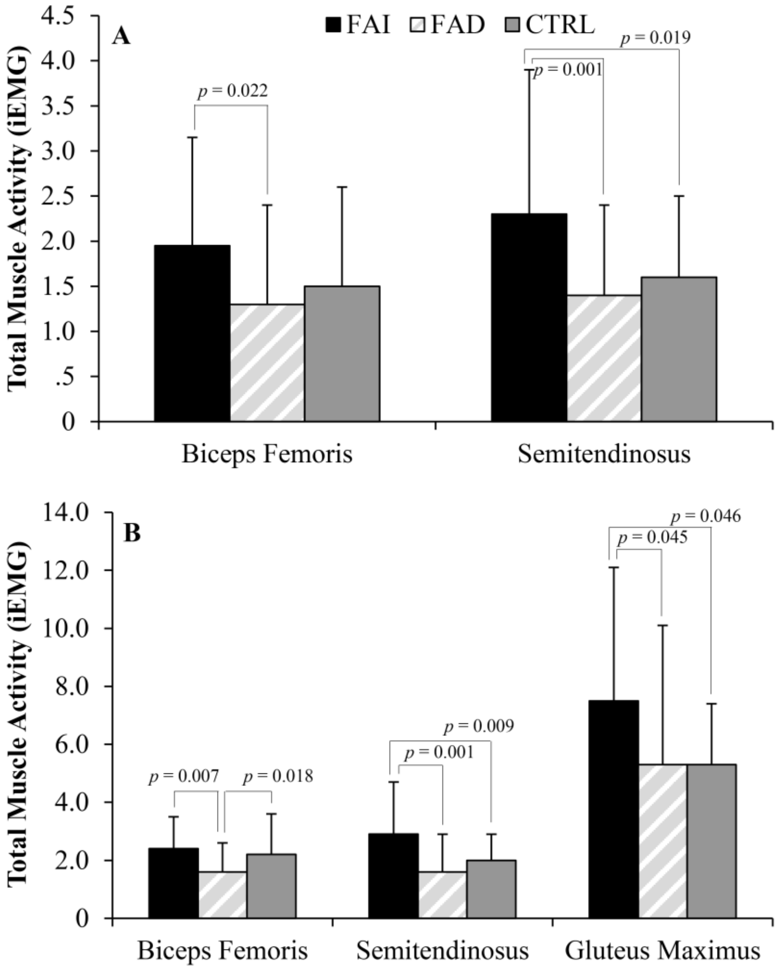


Figure 4.5. Total muscle activity for the femoroacetabular impingement (FAI), femoroacetabular deformity (FAD), and control (CTRL) groups during the (A) squat descent and (B) squat ascent tasks. iEMG, total muscle activity.

4.4. Discussion

Our study hypothesis was that the FAI group would show decreased hip flexion and hip abduction during the squat task when compared with the FAD and CTRL groups. The hypothesis also stated that patients with FAI would have weaker hip flexor muscles and that higher normalized activation for hip flexors would compensate for their weakness while performing the dynamic task.

Although we focused the hypothesis on the hip joint, and differences in hip sagittal ROM were found when comparing the FAI and CTRL groups, the most relevant finding was regarding pelvic tilt, as the FAI group had significantly less ROM when compared with the other 2 groups, specifically in the descending phase of the squat. The muscle strength analyses showed that patients with FAI not only had weaker hip flexors than the other groups but also that participants with FAD demonstrated stronger hip extensors when compared with the symptomatic patients. This strength difference was indicated in the EMG analyses, as the FAI group showed higher muscular activation during the squat task compared with the other groups.

Squat tasks have previously been proposed as a diagnostic tool for assessing FAI, as squatting requires large sagittal hip and pelvic ROM, attainable by many healthy participants but few patients with FAI.[30] It has already been shown that patients with FAI are unable to squat as deeply as their healthy peers because of mobility restrictions at the hip and pelvis.[30] However, no research has included participants with FAD in their comparison; these individuals have the same cam morphology but do not experience any of the pain symptoms. If the restriction in mobility was caused by the presence of femoral cam morphology, then participants with FAD should have achieved a squat depth similar to that of the FAI group; however, their movement was similar to that of the CTRL group (Figure 4.1). It is still elusive that a soft tissue abnormality (i.e. labrum, capsule) could be a cause of the mobility restriction in patients with FAI, but more research is needed to investigate the effect of soft tissue abnormalities on hip motion restriction.

As observed in previous research, a lower neck-shaft angle also differentiates symptoms for patients with cam-type FAI.[37,38] A greater pelvic incidence is also another morphological parameter that can contribute as a predictor of symptomatic cam-type FAI.[16] The differences in motion and symptoms between the FAI and FAD groups cannot be explained by static bony geometry but rather implicate dynamic motion of the femur and pelvis, which are affected by the soft tissues around the hip. In this study, we examined the role of various muscles in dynamic hip movement and found that, as observed in other studies,[9] patients with FAI did not perform as well in hip flexion tasks compared with participants with FAD and healthy controls. In this study, there were no differences in abduction and abduction with hip flexion strength. Interestingly, the FAD group had significantly stronger hip extensor strength (see Table 4.2). As a mechanism to compensate for muscular weakness during a dynamic task, higher normalized EMG activity was expected from the weak muscle when compared with its stronger counterpart. Although the FAI group was shown to have weaker hip flexors, higher EMG activity was not found for the hip flexor measured in this study when compared with the CTRL group. Perhaps one of the deeper hip flexors not measured in this study (e.g. iliacus or psoas major) may be primarily responsible for the lack of strength in the FAI group. Future studies should therefore include muscle activity for more hip flexors, either through indwelling EMG or optimization, to determine which hip flexors are weaker.

EMG activity for all hip extensor muscles was recorded in this study, with the exception of the semimembranosus muscle. As muscle activity results were collected through surface EMG and the semimembranosus lies deep to the ST, the signal output might have been compromised by muscle crosstalk. Moreover, the SENIAM guideline,[19,20] which was used in this study, does not have any placement recommendation regarding the semimembranosus. An examination of the PeakLE during the descent phase of the squat (eccentric) highlighted hip extensor strength in the FAD group, as both the BF and ST muscles had lower peaks compared with the other 2 groups. As for the ascent phase (concentric), higher peaks in the FAI group showed that these patients had to activate these muscles to a greater extent while standing up compared with the FAD and CTRL groups (Figure 4.4).

Sparse research exists that quantifies the muscle activity of patients with FAI. A previous study that compared hip muscle strength between patients with FAI and healthy controls found that the FAI group had weaker hip muscles, with the exception of the internal rotators and extensors.[9] When comparing the FAI and CTRL groups, we also found hip flexor weakness, comparable hip extensor strength, and no differences in TFL strength. The major difference in this study was that we were able to compare these data for the FAD group, which had significantly stronger hip extensors than the FAI and CTRL groups (see Table 4.2). Hip extensor strength may play a significant role in preventing symptomatic cam impingement.

Hip motion is complex from a biomechanical perspective, and its kinematics are influenced by many factors, including osseous, ligamentous, and muscular structures.[5] Hip flexion and extension are often thought of as movement of the femoral head within the acetabulum. Our research highlights the importance of pelvic motion (i.e. anterior and posterior pelvic tilt) in hip kinematics and the role of the hip flexors and extensors in changing the orientation of the acetabulum in various activities, such as squatting. Weakness in hip flexors and extensors may result in restricted ability to control the pelvis for good posture and stabilization. While weakness in the hip extensors would result in anterior pelvic tilt, strong hip extensors can contribute to improved posterior pelvic tilt. Patients with FAI experience painful impingement when their femoral cam deformity abuts against the acetabulum. Increased anterior pelvic tilt can lead to early impingement, which could explain why patients with FAI could not squat as deeply as their counterparts with FAD (see Figure 4.1), who were able to posteriorly tilt their pelvis and avoid painful impingement. The result of this muscle imbalance was restricted pelvis ROM in the FAI group during the squat task (see Figure 4.2).

During both squat phases, the FAI group had to activate the ST and BF muscles to a greater degree than the FAD group (Figure 4.4A and 4.4B). The same occurred for the GMax muscle during squat ascent (Figure 4.4B). It was speculated that decreased activation of the GMax and hamstring muscles could contribute to the lack of posterior tilt during a deep squat[2]; however, our findings show the exact opposite. This is because of the origin of these muscles, as any anterior pelvic tilt increases the moment arm of these muscles, requiring the FAI group to activate to a greater extent to achieve the same movement as the FAD group, which was in a posterior pelvic tilt position.

Because the ST and BF muscles originate on the ischial tuberosity and the GMax originates on the gluteal surface of the ilium, anterior pelvic tilt would increase the length of these muscles, as shown by the FAI group. To overcome this lengthened position, these muscles had greater peak activation (Figure 4.4) and iEMG (Figure 4.5), resulting in a less efficient squat performance. Stronger hip extensors would allow the pelvis to be brought to a posterior position during the midphase of the squat, perhaps avoiding impingement.

Possible overdiagnosis and overtreatment of patients with suspected FAI have always been a concern.[18,43] Surgical procedures for correcting FAI have increased by as much as 18-fold in the United States from 1999 to 2009.[10] Such surgical interventions are associated with many possible complications, including prolonged pain, nerve damage, fractures, and the development of hip osteoarthritis.[13,36] Despite these surgical interventions to address FAI, some patients may still need total hip arthroplasty.[21,44] Current evidence for the conservative treatment of FAI is limited to case series[12,22] (level 4 evidence) and is affected by patient demand for surgical treatment as well as a paucity of effective exercises.[47] In these studies, however, a staged physical therapy approach with activity modification and exercise led to improved patient-reported outcomes.[12,22] The use of conservative treatment in patients with FAI could be a strategy for avoiding surgery to ease hip pain.

Previous research suggests that hip extensors are not weaker in patients with FAI compared with healthy control participants[9]; therefore, hip extensor therapy was not a vital component in physical therapy protocols. However, healthy participants do not have the same cam deformity as patients with FAI, so they are still able to achieve a low, pain-free squat without controlling the pelvis in the same manner as participants with FAD. Compared with the FAI group, the FAD group in our cohort had significantly stronger hip extensors (see Table 4.2) and significantly better posterior pelvic tilt at the bottom of the squat (see Figure 4.2). We believe that these stronger hip extensors play an important role in sagittal pelvic ROM, allowing the FAD group to avoid impingement between the acetabular rim and femoral head-neck junction and allowing them to achieve a much lower squat depth than the FAI group (see Figure 4.1). Thus, improving hip extensor strength and pelvic mobility may affect symptoms for patients with FAI. Future studies should perform an intervention that tests this hypothesis as a way to possibly prevent some corrective surgical procedures. As the participants with FAD were younger than the patients with FAI and had similar cam-type morphology, it can be speculated that they were in the early progression of FAI. It may not be until labral tears or capsular lesions occur as a result of cam impingement that these individuals progress to symptoms and into the FAI group. Therefore, a longitudinal study of FAD is necessary to accept or reject the hypothesis that some participants with FAD will progress to the symptomatic group.

Some limitations of this study must be acknowledged. One limitation was the small cohort, which included 16 patients with symptomatic cam-type FAI. The recruitment of 18 participants with an asymptomatic cam deformity can be considered a challenge, as they do not present pain and are difficult to track in the clinical practice. The strength measurements were performed in an isometric condition, and the analyses were extrapolated to the dynamic squat task. A better condition to assess the participants' strength could be achieved on an isokinetic device; however, this would have drastically increased the overall data collection time. Although EMG data collection was rigorously completed and followed all requirements[19] regarding skin preparation and probe placement and was also normalized by MVIC, muscle activation signal variability among participants must be noted, as individual skin adipose tissue and motor point location may vary.

This study provides a greater understanding of the role of the muscles and soft tissues around the hip that contribute to the possible development of symptomatic FAI. The patients with symptomatic cam-type FAI were unable to achieve as deep of a squat as those in the FAD and the CTRL groups, and the asymptomatic participants with FAD had significantly stronger hip extensors and greater pelvic mobility compared with patients with FAI. Future research should investigate rehabilitation and conservative treatments that focus on both strengthening hip extensor muscles and increasing pelvic mobility for their potential to reduce symptoms or even delay or avoid corrective surgery in patients with FAI.

4.5. References

1. Allen D, Beaulé PE, Ramadan O, Doucette S. Prevalence of Associated Deformities and Hip Pain in Patients with Cam-Type Femoroacetabular Impingement. J Bone Jt Surg - Br Vol. 2009;91-B(5):589-594. doi:10.1302/0301-620X.91B5.22028.

2. Bagwell JJ, Snibbe J, Gerhardt M, Powers CM. Hip Kinematics and Kinetics in Persons with and Without Cam Femoroacetabular Impingement during a Deep Squat Task. Clin Biomech. 2016;31:87-92. doi:10.1016/J.Clinbiomech.2015.09.016.

3. Barton C, Salineros MJ, Rakhra KS, Beaulé PE. Validity of the Alpha Angle Measurement on Plain Radiographs in the Evaluation of Cam-Type Femoroacetabular Impingement. Clin Orthop Relat Res. 2011;469(2):464-469. doi:10.1007/S11999-010-1624-X.

4. Beck M, Kalhor M, Leuning M, Ganz R. Hip Morphology Influences The Pattern Of Damage To The Acetabular Cartilage: Femoroacetabular Impingement As A Cause Of Early Osteoarthritis Of The Hip. J Bone Jt Surg - Br Vol. 2005;87-B(7):1012-1018. doi:10.1302/0301-620X.87B7.15203.

5. Bowman KF, Fox J, Sekiya JK. A Clinically Relevant Review of Hip Biomechanics. Arthrosc - J Arthrosc Relat Surg. 2010;26(8). doi:10.1016/J.Arthro.2010.01.027.

6. Brisson N, Lamontagne M, Kennedy MJ, Beaulé PE. The Effects of Cam Femoroacetabular Impingement Corrective Surgery on Lower-Extremity Gait Biomechanics. Gait Posture. 2013;37(2):258-263. doi:10.1016/J.Gaitpost.2012.07.016.

7. Byrd JWT. Femoroacetabular Impingement in Athletes: Current Concepts. Am J Sports Med. 2014;42(3):737-751. doi:10.1177/0363546513499136.

8. Casartelli NC, Leunig M, Item-Glatthorn JF, Lepers R, Maffiuletti NA. Hip Flexor Muscle Fatigue in Patients with Symptomatic Femoroacetabular Impingement. Int Orthop. 2012;36(5):967-973. doi:10.1007/S00264-011-1385-5.

9. Casartelli NC, Maffiuletti NA, Item-Glatthorn JF, Et Al. Hip Muscle Weakness in Patients with Symptomatic Femoroacetabular Impingement. Osteoarthr Cartil. 2011;19(7):816-821. doi:10.1016/J.Joca.2011.04.001.

10. Colvin AC, Harrast J, Harner C, Et Al. Trends in Hip Arthroscopy. J Bone Joint Surg Am. 2012;94(4):E23. doi:10.2106/JBJS.J.01886.

11. Diamond LE, Van Den Hoorn W, Bennell KL, Et Al. Coordination Of Deep Hip Muscle Activity Is Altered In Symptomatic Femoroacetabular Impingement. J Orthop Res. 2017;35(7):1494-1504. doi:10.1002/Jor.23391.

12. Emara K, Samir W, Motasem EH, Ghafar KA El. Conservative Treatment for Mild Femoroacetabular Impingement. J Orthop Surg (Hong Kong). 2011;19(1):41-45. doi:10.1177/230949901101900109.

13. Ganz R, Gill T, Gautier E, Ganz K, Krügel N, Berlemann U. Surgical Dislocation Of The Adult Hip: A Technique With Full Access To The Femoral Head And Acetabulum Without The Risk Of Avascular Necrosis. J Bone Jt Surg. 2001:1119-1124. doi:10.1007/978-1-4471-5451-8_144.

14. Ganz R, Parvizi J, Beck M, Leunig M, Nötzli H, Siebenrock KA. Femoroacetabular Impingement: A Cause for Osteoarthritis of the Hip. Clin Orthop Relat Res. 2003;(417):112-120. doi:10.1097/01.Blo.0000096804.78689.C2.

15. Gosvig KK, Jacobsen S, Sonne-Holm S, Gebuhr P. The Prevalence Of Cam-Type Deformity Of The Hip Joint : A Survey Of 4151 Subjects Of The Copenhagen Osteoarthritis Study. Acta Radiol. 2008;4:436-441. doi:10.1080/02841850801935567.

16. Grammatopoulos G, Speirs AD, Ng KCG, Et Al. The Acetabular And Spino-Pelvic Morphologies Are Different In Subjects With Symptomatic Cam Femoro-Acetabular Impingement. J Orthop Res. 2018:1-9. doi:10.1002/Jor.23856.

17. Hack K, Di Primio G, Rakhra K, Beaulé PE. Prevalence of Cam-Type Femoroacetabular Impingement Morphology in Asymptomatic Volunteers. J Bone Jt Surg. 2010;92(14):2436. doi:10.2106/JBJS.J.01280.

18. Hartofilakidis G, Bardakos N, Babis G, Georgiades G. An Examination of the Association between Different Morphotypes of Femoroacetabular Impingement in Asymptomatic Subjects and the Development of Osteoarthritis of the Hip. J Bone Jt Surg [Br]. 2011;93(5):580-586. doi:10.1302/0301-620X.93B5.

19. Hermens HJ, Freriks B, Disselhorst-Klug C, Rau G. Development of Recommendations for SEMG Sensors and Sensor Placement Procedures. J Electromyogr Kinesiol. 2000;10:361-374. doi:10.1016/S1050-6411(00)00027-4.

20. Hermens HJ, Freriks B, Merletti R, Et Al. SENIAM 8: European Recommendations for Surface Electromyography. Roessingh Research and Development; 1999. ISBN: 90-75452-15-2.

21. Horisberger M, Brunner A, Herzog RF. Arthroscopic Treatment of Femoroacetabular Impingement of the Hip: A New Technique to Access the Joint. Clin Orthop Relat Res. 2010;468(1):182-190. doi:10.1007/S11999-009-1005-5.

22. Hunt D, Prather H, Hayes MH, Clohisy JC. Clinical Outcomes Analysis of Conservative and Surgical Treatment of Patients with Clinical Indications of Prearthritic, Intra-Articular Hip Disorders. Pm&R. 2012;4(7):479-487. doi:10.1016/J.Pmrj.2012.03.012.

23. Hunt MA, Gunether JR, Gilbart MK. Kinematic and Kinetic Differences during Walking In Patients with and Without Symptomatic Femoroacetabular Impingement. Clin Biomech. 2013;28(5). doi:10.1016/J.Clinbiomech.2013.05.002.

24. Ito K, Leunig M, Ganz R. Histopathologic Features of the Acetabular Labrum in Femoroacetabular Impingement. Clin Orthop Relat Res. 2004;429:262-271. doi:10.1097/01.Blo.0000144861.11193.17.

25. Ito K, Minka M, Leunig M, Werlen S, Ganz R. Femoroacetabular Impingement and the Cam-Effect. A MRI-Based Quantitative Anatomical Study of the Femoral Head-Neck Offset. J Bone Joint Surg Br. 2001;83-B(2):171-176. doi:10.1302/0301-620X.83B2.11092.

26. Jung KA, Restrepo C, Hellman M, Abdelsalam H, Morrison W, Parvizi J. The Prevalence of Cam-Type Femoroacetabular Deformity in Asymptomatic Adults. Bone Joint J. 2011;93-B(10):1303-1307. doi:10.1302/0301-620X.93B10.26433.

27. Kennedy MJ, Lamontagne M, Beaulé PE. Femoroacetabular Impingement Alters Hip and Pelvic Biomechanics during Gait: Walking Biomechanics of FAI. Gait Posture. 2009;30(1):41-44. doi:10.1016/J.Gaitpost.2009.02.008.

28. Khanna V, Caragianis A, Diprimio G, Rakhra K, Beaulé PE. Incidence Of Hip Pain In A Prospective Cohort Of Asymptomatic Volunteers: Is The Cam Deformity A Risk Factor For Hip Pain? Am J Sports Med. 2014;42(4):793-797. doi:10.1177/0363546513518417.

29. Lamontagne M, Brisson N, Kennedy MJ, Beaulé PE. Preoperative and Postoperative Lower-Extremity Joint and Pelvic Kinematics during Maximal Squatting Of Patients with Cam Femoroacetabular Impingement. J Bone Jt Surg. 2011;93(2):40-45. doi:10.2106/JBJS.J.01809.

30. Lamontagne M, Kennedy MJ, Beaulé PE. The Effect of Cam FAI on Hip and Pelvic Motion during Maximum Squat. Clin Orthop Relat Res. 2009;467:645-650. doi:10.1007/S11999-008-0620-X.

31. Lamontagne M, Ng KG, Mantovani G, Catelli DS. Biomechanics of Femoroacetabular Impingement. In: Sports Injuries. 2a. 62: Springer-Verlag Berlin Heidelberg; 2015:783-795. doi:10.1007/978-3-642-36801-1.

32. Laude F, Boyer T, Nogier A. Anterior Femoroacetabular Impingement. Jt Bone Spine. 2007;74(2):127-132. doi:10.1016/J.Jbspin.2007.01.001.

33. Leunig M, Beaulé PE, Ganz R. The Concept of Femoroacetabular Impingement: Current Status and Future Perspectives. Clin Orthop Relat Res. 2009;467(3):616-622. doi:10.1007/S11999-008-0646-0.

34. Mantovani G, Lamontagne M. How Different Marker Sets Affect Joint Angles In Inverse Kinematics Framework. J Biomech Eng. 2016;139(4):44503. doi:10.1115/1.4034708.

35. Martin HD, Kelly BT, Leunig M, Et Al. The Pattern and Technique in the Clinical Evaluation of the Adult Hip: The Common Physical Examination Tests of Hip Specialists. Arthrosc - J Arthrosc Relat Surg. 2010;26(2):161-172. doi:10.1016/J.Arthro.2009.07.015.

36. Matsuda DK, Carlisle JC, Arthurs SC, Wierks CH, Philippon MJ. Comparative Systematic Review of the Open Dislocation, Mini-Open, and Arthroscopic Surgeries for Femoroacetabular Impingement. Arthrosc - J Arthrosc Relat Surg. 2011;27(2):252-269. doi:10.1016/J.Arthro.2010.09.011.

37. Ng KCG, Lamontagne M, Jeffers JRT, Grammatopoulos G, Beaulé PE. Anatomic Predictors of Sagittal Hip and Pelvic Motions in Patients with a Cam Deformity. Am J Sports Med. 2018:1-12. doi:10.1177/0363546518755150.

38. Ng KCGG, Mantovani G, Lamontagne M, Labrosse MR, Beaulé PE. Increased Hip Stresses Resulting From a Cam Deformity and Decreased Femoral Neck-Shaft Angle during Level Walking. Clin Orthop Relat Res. 2017;475(4):998-1008. doi:10.1007/S11999-016-5038-2.

39. Ng KG, Lamontagne M, Adamczyk AP, Rahkra KS, Beaulé PE. Patient-Specific Anatomical and Functional Parameters Provide New Insights into the Pathomechanism of Cam FAI. Clin Orthop Relat Res. 2015;473(4):1289-1296. doi:10.1007/S11999-014-3797-1.

40. Ng KG, Lamontagne M, Beaulé PE. Differences in Anatomical Parameters between the Affected and Unaffected Hip in Patients with Bilateral Cam-Type Deformities. Clin Biomech. 2016;33:13-19. doi:10.1016/J.Clinbiomech.2016.01.007.

41. Nötzli H, Wyss T, Stoecklin C, Schmid M, Treiber K, Hodler J. The Contour of the Femoral Head-Neck Junction as a Predictor for the Risk of Anterior Impingment: Commentary. J Bone Jt Surg. 2002;84(B):556-560. doi:10.1024/0369-8394.92.11.475.

42. Nussbaumer S, Leunig M, Glatthorn JF, Stauffacher S, Gerber H, Maffiuletti NA. Validity and Test-Retest Reliability of Manual Goniometers for Measuring Passive Hip Range of Motion in Femoroacetabular Impingement Patients. BMC Musculoskelet Disord. 2010;11:194. doi:10.1186/1471-2474-11-194.

43. Palmer WE. Femoroacetabular Impingement: Caution Is Warranted in Making Imaging-Based Assumptions and Diagnoses. Radiology. 2010;257(1):4-7. doi:10.1148/Radiol.10091298.

44. Philippon M, Briggs K, Yen Y-M, Kuppersmith D. Outcomes Following Hip Arthroscopy for Femoroacetabular Impingement with Associated Chondrolabral Dysfunction: Minimum Two-Year Follow-Up. J Bone Jt Surg [Br]. 2009;91:16-23. doi:10.1302/0301-620X.91B1.

45. Rakhra KS, Sheikh AM, Allen D, Beaulé PE. Comparison of MRI Alpha Angle Measurement Planes in Femoroacetabular Impingement. Clin Orthop Relat Res. 2009;467(3):660-665. doi:10.1007/S11999-008-0627-3.

46. Ranawat AS, Schulz B, Baumbach SF, Meftah M, Ganz R, Leunig M. Radiographic Predictors of Hip Pain in Femoroacetabular Impingement. HSS J. 2011;7(2):115-119. doi:10.1007/S11420-010-9192-X.

47. Reiman MP, Thorborg K. Femoroacetabular Impingement Surgery: Are We Moving Too Fast and Too Far Beyond the Evidence? Br J Sports Med. 2015;49:782-784. doi:10.1136/Bjsports-2014-093821.

48. Rylander J, Shu B, Favre J, Safran M, Andriacchi T. Functional Testing Provides Unique Insights into the Pathomechanics of Femoroacetabular Impingement and an Objective Basis For Evaluating Treatment Outcome. J Orthop Res. 2013;31(9):1461-1468. doi:10.1002/Jor.22375.

49. Santori N, Villar RN. Arthroscopic Findings in the Initial Stages of Hip Osteoarthritis. Orthopedics. 1999;22(4):405-409. doi:10.3928/0147-7447-19990401-09.

50. Siebenrock K, Wahab K, Werlen S, Kalhor M, Leunig M, Ganz R. Abnormal Extension of The Femoral Head Epiphysis as a Cause of Cam Impingement. Clin Orthop Relat Res. 2004;(418):54-60.

51. Speirs A, Beaulé P, Rakhra K, Schweitzer M, Frei H. Increased Acetabular Subchondral Bone Density Is Associated with Cam-Type Femoroacetabular Impingement. Osteoarthr Cartil. 2013;21(4):551-558. doi:10.1016/J.Joca.2013.01.012.

52. Sutter R, Dietrich TJ, Zingg PO, Pfirrmann CWA. How Useful Is the Alpha Angle for Discriminating between Symptomatic Patients with Cam-Type Femoroacetabular Impingement and Asymptomatic Volunteers? Radiology. 2012;264(2):514-521. doi:10.1148/Radiol.12112479.

53. Tateuchi H, Taniguchi M, Mori N, Ichihashi N. Balance of Hip and Trunk Muscle Activity Is Associated with Increased Anterior Pelvic Tilt During Prone Hip Extension. J Electromyogr Kinesiol. 2012;22(3):391-397. doi:10.1016/J.Jelekin.2012.03.003.

54. Wells KF, Dillon EK. The Sit and Reach—A Test of Back and Leg Flexibility. Res Quarterly Am Assoc Heal Phys Educ Recreat. 1952;23(1):115-118. doi:10.1080/10671188.1952.10761965.

III Impact of Hip Surgeries on the Squat Performance

5 Increased Pelvic Mobility and Altered Hip Muscles Contraction Patterns: Two-Year Follow-Up Cam FAI Corrective Surgery

Danilo S. Catelli [12] | Erik Kowalski [1] | Paul E. Beaulé [3] | Mario Lamontagne [134]

[1]School of Human Kinetics, University of Ottawa, Ottawa, Ontario, Canada.
[2]CAPES Foundation, Ministry of Education of Brazil, Brasilia, Brazil.
[3]Division of Orthopaedic Surgery, University of Ottawa, Ottawa, Ontario, Canada.
[4]Department of Mechanical Engineering, University of Ottawa, Ottawa, Ontario, Canada.

Contents of this chapter submitted as an article in *Journal of Hip Preservation Surgery®* | September 2018

Catelli DS, Kowalski E, Beaulé PB, Lamontagne M, Increased Pelvic Mobility and Altered Hip Muscles Contraction Patterns: Two-Year Follow-Up Cam FAI Corrective Surgery. 2018; Submitted.

5.0. Abstract

Femoroacetabular impingement (FAI) surgery can produce improvements in function and patient satisfaction; however, data on muscle assessment and kinematics of high mobility tasks of postoperative patients is limited. The purpose of this study was to evaluate kinematics and muscle activity during a deep squat task, as well as muscle strength in a 2-year follow-up FAI corrective surgery. Eleven cam FAI patients underwent motion and electromyography capture while performing a squat task prior and 2-years after osteochondroplasty and were BMI- age- and sex-matched to 11 healthy control participants (CTRL). Isometric muscle strength, flexibility, and patient-reported outcome measures (PROMs) were also evaluated. Postoperative FAI were significantly weaker during hip flexion (21.3%) and hip flexion-with-abduction (21.1%) movements when compared to CTRL, no improvements in squat depth was observed. However, postoperative FAI increased the pelvic range of motion during the squat descent ($p=0.016$) and ascent ($p=0.047$). They had greater peak activity for the semitendinosus and total muscle activity for the gluteus medius, but decreased peak activity for the glutei and rectus femoris during squat descent; greater total muscle activity for the tensor fascia latae was observed during squat ascent ($p=0.005$). Although not improving squat depth, postoperative patients increased pelvic ROM and showed positive PROMs. The muscle weakness associated with hip flexion and flexion-with-abduction observed at the follow-up can be associated with the alterations in the muscle activity and neuromuscular patterns. Rehabilitation programs should focus on increasing pelvis and hip muscles flexibility and strength, with special attention on strengthening the hip flexors and TFL muscles.

5.1. Introduction

Cam-type femoroacetabular impingement (FAI) is typically observed when patients test positive during a flexion, adduction and internal rotation (FADIR) physical examination.[1–3] The cam FAI can be confirmed with computed tomography (CT) imaging; with axial and radial alpha angles greater than 50.5° or 60°, respectively.[4–9] The presence of the cam deformity along with a lower femoral neck-shaft angle can contribute to the onset of symptoms including pain in the groin area or buttocks area, radiating down the iliotibial band.[8,10] This pain is often constant and is most commonly felt during activity, but many also are experienced during sustained periods of hip flexion, such as sitting.[11–13]

FAI symptoms are first treated with conservative methods;[14–17] however, once all conservative methods are exhausted, surgical correction, an osteochondroplasty of the femoral head-neck junction,[18] is often required.[14,19,20] The surgery for FAI is done through either open[21–23] or arthroscopic[24–28] procedures. Patient-reported outcome measures (PROMs) have indicated that both surgical methods are effective at reducing pain and improving quality of life.[29–32]

Although PROMs have given insight into the success of surgical correction on patients with FAI, only a limited number of studies have objectively compared patients before and after surgery using biomechanical outcomes[33–39] mainly during gait. This task does not place patients in a near impinged position. Only three studies have compared FAI patients pre- and postoperatively during tasks with extreme hip flexion.[34,39,40] For the surgical treatment of femoroacetabular impingement, it is unknown how this affects the muscle strength at the hip during isometric contraction or joint biomechanics and muscle activity of the hip muscles during activities with a large range of motion. Therefore, comparing the strength, kinematics and muscle activity during a deep squat task in FAI patients before and at two-year following surgery can provide insight into optimizing function for those suffering from FAI.

The purpose of this study was to examine if postoperative FAI patients have improved the squat depth, pelvic and hip range of motion, hip muscle strength, or differ their hip muscle activity pattern compared to their preoperative condition.

5.2. Methods

This study had a prospective, matched cohort design (level II evidence). Eleven male patients with unilateral symptomatic cam FAI were compared to eleven male body mass index (BMI)-, age-matched healthy controls (CTRL) – Table 5.1. Symptomatic patients had a positive impingement test and presented a cam deformity greater than 50.5° and 60° in the oblique-axial and radial planes, respectively.[4–9] For this study, CTRL participants were submitted to CT scan previous to their participation to secure that they were not asymptomatic participants with a cam deformity. Participants were also excluded if they had any musculoskeletal or neurological disorders, degenerative diseases, previous major lower limb injuries, or a BMI greater than 30 kg/m^2. FAI participants went for motion analysis testing before receiving surgery and at minimum two years postoperatively (25.05±1.13 months), whereas CTRL participants performed the testing protocol once. The study was approved by the hospital and university research ethics boards, and all participants signed and provided informed consent before their participation in the study.

Table 5.1. Group demographics and patient-reported outcome measures for HOOS questionnaire.

Groups		FAI Preoperative	FAI Postoperative	Control
Group Size (n)		11		11
Age (years)		34.1 ± 7.4	36.2 ± 7.4	33.1 ± 7.2
Height (m)		1.77 ± 0.1	1.78 ± 0.1	1.74 ± 0.1
Weight (kg)		80.0 ± 10.3	81.0 ± 10.4	77.3 ± 13.9
BMI (kg/m^2)		25.4 ± 2.7	25.6 ± 3.6	25.4 ± 3.2
α-Angle (°)	3:00 position	54.0 ± 7.2†*	45.6 ± 6.7	43.3 ± 4.7
	1:30 position	66.3 ± 5.4†*	52.5 ± 9.1	53.0 ± 4.9
Sit-and-Reach test		29.8 ± 8.4	25.8 ± 9.4	24.16 ± 8.3
HOOS Symptoms		70.0 ± 10.7†*	81.4 ± 10.0†	99.1 ± 2.0
HOOS Pain		70.0 ± 16.9†*	90.0 ± 8.3	98.9 ± 3.8
HOOS ADL		81.7 ± 15.1†*	95.4 ± 6.6	99.6 ± 1.3
HOOS Sport/Recreation		56.8 ± 25.1†*	83.0 ± 13.7	98.3 ± 5.7
HOOS Quality of Life		39.2 ± 21.8†*	65.9 ± 21.5†	97.2 ± 9.4

Data are reported as mean ± SD
* significant difference (p<0.05) compared with FAI post-op
† significant difference (p<0.05) compared with CTRL

Four of the FAI patients underwent corrective surgery, an osteochondroplasty of the femoral head-neck junction, with an open approach with surgical dislocation and 7 had surgery with an arthroscopic approach, all performed by the same surgeon. A standard 6-week physiotherapy program followed surgeries.

After completing the CT scan examination, all participants were transferred to the motion capture laboratory at the local university where they completed the Hip Disability and Osteoarthritis Outcome Score (HOOS) questionnaire and performed two trials of sit-and-reach flexibility test with the feet level at 20 cm.[41] Wireless electromyography (EMG) probes (BTS FreeEMG 300, Padova, Italy) were placed on the *rectus femoris* (RF), *biceps femoris* long head (BF), *semitendinosus* (ST), *gluteus medius* (GMed), *gluteus maximus* (GMax), and *tensor fasciae latae* (TFL) muscles of both limbs to record muscle activity. Two maximal voluntary isometric contractions (MVICs) were captured using a hand-held dynamometer (model 01163, Lafayette Instrument, Lafayette, USA) for each task and were separated by a 30s resting interval (Table 5.2).

Three-dimensional motion analysis was collected using ten infrared cameras (MX-13, Vicon, Oxford, UK) and 45 retroreflective skin markers placed on anatomical landmarks as the UOMAM marker set.[42] Participants completed five deep squats to their maximal depth at a controlled and self-selected pace. They were instructed to place their feet hip-width apart, directed anteriorly, with toes and heels in full contact with the ground during the entire squat cycle. The squat trials were separated into descending and ascending phases, and squat depth was normalized with respect to their leg length; the distance between the anterior superior iliac-spine (ASIS) to the medial malleolus. EMG and motion data were exported into a custom built Matlab script (MathWorks, Natick, USA) for extraction and processing. Motion trajectories were filtered using a Woltring filter (MSE=15 mm^2). Pelvic and hip sagittal ROM, along with peak hip flexion, peak hip abduction and peak knee flexion were extracted and averaged between the five trials. EMG data was filtered using a bandpass filter (20-450Hz) and rectified. From the normalized signal, peak linear envelope (PeakLE) and total muscle activity (iEMG) were determined for each muscle during each phase of the squat and normalized by their MVIC. The data were then averaged between the five trials and concerning each group in order to be analyzed.

All data were explored for normality. Comparisons between pre- and postoperative were made using either a paired t-test or its non-parametric equivalent Wilcoxon signed rank test. To compare differences between the FAI conditions and the CTRL group, a one-way ANOVA was used with a Bonferroni post-hoc comparison, to determine where significant differences occurred (CI=95%). The effect size was calculated with Cohen's d and was considered as either small ($d=0.2$), medium ($d=0.5$) and large ($d=0.8$).

5.3. Results

The FAI patients reported significantly improved HOOS on all measures on their follow-up compared to their preoperative values (Table 5.1). No significant differences amongst the groups were found in the sit-and-reach flexibility test.

Squat depth for the three groups is illustrated in Figure 5.1. Between the preoperative and postoperative conditions for the FAI group, there was no change in squat depth as they achieved a squat depth of 33.1% leg length and 34.2% leg length (0% representing the lowest squat depth), for the pre- and postoperative conditions, respectively. The CTRL group was able to squat significantly lower (27.0%) than both the pre- and postoperative FAI conditions ($p < 0.01$; $d > 0.65$), suggesting a moderate to high significance.

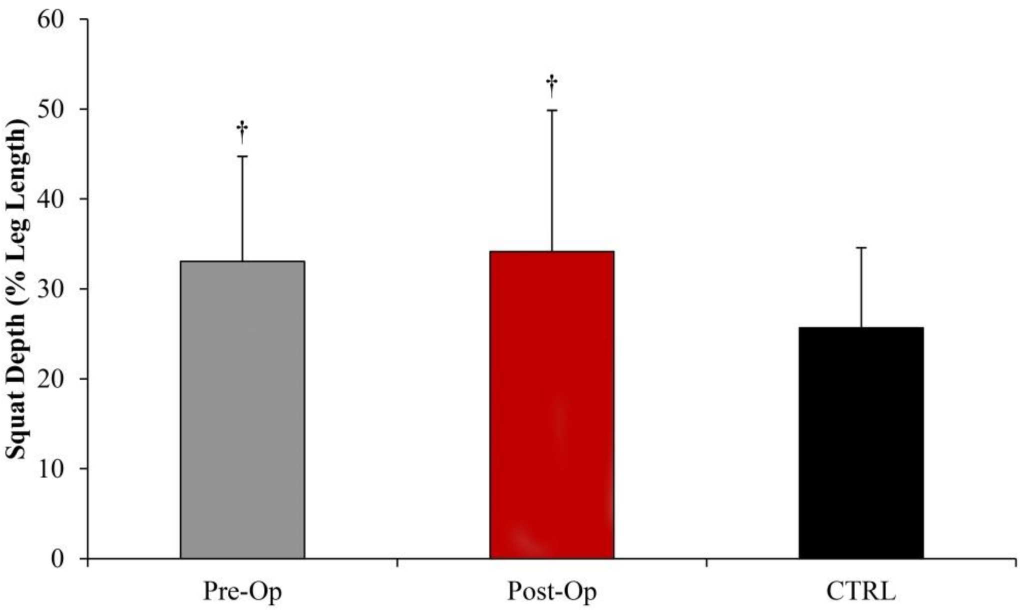
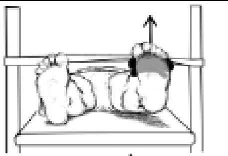
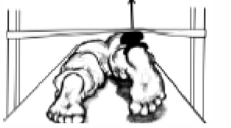
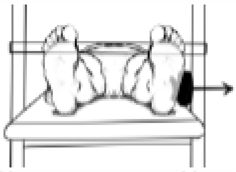


Figure 5.1. Squat depth normalized to the percentage of leg length for the FAI group before and after surgery, compared to the healthy CTRL. † significant difference ($p<0.01$) compared with CTRL.

At two-year follow-up, FAI patients showed decreased muscle strength compared to their preoperative values for all hip MVIC tasks, but the results were not statistically significant. When comparing strength measures to the CTRL group, while there were no differences with the preoperative condition, post-surgery the FAI patients were significantly weaker during hip flexion ($p=0.032$, $d=0.73$) and hip abduction with flexion ($p=0.048$, $d=0.69$) tasks (Table 5.2). A comparison between flexion/extension strength ratios was also done (FAI pre-op 1.08±0.46; FAI post-op 1.28±1.00 and CTRL 1.72±1.02), and a significant difference arose when comparing the preoperative patients with the CTRL ($p = 0.015$, $d = 0.78$).

Table 5.2. Hip muscle strength produced during MVIC and hand-held dynamometer (HHD) placement. The arrow represents the location of the HHD and the direction of the force vector.

Movement	Muscles	Illustration	Normalized Torque (Nm/kg) Mean ± SD		
			Pre-Op	Post-Op	CTRL
Hip Flexion†	*rectus femoris*		1.78±0.51	1.70±0.68	2.16±0.60
Hip Extension	*gluteus maximus* *biceps femoris* *semitendinosus*		1.84±0.56	1.70±0.71	1.47±0.46
Hip Abduction	*gluteus medius*		1.54±0.31	1.47±0.41	1.59±0.47
Hip Flexion with Hip Abduction†	*tensor fasciae latæ*		1.49±0.40	1.27±0.53	1.61±0.48

† The Post-Op group differed significantly from the CTRL group (p < 0.05).

Pelvic sagittal ROM was significantly greater postoperatively compared to the preoperative condition for both descent (p=0.016, d=0.87) and ascent (p=0.047, d=0.68) phases of the squat in both cases suggesting a high to moderate significance (Table 5.3). Although the FAI group was able to achieve greater peak hip flexion following surgery, the difference was not significant but showed a trend (p=0.054, d=0.66) between pre- and postoperative conditions, nor for hip sagittal ROM. No significant differences were observed in hip abduction, and knee flexion (Table 3), as well as when compared to the CTRL group.

Table 5.3. Hip and pelvis kinematics during the descent and ascent phases of the squat.

	FAI Pre-Op	FAI Post-Op	CTRL
Pelvic ROM (°) – squat descent*	9.0 ± 4.5	16.0 ± 6.2	11.7 ± 7.8
Pelvic ROM (°) – squat ascent*	8.9 ± 3.4	14.7 ± 7.3	10.4 ± 7.3
Hip ROM (°)	91.4 ± 24.2	101.3 ± 7.2	101.1 ± 7.3
Peak hip flexion (°)	95.4 ± 19.5	104.1 ± 8.8	103. ± 8.6
Peak hip abduction (°)	13.3 ± 6.2	11.6 ± 4.8	13.5 ± 3.3
Peak knee flexion (°)	123.0 ± 15.1	121.3 ± 20.3	118.2 ± 7.3

Data are reported as mean ± SD

* significant difference (p<0.05) between FAI pre- and post-op

During the squat descent phase, the postoperative FAI patients had decreased PeakLE for the glutei (GMax and GMed) and RF muscles, but increased for the ST muscle, when compared to their preoperative values (Figure 5.2A). No differences in PeakLE existed between preoperative and postoperative conditions during the squat ascent phase (Figure 5.2B). The postoperative FAI patients had increased iEMG for the GMed and the TFL muscles during the squat descent (Figure 5.2C) and ascent (Figure 5.2D) phases, respectively when compared to their preoperative condition. Both hamstring muscles (BF and ST) had an increase in iEMG for both phases of the squat; however, it did not reach significance.

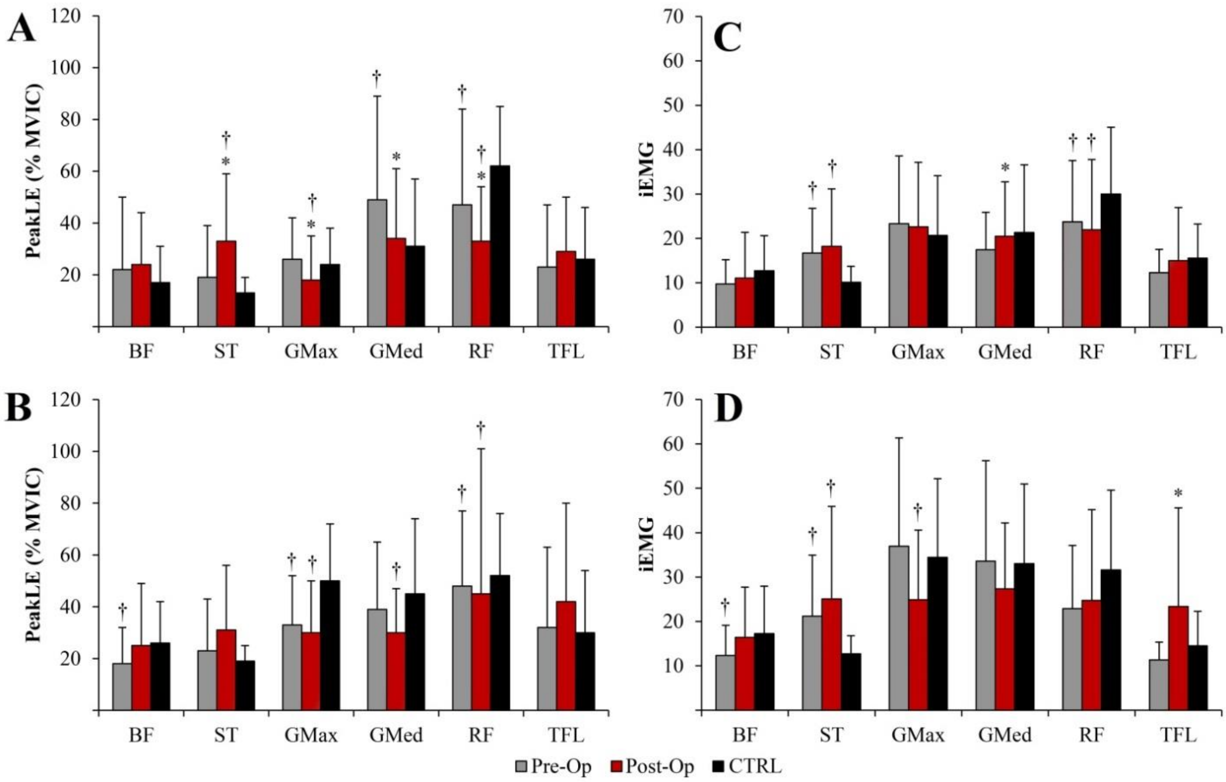


Figure 5.2. EMG peak linear envelope (PeakLE) and total activity (iEMG) for the *biceps femoris* (BF), *semitendinosus* (ST), *gluteus maximus* (Gmax), *gluteus medius* (GMed), *rectus femoris* (RF) and *tensor fasciae latea* (TFL) muscles during descent (A & C) and ascent (B & D) phases of the squat task. * significant difference (p<0.05) compared with FAI pre-op. † significant difference (p<0.05) compared with CTRL.

5.4. Discussion

The osteochondroplasty of the femoral head-neck junction has been chosen as an option to relieve pain for cam FAI patients; however, with only a few studies objectively measuring the biomechanical outcomes before and after surgery,[34–36,38,43] the evidence is primarily limited to PROMs.

Of the few biomechanical studies, many have done gait comparisons which do not require an extreme ROM of the hip, such as a deep squatting task. Using this task to compare the biomechanics and muscle activity in patients before and after surgery can empirically evaluate if the patient improves their function and mobility at near impingement. The findings of our study show that the post-surgery participants have improved pelvic mobility, but this did not translate to improved squat depth or an increase in hip ROM. At the two-year follow-up, postoperative patients were weaker when compared to their matched-healthy CTRLs when tested isometrically for hip flexion and hip flexion combined with abduction. Also, the maximum muscle activity and total muscle activity changed during the squat performance.

Few studies examined the biomechanics of a squat movement or have reported squat data in patients with FAI.[10,34,39,40,44–46] Four of the previous studies examined maximal squat depth,[10,40,45,46] whereas the other study had patients perform a squat to 25% of the total body height at a controlled speed;[44] thus making it difficult to compare our results to the latter. Only two studies had compared squat depth between pre- and postoperative conditions.[34,39] Our study showed a deep squat of 33.1% (pre-op) and 34.2% (post-op) of leg length, which are similar values measured by previous studies.[34,39] Only one of these studies found a significant difference between pre- and postoperative conditions[34] whereas the most recent study did not show any statistical significance between the conditions[39] like the present work. For both studies[34,39] their cohort had similar mixed surgical approaches as the current study. We cannot conclusively determine that squat depth is changed following corrective surgery. The maximum squat depth on the postoperative patients did not improve from the preoperative condition, and it remained still higher than on the controls suggesting a reduced joint ROM from the lower limb joints.

At the two-year follow-up, the FAI patients have shown increased pelvic mobility during the squat task, although no differences in peak hip flexion or hip ROM were observed. Previous studies also did not find any significant differences in hip kinematics during a squat task.[34] An improving in the pelvic ROM, but not at the hip joint, could help to justify why the ultimate measurement, the squat depth, was not improved, although being performed with different kinematics. Perhaps less pain and better proprioception of the joint had a positive effect on the pelvic ROM that was not transferred to the hip, limiting the squat depth performance. The hypothesis raised is that after many years of dealing with pain at end ranges of motion while waiting for corrective surgery may have caused soft tissue stiffening and contraction imbalance of the muscles surrounding the hip joint. Which was verified by the flexion/extension hip strength ratio analysis, where the preoperative patients showed an anteroposterior muscle force imbalance when compared with the healthy CTRLs. A recent systematic review has suggested that in patients with hip stiffness, a capsular release may be appropriate[47] during an FAI corrective surgery. Soft tissue stiffness would affect movement, especially during closed chain tasks such as the squat. Perhaps the release of stiff soft-tissue structures while performing the osteochondroplasty may be a strategy to allow also hip ROM improvements. We measured sit-and-reach flexibility and found no difference between the pre- and postoperative values. As this test limits the flexibility measurement of the back and hamstrings only, future research should compare the active and passive ROM of the hip in patients before and after the FAI corrective surgery. Also, aftercare rehabilitation should aim at improving the flexibility of the soft tissues and muscles surrounding the pelvis and hip, which will improve mobility during open and closed chain tasks.

Muscle strength is another important factor that will need to be addressed following surgery in FAI individuals. The postoperative FAI patients were weaker during pure hip flexion and hip flexion-with-abduction compared to the CTRL participants during the isometric test. Therefore, aftercare rehabilitation should focus on improving the strength of the TFL and hip flexor muscles, as the strength gain in combination with flexibility will improve hip mobility. Previous studies on hip muscle strength in preoperative FAI patients during isometric tasks showed that hip flexors and TFL muscles were also significantly weaker compared to a control group.[48] Although the differences in this study arose only at the two-year follow-up patients, the conclusions remain the same, as patients with symptomatic FAI present muscle weakness for the hip flexors muscle groups; which also led to a muscle imbalance regarding the ratio between hip flexion and extension strength, when compared to the CTRL participants. As muscle weakness in osteoarthritis (OA) individuals can be an indicator of progression,[48,49] the assumption that FAI could potentially lead to hip OA[11,12,50] is asserted. Therefore the findings of this study support the assessment of hip muscle strength in routine clinical examinations to help diagnose FAI.[48,51] The differences observed in muscle strength are best explained by EMG information.

Postoperative FAI patients used differently the muscle activity synergies in the hip compared to preoperative conditions. PeakLE EMG has shown a higher peak activation of the ST, while both glutei muscle and the RF lowered during the descent phase of the squat. The total muscle activation also demonstrated an increase for the GMed and the TFL during squat descending and ascending, respectively. Excessive ST activation in symptomatic cam FAI during the squat has already been reported[45] (Chapter 4), and it can be associated with the lack of force of the hip flexors. During the descent phase of the squat, the ST acts eccentrically to control the movement, in the FAI the ST over activates to compensate the hip flexor weakness and allow the task to be performed; as a biarticular muscle, it can be associated with the limited preoperative pelvic tilt and highlights the muscle unbalance. GMax total muscle activation was also significantly lower at the two-year follow-up when compared to the CTRL group. Perhaps the muscle peak activation reduction observed in the glutei, and RF muscles have been caused by an accumulation of adipose tissue within the muscle. The intermuscular adipose tissue (IMAT) or fatty infiltration has been previously reported in orthopedics in post-surgical cases following rotator cuff repairs.[52,53] Macroarchitectural changes, such as the trauma caused by surgery, may facilitate fatty infiltration between muscle fibers,[54] which may cause a rearrangement of the muscular tissue.[54] The cells responsible for causing IMAT, fibro/adipogenic progenitors, replace muscle cells with a mix of fibrous tissue and white adipocytes following surgery, compromising the muscle function.[55] This could explain why we observed a decreased activation in the glutei and RF muscles postoperatively, compensated by an increase in ST activation.

High levels of IMAT have been shown to lower muscle strength in the legs.[56] Moreover, although not significant, at the follow-up, all hip muscles examined in the FAI patients were weaker postoperatively (Table 5.2). Increased levels of IMAT has been found in young adults who had reduced physical activity, causing an increase of 15-20% in IMAT and a strength reduction of 4-6%.[57] Future research should examine muscle fiber composition and architecture in FAI patients pre- and postoperatively to confirm if IMAT of the hip muscles does increase following surgery.[52] If this is the case, FAI treatment should be reconsidered.

This study had certain limitations. First, our cohort consisted of only male participants; however, as the cam morphology is statistically more prevalent in males,[58-60] sex comparison was not warranted. Second, this research did not focus on comparing surgical approaches since our cohort was not large enough for achieving a meaningful power. However non-parametric analysis showed no differences in the analyzed variables amongst the approaches. It has been suggested that the arthroscopic approach offers better muscle preservation which could provide a better joint function.[30,32,61] As a minimum of two years was used for the patients' reassessment, we believe that no short-term benefits between the two approaches would have arisen after two years, also in a mixed approach cohort, a two-year follow-up have showed that FAI surgical correction was associated with decreased T1ρ and bone mass density, improving the overall health of the hip joint.[39] Additionally, one systematic review has shown that one approach is not significantly superior to the other.[32] Third, we did not control for the speed of the squat between participants. Participants were instructed to squat at a controlled and self-selected pace. The speed of movement could affect the joint moments, however, we believe that speed would have minimal effect on kinematics variables. Also, all trials have been time normalized during its processing.

5.5. Conclusions

Although at the two-year follow-up analysis the cam FAI patients did not improve the squat depth, they have shown increased pelvic ROM and positive PROMs. The weakness of muscles associated with hip flexion and flexion-with-abduction were also observed at the follow-up, which may be associated with the alterations in the muscle activity and neuromuscular patterns. The use of squat test pre- and post-surgical correction of cam FAI can provide valuable information for the clinical practice while identifying pelvic mobility in a dynamic task. The rehabilitation program should focus on increasing the flexibility and strength of muscles around the pelvis and hip, with special attention on strengthening the hip flexors and TFL muscles. Increasing flexibility of other lower extremity muscles should not be overlooked as it will improve mobility during closed kinetic/kinematic chain tasks. Implementation of hip muscular strength measurement before and after surgery may provide additional insights into the rehabilitation program, as muscle weakness may have caused a change in the muscular contraction strategy, and also as a tool to evaluate muscle strength balance. Further research needs to examine medical images of FAI patients before and after surgery to determine if the intervention can be a cause for intermuscular adipose tissue.

5.6. References

1. Martin DE, Tashman S. 2010. The Biomechanics of Femoroacetabular Impingement. Oper. Tech. Orthop. 20(4):248–254.

2. Nussbaumer S, Leunig M, Glatthorn JF, et al. 2010. Validity and test-retest reliability of manual goniometers for measuring passive hip range of motion in femoroacetabular impingement patients. BMC Musculoskelet. Disord. 11:194.

3. Byrd JWT. 2014. Femoroacetabular Impingement in Athletes: Current Concepts. Am. J. Sports Med. 42(3):737–51.

4. Beaulé PE, Hynes K, Parker G, Kemp KA. 2012. Can the alpha angle assessment of cam impingement predict acetabular cartilage delamination? Clin Orthop Relat Res 470(12):3361–3367.

5. Nötzli H, Wyss T, Stoecklin C, et al. 2002. The Contour of the Femoral Head-Neck Junction as a Predictor for the Risk of Anterior Impingment: Commentary. J. bone Jt. Surg. 84(B):556–60.

6. Rakhra KS, Sheikh AM, Allen D, Beaulé PE. 2009. Comparison of MRI alpha angle measurement planes in femoroacetabular impingement. Clin. Orthop. Relat. Res. 467(3):660–665.

7. Nouh MR, Schweitzer ME, Rybak L, Cohen J. 2008. Femoroacetabular impingement: Can the alpha angle be estimated? Am. J. Roentgenol. 190(5):1260–1262.

8. Ng KG, Lamontagne M, Beaulé PE. 2016. Differences in Anatomical Parameters between the Affected and Unaffected Hip in Patients with Bilateral Cam-Type Deformities. Clin. Biomech. 33:13–19.

9. Hack K, Di Primio G, Rakhra K, Beaulé PE. 2010. Prevalence of Cam-Type Femoroacetabular Impingement Morphology in Asymptomatic Volunteers. J. Bone Jt. Surg. 92(14):2436.

10. Ng KG, Lamontagne M, Adamczyk AP, et al. 2015. Patient-Specific Anatomical and Functional Parameters Provide New Insights into the Pathomechanism of Cam FAI. Clin. Orthop. Relat. Res. 473(4):1289–1296.

11. Ganz R, Parvizi J, Beck M, et al. 2003. Femoroacetabular Impingement: a Cause for Osteoarthritis of the Hip. Clin. Orthop. Relat. Res. (417):112–120.

12. Leunig M, Ganz R. 2005. Femoroacetabular impingement. A common cause of hip complaints leading to arthrosis. Unfallchirurg 108(1):9–10, 12–7.

13. Crawford JR, Villar RN. 2005. Current concepts in the management of femoroacetabular impingement. J. Louisiana State Med. Soc. 144:1459–1462 Available from: http://www.ncbi.nlm.nih.gov/pubmed/16260657.

14. Loudon JK, Reiman MP. 2014. Conservative management of femoroacetabular impingement (FAI) in the long distance runner. Phys. Ther. Sport 15(2):82–90 Available from: http://dx.doi.org/10.1016/j.ptsp.2014.02.004.

15. Griffin DR, Dickenson EJ, O'Donnell J, et al. 2016. The Warwick Agreement on femoroacetabular impingement syndrome (FAI syndrome): an international consensus statement. Br. J. Sports Med. 50(19):1169–76.

16. Wall PDH, Fernandez M, Griffin DR, Foster NE. 2013. Nonoperative Treatment for Femoroacetabular Impingement: A Systematic Review of the Literature. PM R 5(5):418–426.

17. Casartelli NC, Bizzini M, Kemp J, et al. 2017. What treatment options exist for patients with femoroacetabular impingement syndrome but without surgical indication? Br. J. Sports Med. :bjsports-2017-098059.

18. Beaulé PE, Le Duff MJ, Zaragoza E. 2007. Quality of life following femoral head-neck osteochondroplasty for femoroacetabular impingement. J. Bone Jt. Surg. - Ser. A 89(4):773–779.

19. Emara K, Samir W, Motasem EH, Ghafar KA El. 2011. Conservative treatment for mild femoroacetabular impingement. J. Orthop. Surg. (Hong Kong) 19(1):41–5.

20. Hunt D, Prather H, Hayes MH, Clohisy JC. 2012. Clinical Outcomes Analysis of Conservative and Surgical Treatment of Patients With Clinical Indications of Prearthritic, Intra-articular Hip Disorders. Pm&R 4(7):479–487.

21. Ribas-Fernandez M, Marin-Peña OR, Regenbrecht B, et al. 2007. Hip osteoplasty by an anterior minimally invasive approach for active patients with femoroacetabular impingement. HIP Int. 17(2):91–98.

22. Lincoln M, Johnston K, Muldoon M, Santore R. 2009. Combined Arthroscopic and Modified Open Approach for Cam Femoroacetabular Impingement: A Preliminary Experience. Arthrosc. - J. Arthrosc. Relat. Surg. 25(4):392–399.

23. Laude F, Sariali E, Nogier A. 2009. Femoroacetabular impingement treatment using Arthroscopy and Anterior approach. Clin. Orthop. Relat. Res. 467(3):747–752.

24. Weiland DE, Philippon MJ. 2005. Arthroscopic Technique of Femoroacetabular Impingement. Oper. Tech. Orthop. 15(3):256–260.

25. Philippon MJ, Stubbs AJ, Schenker ML, et al. 2007. Arthroscopic management of femoroacetabular impingement: osteoplasty technique and literature review. Am. J. Sports Med. 35(9):1571–1580.

26. Byrd JWT, Jones KS. 2009. Arthroscopic Femoroplasty in the management of cam-type Femoroacetabular impingement. In: Clinical Orthopaedics and Related Research. p 739–746.

27. Philippon MJ, Schenker ML. 2006. Arthroscopy for the Treatment of Femoroacetabular Impingement in the Athlete. Clin. Sports Med. 25(2):299–308.

28. Horisberger M, Brunner A, Herzog RF. 2010. Arthroscopic treatment of femoroacetabular impingement of the hip: A new technique to access the joint. Clin. Orthop. Relat. Res. 468(1):182–190.

29. Joseph R, Pan X, Cenkus K, et al. 2016. Sex Differences in Self-Reported Hip Function Up to 2 Years After Arthroscopic Surgery for Femoroacetabular Impingement. Am J Sport. Med 44(1):54–59.

30. Philippon M, Briggs K, Yen Y-M, Kuppersmith D. 2009. Outcomes following hip arthroscopy for femoroacetabular impingement with associated chondrolabral dysfunction: minimum two-year follow-up. J Bone Jt. Surg [Br] 91:16–23.

31. Larson CM, Giveans MR. 2008. Arthroscopic Management of Femoroacetabular Impingement: Early Outcomes Measures. Arthrosc. - J. Arthrosc. Relat. Surg. 24(5):540–546.

32. Botser IB, Smith TW, Nasser R, Domb BG. 2011. Open surgical dislocation versus arthroscopy for femoroacetabular impingement: A comparison of clinical outcomes. Arthrosc. - J. Arthrosc. Relat. Surg. 27(2):270–278.

33. Malagelada F, Del Carmen VA, Barke SJ, et al. 2015. The Anterior Mini-Open Approach for Femeroacetabular Impingement: Gait and Functional Assessment at One Year Post-Surgery. Ann. Phys. Rehabil. Med. 58(2):60–65.

34. Lamontagne M, Brisson N, Kennedy MJ, Beaulé PE. 2011. Preoperative and Postoperative Lower-Extremity Joint and Pelvic Kinematics during Maximal Squatting of Patients with Cam Femoroacetabular Impingement. J. Bone Jt. Surg. 93(2):40–5.

35. Brisson N, Lamontagne M, Kennedy MJ, Beaulé PE. 2013. The Effects of Cam Femoroacetabular Impingement Corrective Surgery on Lower-Extremity Gait Biomechanics. Gait Posture 37(2):258–263.

36. Rylander JH, Shu B, Andriacchi TP, Safran MR. 2011. Preoperative and Postoperative Sagittal Plane Hip Kinematics in Patients with Femoroacetabular Impingement during Level Walking. Am. J. Sports Med. 39:36S–42S.

37. Rylander J, Shu B, Favre J, et al. 2013. Functional testing provides unique insights into the pathomechanics of femoroacetabular impingement and an objective basis for evaluating treatment outcome. J. Orthop. Res. 31(9):1461–1468.

38. Bedi A, Dolan M, Hetsroni I, et al. 2011. Surgical Treatment of Femoroacetabular Impingement Improves Hip Kinematics. Am. J. Sports Med. 39(Supplement 1):43S–49S.

39. Beaulé PE, Speirs AD, Anwander H, et al. 2017. Surgical Correction of Cam Deformity in Association with Femoroacetabular Impingement and Its Impact on the Degenerative Process within the Hip Joint. J. Bone Jt. Surg. 99(16):1373–1381.

40. Diamond LE, Bennell KL, Wrigley T V., et al. 2017. Squatting biomechanics in individuals with symptomatic femoroacetabular impingement. Med. Sci. Sports Exerc. 49(8):1520–1529.

41. Wells KF, Dillon EK. 1952. The Sit and Reach—A Test of Back and Leg Flexibility. Res. Quarterly. Am. Assoc. Heal. Phys. Educ. Recreat. 23(1):115–118 Available from: http://www.tandfonline.com/doi/abs/10.1080/10671188.1952.10761965.

42. Mantovani G, Lamontagne M. 2016. How different marker sets affect joint angles in inverse kinematics framework. J. Biomech. Eng. 139(4):044503.

43. Kennedy MJ, Lamontagne M, Beaulé PE. 2009. Femoroacetabular Impingement Alters Hip and Pelvic Biomechanics during Gait: Walking Biomechanics of FAI. Gait Posture 30(1):41–44.

44. Kumar D, Dillon A, Nardo L, et al. 2014. Differences in the association of hip cartilage lesions and cam-type femoroacetabular impingement with movement patterns: A preliminary study. PM R 6(8):681–689.

45. Catelli DS, Kowalski E, Beaulé PE, et al. 2018. Asymptomatic Individuals with Femoroacetabular Deformity Demonstrate Stronger Hip Extensors and Greater Pelvis Mobility during the Deep Squat Task. Orthop. J. Sport. Med. 6(7):1–10.

46. Lamontagne M, Kennedy MJ, Beaulé PE. 2009. The Effect of Cam FAI on Hip and Pelvic Motion during Maximum Squat. Clin Orthop Relat Res 467:645–650.

47. Ortiz-Declet V, Mu B, Chen AW, et al. 2018. Should the Capsule Be Repaired or Plicated After Hip Arthroscopy for Labral Tears Associated With Femoroacetabular Impingement or Instability? A Systematic Review. Arthroscopy 34(1):303–318.

48. Casartelli NC, Maffiuletti NA, Item-Glatthorn JF, et al. 2011. Hip muscle weakness in patients with symptomatic femoroacetabular impingement. Osteoarthr. Cartil. 19(7):816–821.

49. Brandt KD. 2003. Is a Strong Quadriceps Muscle Bad for a Patient with Knee Osteoarthritis? Ann. Intern. Med. 138(8):678–679.

50. Leunig M, Beaulé PE, Ganz R. 2009. The Concept of Femoroacetabular Impingement: Current Status and Future Perspectives. Clin. Orthop. Relat. Res. 467(3):616–622.

51. Kelly BT, Williams RJ, Philippon MJ. 2003. Hip arthroscopy: current indications, treatment options, and management issues. Am. J. Sports Med. 31(6):1020–37.

52. Meyer DC, Hoppeler H, von Rechenberg B, Gerber C. 2004. A pathomechanical concept explains muscle loss and fatty muscular changes following surgical tendon release. J. Orthop. Res. 22(5):1004–1007.

53. Goutallier D, Postel JM, Bernageau J, et al. 1994. Fatty muscle degeneration in cuff ruptures. Pre- and postoperative evaluation by CT scan. Clin. Orthop. Relat. Res. (304):78–83.

54. Addison O, Marcus RL, Lastayo PC, Ryan AS. 2014. Intermuscular fat: A review of the consequences and causes. Int. J. Endocrinol. 2014.

55. Joe AWB, Yi L, Natarajan A, et al. 2010. Muscle injury activates resident fibro/adipogenic progenitors that facilitate myogenesis. Nat. Cell Biol. 12(2):153–163.

56. Goodpaster BH, Carlson CL, Visser M, et al. 2001. Attenuation of skeletal muscle and strength in the elderly: The Health ABC Study. J. Appl. Physiol. 90:2157–2165.

57. Manini TM, Clark BC, Nalls M a, et al. 2007. Reduced physical activity increases intermuscular adipose tissue in healthy young adults. Am. J. Clin. Nutr. 85(2):377–384.

58. Kassarjian A, Brisson M, Palmer WE. 2007. Femoroacetabular impingement. Eur. J. Radiol. 63(1):29–35.

59. Allen D, Beaule PE, Ramadan O, Doucette S. 2009. Prevalence of associated deformities and hip pain in patients with cam-type femoroacetabular impingement. J. Bone Jt. Surg. - Br. Vol. 91–B(5):589–594.

60. Beck M, Kalhor M, Leuning M, Ganz R. 2005. Hip Morphology Influences the Pattern of Damage to the Acetabular Cartilage: Femoroacetabular Impingement as a Cause of Early Osteoarthritis of the Hip. J. Bone Jt. Surg. - Br. Vol. 87–B(7):1012–1018.

61. Papalia R, Del Buono A, Franceschi F, et al. 2012. Femoroacetabular impingement syndrome management: Arthroscopy or open surgery? Int. Orthop. 36(5):903–914.

6 Does the Dual-Mobility Hip Prosthesis Produce Better Joint Kinematics during Extreme Hip Flexion Task?

Danilo S. Catelli [12] | Erik Kowalski [1] | Paul E. Beaulé [3] | Mario Lamontagne [134]

[1]School of Human Kinetics, University of Ottawa, Ottawa, Ontario, Canada.
[2]CAPES Foundation, Ministry of Education of Brazil, Brasilia, Brazil.
[3]Division of Orthopaedic Surgery, University of Ottawa, Ottawa, Ontario, Canada.
[4]Department of Mechanical Engineering, University of Ottawa, Ottawa, Ontario, Canada.

Contents of this chapter published as an article in *The Journal of Arthroplasty®* | May 2017

Catelli DS, Kowalski E, Beaulé PB, Lamontagne M, Does the Dual-Mobility Hip Prosthesis Produce Better Joint Kinematics during Extreme Hip Flexion Task? *J Arthroplasty*, 2017; 32(10):3206-3212.

DOI: 10.1016/j.arth.2017.04.049

6.0. Abstract

Background: Total hip arthroplasty (THA) using dual-mobility (DM) design permits larger hip range of motion. However, it is unclear how it benefits the patients during activities of daily living. The purpose was to compare kinematic variables of the operated limb between THA patients using either DM or single-bearing (SB) implants during a squat task.

Methods: Twenty-four THA patients were randomly assigned to either a DM or SB implant and matched to 12 healthy controls (CTRLs). They underwent 3-dimensional squat motion analysis before and 9 months after surgery. Sagittal and frontal plane angles of the pelvis and the hip were analyzed using statistical parametric mapping. Paired analyses compared presurgery and postsurgery squat depth.

Results: Peak sagittal pelvis angle of DM was closer to normal compared with that of SB. Both implant groups had similar hip angle patterns and magnitude but significantly lower than the CTRLs. SB reached a much large hip abduction compared with the other groups. Both surgical groups had significantly worst squat depth than the CTRLs.

Conclusion: Neither THA implant groups were able to return pelvis and hip kinematics to the level of CTRLs. The deficit of DM implants at the pelvis combined with the poorer functional scores should caution clinicians to use this implant design in active patients. SB design causes a larger hip abduction to reach their maximum squat depth. Post-THA rehabilitation should focus on improving joint range of motion and strength.

6.1. Introduction

Dual-mobility (DM) acetabular cups have been shown to reduce the risk of dislocation and have been typically used in revision surgeries [1,2], primary cases for femoral neck fractures [3],or patients at high risk of dislocation (ie, patients with previous spinal fusion) [4]. The DM acetabular cup allows for increased range of motion (ROM) [5]. However, the benefit of this increased ROM during activities of daily living (ADLs) is unclear. Motion analysis is a valuable tool to measure the effectiveness of surgery to compare the improvement of patient's movements after surgery. Since the DM hip implant has typically been used for revision surgeries, its potential advantages during primary total hip arthroplasty (THA) have not been examined. In addition, regarding DM cups as primary THA, there are concerns such as the potential for increased wear, intraprosthetic dislocation, and risk of groin pain [6-9]. Recent wear study has found no difference in implant wear between DM and the traditional 22-mm metal-polyethylene (PE) bearings [10]. Biomechanical studies have shown that THA patients with DM or SB implants have significantly improved their gait and kinematic parameters from the presurgery, but did not reach the level of the controls (CTRLs) [11,12]. To determine if benefits exist with DM implants during primary THA, a squat task would allow comparison at extreme ROM with traditional standard/single- bearing (SB) implants.

The purpose of the present study was to analyze and compare lower limb joint kinematics of patients undergoing primary THA with either a DM or SB implants during a deep squatting task. In this study, we addressed 2 research questions: (1) Does THA surgery with either a DM or SB implant result in pelvis and hip joint kinematics closer to healthy CTRLs during a deep squatting task? and (2) Does THA surgery with a DM implant allow for patients to achieve a deeper squat depth than with an SB implant?

6.2. Material and Methods

The study design was a randomized controlled study. Surgeries were performed at the local hospital, and testing was performed at the local university. Participants provided informed consent before the study, and all investigations were conducted ethically in conformity with research principles.

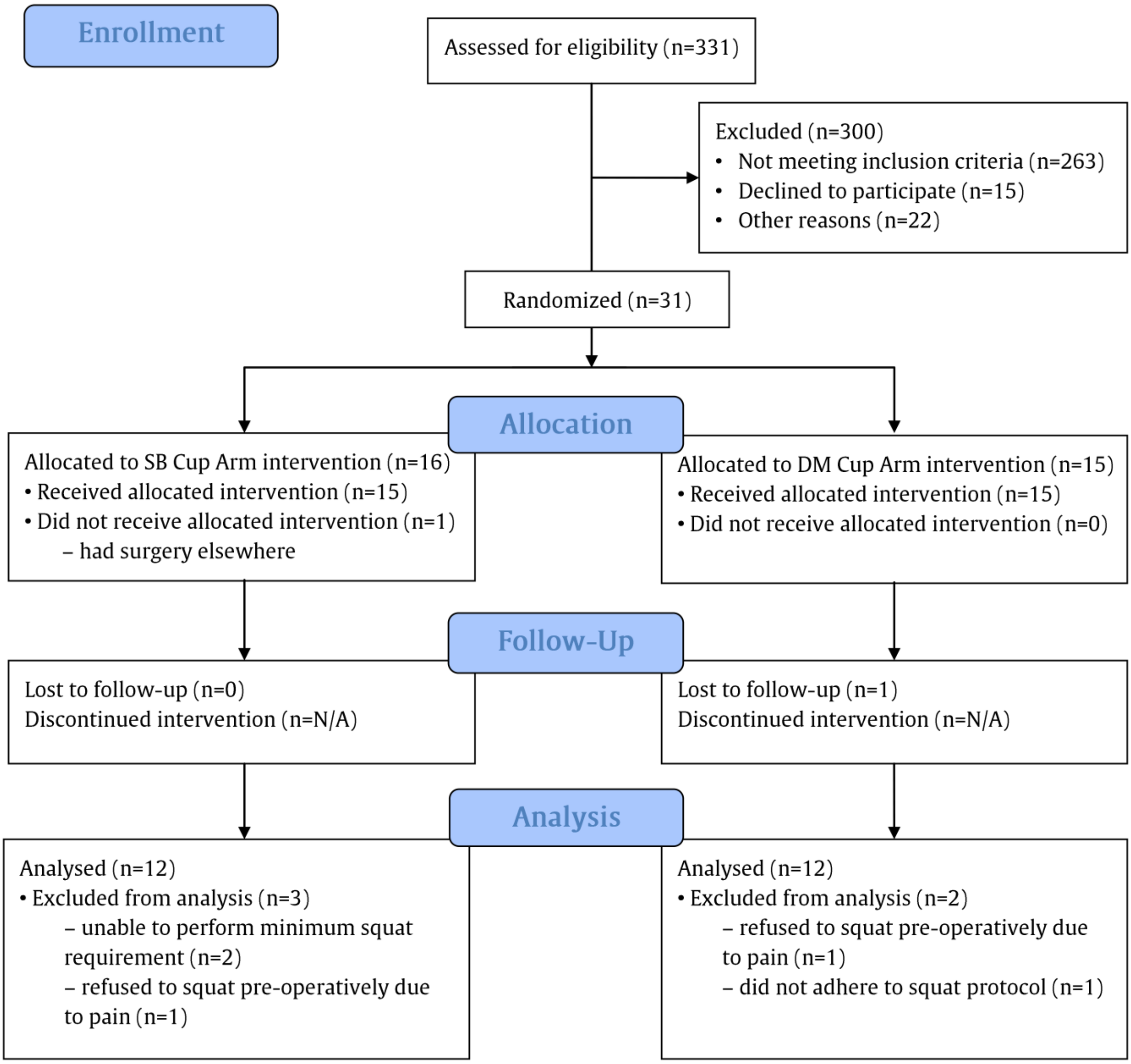


Figure 6.1. CONSORT flowchart describing the patients' study enrollment.

Three hundred thirty-one patients were screened using the CONSORT guidelines for randomized clinical trials (Figure 6.1). Thirty- one patients undergoing unilateral primary THA for noninflammatory degenerative osteoarthritis between 55 and 79 years of age were recruited from the local university hospital. Although DM design is frequently recommended in cases where patients are at high risk of dislocation [3,4] to provide a more fair comparison between the 2 designs in primary THA, the same exclusion criteria were applied to both groups of patients. The patients suffering from any other lower limb joint disorders other than the one for THA, with any other joint arthroplasty at the ipsilateral or contralateral limb, with evidence of active infection, with neurologic or musculoskeletal disease that may adversely affect gait or weight bearing, with neuropathic joints, and with requiring structural bone grafts were excluded from participating in the study. THA patients were prospectively randomized to either a DM implant (Versafitcup DM; Medacta) or SB acetabular component (Versafitcup CC; Medacta) using the Medacta system (Medacta International, Switzerland). All patients underwent direct anterior approach using the positioning table [13]. Finally, 12 THA patients in each implant group were tested and matched by age, gender, and body mass index to 12 healthy participants (CTRL; Table 6.1). The reason for the excluded participants post- processing was mainly either because they were unable to reach the minimum squat depth of 90% of their leg length or because the patients were unwilling in performing the task because of either fear or pain.

Table 6.1. Demographics of the Both THA Implants and CTRL Groups, Including the HOOS for Symptoms, Pain, Function-DL, Function-SRA, and QOL.

Demographics	**DM**		**SB**		**CTRL**
	Preop	Postop	Preop	Postop	
Number (M/F)	8/4		10/2		6/6
Age (y)	63.1 ± 5.6	63.6 ± 5.6	62.7 ± 4.9	63.3 ± 4.9	62.5 ± 9.5
BMI (kg/m^2)	28.1 ± 2.7	28.3 ± 0.9	29.6 ± 4.8	29.4 ± 3.9	25.8 ± 3.8
Follow-up (mo)	6.8 ± 0.8		6.8 ± 1.3		N/A
HOOS Symptoms*†	39.2 ± 14.6	90.5 ± 9.1	41.5 ± 18.4	96.1 ± 5.5	97.9 ± 4.0
HOOS Pain*†	42.3 ± 15.0	87.3 ± 17.3	48.2 ± 18.5	98.9 ± 1.8	99.1 ± 3.0
HOOS DL*†	47.1 ± 18.9	92.6 ± 9.9	54.1 ± 19.6	96.1 ± 5.1	99.2 ± 1.9
HOOS SRA*†	21.6 ± 17.1	79.2 ± 17.1	32.5 ± 24.9	88.9 ± 12.4	99.5 ± 1.8
HOOS QOL*†‡	26.0 ± 15.9	80.1 ± 16.9	23.3 ± 16.8	86.8 ± 14.5	98.4 ± 3.9

Data are reported as mean ± SD. BMI, body mass index; CTRL, control; DL, daily living; DM, dual-mobility; F, female; HOOS, Hip Disability and Osteoarthritis Outcome Score; M, male; N/A, not applicable; Postop, postoperative; Preop, preoperative; QOL, quality of life; SB, single bearing; SRA, sports and recreational activities; THA, total hip arthroplasty.
* DM pre-op vs CTRL ($P \leq .001$)
† SB pre-op vs CTRL ($P \leq .001$)
‡ DM post-op vs CTRL ($P = .036$)

For the motion analysis sessions, participants were outfitted with 45 reflective markers affixed on bony landmarks as the University of Ottawa motion analysis model marker set [14] before performing a minimum of 5 deep squat trials at a self-selected pace. During the squat task, all participants stood with each foot on a force platform (Bertec Corporation, Columbus, OH), feet hip-width apart, and were instructed to squat as low as possible without lifting their feet from the floor. Three-dimensional joint kinematics of the lower limbs was collected at 200 Hz using a 10-camera (MX- 13) infrared motion analysis system (Vicon; Oxford Metrics, UK). Motion analysis and Hip disability Osteoarthritis Outcome Score (HOOS) were done preoperatively (preop) and at a minimum of 6 months postoperatively (postop) for THA patients, whereas CTRL participants completed only 1 motion analysis session.

Kinematic data were processed in Nexus 1.8.3 (Vicon; Oxford Metrics, UK) by running the University of Ottawa motion analysis model and then exported in a custom MATLAB script (MATLAB 2015b;MathWorks, Natick, MA) to calculate group averages, extract relevant variables, and run statistics. Variables of interest included sagittal kinematics of the pelvis and sagittal and frontal planes of the hip joint (Figure 6.2), as well as maximal squat depth. All trials were time normalized based on the descending and ascending squat cycle; maximum hip extension point (standing) and lowest depth point (squatted) were used as a reference; and individual averages for each participant were calculated across the 5 trials.

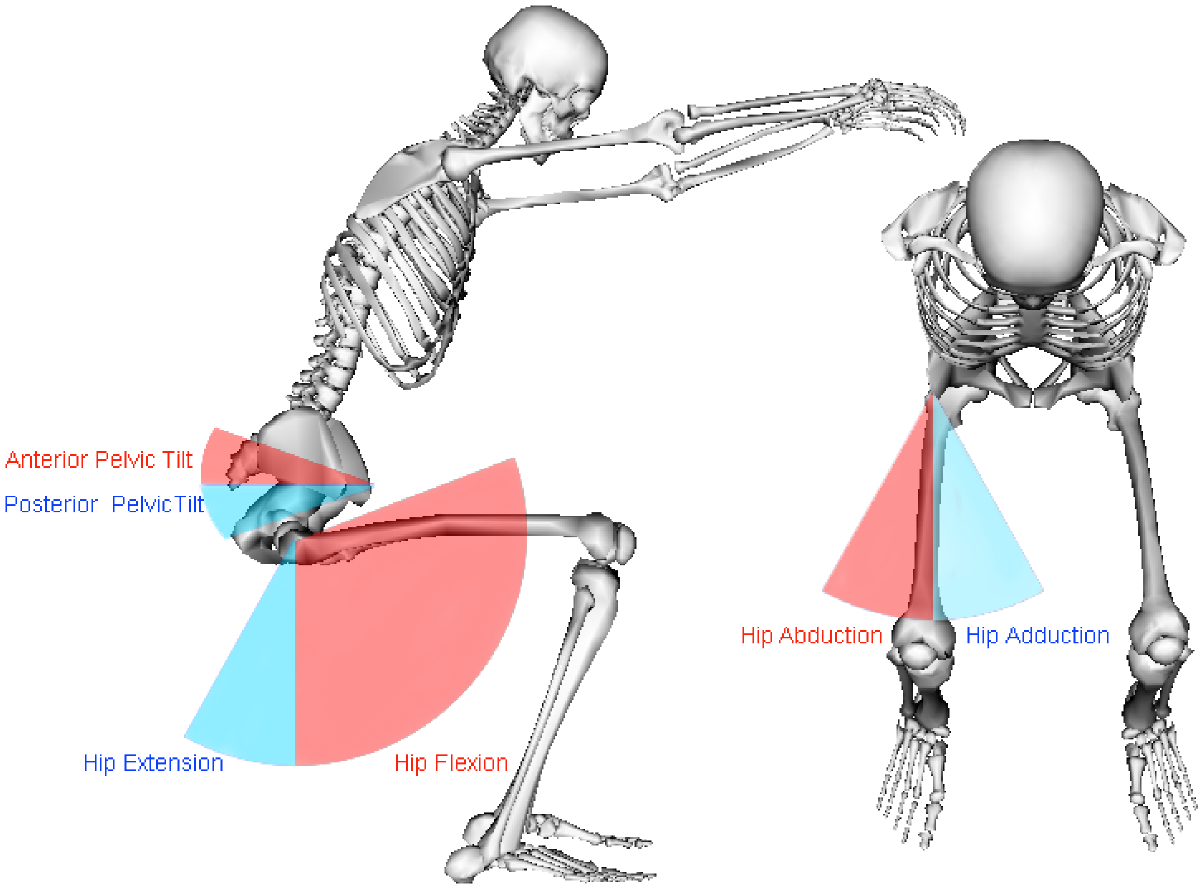


Figure 6.2. Representation of pelvic tilt, hip sagittal, and hip frontal angles during squatting.

Maximum squat depth was calculated based on the height of the center of the pelvis at its lowest point in the squat cycle divided by the length of the leg. The maximum squat depth is reported as a percentage of leg length (measured from the affected anterior superior iliac spine to the medial malleolus), with a value of 0% representing a maximum squat.

Joint kinematic waveform of all 3 groups were compared at each point of the cycle using statistical parametric mapping [15], which uses random field theory to identify field regions that significantly covary with a general linear model [16]. Squat depths between presurgery and postsurgery were compared using a paired t test, and 1-way analysis of variance with a Bonferroni post hoc comparison to determine where significant differences occurred between groups (confidence interval = 95%).

6.3. Results

At a mean follow-up of 6.8 ± 1.0 months, functional scores improved significantly in both groups with the SB group having overall better function than the DM group (Table 6.1).

Sagittal pelvic tilt was only significantly different between the DM and SB groups for the first 12% of the squat descent phase (Figure 6.3A). Compared with the CTRL group, the SB group was significantly different during the first 44% of the squat descent phase, whereas the DM group was significantly different during the first 5% of the squat ascent phase (Fig. 6.3B).

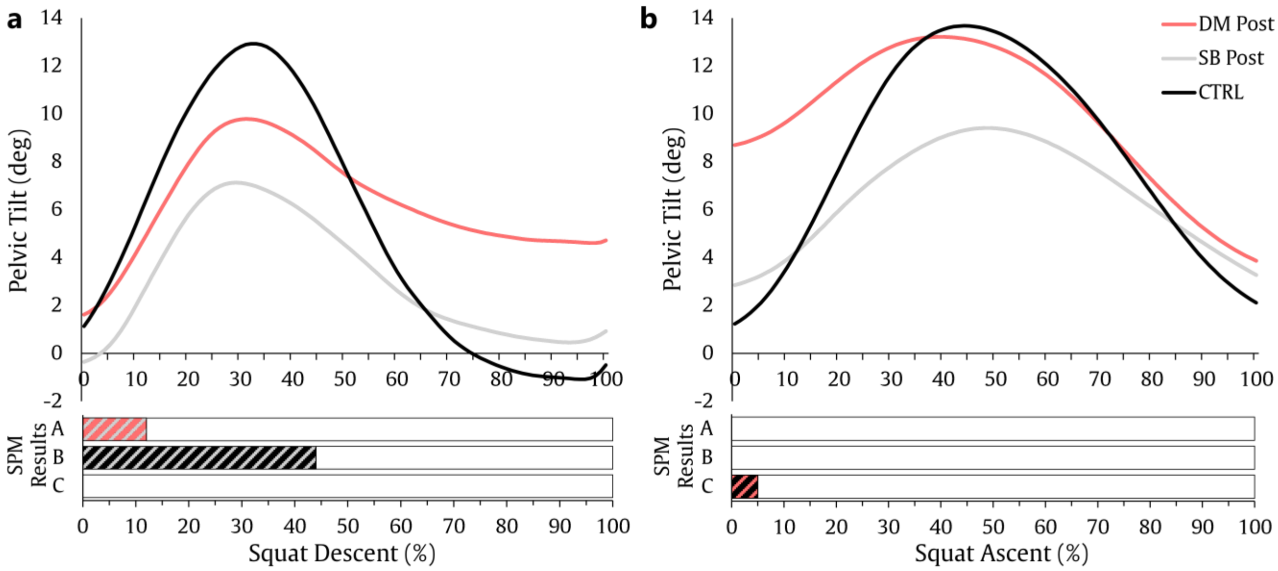


Figure 6.3. Mean of pelvic tilt angles during squatting task: (A) descent phase and (B) ascent phase. SPM results are displayed below the figure and indicate significant (P < .05) differences between (a) DM post vs SB post, (b) SB post vs CTRL, and (c) DM post vs CTRL. CTRL, control; DM, dual-mobility; SB, single bearing; SPM, statistical parametric mapping.

No significant differences in sagittal hip angles existed between the DM and SB groups throughout the squat descent or ascent phase. Compared with the CTRL group, the DM group was significantly different 3%-16% and 27%-100% of the squat descent phase, whereas the SB group was significantly different from 3%-100% of the squat descent phase (Figure 6.4A). For the squat ascent phase, the DM group was significantly different from the CTRL for the first 81% of the cycle, whereas the SB group was significantly different for the first 73% of the squat ascent cycle (Figure 6.4B).

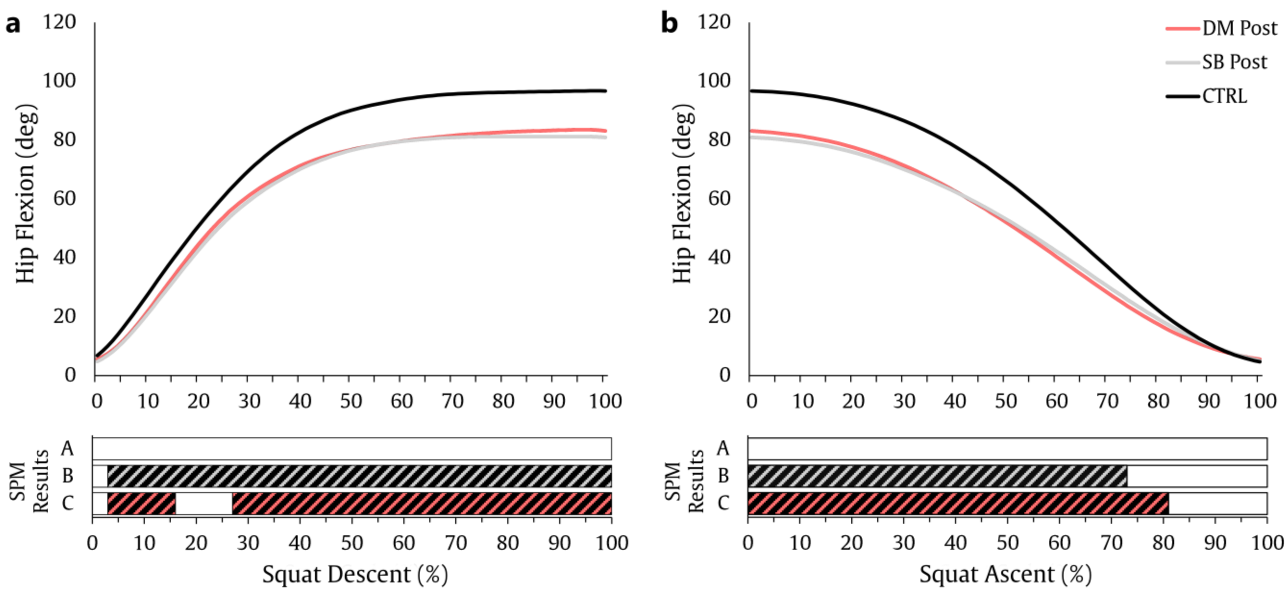


Figure 6.4. Mean of sagittal hip angles during squatting task: (A) descent phase and (B) ascent phase. SPM results are displayed below the figure and indicate significant (P < .05) differences between (a) DM post vs SB post, (b) SB post vs CTRL, and (c) DM post vs CTRL.

In the frontal plane of the hip, the DM and SB groups were significantly different from each other from 48% to 62% of the squat ascent phase (Figure 6.5B). The DM group was not significantly different from the CTRL group for either squat cycle, whereas the SB group was significantly different from the CTRL from 21% to 60% of the squat descent phase (Figure 6.5A) and from 2% to 43% of the squat ascent phase (Figure 6.5B).

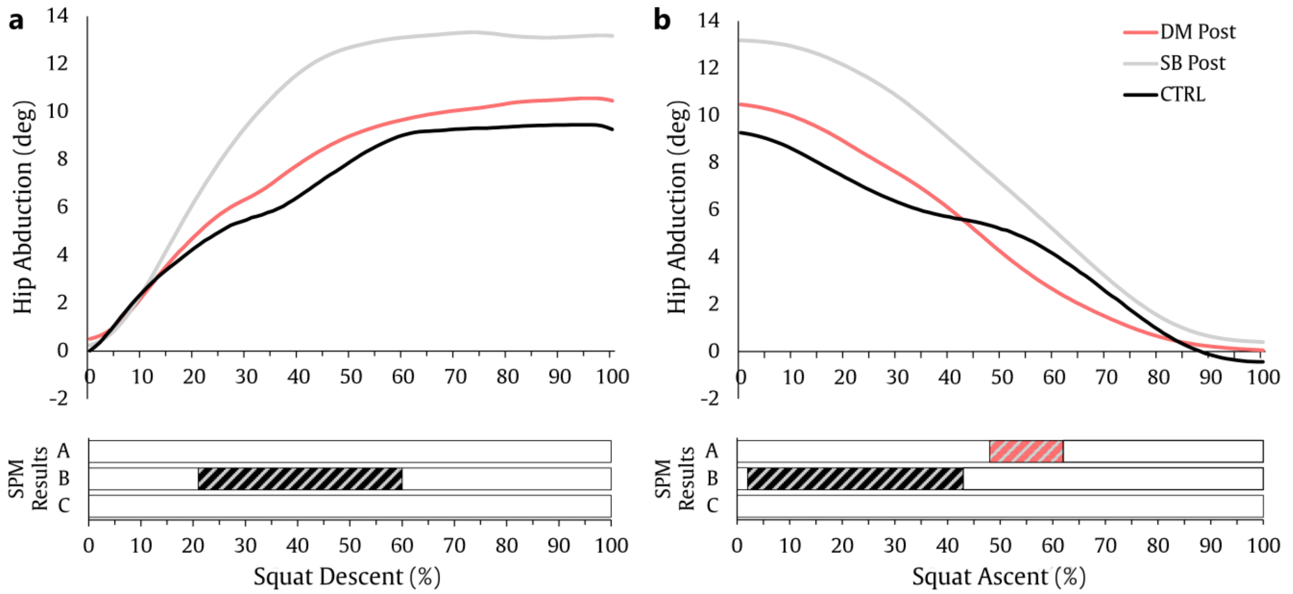


Figure 6.5. Mean of frontal hip angles during squatting task: (A) descent phase and (B) ascent phase. SPM results are displayed below the figure and indicate significant (P < .05) differences between (a) DM post vs SB post, (b) SB post vs CTRL, and (c) DM post vs CTRL.

Although squat depth did not improve postoperatively for the DM implant compared with its preoperative values (preop, 74.95% ± 13.55%; postop, 73.11% ± 10.94%; $P = .54$), the SB showed a significant improvement after surgery (preop, 72.73% ± 15.45%; postop, 68.96% ± 13.19%; $P = .04$; power = 0.442). When compared with the CTRL group, both groups remained significantly different ($P < .01$; power = 0.883).

6.4. Discussion

Patient-reported outcomes and overall health-related quality of life from THA patients demonstrated definite benefits from pre- surgery to postsurgery. The DM group had a lower score than the SB group, and the DM group was significantly different than the CTRL group. This means the DM group did not reach the level of expected function. Factors influencing a successful THA range from safety (ie, avoiding infection) and efficacy (maximize function) to effective- ness (avoiding revision). One key aspect of efficacy is the influence of prosthetic design/positioning on patient function [17,18]. Although most commonly, patient-reported outcomes are used to assess patient function, gait analysis can provide greater insight and sensitivity to better understand implant performance [19]. Most biomechanical gait studies examine gait-related ADLs [12] ranging from walking [20], stair ascent [21], and rising from a chair [22]. In general, patients' abilities to perform ADLs improve after surgery; however, more demanding activities, such as stair climbing and chair rising may still be difficult [23, 24]. Also, it is still unclear which ADLs put patients at risk of hip dislocation [25]. Combining this with the difficulties in defining optimal acetabular component orientation [26] and adjusting for sagittal pelvic alignment [27] have led surgeons to use larger head sizes as well as DM designs [28]. Although the basic biomechanical advantages of larger femoral heads (ie, decreased risk of impingement and greater jumping distance are evident) in improving hip stability are well known, this is not the case for DM designs.

In our prospective, randomized study comparing standard bearing designs with a DM design group and a CTRL group, we found that kinematics of both groups remained significantly different from the CTRL group during the squatting task. The most evident motion was for the sagittal hip motion (Fig. 6.4), where both groups were significantly different from the CTRL group for the majority of the movement. Similar findings existed in gait studies, which found that hip ROM in the sagittal plane was reduced post-THA compared with that of a CTRL group [20, 29-34]. This reduction in hip motion may be caused by a flexion contracture owing to increased passive resistance from the hip flexors, which is relatively prevalent in patients with THA [35]. Alternatively, this reduced hip motion may be due to a strength deficit of the hip flexors in THA patients compared with CTRLs [36]. Differences existed in the frontal hip kinematics as the SB group abducted more to complete the squat task (Fig. 6.5). The DM patients are perhaps better able to maintain their center of rotation owing to the large head size better controlling their squat strategy than the SB group. Similar findings were reported by Bouffard et al (2011) [37] when comparing hip resurfacing with other bearing designs. This requires further exploration, and this may have important long-term implications for patient function.

During the squat descent phase, the SB implant showed a sagittal pelvic tilt significantly different from the DM and CTRL groups for the first 12% of the cycle and continued to be significantly different from the CTRL group until 44% of the descent cycle (Fig. 6.3A). However, the SB implant was able to return to their neutral pelvis position at the bottom of the squat (100% descent phase, 0% ascent phase), whereas the same is not true for the DM group. The pelvis of the DM group is significantly more anteriorly tilted than the CTRL group for the first 5% of the ascent phase (Fig. 6.3B). The DM group is unable to return to the neutral pelvis position at the deepest part of the squat, which may be due to the design of the implant. This may reflect the bearing motions between the 2 different couples: the larger PE bearing surface and the monoblock acetabular shell and motion of the smaller head within the PE bearing. In other words, when going into squat, the coefficient of friction being higher in the DM results in preventing the pelvis to decrease its tilt passively until enough muscles can be engaged in the later phases of the ascent for the pelvic to resume its normal motion. Because the pelvis is unable to adjust accordingly, this could explain some of the psoas tendon irritation/groin pain noted with DM implants [38] as the iliopsoas tendon unit is under tension during a greater portion of the squat cycle and more likely to catch on larger PE bearing [39]. More importantly, patients with DM implant had overall lower functional scores (Table 6.1) which could be due to psoas tendon irritation; however, since there were improvements in pain but not squat depth after the surgery, it is believed that pain is not the limiting factor.

Although the SB group was able to squat to a greater depth than the DM group, the difference was not significant, and both groups remained significantly different from the CTRL group (Figure 6.6). This suggests that it is not the implant design that affects the mobility postsurgery, as both groups were unable to reach normal squat depth. This could be due to higher coefficient of friction at the interface causing greater cocontraction of the hip flexors and hip extensors predisposing to hip pain. Also, this could indicate that there are flexibility issues in both groups from increased passive resistance of the hip extensors combined with weak flexors [35]. Years of decreased mobility while awaiting THA would have tightened all the muscles surrounding the hip joint. Evidence of this was found in gait studies of THA patients, which found that stride length was shorter than in CTRLs which resulted in reduced hip ROM [20, 30, 40]. Explanations have been given for this reduction in hip mobility, which include pain [41], muscle weakness [32, 42], unrecovered soft-tissue damage [32], or a physical barrier to further movement [43].

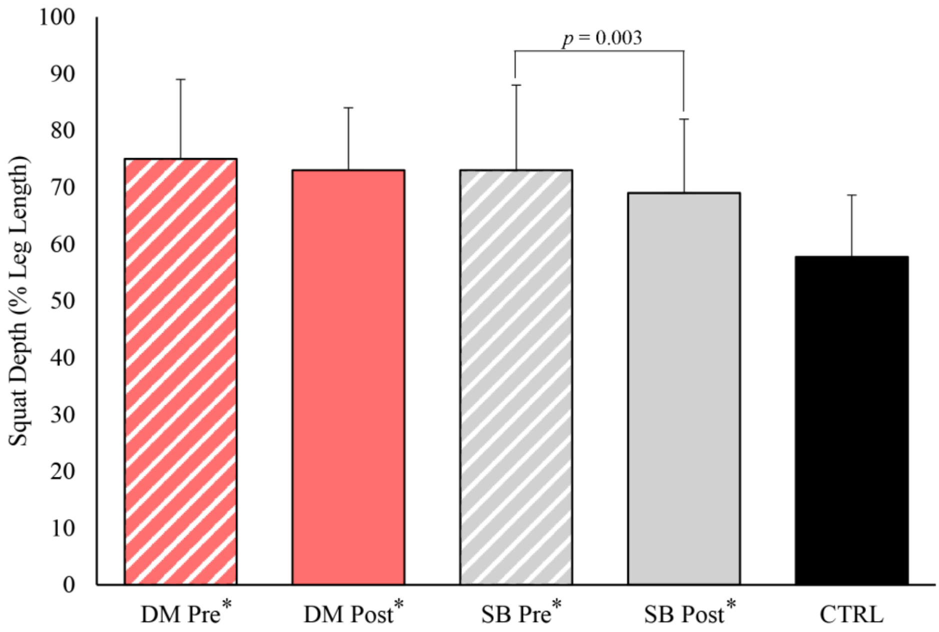


Figure 6.6. Mean maximal squat depth represented as percentage of leg length. A lower value represents a deeper squat depth. *Represents a statistical significance to the CTRL group ($P < .05$).

Our study did not include strength measurement; however, studies have shown significant improvements after THA at 24 weeks [44] and at 1 year [45], although the values were still 20% lower than the nonoperated limb. At the deepest part of the squat, the muscles are being loaded eccentrically. Therefore, if patients are unable to achieve a deep squat, it is most likely due to unrecovered soft-tissue damage or due to a physical barrier such as tightness of the soft tissue. It is unlikely that there were soft-tissue damages 6 months after surgery; therefore, postsurgery rehabilitation should focus on improving the flexibility and strength of the muscles surrounding the hip. Improvements in the strength and flexibility of the hip muscles may translate to a deeper squat for both implant types.

The focus of the study was to compare the joint kinematics during a deep squat task; and although no study has reported gender differences during a 2-legged squat, not controlling for gender may have introduced variances in squat kinematics, which may have prevented significant findings from emerging. However, as the same surgical approach was used in all patients - not compromising the abductors and short external rotators - our study targets the influence of prosthetic head design.

The time duration of the squat was not controlled, which may introduce variance in the findings. To account for this, the task was time-normalized separately into descent and ascent phases. However, some variations of acceleration within the task may modify the data interpretation of a subphase (eg, pelvis flexion and pelvis extension). Another limitation of the study was that we did not control the patients' squatting strategy, only the placement of their feet. This meant that excessive trunk flexion and frontal plane variations of the knee were permitted. Although this made the task familiar for all participants as it allowed them to use the squat strategy they are most used to, it may have created variations in joint kinematics within the groups. Again, these limitations are somewhat mitigated because of randomization.

6.5. Conclusion

Our findings are in line with the literature, which indicated that joint kinematics of the pelvis and hip do not return to the level of controls after THA. Although hip mobility remained the furthest from the CTRLs, differences still existed at the pelvis. The DM implant was unable to return to its neutral pelvis position at the deepest part of the squat which combined with the poorer functional scores should caution clinicians to use this implant design in active patients. The SB design caused the patients to over abduct their hip to reach their maximum squat depth when compared with the other groups. Post-THA rehabilitation should focus on improving hip joint mobility and strength.

6.6. References

1 Langlais FL, Ropars M, Gaucher F, Musset T, Chaix O. Dual mobility cemented cups have low dislocation rates in THA revisions. Clin Orthop Relat Res 2008;466:389–95. doi:10.1007/s11999-007-0047-9.

2 Philippot R, Adam P, Reckhaus M, Delangle F, Verdot F, Curvale G, et al. Prevention of dislocation in total hip revision surgery using a dual mobility design. Orthop Traumatol Surg Res 2009;95:407–13. doi:10.1016/j.otsr.2009.04.016.

3 Tarasevicius S, Busevicius M, Robertsson O, Wingstrand H. Dual mobility cup reduces dislocation rate after arthroplasty for femoral neck fracture. BMC Musculoskelet Disord 2010;11:175. doi:10.1186/1471-2474-11-175.

4 Mudrick CA, Melvin JS, Springer BD. Late posterior hip instability after lumbar spinopelvic fusion. Arthroplast Today 2015;1:25–9. doi:10.1016/j.artd.2015.05.002.

5 Guyen O, Chen QS, Bejui-Hugues J, Berry DJ, An K-N. Unconstrained tripolar hip implants. Clin Orthop Relat Res 2007;455:202–8. doi:10.1097/01.blo.0000238796.59596.1f.

6 Swiontkowski M. Concern about early intraprosthetic dislocation in dual-mobility implants. JBJS Case Connect 2013;3:e91. doi:10.2106/JBJS.CC.M.00212.

7 De Martino I, D'Apolito R, Waddell BS, McLawhorn AS, Sculco PK, Sculco TP. Early intraprosthetic dislocation in dual-mobility implants: a systematic review. Arthroplast Today 2017:6–11. doi:10.1016/j.artd.2016.12.002.

8 Waddell BS, De Martino I, Sculco T, Sculco P. Total hip arthroplasty dislocations are more complex than they appear: A case report of intraprosthetic dislocation of an anatomic dual-mobility implant after closed reduction. Ochsner J 2016;16:185–90.

9 Hamadouche M, Arnould H, Bouxin B. Is a cementless dual mobility socket in primary THA A reasonable option? Clin Orthop Relat Res 2012;470:3048–53. doi:10.1007/s11999-012-2395-3.

10 Adam P, Farizon F, Fessy MH. Dual mobility retentive acetabular liners and wear: Surface analysis of 40 retrieved polyethylene implants. Orthop Traumatol Surg Res 2014;100:85–91. doi:10.1016/j.otsr.2013.12.011.

11 Martz P, Bourredjem A, Laroche D, Arcens M, Labattut L, Binquet C, et al. Röttinger approach with dual-mobility cup to improve functional recovery in hip osteoarthritis patients: biomechanical and clinical follow-up. Int Orthop 2016:1–7. doi:10.1007/s00264-016-3245-9.

12 Kolk S, Minten MJM, Van Bon GEA, Rijnen WH, Geurts ACH, Verdonschot N, et al. Gait and gait-related activities of daily living after total hip arthroplasty: A systematic review. Clin Biomech 2014;29:705–18. doi:10.1016/j.clinbiomech.2014.05.008.

13 Matta JM, Shahrdar C, Ferguson T. Single-incision anterior approach for total hip arthroplasty on an orthopaedic table. Clin Orthop Relat Res 2005;441:115–24. doi:10.1097/01.blo.0000194309.70518.cb.

14 Mantovani G, Lamontagne M. How different marker sets affect joint angles in inverse kinematics framework. J Biomech Eng 2016. doi:10.1115/1.4034708.

15 Pataky TC. Generalized n-dimensional biomechanical field analysis using statistical parametric mapping. J Biomech 2010;43:1976–82. doi:10.1016/j.jbiomech.2010.03.008.

16 Pataky TC, Robinson MA, Vanrenterghem J. Vector field statistical analysis of kinematic and force trajectories. J Biomech 2013;46:2394–401. doi:10.1016/j.jbiomech.2013.07.031.

17 Charles MN, Bourne RB, Davey JR, Greenwald AS, Morrey BF, Rorabeck CH. Soft-tissue balancing of the hip: The role of femoral offset restoration. J Bone Jt Surgery-American Vol 2004;86:1078–88. doi:10.2106/00004623-200405000-00030.

18 Lavigne M, Therrien M, Nantel J, Roy A, Prince F, Vendittoli PA. The John Charnley Award: The functional outcome of hip resurfacing and large-head THA is the same: A randomized, double-blind study. Clin Orthop Relat Res 2010;468:326–36. doi:10.1007/s11999-009-0938-z.

19 Lamontagne M, Beaulieu ML, Varin D, Beaulé PE. Gait and motion analysis of the lower extremity after total hip arthroplasty: What the orthopedic surgeon should know. Orthop Clin North Am 2009;40:397–405. doi:10.1016/j.ocl.2009.02.001.

20 Beaulieu ML, Lamontagne M, Beaulé PE. Lower limb biomechanics during gait do not return to normal following total hip arthroplasty. Gait Posture 2010;32:269–73. doi:10.1016/j.gaitpost.2010.05.007.

21 Lamontagne M, Varin D, Beaulé PE. Does the anterior approach for total hip arthroplasty better restore stair climbing gait mechanics? J Orthop Res 2011;29:1412–7. doi:10.1002/jor.21392.

22 Lamontagne M, Beaulieu ML, Varin D, Beaulé PE. Lower-limb joint mechanics after total hip arthroplasty during sitting and standing tasks. J Orthop Res 2012;30:1611–7. doi:10.1002/jor.22127.

23 Foucher KC, Hurwitz DE, Wimmer MA. Do gait adaptations during stair climbing result in changes in implant forces in subjects with total hip replacements compared to normal subjects? Clin Biomech 2008;23:754–61. doi:10.1016/j.clinbiomech.2008.02.006.

24 Talis VL, Grishin AA, Solopova IA, Oskanyan TL, Belenky VE, Ivanenko YP. Asymmetric leg loading during sit-to-stand, walking and quiet standing in patients after unilateral total hip replacement surgery. Clin Biomech 2008;23:424–33. doi:10.1016/j.clinbiomech.2007.11.010.

25 Elkins JM, Callaghan JJ, Brown TD. The 2014 Frank Stinchfield Award: The “landing zone” for wear and stability in total hip arthroplasty is smaller than we thought: A computational analysis. Clin Orthop Relat Res 2014;473:441–52. doi:10.1007/s11999-014-3818-0.

26 Callanan MC, Jarrett B, Bragdon CR, Zurakowski D, Rubash HE, Freiberg AA, et al. The John Charnley Award: Risk factors for cup malpositioning: Quality improvement through a joint registry at a tertiary hospital. Clin Orthop Relat Res 2011;469:319–29. doi:10.1007/s11999-010-1487-1.

27 Pierrepont J, Hawdon G, Miles BP, Connor BO, Baré J, Walter LR, et al. Variation in functional pelvic tilt in patients undergoing total hip arthroplasty. Bone Joint J 2017;99–B:184–91. doi:10.1302/0301-620X.99B2.BJJ-2016-0098.R1.

28 De Martino I, Triantafyllopoulos GK, Sculco PK, Sculco TP. Dual mobility in total hip arthroplasty. World J Orthop 2014;45:1–8. doi:10.1016/j.ocl.2013.08.004.

29 Bennett D, Humphreys L, O’Brien S, Kelly C, Orr J, Beverland DE. The influence of wear paths produced by hip replacement patients during normal walking on wear rates. J Orthop Res 2008;26:1210–7. doi:10.1002/jor.20583.

30 Foucher KC, Hurwitz DE, Wimmer MA. Preoperative gait adaptations persist one year after surgery in clinically well-functioning total hip replacement patients. J Biomech 2007;40:3432–7. doi:10.1016/j.jbiomech.2007.05.020.

31 Kiss RM, Illyés Á. Comparison of gait parameters in patients following total hip arthroplasty with a direct-lateral or antero-lateral surgical approach. Hum Mov Sci 2012;31:1302–16. doi:10.1016/j.humov.2012.02.004.

32 Madsen MS, Ritter MA, Morris HH, Meding JB, Berend ME, Faris PM, et al. The effect of total hip arthroplasty surgical approach on gait. J Orthop Res 2004;22:44–50. doi:10.1016/S0736-0266(03)00151-7.

33 Perron M, Malouin F, Moffet H, McFadyen BJ. Three-dimensional gait analysis in women with a total hip arthroplasty. Clin Biomech 2000;15:504–15. doi:10.1016/S0268-0033(00)00002-4.

34 Varin D, Lamontagne M, Beaulé PE. Does the anterior approach for tha provide closer-to-normal lower-limb motion? J Arthroplasty 2013;28:1401–7. doi:10.1016/j.arth.2012.11.018.

35 Davis KE, Ritter M a, Berend ME, Meding JB. The importance of range of motion after total hip arthroplasty. Clin Orthop Relat Res 2007;465:180–4. doi:10.1097/BLO.0b013e31815c5a64.

36 Frost KL, Bertocci GE, Wassinger C a, Munin MC, Burdett RG, Fitzgerald SG. Isometric performance following total hip arthroplasty and rehabilitation. J Rehabil Res Dev 2006;43:435–44. doi:10.1682/JRRD.2005.06.0100.

37 Bouffard V, Nantel J, Therrien M, Vendittoli P-A, Lavigne M, Prince F. Center of mass compensation during gait in hip arthroplasty patients: Comparison between large diameter head total hip arthroplasty and hip resurfacing. Rehabil Res Pract 2011;2011:586412. doi:10.1155/2011/586412.

38 Lachiewicz PF, Watters TS. The use of dual-mobility components in total hip arthroplasty. J Am Acad Orthop Surg 2012;20:481–6. doi:10.5435/JAAOS-20-08-481.

39 Varadarajan KM, Zumbrunn T, Duffy MP, Patel R, Freiberg AA, Rubash HE, et al. Reducing the distal profile of dual mobility liners can mitigate soft-tissue impingement and liner entrapment without affecting mechanical performance. J Orthop Res 2016;34:889–97. doi:10.1002/jor.23078.

40 Hodge WAA, Andriacchi TPP, Galante JOO. A relationship between stem orientation and function following total hip arthroplasty. J Arthroplasty 1991;6:229–35. doi:10.1016/S0883-5403(06)80169-5.

41 Bennett D, Ogonda L, Elliott D, Humphreys L, Lawlor M, Beverland D. Comparison of immediate postoperative walking ability in patients receiving minimally invasive and standard-incision hip arthroplasty: A prospective blinded study. J Arthroplasty 2007;22:490–5. doi:10.1016/j.arth.2006.02.173.

42 Grimaldi A, Richardson C, Durbridge G, Donnelly W, Darnell R, Hides J. The association between degenerative hip joint pathology and size of the gluteus maximus and tensor fascia lata muscles. Man Ther 2009;14:611–7. doi:10.1016/j.math.2008.11.002.

43 Vendittoli PA, Ganapathi M, Nuño N, Plamondon D, Lavigne M. Factors affecting hip range of motion in surface replacement arthroplasty. Clin Biomech 2007;22:1004–12. doi:10.1016/j.clinbiomech.2007.07.007.

44 Andersson L, Wesslau A, Bodén H, Dalén N. Immediate or late weight bearing after uncemented total hip arthroplasty: A study of functional recovery. J Arthroplasty 2001;16:1063–5. doi:10.1054/arth.2001.27253.

45 Shih C-H, Du Y-K, Lin Y-H, Wu C-C. Muscular recovery around the hip-joint after total hip-arthroplasty. Clin Orthop Relat Res 1994;302:115–20. doi:10.1097/00003086-199405000-00019.

IV Musculoskeletal Modelling: from Gait to Squat

7 Surgical Correction for Cam Femoroacetabular Impingement Decreases Hip Muscle Forces during Gait at Two-Year Follow-Up

Danilo S. Catelli MSc [15] | K. C. Geoffrey Ng PhD [2] | Erik Kowalski MSc [1] | Paul E Beaulé MD FRCSC [3] | Mario Lamontagne PhD [14]

[1]School of Human Kinetics, University of Ottawa, Canada
[2]Department of Mechanical Engineering, Imperial College London, UK
[3]Division of Orthopaedic Surgery, University of Ottawa, Canada
[4]Department of Mechanical Engineering, University of Ottawa, Canada
[5]CAPES Foundation, Ministry of Education of Brazil, Brasilia DF, Brazil

Contents of this chapter submitted as an article in *Gait & Posture*® | September 2018

Catelli DS, Ng KCG, Kowalski E, Beaulé PE, Lamontagne M, Surgical Correction for Cam Femoroacetabular Impingement Decreases Hip Muscle Forces during Gait at Two-Year Follow-Up. 2018; Submitted.

7.0. Abstract

Background: In order to reduce the development of hip osteoarthritis, cam-type femoroacetabular impingement (FAI) corrective surgery has become the treatment of choice. As pain rarely influences dynamic motions during low hip impinging activities, it is unclear how muscle and hip contact forces, during level walking, are affected by corrective surgery.

Research question: The purpose was to compare the muscle force contributions and hip contact forces during level walking in individuals before and after surgical correction for cam FAI, at 2-year follow-up.

Methods: Eleven male patients with symptomatic cam FAI, who underwent hip osteoplasty, had their level walking recorded pre- and postoperatively. Sagittal and frontal hip kinematics and kinetics were computed and, subsequently, muscle and hip contact forces were estimated using musculoskeletal modelling and static optimization.

Results: Patient-reported outcomes improved postoperatively. Postoperative biceps femoris (pre-op: 0.28±0.11 N/BW; post-op: 0.20±0.07 N/BW) and semimembranosus forces (pre-op: 0.66±0.24 N/BW; post-op: 0.41±0.14 N/BW) were lower at ipsilateral foot-strike as well as postoperative rectus femoris force at contralateral foot-strike (pre-op: 1.44 ±0.24 N/BW; post-op: 1.18 ±0.23 N/BW). Frontal plane hip loading analysis showed a more medialized hip contact forces in postoperative patients during the entire stance phase.

Significance: Medialized hip contact forces indicates that patients enhance medial-lateral stability postoperatively. The reduced dynamic muscle forces of the biceps femoris, semimembranosus and rectus femoris can be an indication that the muscles associated with the sagittal aspect of the gait suggest a less compensatory strategy while trying to assist the hip stability.

7.1. Introduction

Cam-type femoroacetabular impingement (FAI) is caused by an aspherical femoral head that abuts against the acetabular rim during hip flexion, which can lead to chondral abrasion and labral detachment, causing pain in the groin and the development of early adult hip osteoarthritis [1,2]. Patients are often treated through conservative nonsurgical methods, but osteochondroplasty of the femoral head-neck junction is often required [3–5]. The surgery is done through either open [6,7] or arthroscopic [8,9] procedures, and patient-reported outcome measures (PROMs) have indicated that both surgical methods are effective at reducing pain and improving quality of life [10].

Prior to surgery, symptomatic cam FAI patients demonstrate altered biomechanics compared to healthy controls during gait [11–14], squatting [15–17], and stairs climbing [18,19], indicating reduced hip and pelvic range of motion (ROM), and hip flexion and external rotation moments compared to healthy controls. Neuromuscular adaptations may influence symptoms, and biomechanical outcomes as patients with FAI demonstrate weaker hip isometric muscle strength during flexion, extension, abduction, adduction, and external rotation movements [13,20–22]. These adaptations were demonstrated in a musculoskeletal modelling study which indicated reduced psoas major and iliacus muscle forces during gait in symptomatic patients compared to controls [23].

There is no clear consensus on the effect of surgery on biomechanics outcomes. Reported outcomes ranged from no improvements at all [24], improved sagittal and internal hip ROM [14,18], and even reduced hip ROM [25]. PROMs have indicated that surgery alleviates pain and improves quality of life [10], but quantitative evidence to support these findings is still lacking. In other words, how FAI correction surgery alters the muscle forces generated at the hip or how it affects hip loading, regarding hip contact forces during a musculoskeletal modelling analysis is relatively not well known. Therefore, the purpose of this study was to compare muscle force contributions and hip contact forces during level walking in patients before and after receiving surgical correction for cam FAI.

7.2. Methods

7.2.1. Participants

Eleven male patients presented themselves to the senior orthopaedic surgeon's clinical practice with persisting unilateral clinical signs of hip pain and positive impingement tests (Table 1). They underwent pelvic and knee computed tomography (CT) scan (Acquilion, Toshiba Medical Systems Corporation, Otawara, Japan; or Discover CT750, GE Healthcare, Mississauga, ON, Canada), to confirm cam-type FAI morphology, as defined by an axial (3:00) or radial (1:30) alpha angle larger than 50.5° and 60°, respectively [26,27]. Participants were excluded if they indicated any other hip morphology, a history of severe lower limb traumas or surgeries, or had a body mass index (BMI) greater than 30 kg/m2. All patients underwent corrective surgery by the same senior surgeon (e.g. osteochondroplasty and labral-chondral debridement) either through an open surgical dislocation (n = 4) or arthroscopic (n = 7) approach. Motion analysis protocol and completion of the Hip Disability and Osteoarthritis Outcome Score (HOOS) questionnaire were performed preoperatively (up to two months before) and at 2-years postoperatively (24.2 ± 1.5 months). The study was approved by the hospital and institution research ethics boards, and all participants provided informed consent.

7.2.2. Motion Analysis

In order to improve marker placement during motion analysis, radiopaque surface markers were placed on the participants before the CT imaging. These markers were placed on the anterior superior iliac spines, posterior superior iliac spines, medial and lateral epicondyles. At the motion capture laboratory, the radiopaque markers were then replaced with retro-reflective markers and outfitted according to the University of Ottawa Motion Analysis Model (UOMAM) marker set [28]. Participants performed five barefoot level walking trials, at a self-selected pace, where marker trajectories were captured using a ten-infrared camera system sampled at 200Hz (Vicon MX-13, VICON, Oxford, UK) and ground reaction forces were captured using two embedded force plates at

1000 Hz (FP4060-08, Bertec Corporation, Columbus, USA). The data were labelled and filtered (zero-lag, fourth order Butterworth filter, cut-off frequency = 6 Hz), walking speed, and stride length were calculated using motion analysis software (Nexus 2.6.1, VICON, Oxford, UK). The gait analyses were performed during the stance phase (i.e. ipsilateral foot-strike to foot-off) and all gait variables were time-normalized to its cycle.

7.2.3. Musculoskeletal Modelling

A generic musculoskeletal model [29] consisting of 37 degrees of freedom, 80 lower-limb Hill-type muscle-tendon units, and 17 torque actuators driving the upper body was used in an open-source musculoskeletal simulation software (OpenSimTM 3.3, Stanford University, Stanford, USA) [30].

The markers trajectories and ground reaction forces were imported [31], and the generic model was scaled for each participant based on their static pose, with the pelvis and knee markers having a ten-times higher weight as their location was previously verified in the CT imaging. Joint kinematics and net joint moment for each degree of freedom were computed using the inverse kinematics and inverse dynamics tools. The muscle forces were calculated while minimizing the sum of squared muscle activation using the static optimization tool [34,36]. Hip contact forces were reported as three-dimensional vectors acting on the acetabulum and expressed in the pelvic coordinate system. Both hip contact and muscle forces were normalized by body weight.

Intrasubject differences (i.e. pre- vs post-op) for demographics and gait parameters were examined using paired samples t-tests. Intrasubject peak sagittal and frontal hip kinematics and kinetics were compared between conditions using a paired t-test, and hip contact and muscle forces were compared using either a paired samples t-test or a Wilcoxon signed-rank test for non-parametric distributions, given that some of the variables failed the Shapiro-Wilk normality test (CI = 95%). Statistical analyses were performed using SPSS Statistics (v.23, IBM Corporation, Armonk, USA).

7.3. Results

7.3.1. Demographics and Patient Reported Outcome Measure

Postoperatively, patients did not differ in BMI from their preoperative values and showed improvements in all HOOS categories when assessed two years following the surgery (Table 7.1).

Table 7.1. Patient demographics, pain questionnaire, and cam deformity measurement, reporting mean ± SD.

Parameter		FAI pre-op	FAI post-op	p-value
Participants (n)		11		
Age (years)		34.1 ±7.4	36.2 ±7.2	
Height (cm)		177.3 ±6.2	178.1 ±7.2	
BMI (kg/m^2)		25.4 ±2.7	25.6 ±3.6	.79
alpha-angle (deg)	3:00 clock face position	54.0 ± 7.2	45.6 ± 6.7	.028
	1:30 clock face position	66.3 ± 5.4	52.5 ± 9.1	.005
HOOS	Symptoms	70.0 ±10.7	81.4 ±10.0	.04
	Pain	70.0 ±16.9	90.0 ±8.3	.001
	Activities of Daily Living	81.7 ±15.1	95.4 ±6.6	.007
	Sports and Recreational Activities	56.8 ±25.1	83.0 ±13.7	.005
	Quality of Life	39.2 ±21.8	65.9 ±21.5	.01

Data are reported as mean ± SD

7.3.2. Gait Parameters

No differences in walking speed or stride length were observed between pre- and postoperative FAI patients (Table 7.2). The postoperative FAI patients demonstrated increased hip abduction at ipsilateral foot-off ($p = 0.026$); as well as lower hip extension moment at ipsilateral foot-strike ($p = 0.007$) and lower hip flexion moment during contralateral foot-strike, at terminal stance phase ($p = 0.003$), compared to their preoperative values (Figure 7.1).

Table 7.2. Mean and standard deviation of walking speed, and stride length parameters of each patient condition

Parameter	FAI pre	FAI post	p-value
Walking Speed (m/s)	1.36 ± 0.14	1.35 ± 0.06	.91
Stride Length (m)	1.46 ± 0.17	1.46 ± 0.11	.97

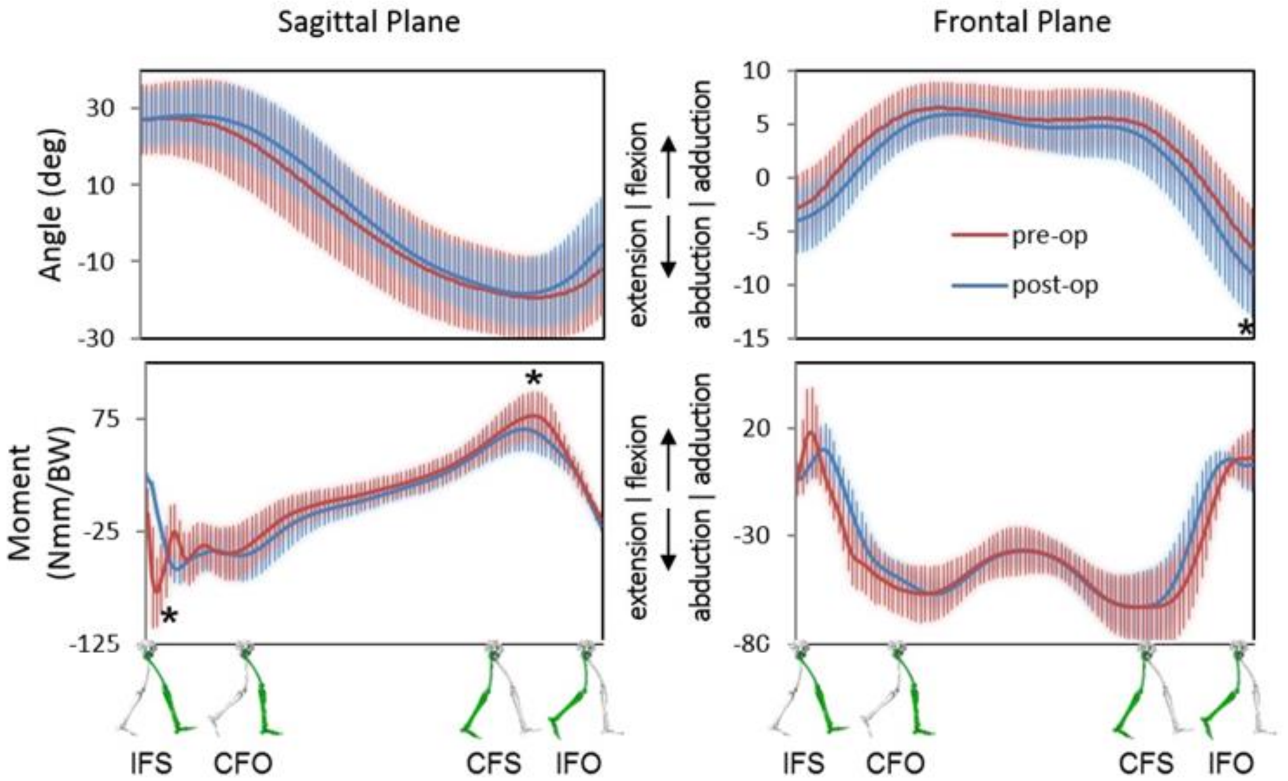


Figure 7.1. Hip joint angles (top row) and moments (bottom row) in the sagittal (left), frontal (right) planes, during the gait cycle, for the FAI preoperative (red) and postoperative conditions (blue). Hip joint moments were normalized by body weight (BW) and the full stance phase was represented at ipsilateral foot-strike (IFS), contralateral foot-off (CFO), contralateral foot-strike (CFS), ipsilateral foot-off (IFO), while the asterisk (*) denotes the statistical difference ($p<0.05$) in hip adduction, and hip flexion and extension moments.

7.3.3. Muscle and Hip Contact Forces

Postoperative FAI demonstrated significantly reduced biceps femoris (long head) ($p = 0.019$, $d = 0.84$) and semimembranosus ($p = 0.008$, $d = 0.99$) forces during ipsilateral foot-strike, and reduced rectus femoris force ($p = 0.039$, $d = 0.72$) at contralateral foot-strike compared to the preoperative. No muscle forces differences arose for the gluteus maximus, iliacus or psoas muscles (Figure 7.2).

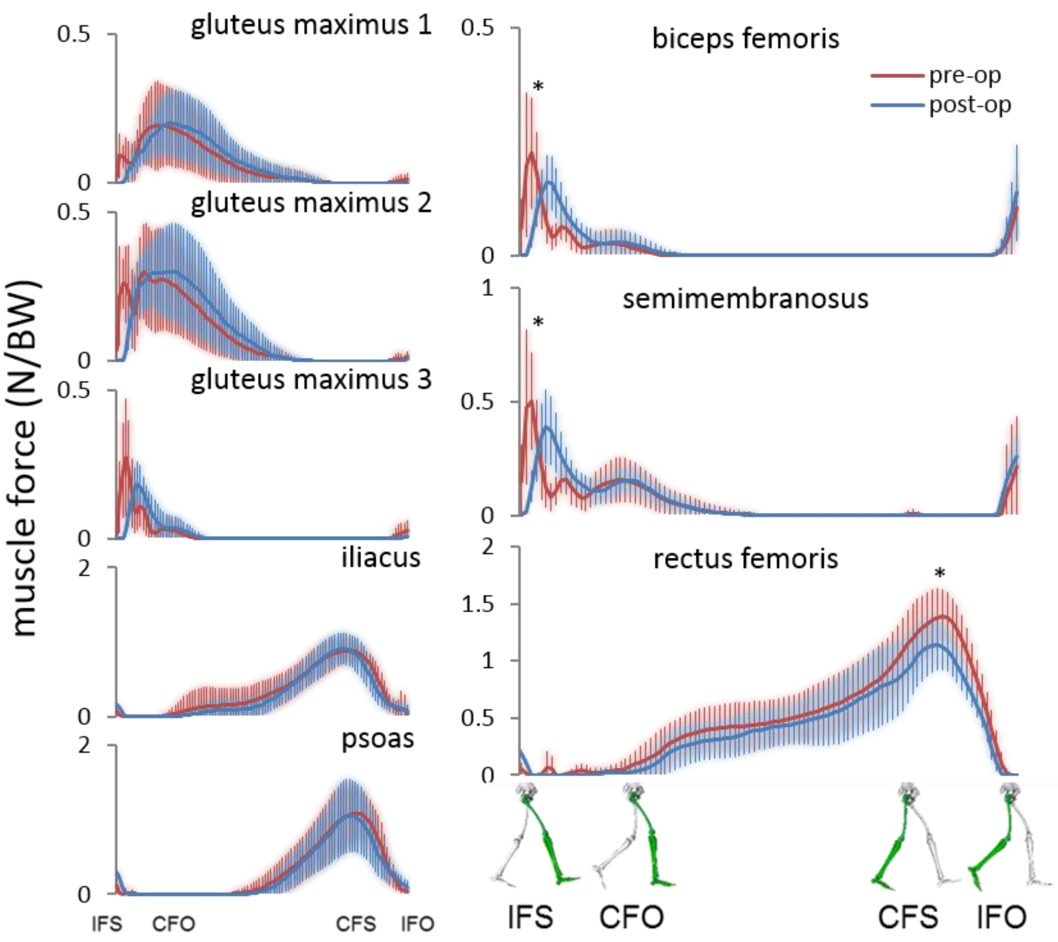


Figure 7.2. Muscle forces during the gait cycle, for the FAI preoperative (red) and postoperative conditions (blue). Muscle forces were normalized by body weight (BW) and determined from static optimization. The full stance phase was represented at ipsilateral foot-strike (IFS), contralateral foot-off (CFO), contralateral foot-strike (CFS), ipsilateral foot-off (IFO). The postoperative FAI demonstrated significantly lower muscle forces for the biceps femoris (long head – top right), semimembranosus (middle right) and rectus femoris (bottom right), denoted by the asterisk (*).

There were no differences between pre- and postoperative peak hip contact anterior ($p = 0.43$), superior ($p = 0.32$), or medial forces ($p = 0.10$; Figure 7.3). During contralateral foot-strike, resultant preoperative peak forces vectors (4.82±1.04 N/BW) were similar to the postoperative results (4.63±0.70 N/BW; $p = 0.20$).

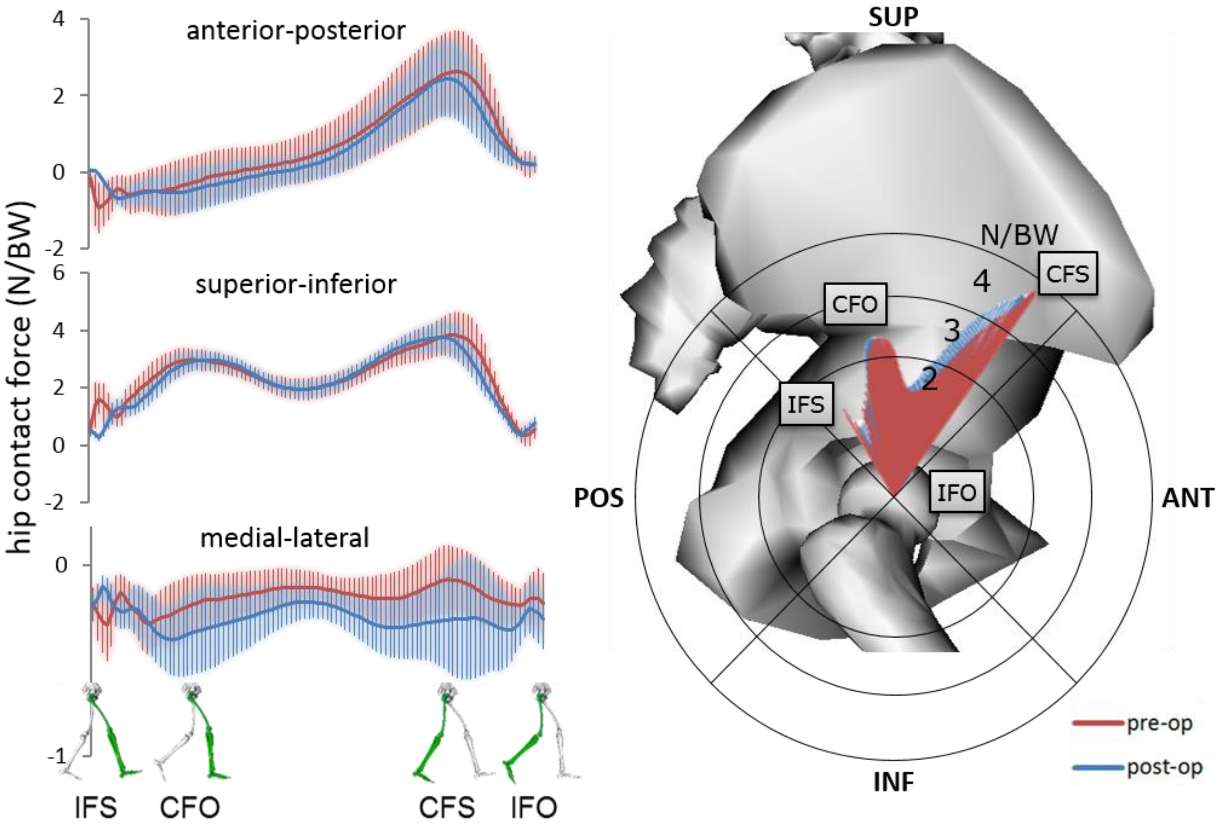


Figure 7.3. Hip contact forces during the gait cycle, for the FAI preoperative (red) and postoperative conditions (blue), in all three planes. The full stance phase was represented at ipsilateral foot-strike (IFS), contralateral foot-off (CFO), contralateral foot-strike (CFS), ipsilateral foot-off (IFO). The resultant average hip contact forces during the gait cycle (right), depicted in the sagittal plane of a right hip. Hip contact forces were normalized by body weight (BW) and reported with a 'butterfly' graph, showing magnitude (radar graph) and direction (acetabular orientation) with respect to the pelvic coordinate system.

7.4. Discussion

At two-year follow-up, patients showed significant improvement scores in the patient-reported outcomes, reduced forces in the long head of the biceps femoris and semimembranosus at ipsilateral foot-strike, and in the rectus femoris at contralateral foot-strike. There were no differences in hip contact force magnitudes before and after surgical correction.

The cam morphology is unlikely to impinge during a low amplitude motion [32], such as level walking; however, musculoskeletal modelling outputs (i.e. muscle and hip contact forces) can be very beneficial to highlight postoperative alterations observed in kinematics and kinetics. Moreover, the understanding of muscle contributions towards joint loading may provide benefits to design a better strategy for the aspects involving the FAI osteoplasty and aftercare programs [33,34]. Although several studies reported postoperative joint kinematics and kinetics associated with cam FAI [14,18,24,25,35], none of them determined muscle or hip contact forces from musculoskeletal modelling.

At two-year follow-up, the postoperative patients did not change their walking speed or stride length. The previous study showed that FAI patients naturally walk slower and with shorter steps, compared to healthy control individuals [23]. Moreover, altered walking biomechanics of FAI participants could be associated to an adaptive mechanism [23,36]. However, it appears that these variables remain unchanged after corrective surgery, perhaps evidencing that these adaptive mechanisms persist after years of dealing with pain or awaiting surgery. Brisson and associates (2013) [25] analyzed level walking in 10 postoperative FAI patients with a mixed gender group (7 men) and with unfixed follow-up (range 10-32 months) and did not find any significant change in hip and pelvic kinematics, or joint torques. The controlled gender inclusion and follow-up time criteria on this current study were able to show the patients reduced hip adduction during ipsilateral foot-off and hip extension moment during contralateral foot-strike. The inconsistency of reported kinematics and kinetic variables of postoperative studies [14,18,24,25] may represent that the PROMs pain reduction can be more likely associated to the repair of the soft-tissue than the cam osteoplasty itself [14]. Still, kinematic and kinetic changes could be better defined when considering muscle modelling analyzes.

To our knowledge, Ng and associates (2018) [23] was the first study to report musculoskeletal modelling outputs in an FAI population. The limited hip mobility was associated with the reduced muscle force pattern of the iliopsoas muscle complex, which caused a reduction of the anterior, superior, and medial hip contact forces [23]. The present study reported the forces of the hip flexors and extensors to assess the follow-up condition of the patients. However, as gait parameters (i.e. walking speed, stride length), joint kinematics (i.e. hip extension, hip and pelvic range of motion), and spinopelvic anatomy (i.e. pelvic tilt and incidence) did not change after surgery, there were marginal effects towards the activation of the primary hip flexors [23,37]. However, although not statistically significant, the medial-lateral (frontal plane) hip contact force component remained more medialized on the postoperative group during almost the entire stance phase (Figure 3). Similarly, Ng and associates (2018) [23] have shown a more medialized medial-lateral hip contact force component in the healthy participants when compared with preoperative cam FAI during contralateral foot-strike. This medialization process is indicative that postoperative patients enhance medial-lateral loading during the gait stance phase. Patients with hip OA are found to alter gait to increase medial-lateral stability thereby decreasing demand on the hip abductors [38]. However no decrease in dynamic hip abductors muscle forces was detected in our simulations. Considering that the postoperative patients had their hip more abducted during the stance phase, which might have affected the hip loading on the medial-lateral aspect.

The decreased muscle forces on the hamstrings (biceps femoris and semimembranosus) and the rectus femoris during ipsilateral and contralateral foot-strike, respectively, suggests that these muscles no longer have to compensate while trying to maintain hip stability. Additionally, the iliopsoas muscle forces did not change postoperatively to reflect more similar muscle forces to healthy control individuals [23], which suggests that the preventive pain mechanism reported preoperatively [23] may have generated neuromuscular adaptations in the long-term that affected muscle contraction strategies even at 2-year follow-up surgery, when pain no longer plays a role in the motion (Table 7.1). The decreased extension moment at the contralateral foot strike is directly associated with the also decreased muscle activation of the hip flexors, and once the dynamic force of the iliopsoas complex was already reduced preoperatively, the co-contraction of the hip flexors during this extension moment can justify, likewise, the force reduction of the rectus femoris. Yet, one may presume that the reduced forces indicate a more effective gait pattern, as they did not affect the kinematics, however, some caution must be kept regarding the interpretation of this lower force, especially because hip flexor strength [20,39] does not improve after surgery [40], and the gait kinematics were similar to the preoperative data.

For the muscles responsible for the hip extension, both the biceps femoris and the semimembranosus generate an extension moment during ipsilateral foot-strike. The reduced peak hip flexion moment observed in the postoperative patients (Figure 7.1) can be considered as the main reason for the optimization calculations to show a reduced force in the biceps femoris and the semimembranosus at this phase of the gait. The gluteus maximus muscle was segmented in three portions (i.e. superior, middle and inferior) and none of them showed postoperative changes.

The small changes in hip contact forces during level walking merit the need of studies focusing the pre-post assessment during a motion involving higher range of motion (e.g. deep squat) or with a higher hip stabilization request (e.g. step down), which may provide a better understanding of muscle forces in a more demanding task and its implications on the hip contact loading.

Some limitations affected this study. First, the sample size was underpowered, even though is expected that this sort of study would presumably have a small cohort of patients. As our patients were all male and had cam-only FAI morphology (no pincer or mixed), the inference from our findings is limited for only this population. Second, our cohort consisted of patients that underwent either a surgical dislocation or arthroscopy approach, although an in-house comparison did not point to any statistical differences between this two postoperative management. Third, the muscle forces were calculated using a static optimization method that may not perfectly express co-contraction mechanisms altered by a joint pathology. However, this method benefits while not requiring invasive access of deep muscle activity to perform EMG-driven simulations [41], and it also provides comparable muscle activations during various walking speeds [42]. Fourth, we modelled a specific hip pathology onto an idealized musculoskeletal model. The effect of the subject-specific hip bony morphology could not be directly assessed or parameterized in the model, which would greatly influence hip contact forces. Fifth, although all patients were instructed to undergo a six-week postsurgical rehabilitation program, the aftercare rehabilitation program was not controlled, which may have affected the benefit of this procedure for the patients the same way. Sixth, with ongoing strengthening and training, the postoperative patients may further adapt their walking mechanics. It would be feasible as a longitudinal study to conduct another follow-up, in efforts to examine if there will be further improvements to gait mechanics or characteristics to healthy, control groups.

7.5. Conclusion

To our knowledge, this is the first study to evaluate muscle and hip contact forces, to compare preoperative and postoperative FAI patients. Hip contact loading showed more medialized forces postoperatively, which is an indication that patients increase medial-lateral loading after the surgery. By having a more medial loading of the hip joint, this might compensate the loading of the formerly cam deformity. The reduced dynamic muscle force of the rectus femoris, biceps femoris and semimembranosus can be an indication that the muscles associated with the sagittal aspect of the gait do not have to compensate anymore while trying to stabilize the hip.

7.6. References

1 R. Ganz, J. Parvizi, M. Beck, M. Leunig, H. Nötzli, K.A. Siebenrock, Femoroacetabular Impingement: a Cause for Osteoarthritis of the Hip, Clin. Orthop. Relat. Res. (2003) 112–120. doi:10.1097/01.blo.0000096804.78689.c2.

2 R. Ganz, M. Leunig, K. Leunig-Ganz, W.H. Harris, The Etiology of Osteoarthritis of the Hip, Clin Orthop Relat Res. 466 (2008) 264–272. doi:10.1007/s11999-007-0060-z.

3 D. Hunt, H. Prather, M.H. Hayes, J.C. Clohisy, Clinical Outcomes Analysis of Conservative and Surgical Treatment of Patients With Clinical Indications of Prearthritic, Intra-articular Hip Disorders, Pm&R. 4 (2012) 479–487. doi:10.1016/j.pmrj.2012.03.012.

4 J.K. Loudon, M.P. Reiman, Conservative Management of Femoroacetabular Impingement (FAI) in the Long Distance Runner, Phys. Ther. Sport. 15 (2014) 82–90. doi:10.1016/j.ptsp.2014.02.004.

5 P.E. Beaulé, J.C. Clohisy, P. Schoenecker, Y.-J. Kim, M. Millis, R.T. Trousdale, Hip Arthroscopy: An Emerging Gold Standard, Arthrosc. J. Arthrosc. Relat. Surg. 23 (2007) 682. doi:10.1016/j.arthro.2007.03.007.

6 M. Lincoln, K. Johnston, M. Muldoon, R. Santore, Combined Arthroscopic and Modified Open Approach for Cam Femoroacetabular Impingement: A Preliminary Experience, Arthrosc. - J. Arthrosc. Relat. Surg. 25 (2009) 392–399. doi:10.1016/j.arthro.2008.12.002.

7 F. Laude, E. Sariali, A. Nogier, Femoroacetabular Impingement Treatment using Arthroscopy and Anterior Approach, Clin. Orthop. Relat. Res. 467 (2009) 747–752. doi:10.1007/s11999-008-0656-y.

8 M.J. Philippon, A.J. Stubbs, M.L. Schenker, R.B. Maxwell, R. Ganz, M. Leunig, Arthroscopic Management of Femoroacetabular Impingement: Osteoplasty Technique and Literature Review., Am. J. Sports Med. 35 (2007) 1571–1580. doi:10.1177/0363546507300258.

9 M. Horisberger, A. Brunner, R.F. Herzog, Arthroscopic Treatment of Femoroacetabular Impingement of the Hip: A New Technique to Access the Joint, Clin. Orthop. Relat. Res. 468 (2010) 182–190. doi:10.1007/s11999-009-1005-5.

10 I.B. Botser, T.W. Smith, R. Nasser, B.G. Domb, Open Surgical Dislocation Versus Arthroscopy for Femoroacetabular Impingement: A Comparison of Clinical Outcomes, Arthrosc. - J. Arthrosc. Relat. Surg. 27 (2011) 270–278. doi:10.1016/j.arthro.2010.11.008.

11 M.A. Hunt, J.R. Gunether, M.K. Gilbart, Kinematic and Kinetic Differences during Walking in Patients with and without Symptomatic Femoroacetabular Impingement, Clin. Biomech. 28 (2013). doi:10.1016/j.clinbiomech.2013.05.002.

12 L.E. Diamond, T. V. Wrigley, K.L. Bennell, R.S. Hinman, J. O’Donnell, P.W. Hodges, Hip Joint Biomechanics during Gait in People with and without Symptomatic Femoroacetabular Impingement, Gait Posture. 43 (2016) 198–203. doi:10.1016/j.gaitpost.2015.09.023.

13 M.J. Kennedy, M. Lamontagne, P.E. Beaulé, Femoroacetabular Impingement Alters Hip and Pelvic Biomechanics during Gait: Walking Biomechanics of FAI, Gait Posture. 30 (2009) 41–44. doi:10.1016/j.gaitpost.2009.02.008.

14 J.H. Rylander, B. Shu, T.P. Andriacchi, M.R. Safran, Preoperative and Postoperative Sagittal Plane Hip Kinematics in Patients with Femoroacetabular Impingement during Level Walking, Am. J. Sports Med. 39 (2011) 36S–42S. doi:10.1177/0363546511413993.

15 M. Lamontagne, M.J. Kennedy, P.E. Beaulé, The Effect of Cam FAI on Hip and Pelvic Motion during Maximum Squat, Clin Orthop Relat Res. 467 (2009) 645–650. doi:10.1007/s11999-008-0620-x.

16 J.J. Bagwell, J. Snibbe, M. Gerhardt, C.M. Powers, Hip Kinematics and Kinetics in Persons with and without Cam Femoroacetabular Impingement during a Deep Squat Task, Clin. Biomech. 31 (2016) 87–92. doi:10.1016/j.clinbiomech.2015.09.016.

17 L.E. Diamond, K.L. Bennell, T. V. Wrigley, R.S. Hinman, J. O’Donnell, P.W. Hodges, Squatting Biomechanics in Individuals with Symptomatic Femoroacetabular Impingement, Med. Sci. Sports Exerc. 49 (2017) 1520–1529. doi:10.1249/MSS.0000000000001282.

18 J. Rylander, B. Shu, J. Favre, M. Safran, T. Andriacchi, Functional Testing Provides Unique Insights into the Pathomechanics of Femoroacetabular Impingement and an Objective Basis for Evaluating Treatment Outcome, J. Orthop. Res. 31 (2013) 1461–1468. doi:10.1002/jor.22375.

19 L.E. Diamond, K.L. Bennell, T. V. Wrigley, R.S. Hinman, M. Hall, J. O'Donnell, P.W. Hodges, Trunk, Pelvis and Hip Biomechanics in Individuals with Femoroacetabular Impingement Syndrome: Strategies for Step Ascent, Gait Posture. (2018). doi:10.1016/j.gaitpost.2018.01.005.

20 N.C. Casartelli, N.A. Maffiuletti, J.F. Item-Glatthorn, S. Staehli, M. Bizzini, F.M. Impellizzeri, M. Leunig, Hip Muscle Weakness in Patients with Symptomatic Femoroacetabular Impingement, Osteoarthr. Cartil. 19 (2011) 816–821. doi:10.1016/j.joca.2011.04.001.

21 L.E. Diamond, W. Van den Hoorn, K.L. Bennell, T. V. Wrigley, R.S. Hinman, J. O'Donnell, P.W. Hodges, Coordination of Deep Hip Muscle Activity is Altered in Symptomatic Femoroacetabular Impingement, J. Orthop. Res. 35 (2017) 1494–1504. doi:10.1002/jor.23391.

22 S. Kierkegaard, I. Mechlenburg, B. Lund, K. Søballe, U. Dalgas, Impaired Hip Muscle Strength in Patients with Femoroacetabular Impingement Syndrome, J. Sci. Med. Sport. 20 (2017) 1062–1067. doi:10.1016/j.jsams.2017.05.008.

23 K.C.G. Ng, G. Mantovani, L. Modenese, P.E. Beaulé, M. Lamontagne, Altered Walking and Muscle Patterns Reduce Hip Contact Forces in Individuals With Symptomatic Cam Femoroacetabular Impingement, Am. J. Sports Med. 46 (2018) 1–9. doi:10.1177/0363546518787518.

24 F. Malagelada, V.A. Del Carmen, S.J. Barke, L. Guirao Cano, E. Pleguezuelos Cobo, The Anterior Mini-Open Approach for Femeroacetabular Impingement: Gait and Functional Assessment at One Year Post-Surgery, Ann. Phys. Rehabil. Med. 58 (2015) 60–65. doi:10.1016/j.rehab.2014.09.013.

25 N. Brisson, M. Lamontagne, M.J. Kennedy, P.E. Beaulé, The Effects of Cam Femoroacetabular Impingement Corrective Surgery on Lower-Extremity Gait Biomechanics, Gait Posture. 37 (2013) 258–263. doi:10.1016/j.gaitpost.2012.07.016.

26 H. Nötzli, T. Wyss, C. Stoecklin, M. Schmid, K. Treiber, J. Hodler, The Contour of the Femoral Head-Neck Junction as a Predictor for the Risk of Anterior Impingment: Commentary, J. Bone Jt. Surg. 84 (2002) 556–60. doi:10.1024/0369-8394.92.11.475.

27 K.S. Rakhra, A.M. Sheikh, D. Allen, P.E. Beaulé, Comparison of MRI Alpha Angle Measurement Planes in Femoroacetabular Impingement, Clin. Orthop. Relat. Res. 467 (2009) 660–665. doi:10.1007/s11999-008-0627-3.

28 G. Mantovani, M. Lamontagne, How Different Marker Sets Affect Joint Angles in Inverse Kinematics Framework, J. Biomech. Eng. 139 (2016) 044503. doi:10.1115/1.4034708.

29 A. Rajagopal, C.L. Dembia, M.S. Demers, D.D. Delp, J.L. Hicks, S.L. Delp, Full body Musculoskeletal Model for Muscle-driven Simulation of Human Gait, IEEE Trans. Biomed. Eng. 63 (2016) 2068–79. doi:10.1109/TBME.2016.2586891.

30 S.L. Delp, F.C. Anderson, A.S. Arnold, P. Loan, A. Habib, C.T. John, E. Guendelman, D.G. Thelen, OpenSim: Open-Source Software to Create and Analyze Dynamic Simulations of Movement, IEEE Trans. Biomed. Eng. 54 (2007) 1940–1950. doi:10.1109/TBME.2007.901024.

31 A. Mantoan, C. Pizzolato, M. Sartori, Z. Sawacha, C. Cobelli, M. Reggiani, MOtoNMS: A MATLAB toolbox to Process Motion Data for Neuromusculoskeletal Modeling and Simulation, Source Code Biol. Med. 10 (2015) 12. doi:10.1186/s13029-015-0044-4.

32 S. Chegini, M. Beck, S.J. Ferguson, The Effects of Impingement and Dysplasia on Stress Distributions in the Hip Joint during Sitting and Walking: A Finite Element Analysis, J. Orthop. Res. 27 (2009) 195–201. doi:10.1002/jor.20747.

33 K.L. Bennell, L. Spiers, A. Takla, J. O'Donnell, J. Kasza, D.J. Hunter, R.S. Hinman, Efficacy of Adding a Physiotherapy Rehabilitation Programme to Arthroscopic Management of Femoroacetabular Impingement Syndrome: A Randomised Controlled Trial (FAIR), BMJ Open. 7 (2017). doi:10.1136/bmjopen-2016-014658.

34 C.L. Lewis, S.A. Sahrmann, D.W. Moran, Effect of Position and Alteration in Synergist Muscle Force Contribution on Hip Forces When Performing Hip Strengthening Exercises, Clin. Biomech. 24 (2009) 35–42. doi:10.1016/j.clinbiomech.2008.09.006.

35 M. Lamontagne, N. Brisson, M.J. Kennedy, P.E. Beaulé, Preoperative and Postoperative Lower-Extremity Joint and Pelvic Kinematics during Maximal Squatting of Patients with Cam Femoroacetabular Impingement, J. Bone Jt. Surg. 93 (2011) 40–5. doi:10.2106/JBJS.J.01809.

36 C.L. Lewis, S.A. Sahrmann, D.W. Moran, Anterior Hip Joint Force Increases with Hip Extension, Decreased Gluteal Force, or Decreased Iliopsoas Force, 40 (2007) 3725–3731. doi:10.1016/j.jbiomech.2007.06.024.

37 K.C.G. Ng, M. Lamontagne, J.R.T. Jeffers, G. Grammatopoulos, P.E. Beaulé, Anatomic Predictors of Sagittal Hip and Pelvic Motions in Patients With a Cam Deformity, Am. J. Sports Med. (2018) 1–12. doi:10.1177/0363546518755150.

38 A.G.M. Christophe, M. Wesseling, K. Corten, A. Nieuwenhuys, D. Monari, J.P. Simon, I. Jonkers, K. Desloovere, Hip Movement Pathomechanics of Patients with Hip Osteoarthritis Aim at Reducing Hip Joint Loading on the Osteoarthritic Side, Gait Posture. 59 (2018) 11–17. doi:10.1016/j.gaitpost.2017.09.020.

39 D.S. Catelli, E. Kowalski, P.E. Beaulé, K. Smit, M. Lamontagne, Asymptomatic Individuals with Femoroacetabular Deformity Demonstrate Stronger Hip Extensors and Greater Pelvis Mobility during the Deep Squat Task, Orthop. J. Sport. Med. 6 (2018) 1–10. doi:10.1177/2325967118782484.

40 N.C. Casartelli, N.A. Maffiuletti, J.F. Item-Glatthorn, F.M. Impellizzeri, M. Leunig, Hip Muscle Strength Recovery After Hip Arthroscopy in a Series of Patients with Symptomatic Femoroacetabular Impingement, HIP Int. 24 (2014) 387–393. doi:10.5301/hipint.5000131.

41 T.S. Buchanan, D.G. Lloyd, K. Manal, T.F. Besier, Neuromusculoskeletal Modeling: Estimation of Muscle Forces and Joint Moments and Movements from Measurements of Neural Command, J. Appl. Biomech. 20 (2004) 367–395. doi:10.1016/j.bbi.2008.05.010.

42 L. Modenese, A.T.M. Phillips, Prediction of Hip Contact Forces and Muscle Activations during Walking at Different Speeds, Multibody Syst. Dyn. 28 (2012) 157–168. doi:10.1007/s11044-011-9274-7.

8 A Musculoskeletal Model Customized for Squatting Task

Danilo S. Catelli [12] | Mariska Wesseling [3] | Ilse Jonkers [3] | Mario Lamontagne[14]

[1]School of Human Kinetics, University of Ottawa, Ottawa, Canada
[2]CAPES Foundation, Ministry of Education of Brazil, Brasilia, Brazil
[3]Department of Human Movement Sciences, KU Leuven, Leuven, Belgium
[4]Department of Mechanical Engineering, University of Ottawa, Ottawa, Canada

Contents of this chapter published as a rapid innovative communication manuscript in *Computer Methods in Biomechanics and Biomedical Engineering®* | November 2018

Catelli DS, Wesseling M, Jonkers I, Lamontagne M, A Musculoskeletal Model Customized for Squatting Task. *Comput Methods Biomech Biomed Engin*, 2018; 6:1-4.

DOI: 10.1080/10255842.2018.1523396.

8.0. Abstract

Most musculoskeletal models (MSKM) are designed to evaluate gait and running, which have limited range of motion (ROM). The purpose of this study was to examine the effect of wrapping surfaces (WS) at the knee and hip joints in a MSKM, on the muscle moment arms (MA) and activations during squatting. The MSKM was then customized by changing parameters of the original WS and by implementing additional WS. The WS prevent muscles from crossing into the bones, providing realistic muscle MA for large ROM. The modified MSKM is suitable for analysis up to 138º hip and 145º knee flexions.

8.1. Introduction

Musculoskeletal models (MSKM) provide a non-invasive mechanism to investigate human movement and predict the effect of interventions on different tasks (Delp et al. 2007). Most of the lower limb MSKM (Delp et al. 1990; Arnold et al. 2010; Rajagopal et al. 2016) are designed for tasks with limited range of motion (ROM) such as walking and running, and consequently less extreme muscle lengths and moment arms (MA) compared to higher flexion tasks, such as a deep squat. The ability to simulate larger ROM tasks is of utmost importance for sports movements (e.g. sprinting block start, race-walk, long jump take-off) (Lai et al. 2017).

A recent study (Lai et al. 2017) adapted the knee muscle paths of the Rajagopal model (Rajagopal et al. 2016), based on cadaver and MRI data, and modified the muscle-tendon properties of eleven muscles, to allow knee flexion up to 140º for pedalling simulations. To better predict muscle and hip contact forces for tasks requiring large ROM at the hip and knee, the MSKM needs reliable and physiological muscle-tendon properties as well as muscle paths. To avoid muscles paths crossing the bones, the appropriate MA lengths are essential. To our knowledge, the muscle geometry in extreme hip flexion positions is not well established, as only Németh & Ohlsén (1985) have reported in-vivo hip muscles MA length during high hip flexion. Thus, the purpose of this study was to examine the effect of including wrapping surfaces (WS) at the knee and hip joints in a MSKM on the muscle MA and muscle activations during deep squatting.

8.2. Methods

The generic MSKM (Rajagopal et al. 2016) consists of 37 degrees of freedom, 80 lower-limb Hill-type muscle-tendon units (MTU), 40 lower-body WS and 17 torque actuators driving the upper body (OpenSimTM 3.3, Stanford University, Stanford, USA) (Delp et al. 2007). The lower extremity muscle architecture was defined by combining cadaver and in-vivo MRI muscular data

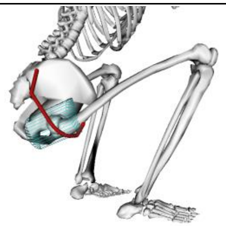
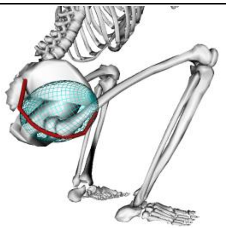
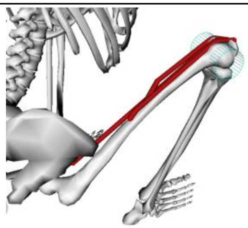
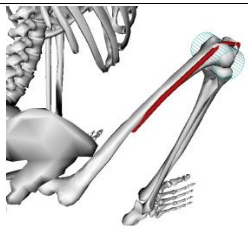
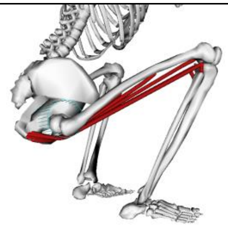
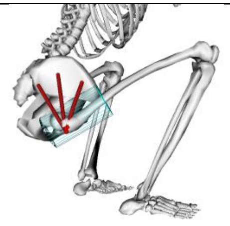
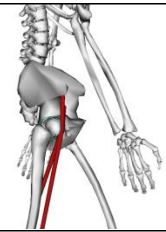
(Rajagopal et al. 2016), and recently updated to allow muscle-driven simulations of higher knee flexion tasks (Lai et al. 2017).

This model was further adjusted by altering the WS parameters through visual assessment of a subject during a deep squatting. Modifications were done respecting the anatomical shape of the muscles, preventing bone crossing and respecting muscle MA reported by Németh & Ohlsén (1985), and the model was then used to simulate walking and deep squat.

8.2.1. Modifications to the Model

To allow simulation-based studies of deep squatting involving high hip and knee flexions, we increased maximal hip flexion from 120º to 138º and maximal knee flexion from 140º to 145º. Four WS of six MTU (superior and middle portions of the gluteus maximus, rectus femoris, vastus intermedius, vastus medialis and vastus lateralis) were updated. Two additional WS were implemented in order to prevent nine MTU (distal, ischial and middle portions of the adductor magnus, biceps femoris long head, semimembranosus, semitendinosus, and anterior, medial and posterior portions of the gluteus medius) from crossing the femur and/or pelvis in deep hip and knee flexion angles (Table 8.1). A third additional WS was implemented at the head of the femur to prevent the rectus femoris and the sartorius muscles to cross the bone during hip extension. Cylindrical WS were used, except for the WS for the middle portion of the gluteus maximus, where an ellipsoidal surface was used instead.

Table 8.1. Details of pelvis and femoral updated and implemented WS – all data is applicable to the right side of the body

		Updated WS				**Implemented WS**		
WS Name		Gmax1_at_pelvis_r	Gmax2_at_pelvis_r	KnExt_at_fem_r	KnExtVL_at_fem_r	Post_at_pelvis_r	Gmed_at_pelvis_r	Flex_at_femhead_r
Associated Joint		pelvis	pelvis	femur	femur	pelvis	pelvis	femur
Associated MTU		Gluteus maximus 1 (superior)	Gluteus maximus 2 (middle)	Rectus Femoris Vastus Intermedius Vastus Medialis	Vastus Lateralis	Adductor magnus (distal) Adductor magnus (ischial) Adductor magnus (middle) Biceps femoris long head Semimembranosus Semitendinosus	Gluteus medius (anterior) Gluteus medius (middle) Gluteus medius (posterior)	Rectus Femoris Sartorius
Geometric Format		cylinder	ellipsoid	cylinder	cylinder	cylinder	cylinder	cylinder
Size (m)	radius	0.039	0.067 / 0.067	0.03	0.03	0.045	0.04	0.024
	length	0.12	0.1	0.1	0.1	0.12	0.15	0.06
Body Rotation (rad) Respect to the pelvis	x	-0.6	-0.5	-0.06234	-0.06234	-0.1	-1	0
	y	0.45	0.39	0.0507601	0.0507601	0	0.6	0
	z	0.5	0	0	0	0	0.7	0
Translation (m) Respect to the pelvis	x	-0.061	-0.058	-0.0019	-0.004	-0.06	-0.03	-0.005
	y	-0.0775	-0.043	-0.40273	-0.4023	-0.07	-0.055	0
	z	0.09	0.09	0.002091	0.002091	0.068	0.1	0.01
								

8.2.2. Motion capture

Subject-specific simulations of squatting were done by tracking experimental data from a healthy male participant (42.5 years, 89.2kg, 186cm). The motion capture system included 10 infrared cameras at 200Hz (MX-13, Vicon, Oxford, UK). For the deep squat, the participant stood with each foot on a force platform (1000Hz, Bertec Corporation, Columbus, OH), feet hip-width apart, and was instructed to squat as low as possible without lifting his feet from the floor. Electromyography (EMG) activity of 8 lower limb muscles was monitored (FreeEMG 300, BTS, Padua, Italy). The squat cycle was analyzed from standing to deepest squat (0-50%) and back to standing position (50-100%). The markers trajectories (Mantovani & Lamontagne 2016) were labelled and filtered in Nexus 2.5 (Vicon, Oxford, UK) and imported into OpenSim (Mantoan et al. 2015). The anterior and posterior superior iliac-spine, and lateral and medial epicondyles markers were placed according to identification through a CT scan (GE Healthcare, Mississauga, Canada). EMG data were normalized to their peak activation during a maximum voluntary isometric contraction.

8.2.3. Model Evaluation

Simulations were performed for the original (Lai et al. 2017) and updated MSKM. The MSKM were first scaled based on a static pose. Joint kinematics and the net joint moments for each degree of freedom were computed using the inverse kinematics and inverse dynamics tools. Muscle activations were calculated using static optimization while minimizing the sum of squared muscle activations. The MA lengths of the quadriceps, hamstrings, glutei and adductor magnus, as well as lower limb muscle activations during squat, were compared between the original and modified MSKM.

8.3. Results

No differences in kinematics were observed between the two models. Visual inspection of the muscle paths shows that the WS prevented all hip and knee muscles from crossing the bony structures during the squatting task (supplemental figure 8.3). Joint angles achieved maximally 120.2º hip flexion, 13.7º hip abduction, 19.6º hip external rotation and 142.2º knee flexion.

Modifying the two knee WS (KnExt_at_fem, KnExtVL_at_fem) increased the MA of the rectus femoris and vastus lateralis for knee flexion angles larger than 141º. The changes in the WS (Gmax2_at_pelvis) corrected the middle portion of the gluteus maximus MA after 50º of hip flexion, resulting in a 25 mm increase at maximum hip flexion. The additional posterior WS (Post_at_pelvis) affected the MA of the adductor magnus, biceps femoris, semimembranosus, and semitendinosus from 96º of hip flexion onwards, increasing the hip MA up to 30 mm. The WS (Flex_at_femhead) at the femoral head affected the sartorius and the rectus femoris only after 13º and 24º of hip extension, respectively (Figure 8.1).

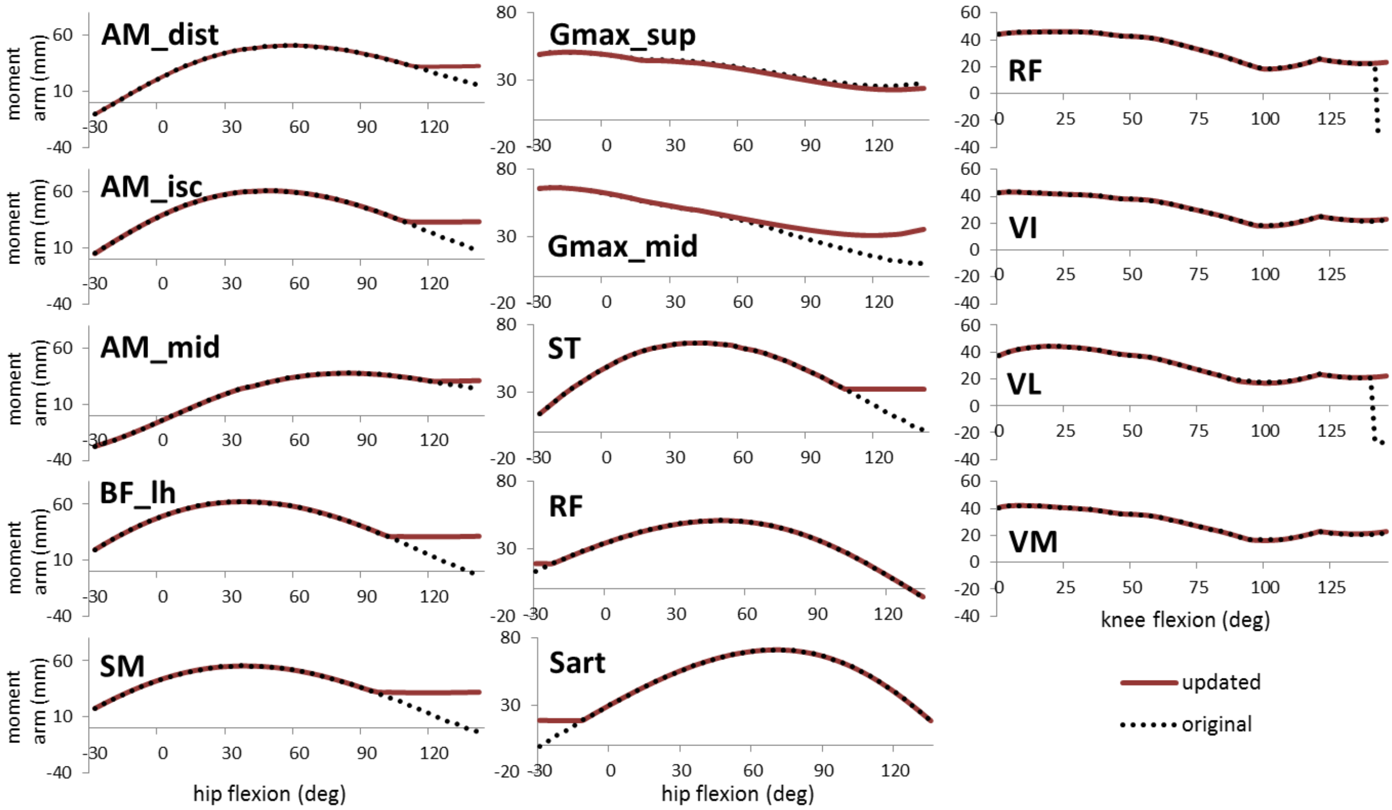


Figure 8.1. Muscle MA between the original (black dotted) and updated models (red) for hip and knee sagittal movement. The fourteen MTU include the three portions of the adductor magnus (distal: AM_dist, ischial: AM_isc, middle: AM_mid) , biceps femoris long head (BF_lh), semimembranosus (SM), two portions of the gluteus maximus (superior: Gmax_sup, middle: Gmax_mid), semitendinosus (ST), rectus femoris (RF), sartorius (Sart), vastus intermedius (VI), vastus lateralis (VL) and vastus medialis (VM).

Muscle activation patterns predicted by the updated model avoided activations dropping to zero or maxing out when the model was in extreme hip and knee flexions (Figure 8.2). Measured EMG had higher activity than predicted models activations.

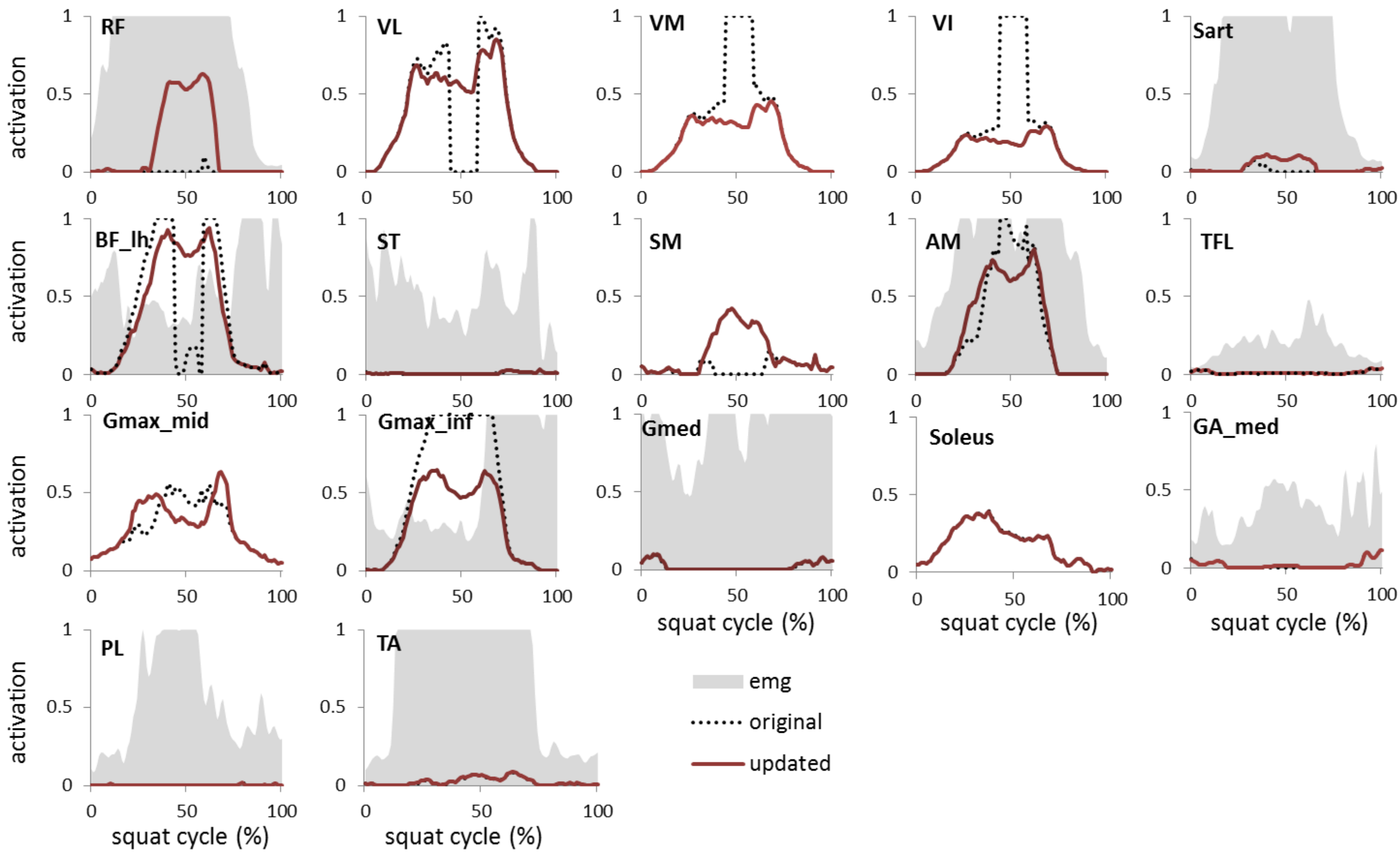


Figure 8.2. Muscle activation patterns predicted from the original (black dotted) and updated (red) models, compared with measured EMG activity (shaded regions) for fifteen MTU during the squat task: rectus femoris (RF), vastius lateralis (VL), vastus medialis (VM), vastus intermedius (VI), sartorius (Sart), biceps femoris long head (BF_lh), semitendinosus (ST), semimembranosus (SM), the proximal portion of the adductor magnus (AM), tensor fascia latea (TFL), the middle portion of the gluteus maximus (Gmax_mid), the inferior portion of the gluteus maximus (Gmax_inf), the anterior portion of the gluteus medium (Gmed), soleus, medial gastrocnemius (GA_med), peroneus longus (PL) and tibialis anterior (TA).

8.4. Discussion

In this study, a newly developed model (Rajagopal et al. 2016; Lai et al. 2017) was modified to allow extreme hip flexion angles. The results show that the updated model avoids very small muscle MA while performing a high ROM task. Mostly cylindrical WS were used, as it improves simulation speed in comparison to ellipsoidal WS (Rajagopal et al. 2016). An ellipsoidal WS in the middle gluteus maximus was necessary, as during a deep squat task, the hip also abducts and internally rotates, making the positioning of a simpler cylindrical WS very challenging.

A qualitative analysis of the modified MSKM showed no muscles crossing the pelvis, femur or tibia, in contrast to the original model. The MA lengths reported in this study are within the range described in the literature (Németh & Ohlsén 1985) – supplemental figure 4 – although the deep squat requires larger hip flexion ROM than the reported studies. MTU that show incorrect MA in deep hip flexion may cause an instantaneous drop in generated activation in an overall muscle group (e.g. VL, BF_lh, SM, figure 8.2).

Our proposed updates extend the model's functional range of motion, making it more applicable to biomechanical studies of movements involving high hip and knee flexions. Still, individual examination when scaling the model is highly recommended, since pelvic geometry may change and affect the location of the WS with respect to the muscle paths. This model is available from SimTK.org (https://simtk.org/projects/high-hip-flex).

8.5. References

Arnold EM, Ward SR, Lieber RL, Delp SL. 2010. A Model of the Lower Limb for Analysis of Human Movement. Ann Biomed Eng. 38:269–279.

Delp SL, Anderson FC, Arnold AS, Loan P, Habib A, John CT, Guendelman E, Thelen DG. 2007. OpenSim: Open-Source Software to Create and Analyze Dynamic Simulations of Movement. IEEE Trans Biomed Eng. 54:1940–1950.

Delp SL, Loan JP, Hoy MG, Zajac FE, Topp EL, Rosen JM. 1990. An Interactive Graphics-Based Model of the Lower Extremity to Study Orthopaedic Surgical Procedures. IEEE Trans Biomed Eng. 37:757–67.

Lai AKM, Arnold AS, Wakeling JM. 2017. Why are Antagonist Muscles Co-activated in My Simulation? A Musculoskeletal Model for Analysing Human Locomotor Tasks. Ann Biomed Eng. 45:2762–74.

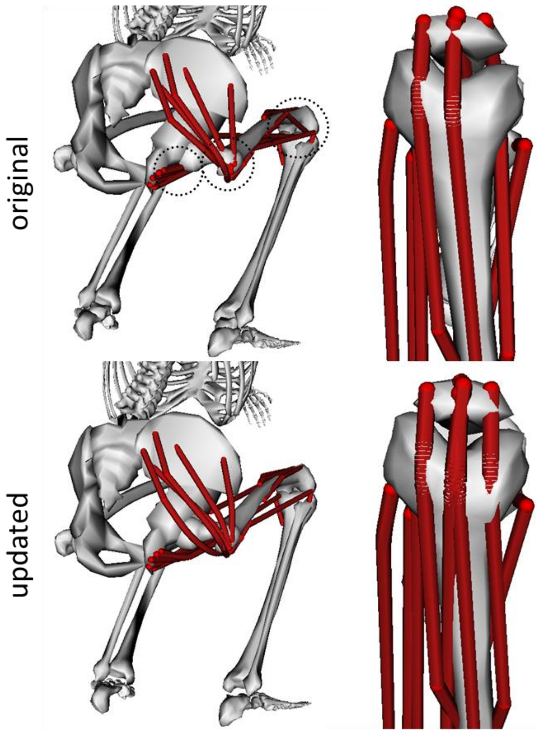
Mantoan A, Pizzolato C, Sartori M, Sawacha Z, Cobelli C, Reggiani M. 2015. MOtoNMS: A MATLAB Toolbox to Process Motion Data for Neuromusculoskeletal Modeling and Simulation. Source Code Biol Med. 10:12.

Mantovani G, Lamontagne M. 2016. How Different Marker Sets Affect Joint Angles in Inverse Kinematics Framework. J Biomech Eng. 139:44503.

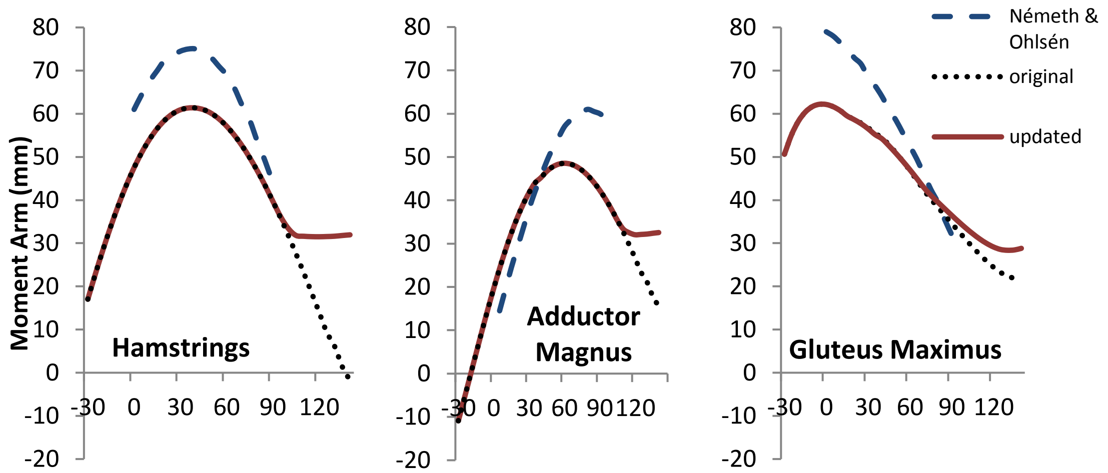
Németh G, Ohlsén H. 1985. In Vivo Moment Arm Lengths for Hip Extensor Muscles at Different Angles of Hip Flexion. J Biomech. 18:129–140.

Rajagopal A, Dembia CL, Demers MS, Delp DD, Hicks JL, Delp SL. 2016. Full Body Musculoskeletal Model for Muscle-Driven Simulation of Human Gait. IEEE Trans Biomed Eng. 63:2068–79.

8.6. Supplementary Material



Supplemental Figure 8.3. Muscle paths during extreme deep squatting in the original (top) and the updated (bottom) models.



Supplemental Figure 8.4. Model MA vs. hip flexion angle for hamstrings (average of biceps femoris long head, semimembranosus and semitendinosus), adductor magnus (average of distal, ischial and middle portions) and the gluteus maximus (average of superior, middle and inferior portions) of the original and updated models, compared to experimental data.

9 Muscle Force Contributions and Hip Contact Forces during a Deep Squatting Task Two-Years after Surgical Correction of Cam Deformity

Danilo S. Catelli [1,6] | K. C. Geoffrey Ng [2] | Mariska Wesseling [3] | Erik Kowalski [1] | Ilse Jonkers [3] | Paul E Beaulé [4] | Mario Lamontagne [1,5]

[1]School of Human Kinetics, University of Ottawa, Canada
[2]Department of Mechanical Engineering, Imperial College London, UK
[3]Department of Human Movement Sciences, KU Leuven, Leuven, Belgium
[4]Division of Orthopaedic Surgery, University of Ottawa, Canada
[5]Department of Mechanical Engineering, University of Ottawa, Canada
[6]CAPES Foundation, Ministry of Education of Brazil, Brasilia, Brazil

Contents of this chapter prepared as an article for submission in *Clinical Biomechanics®* | December 2018

Catelli DS, Ng KCG, Wesseling M, Kowalski E, Jonkers I, Beaulé PE, Lamontagne M, Muscle Force Contributions and Hip Contact Forces during a Deep Squatting Task after Two-Years the Surgical Correction of Cam Deformity. 2018; Pending Submission.

9.0. Abstract

Background: Corrective hip preservation surgery for cam-type femoroacetabular impingement (FAI) aims to avoid joint degeneration. Although medium-term outcomes and gait kinematics have been reported, it is unclear how it would translate to better hip muscle and joint loading in high demanding tasks. The purpose of this study was to compare the muscle forces and hip joint contact forces during a deep squatting task in individuals before and two years after cam FAI surgical correction.

Methods: Motion analysis data of ten male patients with symptomatic cam FAI performing a deep squatting task, was recorded pre- and at two years post-hip osteochondroplasty. Muscle and hip contact forces (HCF) were estimated using musculoskeletal modelling. Statistical parametric mapping (SPM) analysis was used to compare the waveforms of kinematics, muscle forces and HCF outputs pre- and post-surgery.

Findings: Postoperative patients squatted down with a higher anterior pelvic tilt and hip flexion. Increased semimembranosus force and higher vertical HCF were observed postoperatively during squat ascent.

Interpretation: A higher anterior pelvic tilt was associated with an innate restoration of the pelvis position, after surgical correction of FAI. The increased force of the semimembranosus muscle while ascending the squat generated higher vertical HCF, which also influenced the increased HCF total magnitude.

9.1. Introduction

Osteochondroplasty of the cam deformity is a joint-preserving procedure that has become very popular in orthopaedics (Beaulé et al., 2009), as it has been providing excellent outcomes on relieving pain, improving quality of life and clinical function in symptomatic femoroacetabular impingement (FAI) patients (Beaulé et al., 2017; Botser et al., 2011; Joseph et al., 2016; Larson and Giveans, 2008; Philippon et al., 2009). Although a high percentage of the general population will not notice any symptoms of this deformity and will go through life without any consequences of its presence (Chakraverty et al., 2013), symptomatic cam morphology has been associated with acetabular cartilage damage (Anderson et al., 2009; Beaulé et al., 2007; Ganz et al., 2001; Khanna et al., 2014; Murphy et al., 2004) and is proposed as a risk factor for hip osteoarthritis (OA) (Agricola et al., 2013; Ganz et al., 2008, 2003).

Patients with cam FAI demonstrate altered gait (Diamond et al., 2016; Hunt et al., 2013; Kennedy et al., 2009; Rylander et al., 2011), squatting (Bagwell et al., 2016; Diamond et al., 2017a; Lamontagne et al., 2009), and stair climbing (Diamond et al., 2018; Rylander et al., 2013) biomechanics, indicating reduced hip and pelvic range of motion (ROM), and reduced hip flexion and external rotation moments compared to healthy controls. Also, neuromuscular adaptations have been reported to influence symptoms and contribute to changes in biomechanical outcomes (Casartelli et al., 2011; Diamond et al., 2017b; Kennedy et al., 2009; Kierkegaard et al., 2017; Ng et al., 2018).
Several studies reported hip biomechanics after joint preservation (Brisson et al., 2013; Lamontagne et al., 2011; Malagelada et al., 2015; Rylander et al., 2013, 2011), however the effect on musculoskeletal loading in terms of the of dynamic muscle forces, and especially, the hip contact force (HCF), is still lacking, therefore proof on the effect of surgical procedures in restoring musculoskeletal loading still has to be provided. Furthermore, the understanding of changes in muscle forces and their consequent effect on joint loading may provide benefits to design better pre- and/or postoperative FAI rehabilitation programs (Bennell et al., 2017; Lewis et al., 2009). Indeed, reduced psoas and iliacus

muscle forces were previously confirmed during gait in preoperative FAI when compared to healthy controls (Ng et al., 2018), demonstrating that neuromuscular adaptations may influence symptoms and biomechanical outcomes. Furthermore, given the large amplitudes of hip motion seen during a deep squat task compared to gait ranges, squat is more likely to elicit an impingement-like condition, and it may therefore provide an even better understanding of the role of FAI in altered muscle and joint.

Therefore, this study compared the pre-post surgical hip joint and muscle forces in a deep squat task in a group of FAI patients.

9.2. Methods

9.2.1. Participants

Twelve male patients were initially recruited from the surgeon's practice enrolled in this study, presenting unilateral clinical signs of hip pain and positive impingement tests. However two participants were removed from the analysis due to obesity and technical issues (Table 9.1). All patients underwent pelvic CT imaging (Acquilion, Toshiba Medical Systems Corporation, Otawara, Japan; or Discover CT750, GE Healthcare, Mississauga, ON, Canada), to confirm the presence of the cam FAI morphology – i.e. axial (3:00) and/or radial (1:30) alpha angle larger than 50.5° and 60°, respectively (Ganz et al., 2003; Nussbaumer et al., 2010; Ribas-Fernandez et al., 2007). During this exam, bony landmarks of the anterior superior and posterior superior iliac spine (ASIS and PSIS), as well as the medial and lateral knee epicondyles were identified and marked with radiopaque surface markers. The participants were then transferred to the local university where they completed the Hip Disability and Osteoarthritis Outcome Score (HOOS) questionnaire and performed the motion analysis protocol. The same senior surgeon performed surgical correction (e.g. osteochondroplasty and labral-chondral debridement) via an open dislocation with an anterior approach (n = 3) or arthroplasty with a mini-open approach (n = 7). The same protocol was performed preoperatively and 2-years postoperatively (25.2 ± 1.1 months). Exclusion criteria consisted of any other hip morphology, a severe history of lower limb traumas or surgeries, or a body mass index (BMI) indicating obesity (> 35 kg/m2). The study was approved by the hospital's and university's research ethics boards, and all participants provided written informed consent for participation.

Table 9.1. Summary of pre- and postoperative patient demographics, as well as pain questionnaire, and cam deformity measurement parameters of the paired affected hips, reporting mean ± SD.

Parameter		FAI pre-op	FAI post-op	p-value
Participants (n)			10	
Age (years)		35 ± 8	37 ± 8	
BMI (kg/m^2)		26 ± 3	26 ± 4	.76
alpha-angle (deg)	3:00 position	54 ± 8	44 ± 3	.002
	1:30 position	66 ± 5	50 ± 7	.001
HOOS	Symptoms	71 ± 11	82 ± 10	.025
	Pain	71 ± 17	91 ± 6	.001
	Activities of Daily Living	81 ± 15	97 ± 2	.006
	Sports and Recreational Activities	57 ± 26	87 ± 14	.004
	Quality of Life	39 ± 23	67 ± 22	.005

9.2.2. Motion Analysis

Forty-five retro-reflective markers were placed on the participants according to the University of Ottawa Motion Analysis Model (UOMAM) marker set (Mantovani and Lamontagne, 2016). Five deep squatting trials were performed at a self-selected pace, with the feet positioned parallel, hip-width apart and the arms stretched out anteriorly. The marker trajectories were captured using a ten-camera infrared system sampled at 200 Hz (Vicon MX-13, VICON, Oxford, UK) and ground reaction forces (GRF) were captured using two embedded force plates sampled at 1000 Hz (FP4060-08, Bertec Corporation, Columbus, USA). The data were labelled and filtered (zero-lag, 6Hz fourth order Butterworth) using Nexus 2.6.1 (VICON, Oxford, UK). The squat analyses were performed on the full squat cycle (defined by the maximum hip extension point – standing – and lowest depth point – squatted – during descending and ascending phases combined) and all variables were time-normalized to its cycle.

9.2.3. Musculoskeletal Modelling

A newly customized musculoskeletal model (MSKM) (Catelli et al., 2018), based on a previous generic MSKM (Lai et al., 2017; Rajagopal et al., 2016) and specifically adapted for high hip and knee flexion ranges (Chapter 8), containing 80 lower-limb Hill-type muscle-tendon units (MTU) with 37 degrees of freedom (DoF), was used in an open-source musculoskeletal simulation software (OpenSimTM 3.3, Stanford University, Stanford, USA) (Delp et al., 2007).

The marker trajectories and GRF dataset were prepared to OpenSim file format (Mantoan et al., 2015) and the models were scaled based on each patient's static anthropometric dimensions. The anterior superior and posterior superior iliac spine (ASIS and PSIS), as well as the medial and lateral knee epicondyles placement were defined according to their identification during CT scanning, therefore pelvis and knee markers had a ten-time higher scaling weight. Inverse kinematics and inverse dynamics tools were used to compute joint angles and net joint moment for each degree of freedom, while the static optimization tool was used to compute muscle forces, while minimizing the sum of squared muscle activation. An optimal force of 10N was defined for the reserve actuators for the three hip coordinates. The 'JointReaction' analyze tool calculated HCF as three-dimensional vectors acting on the acetabulum expressed in the femoral coordinate system. The hip muscle forces, each HCF component (i.e. x: anterior-posterior, y: superior-inferior, z: medial-lateral) and resultant magnitude were normalized to body weight and were selected as variables, along with the direction of the HCF vector in the sagittal and frontal planes – Figure 9.1.

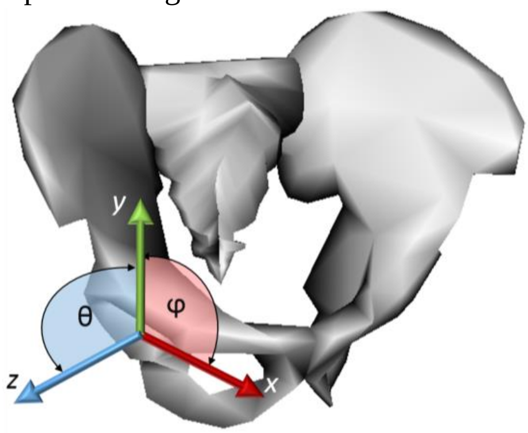


Figure 9.1. Three-dimensional HCF vector directions for the right hip that was expressed in the hip joint centre along the anterior-posterior (x), superior-inferior (y) and medial-lateral (z) axis, along with the sagittal (ϕ) and frontal (θ) plane angles.

9.2.4. Data analysis

Squat time execution was determined for both phases separately, and squat depth was calculated based on the height of the center of the pelvis at its lowest point in the squat cycle divided by the length of the leg. The maximum squat depth is reported as a percentage of leg length (measured from the affected anterior superior iliac spine to the medial malleolus), with a value of 0% representing a maximum squat. Data from the five trials performed per patient were averaged, and only the affected (surgical) side was analyzed.

A Statistical Parametric Mapping (SPM) (Pataky, 2010) two-tail paired t-test was performed to compare the kinematics, muscle forces and the HCF outputs (95% CI) in the time-normalized to the full squat cycle (0-100%) data. This statistical analysis considers entire waveform data and therefore does consider for differences in movement speed. A statistical parametric map, SPMt was calculated, which represents the traditional univariate t-statistic being calculated at each point of the waveform. If SPMt exceeds the critical threshold t, the two groups are considered significantly different in that part of the waveform. Although all 40 MTU were analyzed, only the hip MTU that presented a force higher than 0.5 BW were plotted.

The demographics and the discrete data were assessed for normality using the Shapiro-Wilk test and paired t-test analyses were performed (95% CI). All analyses were performed in a custom Matlab script (v. R2017b, MathWorks Inc, Natick, USA).

9.3. Results

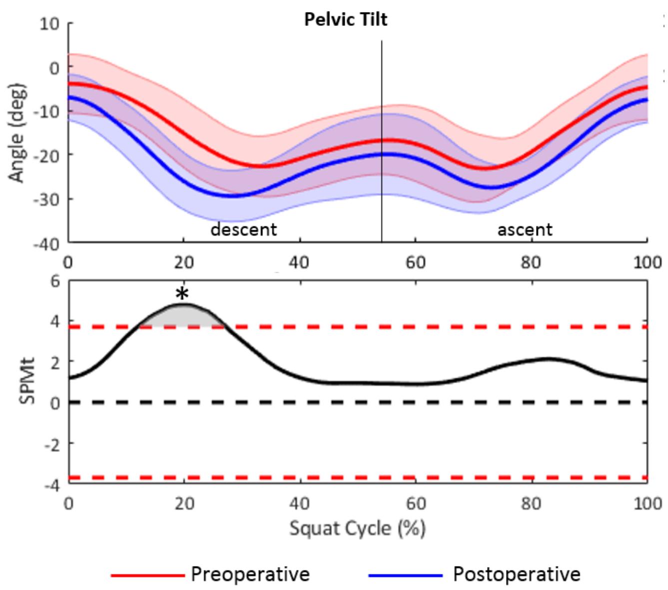
9.3.1. Demographics and Patient Reported Outcome Measure

Postoperative patients showed outcome improvements in all five HOOS categories, and their follow-up BMI remain unaltered from their preoperative values (Table 9.1).

9.3.2. Squat parameters and Kinematics

No differences on time execution was found between pre- (descent: 2.5 ± 0.8 s, ascent: 2.0 ± 0.5 s) and postoperative (descent: 2.0 ± 0.5 s, ascent: 1.9 ± 0.5 s) patients on both phases of the squat. Also, no differences in squat depth were detected from pre- (31 ± 9 % leg length) to postoperative (31 ± 14 % leg length) values.

Lower limb kinematics analysis showed higher anterior pelvic tilt on the postoperative patients during the whole task, with a significant difference being reached at the descending phase between 12 to 27% of the squat cycle ($p < 0.05$). Higher hip flexion was also found at about the same time-point 11 to 27% of the squat cycle ($p < 0.05$). No differences in the hip abduction were confirmed between the groups ($p > 0.05$) – Figure 9.2.



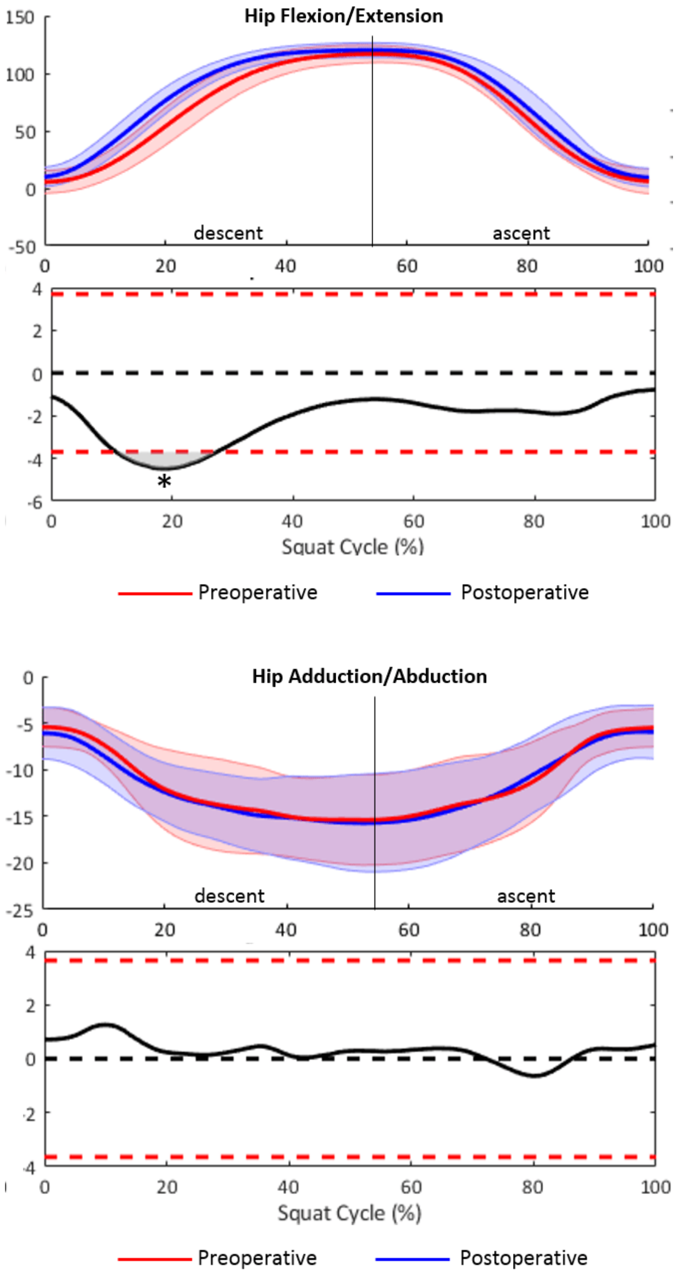


Figure 9.2. Mean joint angles and SPM analysis of the pelvis and hip muscle forces waveforms. A negative value for the 'Pelvic Tilt' plot represents anterior tilt. The top plot represents averaged kinematics per group while the bottom plot represents the t-statistic as a function of time (SPMt). When SPMt goes over the threshold (red dashed line), the significance is reached (*).

9.3.3. Muscle Forces

The SPM analysis of the muscle force magnitude demonstrated no differences between preoperative and postoperative patients for all muscles, except for the semimembranosus, in which the force generated between 68-71% of the squat cycle were increased in the postoperative patients (p = 0.014). Six hip muscles had a peak force higher than 0.5 N/BW and were plotted below – Figure 9.3.[3]

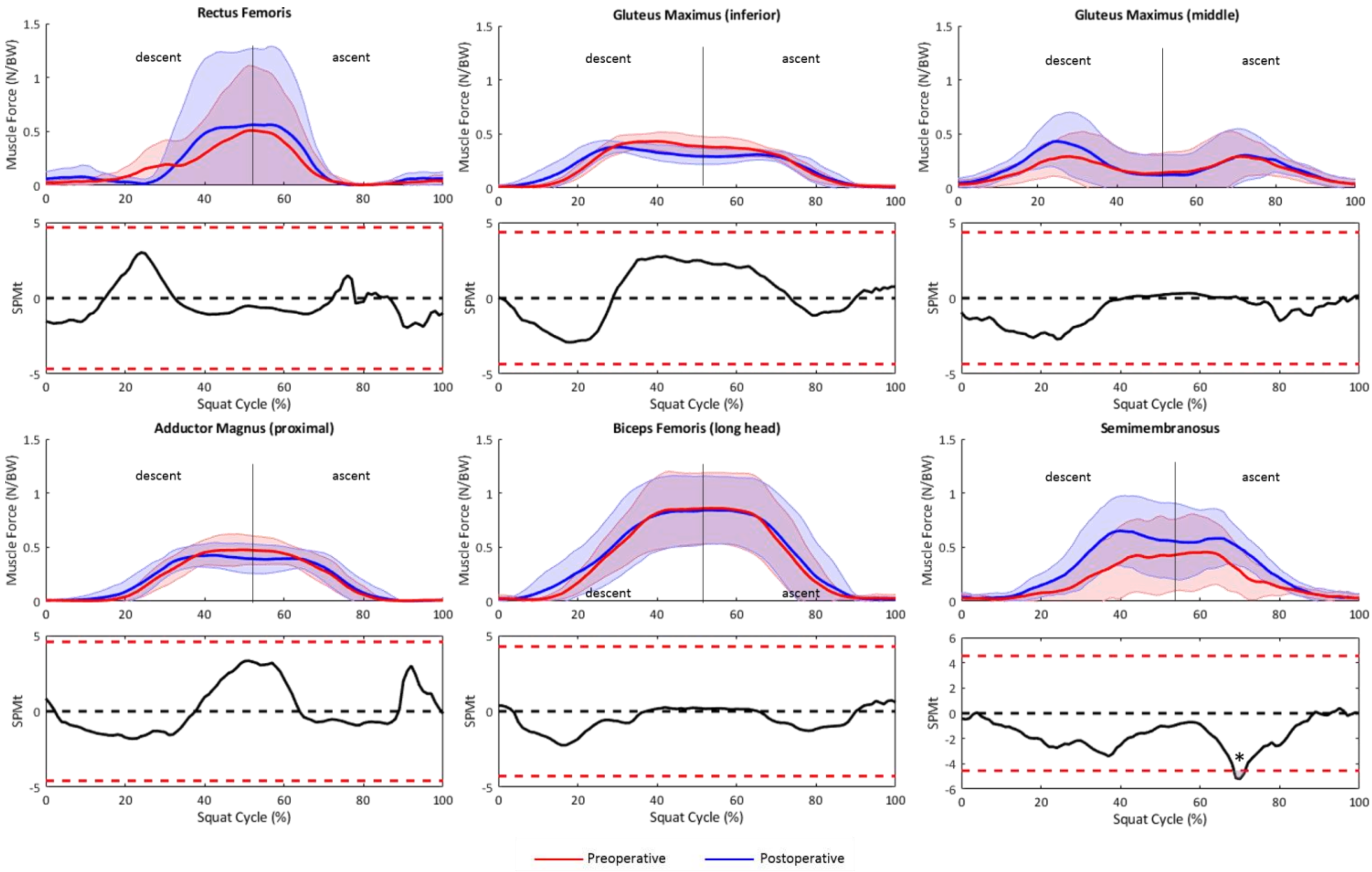


Figure 9.3. Mean forces and SPM analysis of the hip muscle forces waveforms. The top plot represents averaged muscle force per group while the bottom plot represents the t-statistic as a function of time (SPMt). When SPMt goes over the threshold (red dashed line), the significance is reached (*).

9.3.4. Hip Contact Forces

The HCF in the three planes of action, as well as the force vector and direction in the sagittal and the frontal planes, are illustrated in Figure 9.4. Patients showed no postoperative differences in the anterior-posterior and the medial-lateral axis. However, the superior-inferior contact forces were higher from 67 to 70% of the squat cycle in the postoperative patients (p = 0.011), which

[3] The coefficient of determination (R^2) of the muscles that were assessed by EMG are available in the appendix – Table A.1.

cause a significant difference in the total magnitude vector at the same time interval (p = 0.015). Sagittal and frontal planes HCF vector direction did not show any statistical significance during the squat – Figure 9.4.

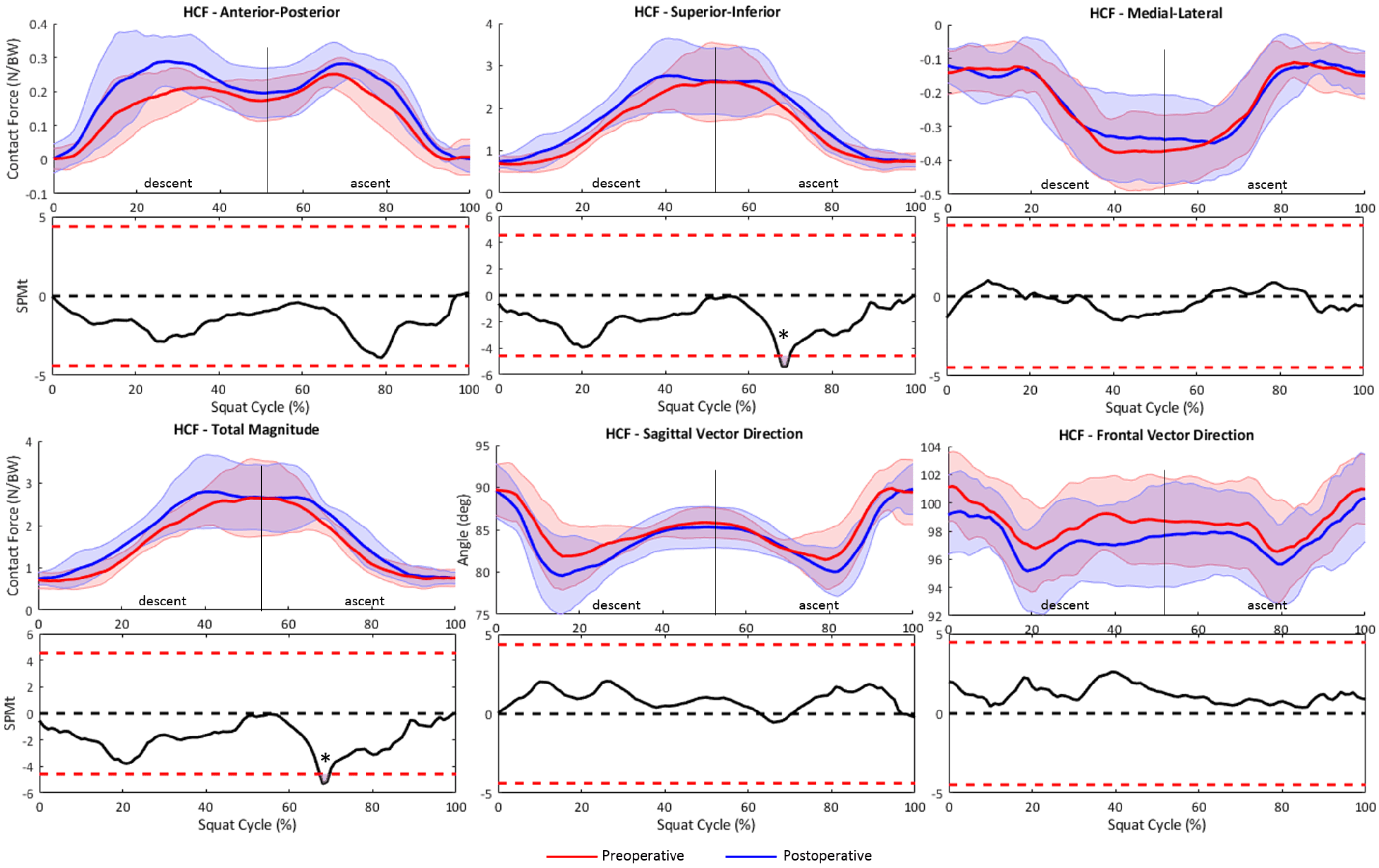


Figure 9.4. Hip contact forces normalized by body weight (BW) and vector direction analysis during the squat task, for the FAI preoperative (red) and postoperative groups (blue), in the three planes of action (top) and its magnitude force (bottom right), as well as the sagittal (bottom centre) and frontal (bottom right) planes vector direction. When SPMt goes over the threshold (red dashed line), the significance is reached (*).

9.4. Discussion

To our knowledge, this is the first study that used musculoskeletal modelling to compare pre- and postoperative patients with cam-type FAI during a deep squatting task, to examine the effects of the osteochondroplasty on musculoskeletal loading as reflected in hip muscle and contact forces. Our findings show that the FAI patients although not improving the squat depth (pre-op: 30.6 ± 8.9%; post-op:31.3 ± 13.4% of leg length), showed altered squat kinematics postoperatively, with increased anterior pelvic tilt and higher hip flexion during the descending phase of the squat. They, presented increased semimembranosus force, which impacted the HCF inducing a higher superior-inferior and resultant force magnitude during the ascending phase of the squat.

Lamontagne et al. (2011) showed increased squat depth due to higher knee flexion and ankle dorsiflexion angles. Moreover, although not showing pelvic tilt improvements, postoperative patients were likely to present higher posterior pelvic tilt at the maximal squat depth than preoperatively. However, our subjects presented a higher anterior tilt during the whole squat cycle postoperatively, although no SPM differences were found at the mid-squat cycle. The differences in the pelvis outcomes for both studies could be justified by the cohort, which in the present study consisted exclusively of males, while they had a slightly younger and mixed composition cohort, or the model's choice. Ultimately, considering the anterosuperior location of the cam-deformity in the femoral head (Chakraverty et al., 2013) is susceptible to impinge when the pelvis is anteriorly tilted (Ross et al., 2014), in their native standing position, our FAI patients were in a more anteriorly tilted position, which would have caused the cam formation during bone maturation (Carsen et al., 2014). However, the development of the cam and the further onset of the symptomatology may have caused the patients to adopt a more posteriorly tilted pelvic position, as well as limiting their hip mobility as a protected mechanism in order to reduce HCF (Ng et al., 2018). Postoperatively (i.e. once the cam was removed), the patients were able to return to the more anteriorly (innate) tilted position as the impingement was no longer affecting their kinematics.

The bi-articular semimembranosus acts eccentrically during the descent phase of the squat (increased muscle-tendon length), and concentrically to rise, in order to superpose the body weight load and extend the hip. Increased activation of the other synergist medial hamstring, the semitendinosus, was already explored postoperatively in the study presented in Chapter 5 (Figure 5.2), during squat descent. However, the increased dynamic force calculated for the semimembranosus was only reached in the ascending phase of the squat. The high standard deviation reached by the semimembranosus during the descending phase in the model, possibly prevented to statistically confirm the increase of the semimembranosus force, as was the case when the patients were squatting down.

Regarding the rectus femoris muscle, a lower postoperative electromyography activation during the squat (Chapter 5) and force production during level walking (Chapter 7) have been reported in the previous phases of this study. The also decreased hamstring force production during walking (biceps femoris and semimembranosus), was associated with no need for these muscles to compensate postoperatively in an open kinetic chain movement, since medialized hip contact forces were detected during the entire stance phase of the gait after cam-deformity removal (Chapter 7). In fact, for being a bi-articular knee extensor, the rectus femoris when activated results in a hip flexion moment, which may not be beneficial if a hip extension moment is required, as for the squatting task. Also, the higher calculated forces of the rectus femoris were reached at the bottom of the squat, where a higher knee lever arm takes effect.

The postoperative higher force of the semimembranosus MTU increased HCF in the inferior-superior aspect of the vector and its resultant magnitude force (Wesseling et al., 2015). Nonetheless, the peak resultant HCF were directed towards the anterior-superior femoral head (Figure 9.4), which are consistent with the chondrolabral damage location in patients with FAI (Beaulé et al., 2005; Beck et al., 2005; Ganz et al., 2008). In case the pelvis remains anteriorly tilted at the bottom of the squat, the femoroacetabular joint may be conducive for a hip impingement (Ross et al., 2014), as the femoral interaction would most likely lead to contact with the acetabular rim. Increasing the posterior pelvic tilt, on the other hand, would avoid bony interaction, which might be the reason for the larger posterior pelvis tilit in the preoperative patients.

There are some limitations to this study that should be addressed. First, we acknowledge the small sample size of our cohort, increasing the number of participants would result in higher predictive power. Second, our cohort consisted of patients that underwent osteochondroplasty with two types of approaches: surgical dislocation or arthroscopy. Although comparable functional (Bedi et al., 2011) and patient outcomes (de Sa et al., 2016) perspectives have been recently reported, the muscle effect of both surgeries could affect the force-length relationship and/or the maximum isometric force in particular MTUs differently, which has not been accounted in this model. Third, the static optimization method to calculate muscle forces may struggle to express co-contraction mechanisms altered by a joint pathology; however, this is still preferable than using invasive needles to access deep muscle activity to perform EMG-driven simulations (Buchanan et al., 2004)[4]. Fourth, the cam-type morphology has not been directly included in the musculoskeletal model as scaled generic models were used. This does not validate the current modeling approach as the kinematics solution as driven by the measured marker positions.

Suggestions for future studies may include: i) the use of FE formulation based on three-dimensional surfaces for the pelvis, femoral head and labrum to visualize and evaluate the loading of the internal hip structures during loaded dynamic motion, while distinguishing subtle differences in intersubject hip morphology (Kapron et al., 2014; Rylander, 2014), ii) the use of magnetic resonance imaging to include the subject-specific cam-type hip bone morphology, to visualize intersubject pre- and postoperative effects of the surgical intervention during loaded dynamic motion; and iii) a controlled clinical trial with a conservative management of FAI pain, which can test if any gain in pelvic mobility may also generate increased muscle force before performing the surgery correction.

9.5. Conclusion

This study provided insights of the mid-term evaluation after surgical correction of cam deformity on muscle and hip contact forces. Although the postoperative patients altered squat kinematics (pelvic tilt and hip flexion) only during the descent phase, the postoperative higher muscle force for the semimembranosus muscle happened during the ascending phase. This increased force

[4] Coefficient of determination (R^2) of the predicted muscle activation from the squat model and the muscle excitations measured by EMG are included in the Appendix section (Table A.1). The predicted muscle activations were shifted back over the excitations signals in order to account for the electrical mechanical delay.

during hip extension also caused an increased contact force of the vertical HCF component and its total magnitude.

9.6. References

Agricola, R., Waarsing, J.H., Arden, N.K., Carr, A.J., Bierma-Zeinstra, S.M.A., Thomas, G.E., Weinans, H., Glyn-Jones, S., 2013. Cam Impingement of the Hip - A Risk Factor for Hip Osteoarthritis. Nat. Rev. Rheumatol. 9, 630–634. https://doi.org/10.1038/nrrheum.2013.114

Anderson, L.A., Peters, C.L., Park, B.B., Stoddard, G.J., Erickson, J.A., Crim, J.R., 2009. Acetabular Cartilage Delamination in Femoroacetabular Impingement. Risk Factors and Magnetic Resonance Imaging Diagnosis. J. Bone Jt. Surg. 91, 305–313. https://doi.org/10.2106/JBJS.G.01198

Bagwell, J.J., Snibbe, J., Gerhardt, M., Powers, C.M., 2016. Hip Kinematics and Kinetics in Persons with and Without Cam Femoroacetabular Impingement during a Deep Squat Task. Clin. Biomech. 31, 87–92. https://doi.org/10.1016/j.clinbiomech.2015.09.016

Beaulé, P.E., Allen, D.J., Clohisy, J.C., Schoenecker, P.L., Leunig, M., 2009. The Young Adult with Hip Impingement: Deciding on the Optimal Intervention. J. Bone Jt. Surg. 91, 210–21. https://doi.org/10.1016/j.jhse.2007.02.052

Beaulé, P.E., Le Duff, M.J., Zaragoza, E., 2007. Quality of Life Following Femoral Head-Neck Osteochondroplasty for Femoroacetabular Impingement. J. Bone Jt. Surg. - Ser. A 89, 773–779. https://doi.org/10.2106/JBJS.F.00681

Beaulé, P.E., Speirs, A.D., Anwander, H., Melkus, G., Rakhra, K., Frei, H., Lamontagne, M., 2017. Surgical Correction of Cam Deformity in Association with Femoroacetabular Impingement and Its Impact on the Degenerative Process within the Hip Joint. J. Bone Jt. Surg. 99, 1373–1381. https://doi.org/10.2106/JBJS.16.00415

Beaulé, P.E., Zaragoza, E., Motamedi, K., Copelan, N., Dorey, F.J., 2005. Three-dimensional computed tomography of the hip in the assessment of femoroacetabular impingement. J. Orthop. Res. 23, 1286–1292. https://doi.org/10.1016/j.orthres.2005.03.011

Beck, M., Kalhor, M., Leuning, M., Ganz, R., 2005. Hip Morphology Influences the Pattern of Damage to the Acetabular Cartilage: Femoroacetabular Impingement as a Cause of Early Osteoarthritis of the Hip. J. Bone Jt. Surg. - Br. Vol. 87–B, 1012–1018. https://doi.org/10.1302/0301-620X.87B7.15203

Bedi, A., Zaltz, I., De La Torre, K., Kelly, B.T., 2011. Radiographic Comparison of Surgical Hip Dislocation and Hip Arthroscopy for Treatment of Cam Deformity in Femoroacetabular Impingement. Am. J. Sports Med. 39 Suppl, 20S–8S. https://doi.org/10.1177/0363546511412734

Bennell, K.L., Spiers, L., Takla, A., O'Donnell, J., Kasza, J., Hunter, D.J., Hinman, R.S., 2017. Efficacy of Adding a Physiotherapy Rehabilitation Programme to Arthroscopic Management of Femoroacetabular Impingement Syndrome: A Randomised Controlled Trial (FAIR). BMJ Open 7. https://doi.org/10.1136/bmjopen-2016-014658

Botser, I.B., Smith, T.W., Nasser, R., Domb, B.G., 2011. Open Surgical Dislocation Versus Arthroscopy for Femoroacetabular Impingement: A Comparison of Clinical Outcomes. Arthrosc. - J. Arthrosc. Relat. Surg. 27, 270–278. https://doi.org/10.1016/j.arthro.2010.11.008

Brisson, N., Lamontagne, M., Kennedy, M.J., Beaulé, P.E., 2013. The Effects of Cam Femoroacetabular Impingement Corrective Surgery on Lower-Extremity Gait Biomechanics. Gait Posture 37, 258–263. https://doi.org/10.1016/j.gaitpost.2012.07.016

Buchanan, T.S., Lloyd, D.G., Manal, K., Besier, T.F., 2004. Neuromusculoskeletal Modeling: Estimation of Muscle Forces and Joint Moments and Movements from Measurements of Neural Command. J. Appl. Biomech. 20, 367–395. https://doi.org/10.1016/j.bbi.2008.05.010

Carsen, S., Moroz, P.J., Rakhra, K., Ward, L.M., Dunlap, H., Hay, J.A., Willis, R.B., Beaulé, P.E., 2014. The Otto Aufranc Award. On the Etiology of the Cam Deformity: A Cross-Sectional Pediatric MRI Study. Clin. Orthop. Relat. Res. 472, 430–436. https://doi.org/10.1007/s11999-013-2990-y

Casartelli, N.C., Maffiuletti, N.A., Item-Glatthorn, J.F., Staehli, S., Bizzini, M., Impellizzeri, F.M., Leunig, M., 2011. Hip Muscle Weakness in Patients with Symptomatic Femoroacetabular Impingement. Osteoarthr. Cartil. 19, 816–821. https://doi.org/10.1016/j.joca.2011.04.001

Catelli, D.S., Wesseling, M., Jonkers, I., Lamontagne, M., 2018. A Musculoskeletal Model Customized for Squatting Task. Comp. Methods Biomech. Biomed. Engin. 6:1-4. https://doi.org/10.1080/10255842.2018.1523396

Chakraverty, J.K., Sullivan, C., Gan, C., Narayanaswamy, S., Kamath, S., 2013. Cam and Pincer Femoroacetabular Impingement: CT Findings of Features Resembling Femoroacetabular Impingement in a Young Population without Symptoms. Am. J. Roentgenol. 200, 389–395. https://doi.org/10.2214/AJR.12.8546 de Sa, D., Horner, N.S., MacDonald, A., Simunovic, N., Slobogean, G., Philippon, M.J., Belzile, E.L., Karlsson, J., Ayeni, O.R., 2016. Evaluating Healthcare Resource Utilization and Outcomes for Surgical Hip Dislocation and Hip Arthroscopy for Femoroacetabular Impingement. Knee Surgery, Sport. Traumatol. Arthrosc. https://doi.org/10.1007/s00167-015-3722-5

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. https://doi.org/10.1109/TBME.2007.901024

Diamond, L.E., Bennell, K.L., Wrigley, T. V., Hinman, R.S., Hall, M., O'Donnell, J., Hodges, P.W., 2018. Trunk, Pelvis and Hip Biomechanics in Individuals with Femoroacetabular Impingement Syndrome: Strategies for Step Ascent. Gait Posture. https://doi.org/10.1016/j.gaitpost.2018.01.005

Diamond, L.E., Bennell, K.L., Wrigley, T. V., Hinman, R.S., O'Donnell, J., Hodges, P.W., 2017a. Squatting Biomechanics in Individuals with Symptomatic Femoroacetabular Impingement. Med. Sci. Sports Exerc. 49, 1520–1529. https://doi.org/10.1249/MSS.0000000000001282

Diamond, L.E., Van den Hoorn, W., Bennell, K.L., Wrigley, T. V., Hinman, R.S., O'Donnell, J., Hodges, P.W., 2017b. Coordination of Deep Hip Muscle Activity Is Altered in Symptomatic Femoroacetabular Impingement. J. Orthop. Res. 35, 1494–1504. https://doi.org/10.1002/jor.23391

Diamond, L.E., Wrigley, T. V., Bennell, K.L., Hinman, R.S., O'Donnell, J., Hodges, P.W., 2016. Hip Joint Biomechanics during Gait in People with and without Symptomatic Femoroacetabular Impingement. Gait Posture 43, 198–203. https://doi.org/10.1016/j.gaitpost.2015.09.023

Ganz, R., Gill, T., Gautier, E., Ganz, K., Krügel, N., Berlemann, U., 2001. Surgical Dislocation of the Adult Hip: A Technique with Full Access to the Femoral Head and Acetabulum without the Risk of Avascular Necrosis. J. Bone Jt. Surg. 1119–24. https://doi.org/10.1007/978-1-4471-5451-8_144

Ganz, R., Leunig, M., Leunig-Ganz, K., Harris, W.H., 2008. The Etiology of Osteoarthritis of the Hip: an Integrated Mechanical Concept. Clin Orthop Relat Res 466, 264–272. https://doi.org/10.1007/s11999-007-0060-z

Ganz, R., Parvizi, J., Beck, M., Leunig, M., Nötzli, H., Siebenrock, K.A., 2003. Femoroacetabular Impingement: a Cause for Osteoarthritis of the Hip. Clin. Orthop. Relat. Res. 112–120. https://doi.org/10.1097/01.blo.0000096804.78689.c2

Hunt, M.A., Gunether, J.R., Gilbart, M.K., 2013. Kinematic and Kinetic Differences during Walking in Patients with and without Symptomatic Femoroacetabular Impingement. Clin. Biomech. 28. https://doi.org/10.1016/j.clinbiomech.2013.05.002

Joseph, R., Pan, X., Cenkus, K., Brown, L., Ellis, T., Di Stasi, S., 2016. Sex Differences in Self-Reported Hip Function Up to 2 Years After Arthroscopic Surgery for Femoroacetabular Impingement. Am J Sport. Med 44, 54–59. https://doi.org/10.1177/0363546515610535

Kapron, A.L., Aoki, S.K., Peters, C.L., Anderson, A.E., 2014. Subject-specific Patterns of Femur-labrum Contact are Complex and Vary in Asymptomatic Hips and Hips with Femoroacetabular Impingement. Clin. Orthop. Relat. Res. 472, 3912–3922. https://doi.org/10.1007/s11999-014-3919-9

Kennedy, M.J., Lamontagne, M., Beaulé, P.E., 2009. Femoroacetabular Impingement Alters Hip and Pelvic Biomechanics during Gait: Walking Biomechanics of FAI. Gait Posture 30, 41–44. https://doi.org/10.1016/j.gaitpost.2009.02.008

Khanna, V., Caragianis, A., Diprimio, G., Rakhra, K., Beaulé, P.E., 2014. Incidence of Hip Pain in a Prospective Cohort of Asymptomatic Volunteers: Is the Cam Deformity a Risk Factor for Hip Pain? Am. J. Sports Med. 42, 793–7. https://doi.org/10.1177/0363546513518417

Kierkegaard, S., Mechlenburg, I., Lund, B., Søballe, K., Dalgas, U., 2017. Impaired Hip Muscle Strength in Patients with Femoroacetabular Impingement Syndrome. J. Sci. Med. Sport 20, 1062–1067. https://doi.org/10.1016/j.jsams.2017.05.008

Lai, A.K.M., Arnold, A.S., Wakeling, J.M., 2017. Why are Antagonist Muscles Co-activated in My Simulation? A Musculoskeletal Model for Analysing Human Locomotor Tasks. Ann. Biomed. Eng. 45, 2762–74. https://doi.org/10.1007/s10439-017-1920-7

Lamontagne, M., Brisson, N., Kennedy, M.J., Beaulé, P.E., 2011. Preoperative and Postoperative Lower-Extremity Joint and Pelvic Kinematics during Maximal Squatting of Patients with Cam Femoroacetabular Impingement. J. Bone Jt. Surg. 93, 40–5. https://doi.org/10.2106/JBJS.J.01809

Lamontagne, M., Kennedy, M.J., Beaulé, P.E., 2009. The Effect of Cam FAI on Hip and Pelvic Motion during Maximum Squat. Clin Orthop Relat Res 467, 645–650. https://doi.org/10.1007/s11999-008-0620-x

Larson, C.M., Giveans, M.R., 2008. Arthroscopic Management of Femoroacetabular Impingement: Early Outcomes Measures. Arthrosc. - J. Arthrosc. Relat. Surg. 24, 540–546. https://doi.org/10.1016/j.arthro.2007.11.007

Lewis, C.L., Sahrmann, S.A., Moran, D.W., 2009. Effect of Position and Alteration in Synergist Muscle Force Contribution on Hip Forces When Performing Hip Strengthening Exercises. Clin. Biomech. 24, 35–42. https://doi.org/10.1016/j.clinbiomech.2008.09.006

Malagelada, F., Del Carmen, V.A., Barke, S.J., Guirao Cano, L., Pleguezuelos Cobo, E., 2015. The Anterior Mini-Open Approach for Femeroacetabular Impingement: Gait and Functional Assessment at One Year Post-Surgery. Ann. Phys. Rehabil. Med. 58, 60–65. https://doi.org/10.1016/j.rehab.2014.09.013

Mantoan, A., Pizzolato, C., Sartori, M., Sawacha, Z., Cobelli, C., Reggiani, M., 2015. MOtoNMS: A MATLAB Toolbox to Process Motion Data for Neuromusculoskeletal Modeling and Simulation. Source Code Biol. Med. 10, 12. https://doi.org/10.1186/s13029-015-0044-4

Mantovani, G., Lamontagne, M., 2016. How Different Marker Sets Affect Joint Angles in Inverse Kinematics Framework. J. Biomech. Eng. 139, 044503. https://doi.org/10.1115/1.4034708

Murphy, S., Tannast, M., Kim, Y.J., Buly, R., Millis, M.B., 2004. Debridement of the Adult Hip for Femoroacetabular Impingement: Indications and Preliminary Clinical Results. Clin. Orthop. Relat. Res. 178–181. https://doi.org/10.1097/01.blo.0000150307.75238.b9

Ng, K.C.G., Mantovani, G., Modenese, L., Beaulé, P.E., Lamontagne, M., 2018. Altered Walking and Muscle Patterns Reduce Hip Contact Forces in Individuals With Symptomatic Cam Femoroacetabular Impingement. Am. J. Sports Med. 46, 1–9. https://doi.org/10.1177/0363546518787518

Nussbaumer, S., Leunig, M., Glatthorn, J.F., Stauffacher, S., Gerber, H., Maffiuletti, N.A., 2010. Validity and Test-Retest Reliability of Manual Goniometers for Measuring Passive Hip Range of Motion in Femoroacetabular Impingement Patients. BMC Musculoskelet. Disord. 11, 194. https://doi.org/10.1186/1471-2474-11-194

Pataky, T.C., 2010. Generalized N-Dimensional Biomechanical Field Analysis Using Statistical Parametric Mapping. J. Biomech. 43, 1976–1982. https://doi.org/10.1016/j.jbiomech.2010.03.008

Philippon, M., Briggs, K., Yen, Y.-M., Kuppersmith, D., 2009. Outcomes Following Hip Arthroscopy for Femoroacetabular Impingement with Associated Chondrolabral Dysfunction: Minimum Two-Year Follow-Up. J Bone Jt. Surg [Br] 91, 16–23. https://doi.org/10.1302/0301-620X.91B1

Rajagopal, A., Dembia, C.L., Demers, M.S., Delp, D.D., Hicks, J.L., Delp, S.L., 2016. Full Body Musculoskeletal Model for Muscle-Driven Simulation of Human Gait. IEEE Trans. Biomed. Eng. 63, 2068–79. https://doi.org/10.1109/TBME.2016.2586891

Ribas-Fernandez, M., Marin-Peña, O.R., Regenbrecht, B., De la Torre, B., Vilarrubias, J.M., 2007. Hip Osteoplasty by an Anterior Minimally Invasive Approach for Active Patients with Femoroacetabular Impingement. Hip Int. 17, 91–98.

Ross, J.R., Nepple, J.J., Philippon, M.J., Kelly, B.T., Larson, C.M., Bedi, A., 2014. Effect of Changes in Pelvic Tilt on Range of Motion to Impingement and Radiographic Parameters of Acetabular Morphologic Characteristics. Am. J. Sports Med. https://doi.org/10.1177/0363546514541229

Rylander, J., Shu, B., Favre, J., Safran, M., Andriacchi, T., 2013. Functional Testing Provides Unique Insights into the Pathomechanics of Femoroacetabular Impingement and an Objective Basis for Evaluating Treatment Outcome. J. Orthop. Res. 31, 1461–1468. https://doi.org/10.1002/jor.22375

Rylander, J.H., 2014. CORR Insights: Subject-specific Patterns of Femur-labrum Contact are Complex and Vary in Asymptomatic Hips and Hips with Femoroacetabular Impingement. Clin. Orthop. Relat. Res. 472, 3912–3922. https://doi.org/10.1007/s11999-014-3919-9

Rylander, J.H., Shu, B., Andriacchi, T.P., Safran, M.R., 2011. Preoperative and Postoperative Sagittal Plane Hip Kinematics in Patients with Femoroacetabular Impingement during Level Walking. Am. J. Sports Med. 39, 36S–42S. https://doi.org/10.1177/0363546511413993

Wesseling, M., Derikx, L.C., De Groote, F., Bartels, W., Meyer, C., Verdonschot, N., Jonkers, I., 2015. Muscle Optimization Techniques Impact the Magnitude of Calculated Hip Joint Contact Forces. J. Orthop. Res. 33, 430–438. https://doi.org/10.1002/jor.22769

V CLOSING

10 Discussion

Summary | Clinical Implications | Limitations | Future Directions | Conclusion

10.1. Summary

With the goal to examine how the THA and the preservation surgery for cam deformity affect the hip biomechanics outcomes, this research program was structured to investigate the hip kinematics, muscle and hip contact forces after surgical intervention using motion capture and computer simulation modelling methods. Specific gaps in the literature were pointed out in the first two chapters of this thesis, and three main areas for research development were highlighted:

1. Examine the muscle strength and lower limb mobility in asymptomatic individuals with a cam-type deformity that may differentiate them from the symptomatic patients;
2. Enhance the understanding of the postoperative effect of both hip surgical procedures, hip preservation and THA on hip biomechanics during a high ROM task such as the deep squat;
3. Simulate the dynamic muscle forces and HCF in pre- and postoperative FAI in different ADLs.

Most of the previous gait analysis studies evaluating postoperative patients who underwent a THA or hip preservation surgeries have been performed during level walking, with limited biomechanical outcomes being assessed. Therefore, the basis of this doctoral study was to evaluate kinematics, kinetics and muscle activation parameters of pre- and postoperative patients while performing a deep squat task, and also adapt and implement a deep squat analysis in computational modelling to evaluate muscle and hip contact forces in a postoperative patients' cohort.

First, the *Asymptomatic Individuals* (Chapter 4) pre-defined a common-ground for preoperative patients, while comparing their kinematic and muscular activity performance during the squat, as well as their muscle strength, with not only healthy control individuals but also their asymptomatic peers (with the cam-deformity). The cohort's asymptomatic participants (FAD) had significantly greater hip extensor strength compared to the FAI and CTRL groups, which may have allowed them to achieve greater pelvic mobility and squat as deep as the CTRL group. Thus, this is raising the hypothesis if a conservative intervention that aims to improve hip extensor strength and pelvic mobility could reduce symptoms for patients with cam FAI and potentially reduce the risk of hip arthroplasty.

Secondly, the studies that assessed *Squatting Motion Analysis after Hip Surgery* (Chapters 5 and 6) assessed joint kinematics, muscle activity and muscle strength of postoperative patients suffering from both cam FAI and OA. While during the follow-up for the hip preservation surgery the patients showed increased pelvic ROM during the squat, weakness associated with hip flexion and hip flexion with abduction were associated with postoperative alterations in the muscle activity. On the other hand, the THA postoperative follow-up analyses focused on comparing two types of prostheses. Our findings showed that the DM patients reached an anterior pelvic tilt similar to the CTRL during the dynamic parts of the squat; however, without returning its neutral position at the bottom of the squat. Moreover, while none of the prosthesis was able to reach the same hip flexion, or improve the squat depth to the level of the CTRLs, the SB seemed to have caused an excessive hip abduction during the trials. Both manuscripts culminated with the recommendation that rehabilitation programs should focus on increasing pelvis and hip muscles flexibility and strength.

Thirdly, the *Musculoskeletal Modelling: from Gait to Squat* (Chapters 7, 8 and 9) incorporated a more robust analysis of the cam FAI participants during their postoperative follow-up to assess muscle forces and HCF. This section included the challenge of adapting a generic full-body MSKM, that are usually developed to perform simulation tasks with limited ROM at the hip and knee joints (e.g. gait and running), to simulate a deep squat task (Chapter 8). The optimized full-body MSKM aimed to represent realistic muscle moment arms during the high joint flexion, thus avoiding that the MTU vectors crossed the bony structures. The gait simulation demonstrated that FAI patients enhance medial-lateral hip loading postoperatively. Also, the reduced dynamic forces of the biceps femoris, semimembranosus and rectus femoris (muscles associated with the sagittal aspect of the gait) may suggest a less compensatory strategy to stabilize the hip. Furthermore, during the deep squat task, a closed kinetic chain motion, postoperative FAI patients showed higher anterior pelvic tilt postoperatively once the cam-deformity was no longer present. Increased semimembranosus force was speculated as a cause to higher vertical HCF and total magnitude during squat ascent.

10.2. Clinical Implications

10.2.1. Patient Reported Outcome Measures

There is no doubt that the surgical interventions on both cam FAI and OA patients caused positive benefits regarding the reduction of pain and general improvement of their reported outcome measures (Tables 5.1 and 6.1). Postoperative HOOS reached higher scores than the preoperative in all categories, for the FAI patients (Chapter 5) and both THA prosthesis designs (Chapter 6). All postoperative HOOS scores were between 83 to 93 for FAI and THA participants, but the 'Quality of Life' sub-score, which have demonstrated values still not comparable with the CTRL individuals (Table 5.1), showing that although an improved was reached, these patients were still aware of their hip problem, and that might still slightly impact their overall quality of life.

However, very similar mechanics were shown when comparing the postoperative outcomes to their preoperative values. Still, the evaluation of the small changes that affected the postoperative patients is crucial to quantify improvements and propose future research directions.

10.2.2. Muscle Strength Analyses

The muscle strength analyses of the *Asymptomatic Individuals* (Chapter 4) showed that patients with FAI not only had weaker hip flexors than the other groups, but also that asymptomatic participants with FAD demonstrated stronger hip extensors when compared with the symptomatic patients. This strength difference was indicated in the EMG analyses, as the FAI group showed higher muscular activation during the squat task compared with the other groups. Previous study compared the preoperative FAI to CTRL participants only and reported no strength differences in hip extensors

(Casartelli et al., 2011). However, when we classified the asymptomatic participants who presented the cam-deformity (i.e. the FAD) from healthy CTRL group, we were able to demonstrate that the asymptomatic may have developed a muscle mechanism to avoid the impingement. The greater muscle strength of the muscle groups responsible for the sagittal aspect of the motion (i.e. hip flexors and extensors) was associated with the greater pelvic ROM reached by the FAD group during the deep squat, which might have allowed them to squat as deep as the CTRL group. This study provided a greater understanding of the role of the muscles and soft tissues around the hip that contribute to the possible development of symptomatic FAI and brought up hypotheses for future studies about conservative treatment for FAI including hip muscle strengthening and pelvis mobility training.

Muscle strength was also evaluated after hip preservation for cam FAI (Chapter 5) and showed that postoperative patients were weaker relatively from the CTRL group for hip flexors and hip flexors-with-abduction muscle groups during the isometric testing. Although these differences were not pointed out on this cohort preoperatively, this finding is still similar as the study from Casartelli et al. (2011) that showed that preoperative FAI patients presented muscle weakness for the hip flexor and hip abduction muscle groups. The muscle weakness also led to a postoperative muscle imbalance regarding the ratio between hip flexion and extension strengths, when compared to the CTRL participants. As muscle weakness in OA individuals can be an indicator of the progression of the disease (Brandt, 2003; Casartelli et al., 2011), the assumption that FAI could potentially lead to hip OA (Ganz et al., 2003; Leunig et al., 2009; Leunig and Ganz, 2005) is asserted. Therefore, the findings of this study supported the introduction of hip muscle strength assessment in routine clinical examinations to help diagnose FAI and as an indicator for aftercare physiotherapy (Casartelli et al., 2011; Kelly et al., 2003).

10.2.3. Pelvic Mobility

The dynamic pelvic tilt was one of the principal variables illustrated in the squat studies in this thesis. Its mobility, 1) from neutral while standing to 2) anterior tilted during the descending, 3) back to neutral at the bottom of the squat, 3) again anterior tilted while raising up and 4) back to neutral with terminal standing, was accomplished by the younger and the older CTRL participants. However, that was not necessarily the case of our symptomatic cohort.

In the THA study, it was possible to see that the postoperative patients did not fully reach the same pelvic ROM, or reach squat depth levels, as the CTRL participants during the deep squat task. The patients who used SB prosthesis were unable to reach an anterior tilt during the descending in the same levels that the CTRL participants were, but on the other hand, they were able to return their pelvis to a neutral position at the bottom of the squat. Contrarily, the patients using the DM prosthesis design moved their pelvis anteriorly as far as the CTRL group during both phases of the squat but were unable to return the pelvis position to neutral at the deepest part of the squat. Two hypotheses were then speculated: 1) that when squatting, the coefficient of friction being higher in the DM results is preventing the pelvis to decrease its tilt passively until enough muscles can be engaged in the later phases of the ascent for the pelvic to resume its normal motion; and 2) the DM design offers a greater mobility (Guyen et al., 2007) that the return to neutral position was not mechanically necessary. Because the pelvis is unable to adjust accordingly, this could explain some of the psoas tendon irritation/groin pain noted with DM implants (Lachiewicz and Watters, 2012) as the iliopsoas tendon unit is under tension during a greater portion of the squat cycle and more likely to catch on larger polyethylene bearing (Varadarajan et al., 2016). The pelvic improvement reached by the SB patients at the bottom of the squat may have helped them to improve their squat depth (although still not reaching CTRL levels), but the prosthesis was also likely responsible for causing an excessive hip abduction during the dynamic task. Although avoiding dislocation (Philippot et al., 2009; Tarasevicius et al., 2010), the lack of pelvic mobility in the DM design combined with the poorer functional scores should caution clinicians to use this implant design in active patients.

The study with the asymptomatic individuals (Chapter 4) showed that while the preoperative patients demonstrated limited dynamic pelvis ROM during both phases of the squat when compared to the CTRL group (Figure 4.2), the asymptomatic FAD participants had similar ROM compared to the CTRLs. The anteroposterior pelvic mobility (tilt) and its association with impingement have shown that dynamic anterior pelvic tilt is predictive to the femoroacetabular impingement to occur, whereas dynamic posterior pelvic tilt would avoid it (Ross et al., 2014). The fact that the FAI patients in our study did not go back to a neutral tilt at the bottom of the squat and remained more anteriorly tilted while standing might be associated with the findings by Ross et al. (2014). The strength advantage of the hip extensors in the FAD group (stronger in comparison with both FAI and CTRL groups) might be associated with its pelvic mobility capability, especially while returning the pelvis to the neutral position at the deepest phase of the squat which might reduce the risk of impingement, and raise the question: could a conservative treatment, involving hip extensor strengthening and pelvic mobility, for people who already developed FAI symptomatology work?

After cam FAI hip preservation surgery, postoperative patients were still not able to reach the squat depth demonstrated for the CTRL group (Chapter 5) or improve their squat depth (Chapter 9). An overall postoperative pelvic ROM improvement was observed during both phases of the squat (Table 5.3). Lamontagne et al. (2011b) however, have not found pelvic tilt improvements in his cohort, their postoperative patients were likely reaching higher posterior pelvic tilt at the bottom of the squat than they were preoperative. However, that was not the case in our cohort (Chapter 9), where the postoperative patients had their pelvis more anteriorly tilted than they were preoperatively, and no SPM differences were found at the deepest part of the squat. The differences in the pelvic outcomes from Lamontagne et al. (2011b) and our study (Chapter 9) could be justified by the cohort, in which theirs had a younger mixed composition (i.e. different pelvis anatomy) and ours consisted exclusively of male participants. Ultimately, considering that the anterosuperior location of the cam-deformity in the femoral head (Chakraverty et al., 2013) – consistent with the location of chondrolabral damage (Beaulé et al., 2005; Beck et al., 2005; Ganz et al., 2008) – is susceptible to impinge when the pelvis is anteriorly tilted (Ross et al., 2014), our FAI patients were, possibly, in a more anteriorly tilted position natively combined with a smaller neck-shaft angle (Ng et al., 2015), which would have caused the cam formation during bone maturation (Carsen et al., 2014). However, the development of the cam and the further onset of the symptomatology may have caused the patients to adopt a more posteriorly tilted pelvic position, as well as limiting their hip mobility as a protected mechanism in order to reduce HCF (Ng et al., 2018b). Postoperatively (i.e. once the cam was removed), the patients were able to return to the more anteriorly (innate) tilted position as the impingement was no longer affecting their kinematics. Hence, the hypothesis formulated in the last paragraph is reinforced: could pelvic mobility be a key factor in the FAI symptomatology initiation? Considering that symptomatic and asymptomatic individuals with the cam-deformity were likely very physically active during their childhood/adolescence (Carsen et al., 2014; Chaudhry and Ayeni, 2014), could a muscle imbalance be the initiation for the symptoms? Would strong hip extensors, creating a better muscle extensor/flexor balance, be able to avoid it?

Likewise, individuals with symptomatic cam FAI have greater pelvic incidence when compared with their asymptomatic peers (Grammatopoulos et al., 2018; Ng et al., 2018a) with a greater amount of superior-posterior coverage at the contact area between the cam morphology and the acetabulum. The pelvic incidence has been identified as an anatomic predictor of hip symptomatology in cam FAI (Ng et al., 2018a).

Perhaps a conservative treatment in cam FAI patients that involve hip flexors and extensors strengthening, associated with flexibility and dynamic mobility of the pelvis, might be a possible indication for those patients who are still borderline in clinical signs and diagnostic imaging.

10.2.4. Dynamic Muscle and Hip Contact Forces

Two studies with the postoperative FAI patients was developed for this thesis in order to simulate muscle forces and HCF, the first one during the level walking task (Chapter 7) and the second during the deep squat task (Chapter 9). An additional methodological paper (Chapter 8) was developed to adapt a generic MSKM in order to control the moment arms of the MTU during high hip and knee flexions to avoid MTUs to cross the bones during a deep squat task.

Usually designed to evaluate walking and running tasks, most MSKMs allow limited ROM. The MSKM optimized in Chapter 8 changed the parameters of the original wrapping surfaces (Lai et al., 2017; Rajagopal et al., 2016) and implemented new wrapping surfaces to prevent the MTU from penetrating the bones while providing realistic muscle moment arms for large hip and knee ROM. This model can be used not only to simulate the deep squat task but likely another range of sportive tasks such as sprint block start, race-walk, long jump take-off, etc.

In chapter 7, the muscle forces and the HCF were simulated during the stance phase of level walking. It was found postoperative reduced dynamic muscle forces of the biceps femoris, semimembranosus and rectus femoris, as well as a noticeable medialized HCF during the entire stance phase of the gait. The medialized forces were associated with the postoperative enhancement of the medial-lateral hip loading, which might have affected the reduced forces of the muscles associated with the sagittal aspect of the gait, suggesting a less compensatory strategy to assist in hip stability. Similarly, Ng et al. (2018b) have shown a more medialized transverse HCF component in the healthy participants when compared with preoperative cam FAI during contralateral foot-strike. They also reported limited hip mobility in preoperative FAI patients that was associated with the reduced muscle force pattern of the iliopsoas muscle complex, causing a reduction of the anterior, superior, and medial hip contact forces. The present study reported the forces of the hip flexors and extensors to assess the postoperative effect. However, as gait parameters (i.e. walking speed, stride length), joint kinematics (i.e. hip extension, hip and pelvic range of motion), and spinopelvic anatomy (i.e. pelvic tilt and incidence) did not change after surgery, there were marginal effects towards the activation of the primary hip flexors (Ng et al., 2018b, 2018a).

The study that simulated the muscle and HCF in the FAI patients during the deep squat task (Chapter 9) showed higher semimembranosus force and vertical HCF component (that also affected the HCF total magnitude) during the ascending phase of the squat. Increased activation of the other medial hamstring, the semitendinosus, was already explored postoperatively in the study presented in Chapter 5 (Figure 5.2). The postoperative higher force of the semimembranosus seemed to have affected the vertical component of the HCF (Wesseling et al., 2015), providing increased superior HCF in the vertical component of the vector and its resultant magnitude force at the same phase of the squat.

10.3. Limitations

10.3.1. Participants

Although it is recognized that biomechanical studies with a medium- or long-term follow-ups will likely have smaller patient cohorts, we must acknowledge that this general limitation affected both studies of this thesis. Difficulties recruiting postoperative patients for the follow-up analyses were the main complication to not reach the desired number of fifteen (n = 15) for the postoperative groups. Although technical difficulties (e.g. missing EMG data, participant refusal/inability to perform the squat tasks) also affected the initial plans.

Additionally, the FAI studies that involved follow-up analysis (Chapters 5, 7 and 9) consisted of a male-only population, as the cam deformity is statistically prevalent in men (Gosvig et al., 2008). The inclusion of female patients could introduce variances in both the anatomical model and the squat kinematics. For that reason, a complementary study to examine postoperative sex-differences could elucidate whether the muscle forces and HCF of FAI are specific to sex.

In respect to the study comparing the symptomatic to the asymptomatic FAI participants (Chapter 4), there was an age difference between both groups with our symptomatic group being slightly older, and one could argue that the FAD participants could be in the process of starting FAI symptomatology, which was not followed-up until now. However, it is important to highlight the fact that there were a few older symptomatic patients who performed deeper squats and wider pelvic motions.

Regarding the postoperative analyses, while for the THA study the follow-up assessment was done around seven months, the FAI patients returned after two years the hip preservation surgery. Therefore, the conclusions of these studies must be considered as short- and medium-term follow-ups, respectively, since ongoing strengthening and training may further improve postoperative mechanics. It would be feasible as a longitudinal study to conduct other follow-ups, an effort to examine if there will be further improvements to gait and squat mechanics in these populations.

10.3.2. Surgical Approach

All patients who underwent THA were accessed through the direct anterior approach using a positioning table (Matta et al., 2005). Therefore, some caution must be taken into consideration while comparing the outcomes of this study with others whose choice of surgical approach was different, or which the surgeon was still in a learning curve process.

Surgical correction of cam deformity in the FAI patients included in all affected hips chondrosteoplasty as well as labral-chondral debridement. However the choice of approach was different among the patients: four underwent a surgical hip dislocation with a direct anterior incision (which only three of them were included for the study on Chapter 9), six underwent a hip arthroscopy combined with mini-anterior arthrotomy (Barton et al., 2009) and one patient was treated purely arthroscopically using intraoperative fluoroscopy. And although recent studies have found comparable functional (Bedi et al., 2011b) and patient-reported outcomes (De Sa et al., 2016) between hip dislocation and arthroscopy approaches, and a mixed linear model post hoc analysis comparing the outputs of Chapter 9 regarding the approaches have not shown significant differences between the different approaches; this is a limitation that must be highlighted.

All surgeries included in this study were performed by the same senior surgeon (P.E.B.) at The Ottawa Hospital, Ottawa. Canada.

10.3.3. Rehabilitation Program

Surgical aftercare is crucial after an orthopaedic surgical intervention and is advised accordingly by the surgical staff. Recommendations are that the patients should do isometric exercises for the glutei and quadriceps on their own for the first four weeks following the surgery. Then, it is recommended that they seek physiotherapist care for another four to six weeks (two to three times a week) where they will be doing active ROM exercises, as well as muscle strengthening against gravity, muscle resistance and gait training.

This study did not provide controlled aftercare to the participants, and due to the limitations of the health care system, we could not guarantee that all the patients underwent the physiotherapy sections for the recommended number of sections or duration. Therefore, the variability within the aftercare program might have also created some variability in our results.

10.3.4. Modelling Parameters

Musculoskeletal models are simplified representations of the MSK system and its motion performance, with many phenomena often being ignored or simplified (Hicks et al., 2015). As models consist of rigid body structures with the connective tissues often neglected, the joint kinematics are simplified, and some degrees of freedom can be absent. The line of action of a muscle is limited to one or more segments of MTU, and multiple MTUs are unable to attach to the same tendon as we find in a real situation (e.g. triceps surae), but the MSKMs consider separate tendons for each MTU. Also several phenomena are ignored in the Hill-type muscle models that are used in OpenSim (Hicks et al., 2015), as for the variable-gearing pennate muscles (Azizi et al., 2008), force enhancement (Herzog et al., 2006), short-range muscle stiffness (Rach and Westbury, 1974) and the muscle dependency variations in force-length curve at submaximal force (Rassier et al., 1999).

Muscle forces simulations were calculated through static optimization that considers each time-frame separately, not factoring in kinematics and muscle states. Static optimization, however, has a cheaper computational cost and it is a more time-efficient approach than other methods such as EMG-driven (Sartori et al., 2012) or hybrid frameworks (Sharif Shourijeh et al., 2016). Still, the static optimization assumes an optimal neuromuscular strategy, while minimizing the muscles' activations to the square power. The verification/validation of our static optimization outputs consisted of correlating the EMG linear envelopes to the muscle activation output of the respective trials. Although this is a common and accepted practice in the MSK modelling community (Hicks et al., 2015; Lund et al., 2012), EMG excitations and the muscle activations (from OpenSim) are not the same phenomena. Static optimization does not consider muscle and kinematic states and assumes that activation force production occurs in the same time point, whereas processes such as the electromechanical delay disturb the relationship between EMG and force production, which is the reason the predicted muscle activations were shifted back over the excitations during the verification procedure (appendix, table A.1).

Another limitation is that the subject-specific hip morphology was not incorporated into the model. The use of individual cam-deformity in the simulations would probably benefit the perceptiveness of this study, especially regarding the kinematics during the deep squat test, as it might have caused an impingement preoperatively.

10.4. Future Directions

Based on the findings of this thesis, future efforts in conservative care, long-term postoperative analysis, characteristics of FAI etiology and other complementary analysis should be investigated.

A recent study compared hip arthroscopy to conservative care treatment of FAI in a large British population (n = 328) and showed that hip arthroscopy led to greater improvement than the conservative after 12 months of treatment (Griffin et al., 2018). However, the indications of this study, especially regarding the knowledge enclosing the asymptomatic population (FAD), may provide us with a better understanding of a personalized hip therapy. Can pelvis flexibility training cause an increase in pelvic mobility dynamically in cam FAI patients? Moreover, can hip extensors strengthening, with greater pelvic mobility help diminish cam FAI symptomatology and OA progression?

Again, our asymptomatic cohort was slightly younger than our preoperative cam FAI patients. A longitudinal study with this type of population is crucial to better understand if they will eventually become symptomatic, and what type of functions could kickoff the symptomatology; or if they have some intrinsic strategies to avoid pain indefinitely.

Also, future studies that could incorporate the contact forces to examine the effects of surgical correction for cam FAI on mechanical hip joint loading, with the use of finite element analysis (Ng et al., 2016a), would help to determine postoperative hip joint stresses.

10.5. Conclusion

This doctoral thesis was comprised of six studies that investigated the motion changes after THA and hip preservation surgeries (for cam FAI) through motion capture and MSK modelling approaches, with the ultimate goal of understanding the changes in muscle performance and HCF during the deep squat task.

Although showing better patient-reported outcomes, the postoperative patients still present very similar mechanics compared to their preoperative values, which can be associated with long-term effect adaptations (e.g. hip flexor weakness, anteriorized pelvic tilt) or perhaps the lack of control of the rehabilitation program.

The evaluation of the pelvic joint has demonstrated to be as crucial as evaluating the hip joint. Pelvic mobility plays an important role in the symptomatology of individuals with the cam deformity, and it might be associated with the strength of the hip extensor muscles. After hip preservation surgery, FAI patients reached pelvic ROM similar to the CTRL, the same was not observed in patients who underwent THA.

One methodological study was able to optimize a generic MSKM to perform simulations of tasks that require very high hip and knee ROM. Now that this optimized model has been developed, more investigations could be carried out, with a broader type of tasks been performed in both clinical and sports research.

This, to our knowledge, represents the first study done on muscle forces and HCF estimation on the two-year postoperative FAI population. And although the changes presented for gait and squat tasks along with the muscle strength did not report higher outcomes as the PROMs; the overall benefit taken from the present research work is the better understanding of the pelvic mobility and hip muscle forces in hip diseases. In addition to the possibility of improving biomechanical assessment of postoperative patients using *in silico* models in order to quantify surgical effectiveness and support clinicians in making subject-specific case decisions. The contributions also lay on the assertion of helping us to formulate future research directions in biomechanics applied to orthopaedics.

11 References

Adam, P., Farizon, F., Fessy, M.H., 2014. Dual mobility retentive acetabular liners and wear: Surface analysis of 40 retrieved polyethylene implants. Orthop. Traumatol. Surg. Res. 100, 85–91. https://doi.org/10.1016/j.otsr.2013.12.011

Agricola, R., Heijboer, M.P., Bierma-Zeinstra, S.M.A., Verhaar, J.A.N., Weinans, H., Waarsing, J.H., 2013a. Cam Impingement Causes Osteoarthritis of the Hip: a Nationwide Prospective Cohort Study. Ann. Rheum. Dis. 72, 918–923. https://doi.org/10.1136/annrheumdis-2012-201643

Agricola, R., Heijboer, M.P., Ginai, A.Z., Roels, P., Zadpoor, A.A., Verhaar, J.A.N., Weinans, H., Waarsing, J.H., 2014. A Cam Deformity Is Gradually Acquired During Skeletal Maturation in Adolescent and Young Male Soccer Players. Am. J. Sports Med. 42, 798–806. https://doi.org/10.1177/0363546514524364

Agricola, R., Heijboer, M.P., Roze, R.H., Reijman, M., Bierma-Zeinstra, S.M.A., Verhaar, J.A.N., Weinans, H., Waarsing, J.H., 2013b. Pincer deformity does not lead to osteoarthritis of the hip whereas acetabular dysplasia does: Acetabular coverage and development of osteoarthritis in a nationwide prospective cohort study (CHECK). Osteoarthr. Cartil. 21, 1514–1521. https://doi.org/10.1016/j.joca.2013.07.004

Agricola, R., Waarsing, J.H., Arden, N.K., Carr, A.J., Bierma-Zeinstra, S.M.A., Thomas, G.E., Weinans, H., Glyn-Jones, S., 2013. Cam Impingement of the Hip - A Risk Factor for Hip Osteoarthritis. Nat. Rev. Rheumatol. 9, 630–634. https://doi.org/10.1038/nrrheum.2013.114

An, K.N., Ueba, Y., Chao, E.Y., Cooney, W.P., and Linscheid, R.L., 1983. Tendon Excursion and Moment Arm of Index Finger Muscles. J. Biomech. 16(6), 419–425.

An, K.N., Takahashi, K., Harrigan, T.P., and Chao, E.Y., 1984. Determination of Muscle Orientations and Moment Arms. ASME J. Bio- mech. Eng. 106(3), 280–282.

Aoki, S.K., Beckmann, J.T., Wylie, J.D., 2016. Arthroscopic Femoral Osteochondroplasty for Cam-Type Femoroacetabular Impingement: The Trough Technique. Arthrosc. Tech. 5, e743–e749. https://doi.org/10.1016/j.eats.2016.02.024

Arnold, A.S., Blemker, S.S., Delp, S.L., 2001. Evaluation of a deformable musculoskeletal model for estimating muscle-tendon lengths during crouch gait. Ann. Biomed. Eng. 29, 263–274. https://doi.org/10.1114/1.1355277

Arnold, A.S., Salinas, S., Asakawa, D.J., Delp, S.L. 2000. Accuracy of Muscle Moment Arms Estimated From MRI-Based Musculoskeletal Models of the Lower Extremity. Comput. Aided Surg. 5(2),108–119. https://doi.org/10.3109/10929080009148877

Arnold, E.M., Ward, S.R., Lieber, R.L., Delp, S.L., 2010. A model of the lower limb for analysis of human movement. Ann. Biomed. Eng. 38, 269–279. https://doi.org/10.1007/s10439-009-9852-5

Azizi, E., Brainerd, E.L., Roberts, T.J., 2008. Variable gearing in pennate muscles. Proc. Natl. Acad. Sci. 105, 1745–1750. https://doi.org/10.1073/pnas.0709212105

Bagwell, J.J., Snibbe, J., Gerhardt, M., Powers, C.M., 2016. Hip kinematics and kinetics in persons with and without cam femoroacetabular impingement during a deep squat task. Clin. Biomech. 31, 87–92. https://doi.org/10.1016/j.clinbiomech.2015.09.016

Barton, C., Banga, K., Beaulé, P.E., 2009. Anterior Hueter Approach in the Treatment of Femoro-Acetabular Impingement: Rationale and Technique. Orthop. Clin. North Am. https://doi.org/10.1016/j.ocl.2009.03.002

Barton, C., Salineros, M.J., Rakhra, K.S., Beaulé, P.E., 2011. Validity of the alpha angle measurement on plain radiographs in the evaluation of cam-type femoroacetabular impingement. Clin. Orthop. Relat. Res. 469, 464–469. https://doi.org/10.1007/s11999-010-1624-x

Beaulé, P.E., Allen, D.J., Clohisy, J.C., Schoenecker, P.L., Leunig, M., 2009. The young adult with hip impingement: deciding on the optimal intervention. J. Bone Jt. Surg. 91, 210–21. https://doi.org/10.1016/j.jhse.2007.02.052

Beaulé, P.E., Clohisy, J.C., Schoenecker, P., Kim, Y.-J., Millis, M., Trousdale, R.T., 2007a. Hip Arthroscopy: An Emerging Gold Standard. Arthrosc. J. Arthrosc. Relat. Surg. 23, 682. https://doi.org/10.1016/j.arthro.2007.03.007

Beaulé, P.E., Hynes, K., Parker, G., Kemp, K.A., 2012. Can the alpha angle assessment of cam impingement predict acetabular cartilage delamination? Clin Orthop Relat Res 470, 3361–3367. https://doi.org/10.1007/s11999-012-2601-3

Beaulé, P.E., Le Duff, M.J., Zaragoza, E., 2007b. Quality of life following femoral head-neck osteochondroplasty for femoroacetabular impingement. J. Bone Jt. Surg. - Ser. A 89, 773–779. https://doi.org/10.2106/JBJS.F.00681

Beaulé, P.E., Speirs, A.D., Anwander, H., Melkus, G., Rakhra, K., Frei, H., Lamontagne, M., 2017. Surgical Correction of

Cam Deformity in Association with Femoroacetabular Impingement and Its Impact on the Degenerative Process within the Hip Joint. J. Bone Jt. Surg. 99, 1373–1381. https://doi.org/10.2106/JBJS.16.00415

Beaulé, P.E., Zaragoza, E., Motamedi, K., Copelan, N., Dorey, F.J., 2005. Three-dimensional computed tomography of the hip in the assessment of femoroacetabular impingement. J. Orthop. Res. 23, 1286–1292. https://doi.org/10.1016/j.orthres.2005.03.011

Beck, M., Kalhor, M., Leuning, M., Ganz, R., 2005. Hip Morphology Influences the Pattern of Damage to the Acetabular Cartilage: Femoroacetabular Impingement as a Cause of Early Osteoarthritis of the Hip. J. Bone Jt. Surg. - Br. Vol. 87–B, 1012–1018. https://doi.org/10.1302/0301-620X.87B7.15203

Beck, M., Leunig, M., Parvizi, J., Boutier, V., Wyss, D., Ganz, R., 2004. Anterior femoroacetabular impingement: part II. Midterm results of surgical treatment. Clin. Orthop. Relat. Res. 418, 67–73. https://doi.org/10.1097/00003086-200401000-00012

Bedi, A., Dolan, M., Hetsroni, I., Magennis, E., Lipman, J., Buly, R., Kelly, B.T., 2011a. Surgical Treatment of Femoroacetabular Impingement Improves Hip Kinematics. Am. J. Sports Med. 39, 43S–49S. https://doi.org/10.1177/0363546511414635

Bedi, A., Kelly, B.T., 2013. Femoroacetabular impingement. J Bone Jt. Surg Am 95, 82–92. https://doi.org/10.2106/JBJS.K.01219

Bedi, A., Zaltz, I., De La Torre, K., Kelly, B.T., 2011b. Radiographic comparison of surgical hip dislocation and hip arthroscopy for treatment of cam deformity in femoroacetabular impingement. Am. J. Sports Med. 39 Suppl, 20S–8S. https://doi.org/10.1177/0363546511412734

Benoit, D.L., Lamontagne, M., Cerulli, G., Liti, A., 2003. The clinical significance of electromyography normalisation techniques in subjects with anterior cruciate ligament injury during treadmill walking. Gait Posture 18, 56–63. https://doi.org/10.1016/S0966-6362(02)00194-7

Bombardier, C., Hawker, G., Mosher, D., 2011. The impact of arthritis in Canada: today and over the next 30 years, Arthritis Alliance of Canada.

Bottai, V., Dell’Osso, G., Celli, F., Bugelli, G., Cazzella, N., Cei, E., Guido, G., Giannotti, S., 2015. Total hip replacement in osteoarthritis: The role of bone metabolism and its complications. Clin. Cases Miner. Bone Metab. 12.

Bozic, K.J., Chan, V., Valone, F.H., Feeley, B.T., Vail, T.P., 2013. Trends in Hip Arthroscopy Utilization in the United States. J. Arthroplasty 28, 140–143. https://doi.org/10.1016/j.arth.2013.02.039

Brand, R., Crowninshield, R., Wittstoc, C., Clark, C., Van Krieken, F., 1982. A Model of Lower Extremity Muscular Anatomy. J. Biomech. Eng. 104, 304–310.

Brand, R.A., Pedersen, D.R., Davy, D.T., Kotzar, G.M., Heiple, K.G., Goldberg, V.M., 1994. Comparison of Hip Force Calculations and Measurements in the Same Patient. J. Arthroplasty 9, 45–51. https://doi.org/10.1016/0883-5403(94)90136-8

Brandt, K.D., 2003. Is a Strong Quadriceps Muscle Bad for a Patient with Knee Osteoarthritis? Ann. Intern. Med. 138, 8, 678-679.

Bresler, B., Frankel, J.P., 1950. The forces and moments in the leg during level walking. Trans. Am. Sot. Mech. Eng. 48, A-62, 27-36.

Brisson, N., Lamontagne, M., Kennedy, M.J., Beaulé, P.E., 2013. The Effects of Cam Femoroacetabular Impingement Corrective Surgery on Lower-Extremity Gait Biomechanics. Gait Posture 37, 258–263. https://doi.org/10.1016/j.gaitpost.2012.07.016

Buchanan, T.S., Lloyd, D.G., Manal, K., Besier, T.F., 2004. Neuromusculoskeletal Modeling: Estimation of Muscle Forces and Joint Moments and Movements from Measurements of Neural Command. J. Appl. Biomech. 20, 367–395. https://doi.org/10.1016/j.bbi.2008.05.010

Buchanan, T.S., Shreeve, D., 1996. An evaluation of optimization techniques for the prediction of muscle activation patterns during isometric tasks. J. Biomech. Eng. 118, 565–574. https://doi.org/10.1115/1.2796044

Byrd, J.W.T., 2014. Femoroacetabular Impingement in Athletes: Current Concepts. Am. J. Sports Med. 42, 737–51. https://doi.org/10.1177/0363546513499136

Byrd, J.W.T., 2006. Hip Arthroscopy: Surgical Indications. Arthrosc. J. Arthrosc. Relat. Surg. 22, 1260–1262. https://doi.org/10.1016/j.arthro.2006.08.021

Canadian Institute for Health Information, 2013. Hip and Knee Replacements in Canada : Canadian Joint Replacement

Registry 2013 Annual Report.

Caplan, N., Stewart, S., Kashyap, S., Banaszkiewicz, P., St Clair Gibson, A., Kader, D., Ewen, A., 2014. The Effect of Total Hip and Hip Resurfacing Arthroplasty on Vertical Ground Reaction Force and Impulse Symmetry during a Sit-to-Stand Task. Clin. Biomech. 29, 1164–9. https://doi.org/10.1016/j.clinbiomech.2014.09.008

Cappozzo, A., Catani, F., Della Croce, U., Leardini, A., 1995. Position and Orientation in Space of Bones During Movement. Clin. Biomech. 10, 171–178.

Carsen, S., Moroz, P.J., Rakhra, K., Ward, L.M., Dunlap, H., Hay, J.A., Willis, R.B., Beaulé, P.E., 2014. The Otto Aufranc Award. On the Etiology of the Cam Deformity: A Cross-Sectional Pediatric MRI Study. Clin. Orthop. Relat. Res. 472, 430–436. https://doi.org/10.1007/s11999-013-2990-y

Casartelli, N.C., Bizzini, M., Kemp, J., Naal, F.D., Leunig, M., Maffiuletti, N.A., 2017. What treatment options exist for patients with femoroacetabular impingement syndrome but without surgical indication? Br. J. Sports Med. bjsports-2017-098059. https://doi.org/10.1136/bjsports-2017-098059

Casartelli, N.C., Maffiuletti, N.A., Bizzini, M., Kelly, B.T., Naal, F.D., Leunig, M., 2016. The management of symptomatic femoroacetabular impingement: what is the rationale for non-surgical treatment? Br. J. Sports Med. 50, 4–6. https://doi.org/10.1136/bjsports-2015-095722

Casartelli, N.C., Maffiuletti, N.A., Item-Glatthorn, J.F., Impellizzeri, F.M., Leunig, M., 2014. Hip muscle strength recovery after hip arthroscopy in a series of patients with symptomatic femoroacetabular impingement. HIP Int. 24, 387–393. https://doi.org/10.5301/hipint.5000131

Casartelli, N.C., Maffiuletti, N.A., Item-Glatthorn, J.F., Staehli, S., Bizzini, M., Impellizzeri, F.M., Leunig, M., 2011. Hip muscle weakness in patients with symptomatic femoroacetabular impingement. Osteoarthr. Cartil. 19, 816–821. https://doi.org/10.1016/j.joca.2011.04.001

CDC, C. for D.C. and P., 2010. Number of all-listed procedures for discharges from short-stay hospitals, by procedure category and age: United States, 2010, National Hospital Discharge Survey: 2010 table.

Chakraverty, J.K., Sullivan, C., Gan, C., Narayanaswamy, S., Kamath, S., 2013. Cam and pincer femoroacetabular impingement: CT findings of features resembling femoroacetabular impingement in a young population without symptoms. Am. J. Roentgenol. 200, 389–395. https://doi.org/10.2214/AJR.12.8546

Chaudhry, H., Ayeni, O.R., 2014. The Etiology of Femoroacetabular Impingement: What We Know and What We Don't. Sport. Heal. A Multidiscip. Approach 6, 157–61. https://doi.org/10.1177/1941738114521576

Chegini, S., Beck, M., Ferguson, S.J., 2009. The Effects of Impingement and Dysplasia on Stress Distributions in the Hip Joint during Sitting and Walking: A Finite Element Analysis. J. Orthop. Res. 27, 195–201. https://doi.org/10.1002/jor.20747

CJRR, 2015. Canadian Joint Replacement Registry 1–63.

Clohisy, J.C., McClure, J.T., 2005. Treatment of Anterior Femoroacetabular Impingement with Combined Hip Arthroscopy and Limited Anterior Decompression. Iowa Orthop. J. 25, 164–171.

Clohisy, J.C., St John, L.C., Schutz, A.L., 2010. Surgical treatment of femoroacetabular impingement: a systematic review of the literature. Clin. Orthop. Relat. Res. 468, 555–564. https://doi.org/10.1007/s11999-009-1138-6; 10.1007/s11999-009-1138-6

Corcos, D. M., Gottlieb, G. L., Latash, M. L., Almeida, G. L., and Agarwal, G. C., 1992. Electromechanical Delay: An Experimental Artifact. J. Electromyogr. Kinesiol. 2,2, 59–68.

Crowninshield, R.D., Johnston, R.C., Andrews, J.G., Brand, R.A., 1978. A Biomechanical Investigation of the Human Hip. J. Biomech. 11, 75–85. https://doi.org/10.1016/0021-9290(78)90045-3

Damsgaard, M., Rasmussen, J., Christensen, S.T., Surma, E., de Zee, M., 2006. Analysis of musculoskeletal systems in the AnyBody Modeling System. Simul. Model. Pract. Theory 14, 1100–1111. https://doi.org/10.1016/j.simpat.2006.09.001

Davy, D.T., Kotzar, G.M., Brown, R.H., Heiple, K.G., Goldberg, V.M., Heiple, K.G., Berilla, J., Bursten, A.H., 1988. Telemetric across Measurements Total the Hip after Arthroplasty. J. Bone Joint Surg. Am. 70, 6.

De Luca, C.J., 1997. The use of surface electromyography in biomechanics. J. Appl. Biomech. 13, 135–163. https://doi.org/citeulike-article-id:2515246

De Luca, C.J., Donald Gilmore, L., Kuznetsov, M., Roy, S.H., 2010. Filtering the surface EMG signal: Movement artifact and baseline noise contamination. J. Biomech. 43, 1573–1579. https://doi.org/10.1016/j.jbiomech.2010.01.027

De Martino, I., D'Apolito, R., Waddell, B.S., McLawhorn, A.S., Sculco, P.K., Sculco, T.P., 2017. Early intraprosthetic dislocation in dual-mobility implants: a systematic review. Arthroplast. Today 6–11. https://doi.org/10.1016/j.artd.2016.12.002

De Martino, I., Triantafyllopoulos, G.K., Sculco, P.K., Sculco, T.P., 2014. Dual mobility in total hip arthroplasty. World J. Orthop. 45, 1–8. https://doi.org/10.1016/j.ocl.2013.08.004

De Sa, D., Horner, N.S., MacDonald, A., Simunovic, N., Slobogean, G., Philippon, M.J., Belzile, E.L., Karlsson, J., Ayeni, O.R., 2016. Evaluating healthcare resource utilization and outcomes for surgical hip dislocation and hip arthroscopy for femoroacetabular impingement. Knee Surgery, Sport. Traumatol. Arthrosc. https://doi.org/10.1007/s00167-015-3722-5

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. https://doi.org/10.1109/TBME.2007.901024

Delp, S.L., Loan, J.P., Hoy, M.G., Zajac, F.E., Topp, E.L., Rosen, J.M., 1990. An Interactive Graphics-Based Model of the Lower Extremity to Study Orthopaedic Surgical Procedures. IEEE Trans. Biomed. Eng. 37, 757–67. https://doi.org/10.1109/10.102791

Desseauve, D., Fradet, L., Lacouture, P., Pierre, F., 2017. Position for labor and birth: State of knowledge and biomechanical perspectives. Eur. J. Obstet. Gynecol. Reprod. Biol. 208, 46–54. https://doi.org/10.1016/j.ejogrb.2016.11.006

Diamond, L.E., Bennell, K.L., Wrigley, T. V., Hinman, R.S., O'Donnell, J., Hodges, P.W., 2017. Squatting biomechanics in individuals with symptomatic femoroacetabular impingement. Med. Sci. Sports Exerc. 49, 1520–1529. https://doi.org/10.1249/MSS.0000000000001282

Diamond, L.E., Wrigley, T. V., Bennell, K.L., Hinman, R.S., O'Donnell, J., Hodges, P.W., 2016a. Hip joint biomechanics during gait in people with and without symptomatic femoroacetabular impingement. Gait Posture 43, 198–203. https://doi.org/10.1016/j.gaitpost.2015.09.023

Diamond, L.E., Wrigley, T. V., Hinman, R.S., Hodges, P.W., O'Donnell, J., Takla, A., Bennell, K.L., 2016b. Isometric and isokinetic hip strength and agonist/antagonist ratios in symptomatic femoroacetabular impingement. J. Sci. Med. Sport 19, 696–701. https://doi.org/10.1016/j.jsams.2015.10.002

Dimmick, S., Stevens, K.J., Brazier, D., Anderson, S.E., 2013. Femoroacetabular Impingement. Radiol Clin N Am 51, 337–352. https://doi.org/http://dx.doi.org/10.1016/j.rcl.2012.12.002

Diulus, C.A., Krebs, V.E., Hanna, G., Barsoum, W.K., 2006. Hip Arthroscopy Technique and Indications. J. Arthroplasty 21, 68–73. https://doi.org/10.1016/j.arth.2006.01.011

Edwards, W.B., Miller, R.H., Derrick, T.R., 2016. Femoral Strain during Walking Predicted with Muscle Forces from Static and Dynamic Optimization. J. Biomech. 49, 1206–1213. https://doi.org/10.1016/j.jbiomech.2016.03.007

Eijer, H., Hogervorst, T., 2017. Femoroacetabular impingement causes osteoarthritis of the hip by migration and micro-instability of the femoral head. Med. Hypotheses 104, 93–96. https://doi.org/10.1016/j.mehy.2017.05.035

Eijer, H., Myers, S.R., Ganz, R., 2001. Anterior femoroacetabular impingement after femoral neck fractures. J. Orthop. Trauma 15, 475–81. https://doi.org/10.1097/00005131-200109000-00003

Ethgen, O., Bruyère, O., Richy, F., Dardennes, C., Reginster, J., 2004. Health-Related Quality of Life in Total Hip and Total Knee Arthroplasty. J. Bone Jt. Surg. 86–A, 963–74.

Farina, D., Merletti, R., Enoka, R.M., 2004. The extraction of neural strategies from the surface EMG Neural Control of Movement The extraction of neural strategies from the surface EMG. J. Appl. Physiol. 96, 1486–1495. https://doi.org/10.1152/japplphysiol.01070.2003

Farkas, G.J., Cvetanovich, G.L., Rajan, K.B., Espinoza Orías, A.A., Nho, S.J., 2015. Impact of Femoroacetabular Impingement Morphology on Gait Assessment in Symptomatic Patients. Sports Health 7, 429–436. https://doi.org/10.1177/1941738115592827

Fayad, T.E., Khan, M.A., Haddad, F.S., 2013. Femoroacetabular Ampingement: An Arthroscopic Solution. Bone Joint J. 95 B, 26–30. https://doi.org/10.1302/0301-620X.95B11.33016

Frank, J.M., Harris, J.D., Erickson, B.J., Slikker, W., Bush-Joseph, C.A., Salata, M.J., Nho, S.J., 2015. Prevalence of femoroacetabular impingement imaging findings in asymptomatic volunteers: A systematic review. Arthrosc. - J. Arthrosc. Relat. Surg. https://doi.org/10.1016/j.arthro.2014.11.042

Freeman, C.R., Azzam, M.G., Leunig, M., 2014. Hip preservation surgery: surgical care for femoroacetabular impingement and the possibility of preventing hip osteoarthritis. J. Hip Preserv. Surg. 1, 46–55. https://doi.org/10.1093/jhps/hnu015

Fridlund, A.J., Cacioppo, J.T., 1986. Guidelines for Human Electromyographic Research. Psychophysiology 23, 567–589. https://doi.org/10.1111/j.1469-8986.1986.tb00676.x

Friederich, J.A., Brand, R.A., 1990. Muscle fiber architecture in the human lower limb. J. Biomech. 23, 91–95. https://doi.org/10.1016/0021-9290(90)90373-B

Ganz, R., Gill, T., Gautier, E., Ganz, K., Krügel, N., Berlemann, U., 2001. Surgical Dislocation of the Adult Hip: A Technique with Full Access to the Femoral Head and Acetabulum without the Risk of Avascular Necrosis. J. Bone Jt. Surg. 1119–24. https://doi.org/10.1007/978-1-4471-5451-8_144

Ganz, R., Leunig, M., Leunig-Ganz, K., Harris, W.H., 2008. The Etiology of Osteoarthritis of the Hip: an Integrated Mechanical Concept. Clin Orthop Relat Res 466, 264–272. https://doi.org/10.1007/s11999-007-0060-z

Ganz, R., Parvizi, J., Beck, M., Leunig, M., Nötzli, H., Siebenrock, K.A., 2003. Femoroacetabular Impingement: a Cause for Osteoarthritis of the Hip. Clin. Orthop. Relat. Res. 112–120. https://doi.org/10.1097/01.blo.0000096804.78689.c2

Garner, B.A., Pandy, M.G., 2003. Estimation of musculotendon properties in the human upper limb. Ann. Biomed. Eng. 31, 207–220. https://doi.org/10.1114/1.1540105

Girard, J., Lavigne, M., Vendittoli, P., 2006. Biomechanical Reconstruction of the Hip: A Randomised Study Comparing Total Hip Resurfacing and Total Hip Arthroplasty. J. Bone Jt. Surg. 88, 721–726. https://doi.org/10.1302/0301-620X.88B6.17447

Glitsch, U., Baumann, W., 1997. The Three-Dimensional Determination of Internal Loads in the Lower Extremity. J. Biomech. 30, 1123–1131. https://doi.org/10.1016/S0021-9290(97)00089-4

Gosvig, K.K., Jacobsen, S., Sonne-Holm, S., Gebuhr, P., 2008. The Prevalence of Cam-Type Deformity of the Hip Joint : A Survey of 4151 Subjects of the Copenhagen Osteoarthritis Study. Acta radiol. 4, 436–441. https://doi.org/10.1080/02841850801935567

Grammatopoulos G., Speirs, A., Ng, K.C.G., Riviere, C., Rakhra, K.S., Lamontagne, M., Beaulé, P.E., 2018. The Acetabular and Spino-Pelvic Morphologies are Different in Subjects with Symptomatic Cam Femoro-Acetabular Impingement. J Orthop. Res. 36, 7, 1840-48. https://doi.org/10.1002/jor.23856

Griffin, D.R., Dickenson, E.J., Wall, P.D.H., Achana, F., Donovan, J.L., Griffin, J., Hobson, R., Hutchinson, C.E., Jepson, M., Parsons, N.R., Petrou, S., Realpe, A., Smith, J., Foster, N.E., 2018. Hip arthroscopy versus best conservative care for the treatment of femoroacetabular impingement syndrome (UK FASHIoN): a multicentre randomised controlled trial. Lancet 391, 2225–2235. https://doi.org/10.1016/S0140-6736(18)31202-9

Grospretre, S., Lepers, R., 2016. Performance characteristics of Parkour practitioners: Who are the traceurs? Eur. J. Sport Sci. 16, 526–535. https://doi.org/10.1080/17461391.2015.1060263

Guanche, C., Bare, A., 2006. Arthroscopic Treatment of Femoroacetabular Impingement. … J. Arthrosc. Relat. Surg. 22, 95–106. https://doi.org/10.1016/j.arthro.2005.10.018

Guyen, O., Chen, Q.S., Bejui-Hugues, J., Berry, D.J., An, K.-N., 2007. Unconstrained tripolar hip implants. Clin. Orthop. Relat. Res. 455, 202–208. https://doi.org/10.1097/01.blo.0000238796.59596.1f

Hack, K., Di Primio, G., Rakhra, K., Beaulé, P.E., 2010. Prevalence of Cam-Type Femoroacetabular Impingement Morphology in Asymptomatic Volunteers. J. Bone Jt. Surg. 92, 2436. https://doi.org/10.2106/JBJS.J.01280

Hainisch, R., Zubayer-Ul-Karim, M., Kranzl, A., Gföhler, M., Pandy, M., 2012. Joint moments in children with cerebral palsy based on biomechanical models. Gait Posture 36, S21. https://doi.org/10.1016/j.gaitpost.2011.10.215

Hamadouche, M., Arnould, H., Bouxin, B., 2012. Is a cementless dual mobility socket in primary THA A reasonable option? Clin. Orthop. Relat. Res. 470, 3048–3053. https://doi.org/10.1007/s11999-012-2395-3

Hammond, C.A., Hatfield, G.L., Gilbart, M.K., Garland, S.J., Hunt, M.A., 2017. Trunk and lower limb biomechanics during stair climbing in people with and without symptomatic femoroacetabular impingement. Clin. Biomech. 42, 108–114. https://doi.org/10.1016/j.clinbiomech.2017.01.015

Hamner, S.R., Seth, A., Delp, S.L., 2010. Muscle Contributions to Propulsion and Support During Running. J. Biomech. 43, 2709–2716. https://doi.org/10.1016/j.jbiomech.2010.06.025

Handsfield, G.G., Meyer, C.H., Hart, J.M., Abel, M.F., Blemker, S.S., 2014. Relationships of 35 lower limb muscles to height and body mass quantified using MRI. J. Biomech. 47, 631–638. https://doi.org/10.1016/j.jbiomech.2013.12.002

Hart, E.S., Metkar, U.S., Rebello, G.N., Grottkau, B.E., 2009. Femoroacetabular Impingement in Adolescents and Young Adults. Orthop. Nurs. 28, 117–26. https://doi.org/10.1097/NOR.0b013e3181a46a1c

Heintzbergen, S., Kulin, N.A., Ijzerman, M.J., Steuten, L.M.G., Werle, J., Khong, H., Marshall, D.A., 2013. Cost-Utility of Metal-on-Metal Hip Resurfacing Compared to Conventional Total Hip Replacement in Young Active Patients with Osteoarthritis. Value Heal. 16, 942–952. https://doi.org/10.1016/j.jval.2013.06.021

Hermens, H.J., Freriks, B., Merletti, R., Stegeman, D., Blok, J., Rau, G., Disselhorst-Klug, C., Hägg, G., 1999. SENIAM 8: European recommendations for surface electromyography. Roessingh Research and Development.

Hernández Gómez, J.J., Marquina, V., Gómez, R.W., 2013. On the performance of Usain Bolt in the 100 m sprint. Eur. J. Phys. 34, 1227–1233. https://doi.org/10.1088/0143-0807/34/5/1227

Herzog, W., Lee, E.J., Rassier, D.E., 2006. Residual force enhancement in skeletal muscle. J. Physiol. https://doi.org/10.1113/jphysiol.2006.107748

Hetsroni, I., Funk, S., Ben-Sira, D., Nyska, M., Palmanovich, E., Ayalon, M., 2015. Femoroacetabular impingement syndrome is associated with alterations in hindfoot mechanics: A three-dimensional gait analysis study. Clin. Biomech. 30, 1189–1193. https://doi.org/10.1016/j.clinbiomech.2015.08.005

Hicks, J.L., Uchida, T.K., Seth, A., Rajagopal, A., Delp, S.L., 2015. Is My Model Good Enough? Best Practices for Verification and Validation of Musculoskeletal Models and Simulations of Movement. J. Biomech. Eng. 137, 020905. https://doi.org/10.1115/1.4029304

Hug, F., Hodges, P.W., Tucker, K., 2015. Muscle Force Cannot Be Directly Inferred From Muscle Activation: Illustrated by the Proposed Imbalance of Force Between Vastus Medialis and Lateralis in People With Patellofemoral Pain. J. Orthop. Sports Phys. Ther. 45, 1–15. https://doi.org/10.2519/jospt.2015.5905

Hunt, M.A., Gunether, J.R., Gilbart, M.K., 2013. Kinematic and kinetic differences during walking in patients with and without symptomatic femoroacetabular impingement. Clin. Biomech. 28. https://doi.org/10.1016/j.clinbiomech.2013.05.002

Huo, B.M.H., Parvizi, J., Bal, B.S., Mont, M.A., 2009. What's New in Total Hip Arthroplasty. J. bone Jt. Surg. 91, 2522–2534. https://doi.org/10.2106/JBJS.I.00801

Kainz, H., Goudriaan, M., Falisse, A., Huenaerts, C., Desloovere, K., De Groote, F., Jonkers, I., 2018. The influence of maximum isometric muscle force scaling on estimated muscle forces from musculoskeletal models of children with cerebral palsy. Gait Posture. https://doi.org/10.1016/j.gaitpost.2018.07.172

Kassarjian, A., Brisson, M., Palmer, W.E., 2007. Femoroacetabular impingement. Eur. J. Radiol. 63, 29–35. https://doi.org/10.1016/j.ejrad.2007.03.020

Kelly, B.T., Williams, R.J., Philippon, M.J., 2003. Hip arthroscopy: current indications, treatment options, and management issues. Am. J. Sports Med. 31, 1020–37.

Kennedy, M.J., Lamontagne, M., Beaulé, P.E., 2009. Femoroacetabular Impingement Alters Hip and Pelvic Biomechanics during Gait: Walking Biomechanics of FAI. Gait Posture 30, 41–44. https://doi.org/10.1016/j.gaitpost.2009.02.008

Khanna, V., Caragianis, A., Diprimio, G., Rakhra, K., Beaulé, P.E., 2014. Incidence of hip pain in a prospective cohort of asymptomatic volunteers: is the cam deformity a risk factor for hip pain? Am. J. Sports Med. 42, 793–7. https://doi.org/10.1177/0363546513518417

King, M.G., Lawrenson, P.R., Semciw, A.I., Middleton, K.J., Crossley, K.M., 2018. Lower limb biomechanics in femoroacetabular impingement syndrome: A systematic review and meta-analysis. Br. J. Sports Med. 52, 566–580. https://doi.org/10.1136/bjsports-2017-097839

Kolk, S., Minten, M.J.M., Van Bon, G.E.A., Rijnen, W.H., Geurts, A.C.H., Verdonschot, N., Weerdesteyn, V., 2014. Gait and gait-related activities of daily living after total hip arthroplasty: A systematic review. Clin. Biomech. 29, 705–718. https://doi.org/10.1016/j.clinbiomech.2014.05.008

Komi, P. V., Belli, A., Huttunen, V., Bennefoy, R., Geyssant, A., Lacour, J.R., 1996. Optic Fibre as a Transducer of Tendomuscular Forces. Eur. J. Appl. Physiol. Occup. Physiol. 72, 278–280. https://doi.org/10.1007/BF00838652

Kumar, D., Dillon, A., Nardo, L., Link, T.M., Majumdar, S., Souza, R.B., 2014. Differences in the association of hip cartilage lesions and cam-type femoroacetabular impingement with movement patterns: A preliminary study. PM R 6, 681–689. https://doi.org/10.1016/j.pmrj.2014.02.002

Kurtz, S., Ong, K., Lau, E., Mowat, F., Halpern, M., 2007. Projections of Primary and Revision Hip and Knee Arthroplasty in the United States from 2005 to 2030. J. Bone Jt. Surg. 89, 780–5. https://doi.org/10.2106/JBJS.F.00222

Lachiewicz, P.F., Watters, T.S., 2012. The use of dual-mobility components in total hip arthroplasty. J Am Acad Orthop Surg 20, 481–486. https://doi.org/10.5435/JAAOS-20-08-481

Lai, A.K.M., Arnold, A.S., Wakeling, J.M., 2017. Why are Antagonist Muscles Co-activated in My Simulation? A Musculoskeletal Model for Analysing Human Locomotor Tasks. Ann. Biomed. Eng. 45, 2762–74. https://doi.org/10.1007/s10439-017-1920-7

Lamontagne, M., Beaulieu, M.L., Beaulé, P.E., 2011a. Comparison of joint mechanics of both lower limbs of tha patients with healthy participants during stair ascent and descent. J. Orthop. Res. 29, 305–311. https://doi.org/10.1002/jor.21248

Lamontagne, M., Beaulieu, M.L., Varin, D., Beaulé, P.E., 2012. Lower-limb joint mechanics after total hip arthroplasty during sitting and standing tasks. J. Orthop. Res. 30, 1611–1617. https://doi.org/10.1002/jor.22127

Lamontagne, M., Brisson, N., Kennedy, M.J., Beaulé, P.E., 2011b. Preoperative and Postoperative Lower-Extremity Joint and Pelvic Kinematics during Maximal Squatting of Patients with Cam Femoroacetabular Impingement. J. Bone Jt. Surg. 93, 40–5. https://doi.org/10.2106/JBJS.J.01809

Lamontagne, M., Kennedy, M.J., Beaulé, P.E., 2009. The Effect of Cam FAI on Hip and Pelvic Motion during Maximum Squat. Clin Orthop Relat Res 467, 645–650. https://doi.org/10.1007/s11999-008-0620-x

Lamontagne, M., Ng, K.C.G., Mantovani, G., Catelli, D.S., 2015. Biomechanics of Femoroacetabular Impingement, in: Sports Injuries. Springer-Verlag Berlin Heidelberg, 62, pp. 783–795. https://doi.org/10.1007/978-3-642-36801-1

Langlais, F.L., Ropars, M., Gaucher, F., Musset, T., Chaix, O., 2008. Dual mobility cemented cups have low dislocation rates in THA revisions. Clin. Orthop. Relat. Res. 466, 389–395. https://doi.org/10.1007/s11999-007-0047-9

Laude, F., Sariali, E., Nogier, A., 2009. Femoroacetabular impingement treatment using Arthroscopy and Anterior approach. Clin. Orthop. Relat. Res. 467, 747–752. https://doi.org/10.1007/s11999-008-0656-y

Lavernia, C.J., Iacobelli, D.A., Brooks, L., Villa, J.M., 2015. The Cost-Utility of Total Hip Arthroplasty: Earlier Intervention, Improved Economics. J. Arthroplasty 30, 945–949. https://doi.org/10.1016/j.arth.2014.12.028

Lavigne, M., Parvizi, J., Beck, M., Siebenrock, K.A., Ganz, R., Leunig, M., 2004. Anterior Femoroacetabular Impingement: Part I. Techniques of Joint Preserving Surgery. Clin. Orthop. Relat. Res. 61–6. https://doi.org/10.1097/00003086-200401000-00011

Learmonth, I.D., Young, C., Rorabeck, C., 2007. The operation of the century: total hip replacement. Lancet. https://doi.org/10.1016/S0140-6736(07)60457-7

Leclercq, S., Benoit, J., de Rosa, J., Euvrard, P., Leteurtre, C., Girardin, P., 2008. Results of the Evora Dual-Mobility Socket after a Minimum Follow-Up of Five Years. Rev. Chir. Orthop. Reparatrice Appar. Mot. 94. https://doi.org/10.1016/j.rco.2007.10.015

Leunig, M., Beaulé, P.E., Ganz, R., 2009. The Concept of Femoroacetabular Impingement: Current Status and Future Perspectives. Clin. Orthop. Relat. Res. 467, 616–622. https://doi.org/10.1007/s11999-008-0646-0

Leunig, M., Ganz, R., 2005. Femoroacetabular impingement. A common cause of hip complaints leading to arthrosis. Unfallchirurg 108, 9–10, 12–7. https://doi.org/10.1007/s00113-004-0902-z

Leunig, M., Podeszwa, D., Beck, M., Werlen, S., Ganz, R., 2004. Magnetic Resonance Arthrography of Labral Disorders in Hips with Dysplasia and Impingement, in: Clinical Orthopaedics and Related Research. pp. 74–80. https://doi.org/10.1097/00003086-200401000-00013

Lincoln, M., Johnston, K., Muldoon, M., Santore, R., 2009. Combined Arthroscopic and Modified Open Approach for Cam Femoroacetabular Impingement: A Preliminary Experience. Arthrosc. - J. Arthrosc. Relat. Surg. 25, 392–399. https://doi.org/10.1016/j.arthro.2008.12.002

Loudon, J.K., Reiman, M.P., 2014. Conservative management of femoroacetabular impingement (FAI) in the long distance runner. Phys. Ther. Sport 15, 82–90. https://doi.org/10.1016/j.ptsp.2014.02.004

Lu, T.W., O'Connor, J.J., 1999. Bone Position Estimation from Skin Marker Co-Ordinates Using Global Optimisation with Joint Constraints. J. Biomech. 32, 129–134. https://doi.org/10.1016/S0021-9290(98)00158-4

Lu, T.W., O'Connor, J.J., Taylor, S.J.G., Walker, P.S., 1998. Validation of a lower limb model with in vivo femoral forces telemetered from two subjects. J. Biomech. 31, 63–69. https://doi.org/10.1016/S0021-9290(97)00102-4

Lund, M.E., De Zee, M., Andersen, M.S., Rasmussen, J., 2012. On validation of multibody musculoskeletal models, in: Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine. pp. 82–94. https://doi.org/10.1177/0954411911431516

Malagelada, F., Del Carmen, V.A., Barke, S.J., Guirao Cano, L., Pleguezuelos Cobo, E., 2015. The Anterior Mini-Open Approach for Femeroacetabular Impingement: Gait and Functional Assessment at One Year Post-Surgery. Ann. Phys. Rehabil. Med. 58, 60–65. https://doi.org/10.1016/j.rehab.2014.09.013

Manal, K., Buchanan, T.S., 2004. Subject-specific estimates of tendon slack length: A numerical method. J. Appl. Biomech. 20, 195–203. https://doi.org/10.1123/jab.20.2.195

Mantoan, A., Pizzolato, C., Sartori, M., Sawacha, Z., Cobelli, C., Reggiani, M., 2015. MOtoNMS: A MATLAB toolbox to process motion data for neuromusculoskeletal modeling and simulation. Source Code Biol. Med. 10, 12. https://doi.org/10.1186/s13029-015-0044-4

Mantovani, G., 2016. Hip Contact Load and Muscle Force in Femoroacetabular Impingement Population. University of Ottawa. PhD Thesis.

Mantovani, G., Lamontagne, M., 2016. How different marker sets affect joint angles in inverse kinematics framework. J. Biomech. Eng. 139, 044503. https://doi.org/10.1115/1.4034708

Mantovani, G., Ng, K.G., Lamontagne, M., 2016. Regression models to predict hip joint centers in pathological hip population. Gait Posture 44, 48–54. https://doi.org/10.1016/j.gaitpost.2015.11.001

Martelli, S., Valente, G., Viceconti, M., Taddei, F., 2015. Sensitivity of a Subject-Specific Musculoskeletal Model to the Uncertainties on the Joint Axes Location. Comput. Methods Biomech. Biomed. Engin. 18, 1555–63. https://doi.org/10.1080/10255842.2014.930134

Martz, P., Bourredjem, A., Laroche, D., Arcens, M., Labattut, L., Binquet, C., Maillefert, J.-F.F., Baulot, E., Ornetti, P., 2016. Röttinger approach with dual-mobility cup to improve functional recovery in hip osteoarthritis patients: biomechanical and clinical follow-up. Int. Orthop. 1–7. https://doi.org/10.1007/s00264-016-3245-9

Matta, J.M., Shahrdar, C., Ferguson, T., 2005. Single-incision anterior approach for total hip arthroplasty on an orthopaedic table. Clin. Orthop. Relat. Res. 441, 115–124. https://doi.org/10.1097/01.blo.0000194309.70518.cb

Mcnitt-Gray, J.L., Yokoi, T., Millward, C., 1993. Landing Strategy Adjustments Made by Female Gymnasts in Response to Drop Height and Mat Composition. J. Appl. Biomech. 9, 173–190.

Millard, M., Uchida, T., Seth, A., Delp, S.L., 2013. Flexing computational muscle: modeling and simulation of musculotendon dynamics. J Biomech Eng 135, 21005. https://doi.org/10.1115/1.4023390

Mirka, G.A., 1991. The quantification of EMG normalization error. Ergonomics 34, 343–352. https://doi.org/10.1080/00140139108967318

Modenese, L., Phillips, A., Bull, A., 2011. An Open Source Lower Limb Model: Hip Joint Validation. J. Biomech. 44, 2185–2193. https://doi.org/10.1016/j.jbiomech.2011.06.019

Morris, M.E., Huxham, F., McGinley, J., Dodd, K., Iansek, R., 2001. The biomechanics and motor control of gait in Parkinson disease. Clin. Biomech. https://doi.org/10.1016/S0268-0033(01)00035-3

Morrow, M.M., Rankin, J.W., Neptune, R.R., Kaufman, K.R., 2014. A comparison of static and dynamic optimization muscle force predictions during wheelchair propulsion. J. Biomech. 47, 3459–3465. https://doi.org/10.1016/j.jbiomech.2014.09.013

Mudrick, C.A., Melvin, J.S., Springer, B.D., 2015. Late posterior hip instability after lumbar spinopelvic fusion. Arthroplast. Today 1, 25–29. https://doi.org/10.1016/j.artd.2015.05.002

Murphy, S.B., Ganz, R., Müller, M.E., 1995. The prognosis in untreated dysplasia of the hip. A study of radiographic factors that predict the outcome. J. Bone Jt. Surg. 77, 985–9.

Murray, R., Duncan, C., 1971. Athletic Activity in Adolescence as an Etiological Factor in Degenerative Hip Disease. J. bone Jt. surgery. 53, 406–419.

Myers, S.R., Eijer, H., Ganz, R., 1999. Anterior femoroacetabular impingement after periacetabular osteotomy. Clin. Orthop. Relat. Res. 363, 93-9.

Németh G, Ohlsén H., 1985. In Vivo Moment Arm Lengths for Hip Extensor Muscles at Different Angles of Hip Flexion. J Biomech. 18, 129–140.

Ng, K.C.G., Lamontagne, M., Jeffers, J.R.T., Grammatopoulos, G., Beaulé, P.E., 2018a. Anatomic Predictors of Sagittal

Hip and Pelvic Motions in Patients With a Cam Deformity. Am. J. Sports Med. 1–12. https://doi.org/10.1177/0363546518755150

Ng, K.C.G., Lamontagne, M., Labrosse, M.R., Beaulé, P.E., 2016a. Hip Joint Stresses Due to Cam-Type Femoroacetabular Impingement: A Systematic Review of Finite Element Simulations. PLoS One 11, 1–18. https://doi.org/10.1371/journal.pone.0147813

Ng, K.C.G., Mantovani, G., Modenese, L., Beaulé, P.E., Lamontagne, M., 2018b. Altered Walking and Muscle Patterns Reduce Hip Contact Forces in Individuals With Symptomatic Cam Femoroacetabular Impingement. Am. J. Sports Med. 46, 1–9. https://doi.org/10.1177/0363546518787518

Ng, K.G., Lamontagne, M., Adamczyk, A.P., Rahkra, K.S., Beaulé, P.E., 2015. Patient-Specific Anatomical and Functional Parameters Provide New Insights into the Pathomechanism of Cam FAI. Clin. Orthop. Relat. Res. 473, 1289–1296. https://doi.org/10.1007/s11999-014-3797-1

Ng, K.C.G., Lamontagne, M., Adamczyk, P., Rakhra, S., Beaule, E., 2014. Discriminant Function Analysis Using Radiographic And Biomechanical Parameters In Determining At-Risk Patients With A Cam Deformity. ORS 2014 Annu. Meet. 9–11.

Ng, K.C.G., Lamontagne, M., Beaulé, P.E., 2016b. Differences in Anatomical Parameters between the Affected and Unaffected Hip in Patients with Bilateral Cam-Type Deformities. Clin. Biomech. 33, 13–19. https://doi.org/10.1016/j.clinbiomech.2016.01.007

Ng, K.C.G., Rouhi, G., Lamontagne, M., Beaulé, P.E., 2012. Finite Element Analysis Examining the Effects of Cam FAI on Hip Joint Mechanical Loading Using Subject-Specific Geometries During Standing and Maximum Squat. HSS J. 8, 206–212. https://doi.org/10.1007/s11420-012-9292-x

Nilsson, M., Ohlsson, C., Mellström, D., Lorentzon, M., 2009. Previous Sport Activity during Childhood and Adolescence Is Associated with Increased Cortical Bone Size in Young Adult Men. J. bone Miner. Res. 24, 125–133. https://doi.org/10.1016/S0162-0908(09)79432-2

Ninomiya, S., Tagawa, H., 1984. Rotational Acetabular Osteotomy for the Dysplatic Hip. J. Bone Jt. Surg. 66–A, 430–436.

Nötzli, H., Wyss, T., Stoecklin, C., Schmid, M., Treiber, K., Hodler, J., 2002. The Contour of the Femoral Head-Neck Junction as a Predictor for the Risk of Anterior Impingment: Commentary. J. bone Jt. Surg. 84, 556–60. https://doi.org/10.1024/0369-8394.92.11.475

Nouh, M.R., Schweitzer, M.E., Rybak, L., Cohen, J., 2008. Femoroacetabular impingement: Can the alpha angle be estimated? Am. J. Roentgenol. 190, 1260–1262. https://doi.org/10.2214/AJR.07.3258

Pandy, M.G., 2001. Computer Modeling and Simulation of Human Movement. Annu. Rev. Biomed. Eng. 3, 245–273. https://doi.org/10.1146/annurev.bioeng.3.1.245

Pandy, M.G., Andriacchi, T.P., 2010. Muscle and Joint Function in Human Locomotion, Annual review of biomedical engineering. https://doi.org/10.1146/annurev-bioeng-070909-105259

Penrod, D.D., Davy, D.T., Singh, D.P., 1974. An optimization approach to tendon force analysis. J. Biomech. 7, 123–129. https://doi.org/10.1016/0021-9290(74)90050-5

Philippon, M.J., Skendze, J.G., Herzog, M., Goljan, P., Briggs, K.K., 2013. Hip arthroscopy for FAI: Predictors of patient satisfaction and conversion to total hip arthroplasty 5 to 7 years following arthroscopy. Orthop. J. Sport. Med. 1. https://doi.org/10.1177/2325967113S00006

Philippon, M.J., Yen, Y.-M., Briggs, K.K., Kuppersmith, D.A., Maxwell, R.B., 2008. Early Outcomes after Hip Arthroscopy for Femoroacetabular Impingement in the Athletic Adolescent Patient: A Preliminary Report. J. Pediatr. Orthop. 28, 705–710. https://doi.org/10.1097/BPO.0b013e318186eb2e

Philippot, R., Adam, P., Reckhaus, M., Delangle, F., Verdot, F., Curvale, G., Farizon, F., 2009. Prevention of dislocation in total hip revision surgery using a dual mobility design. Orthop. Traumatol. Surg. Res. 95, 407–413. https://doi.org/10.1016/j.otsr.2009.04.016

Pivec, R., Johnson, A.J., Mears, S.C., Mont, M.A., 2012. Hip Arthroplasty. Lancet 380, 1768–1777. https://doi.org/10.1016/S0140-6736(12)60607-2

Pollard, T.C.B., 2011. A perspective on femoroacetabular impingement. Skeletal Radiol. https://doi.org/10.1007/s00256-011-1137-z

Pollard, T.C.B., Villar, R.N., Norton, M.R., Fern, E.D., Williams, M.R., Murray, D.W., Carr, a J., 2010. Genetic Influences in the Aetiology of Femoroacetabular Impingement: A Sibling Study. J. Bone Joint Surg. Br. 92, 209–216. https://doi.org/10.1302/0301-620X.92B2.22850

Pourcelot, P., Defontaine, M., Ravary, B., Lemâtre, M., Crevier-Denoix, N., 2005. A Non-Invasive Method of Tendon Force Measurement. J. Biomech. 38, 2124–2129. https://doi.org/10.1016/j.jbiomech.2004.09.012

Pun, S., Kumar, D., Lane, N., 2015. Review: Femoroacetabular Impingement. Arthritis and Rheumatology 67, 17-27. https://doi.org/10.1002/art.38887

Rach, P., Westbury, D., 1974. The Short Range Stoffness of Active Mammalian Muscle and its Effect on Mechanical Properties. J Physiol 331–350.

Raikova, R.T., Aladjov, H.T., 2002. Hierarchical genetic algorithm versus static optimization-investigation of elbow flexion and extension movements. J. Biomech. 35, 1123–35. https://doi.org/S0021929002000313 [pii]

Rajagopal, A., Dembia, C.L., DeMers, M.S., Delp, D.D., Hicks, J.L., Delp, S.L., 2016. Full body musculoskeletal model for muscle-driven simulation of human gait. IEEE Trans. Biomed. Eng. 9294, 1–1. https://doi.org/10.1109/TBME.2016.2586891

Rane, A., Corstiaans, A., 2008. Does micturition improve in the squatting position? J. Obstet. Gynaecol. (Lahore). 28, 317–319. https://doi.org/10.1080/01443610802044726

Rassier, D., MacIntosh, B., Herzog, W., 1999. Length dependence of active force production in skeletal muscle. J. Appl. Physiol. 86, 1445–1457.

Reaz, M.B.I., Hussain, M.S., Mohd-Yasin, F., 2006. Techniques of EMG signal analysis: Detection, processing, classification and applications. Biol. Proced. Online 8, 11–35. https://doi.org/10.1251/bpo115

Reiman, M.P., Peters, S., Sylvain, J., Hagymasi, S., Mather, R.C., Goode, A.P., 2018. Femoroacetabular impingement surgery allows 74% of athletes to return to the same competitive level of sports participation but their level of performance remains unreported: A systematic review with meta-analysis. Br. J. Sports Med. 52, 972–981. https://doi.org/10.1136/bjsports-2017-098696

Reiman, M.P., Thorborg, K., Hölmich, P., 2016. Femoroacetabular Impingement Surgery Is on the Rise — But What Is the Next Step? J. Orthop. Sport. Phys. Ther. 46, 406–408. https://doi.org/10.2519/jospt.2016.0605

Robertson, D., Caldwell, G., Hamill, J., Kamen, G., Whittlesey, S., 2004. Research Methods in Biomechanics.

Ross, J.R., Nepple, J.J., Philippon, M.J., Kelly, B.T., Larson, C.M., Bedi, A., 2014. Effect of changes in pelvic tilt on range of motion to impingement and radiographic parameters of acetabular morphologic characteristics. Am. J. Sports Med. https://doi.org/10.1177/0363546514541229

Rutherford, D.J., Moreside, J., Wong, I., 2018. Differences in Hip Joint Biomechanics and Muscle Activation in Individuals With Femoroacetabular Impingement Compared With Healthy, Asymptomatic Individuals: Is Level-Ground Gait Analysis Enough? Orthop. J. Sport. Med. 6, 1–10. https://doi.org/10.1177/2325967118769829

Rylander, J.H., Shu, B., Favre, J., Safran, M.R., Andriacchi, T.P., 2013. Functional testing provides unique insights into the pathomechanics of femoroacetabular impingement and an objective basis for evaluating treatment outcome. J. Orthop. Res. 31, 1461–1468. https://doi.org/10.1002/jor.22375

Rylander, J.H., Shu, B., Andriacchi, T.P., Safran, M.R., 2011. Preoperative and Postoperative Sagittal Plane Hip Kinematics in Patients with Femoroacetabular Impingement during Level Walking. Am. J. Sports Med. 39, 36S–42S. https://doi.org/10.1177/0363546511413993

Samaan, M.A., Weinhandl, J.T., Bawab, S.Y., Ringleb, S.I., 2016. Determining residual reduction algorithm kinematic tracking weights for a sidestep cut via numerical optimization. Comput. Meth. Biomech. Biomed. Eng. 19, 16, 1721-29. http://dx.doi.org/10.1080/10255842.2016.1183123

Samaan, M.A., Schwaiger, B.J., Gallo, M.C., Link, T.M., Zhang, A.L., Majumdar, S., Souza, R.B., 2017. Abnormal Joint Moment Distributions and Functional Performance During Sit-to-Stand in Femoroacetabular Impingement Patients. PM R 9, 563–570. https://doi.org/10.1016/j.pmrj.2016.10.002

Samaan, M.A., Schwaiger, B.J., Gallo, M.C., Sada, K., Link, T.M., Zhang, A.L., Majumdar, S., Souza, R.B., 2017. Joint Loading in the Sagittal Plane During Gait Is Associated With Hip Joint Abnormalities in Patients With Femoroacetabular Impingement. Am. J. Sports Med. 45, 810–818. https://doi.org/10.1177/0363546516677727

Sampson, J.D., Safran, M.R., 2015. Biomechanical implications of corrective surgery for FAI: An evidence-based review. Sports Med. Arthrosc. 23, 169–173. https://doi.org/10.1097/JSA.0000000000000092

Sartori, M., Gizzi, L., Lloyd, D.G., Farina, D., 2013. A musculoskeletal model of human locomotion driven by a low dimensional set of impulsive excitation primitives. Front. Comput. Neurosci. 7, 79. https://doi.org/10.3389/fncom.2013.00079

Sartori, M., Reggiani, M., Farina, D., Lloyd, D.G., 2012. EMG-Driven Forward-Dynamic Estimation of Muscle Force and Joint Moment about Multiple Degrees of Freedom in the Human Lower Extremity. PLoS One 7. https://doi.org/10.1371/journal.pone.0052618

Seth, A., Hicks, J.L., Uchida, T.K., Habib, A., Dembia, C.L., Dunne, J.J., Ong, C.F., DeMers, M.S., Rajagopal, A., Millard, M., Hamner, S.R., Arnold, E.M., Yong, J.R., Lakshmikanth, S.K., Sherman, M.A., Ku, J.P., Delp, S.L., 2018. OpenSim: Simulating musculoskeletal dynamics and neuromuscular control to study human and animal movement. PLOS Comput. Biol. 14, e1006223. https://doi.org/10.1371/journal.pcbi.1006223

Sharif Shourijeh, M., Smale, K.B., Potvin, B.M., Benoit, D.L., 2016. A Forward-Muscular Inverse-Skeletal Dynamics Framework for Human Musculoskeletal Simulations. J. Biomech. https://doi.org/10.1016/j.jbiomech.2016.04.007

Siebenrock, K.A., Behning, A., Mamisch, T.C., Schwab, J.M., 2013. Growth plate alteration precedes cam-type deformity in elite basketball players hip. Clin. Orthop. Relat. Res. 471, 1084–1091. https://doi.org/10.1007/s11999-012-2740-6

Siebenrock, K.A., Ferner, F., Noble, P.C., Santore, R.F., Werlen, S., Mamisch, T.C., 2011. The Cam-Type Deformity of the Proximal Femur Arises in Childhood in Response to Vigorous Sporting Activity. Clin. Orthop. Relat. Res. 469, 3229–3240. https://doi.org/10.1007/s11999-011-1945-4

Siebenrock, K.A., Schwab, J.M., 2013. The Cam-Type Deformity - What Is It: SCFE, Osteophyte, or a New Disease? J Pediatr Orthop 33, S121-5. https://doi.org/10.1097/BPO.0b013e3182771782

Smale, K.B., 2018. Relating subjective and objective knee function after anterior cruciate ligament injury through biomechanical and neuromusculoskeletal modelling approaches. University of Ottawa.

Speirs, A., Beaulé, P., Rakhra, K., Schweitzer, M., Frei, H., 2013a. Increased Acetabular Subchondral Bone Density Is Associated with Cam-Type Femoroacetabular Impingement. Osteoarthr. Cartil. 21, 551–558. https://doi.org/10.1016/j.joca.2013.01.012

Speirs, A., Beaulé, P., Rakhra, K., Schweitzer, M., Frei, H., 2013b. Bone density is higher in cam-type femoroacetabular impingement deformities compared to normal subchondral bone. Osteoarthr. Cartil. 21, 1068–1073. https://doi.org/10.1016/j.joca.2013.04.016

Spoor, C. W., and van Leeuwen, J. L., 1992. Knee Muscle Moment Arms From MRI and From Tendon Travel. J. Biomech. 25(2), 201–206.

Stansfield, B., Nicol, A., Paul, J., Kelly, I., Graichen, F., Bergmann, G., 2003. Direct Comparison of Calculated Hip Joint Contact Forces with Those Measured Using Instrumented Implants. An evaluation of a Three-Dimensional Mathematical Model of the Lower Limb. J. Biomech. 36, 929–936. https://doi.org/10.1016/S0021-9290(03)00072-1

Stansfield, B.W., Nicol, A.C., 2002. Hip joint contact forces in normal subjects and subjects with total hip prostheses: Walking and stair and ramp negotiation. Clin. Biomech. 17, 130–139. https://doi.org/10.1016/S0268-0033(01)00119-X

Steele, K.M., DeMers, M.S., Schwartz, M.H., Delp, S.L., 2012. Compressive tibiofemoral force during crouch gait. Gait Posture 35, 556–560. https://doi.org/10.1016/j.gaitpost.2011.11.023

Steele, K.M., Seth, A., Hicks, J.L., Schwartz, M.S., Delp, S.L., 2010. Muscle Contributions to Support and Progression during Single-Limb Stance in Crouch Gait. J Biomech. 43, 2099–2105. https://doi.org/10.1016/j.surg.2006.10.010.Use

Stevens, M.S., Legay, D.A., Glazebrook, M.A., Amirault, D., 2010. The evidence for hip arthroscopy: Grading the current indications. Arthrosc. - J. Arthrosc. Relat. Surg. https://doi.org/10.1016/j.arthro.2010.07.016

Sutherland, D., 1997. The development of mature gait. Gait Posture 6, 163–170. https://doi.org/10.1016/S0966-6362(97)00029-5

Swiontkowski, M., 2013. Concern about early intraprosthetic dislocation in dual-mobility implants. JBJS Case Connect. 3, 3 Suppl 9, e911-7. https://doi.org/10.2106/JBJS.CC.M.00212

Tannast, M., Siebenrock, K.A., Anderson, S.E., 2007. Femoroacetabular impingement: Radiographic diagnosis - What the radiologist should know. Am. J. Roentgenol. 188, 1540-52. https://doi.org/10.2214/AJR.06.0921

Tarasevicius, S., Busevicius, M., Robertsson, O., Wingstrand, H., 2010. Dual mobility cup reduces dislocation rate after arthroplasty for femoral neck fracture. BMC Musculoskelet. Disord. 11, 175. https://doi.org/10.1186/1471-2474-11-175

Taruc-Uy, R.L., Lynch, S. a, 2013. Diagnosis and Treatment of Osteoarthritis. Prim. Care 40, 821–36. https://doi.org/10.1016/j.pop.2013.08.003

Teratani, T., Naito, M., Kiyama, T., Maeyama, A., 2010. Periacetabular Osteotomy in Patients Fifty Years of Age or Older. J. Bone Jt. Surg. 92, 31–41. https://doi.org/10.2106/JBJS.H.01556

Todorov, E., Jordan, M.I., 2002. Optimal feedback control as a theory of motor coordination. Nat. Neurosci. 5, 1226–1235. https://doi.org/10.1038/nn963

Tönnis, D., Heinecke, A., 1999. Acetabular and femoral anteversion: relationship with osteoarthritis of the hip. J. bone Jt. Surg. 81, 1747–1770. https://doi.org/10.2106/JBJS.L.00710

Vainionpää, A., Korpelainen, R., Leppäluoto, J., Jämsä, T., 2005. Effects of high-impact exercise on bone mineral density: A randomized controlled trial in premenopausal women. Osteoporos. Int. 16, 191–197. https://doi.org/10.1007/s00198-004-1659-5

Valente, G., Crimi, G., Vanella, N., Schileo, E., Taddei, F., 2017. NMSBUILDER: Freeware to create subject-specific musculoskeletal models for OpenSim. Comput. Methods Programs Biomed. 152, 85–92. https://doi.org/10.1016/j.cmpb.2017.09.012

Valente, G., Pitto, L., Testi, D., Seth, A., Delp, S.L., Stagni, R., Viceconti, M., Taddei, F., 2014. Are Subject-Specific Musculoskeletal Models Robust to the Uncertainties in Parameter Identification? PLoS One 9. https://doi.org/10.1371/journal.pone.0112625

van Bolhuis, B.M., Gielen, C.C., 1999. A comparison of models explaining muscle activation patterns for isometric contractions. Biol. Cybern. 81, 249–261. https://doi.org/10.1007/s004220050560

Varadarajan, K.M., Zumbrunn, T., Duffy, M.P., Patel, R., Freiberg, A.A., Rubash, H.E., Malchau, H., Muratoglu, O.K., 2016. Reducing the distal profile of dual mobility liners can mitigate soft-tissue impingement and liner entrapment without affecting mechanical performance. J. Orthop. Res. 34, 889–897. https://doi.org/10.1002/jor.23078

Vaughan, C.L., 1984. Computer Simulation of Human Motion in Sports. Exerc. Sport Sci. Rev. 12, 373–398.

Waddell, B.S., De Martino, I., Sculco, T., Sculco, P., 2016. Total hip arthroplasty dislocations are more complex than they appear: A case report of intraprosthetic dislocation of an anatomic dual-mobility implant after closed reduction. Ochsner J. 16, 185–90.

Ward, S.R., Eng, C.M., Smallwood, L.H., Lieber, R.L., 2009. Are current measurements of lower extremity muscle architecture accurate? Clin. Orthop. Relat. Res. 467, 1074–1082. https://doi.org/10.1007/s11999-008-0594-8

Weiland, D.E., Philippon, M.J., 2005. Arthroscopic Technique of Femoroacetabular Impingement. Oper. Tech. Orthop. 15, 256–260. https://doi.org/10.1053/j.oto.2005.07.002

Wesseling, M., Derikx, L.C., De Groote, F., Bartels, W., Meyer, C., Verdonschot, N., Jonkers, I., 2015. Muscle optimization techniques impact the magnitude of calculated hip joint contact forces. J. Orthop. Res. 33, 430–438. https://doi.org/10.1002/jor.22769

Wesseling, M., Meyer, C., Corten, K., Desloovere, K., Jonkers, I., 2018. Longitudinal joint loading in patients before and up to one year after unilateral total hip arthroplasty. Gait Posture 61, 117–124. https://doi.org/10.1016/j.gaitpost.2018.01.002

Wickiewicz, T.L., Roy, R.R., Powell, P.L., Edgerton, V.R., 1983. Muscle architecture of the human lower limb. Clin. Orthop. Relat. Res. https://doi.org/009-921X/83/1000/275

Wilson, J.J., Furukawa, M., 2014. Evaluation of the Patient with Hip Pain. Am. Fam. Physician 89, 27–34. https://doi.org/10.1016/S1002-0160(13)60089-2

Winby, C.R., Lloyd, D.G., Kirk, T.B., 2008. Evaluation of different analytical methods for subject-specific scaling of musculotendon parameters. J. Biomech. 41, 1682–1688. https://doi.org/10.1016/j.jbiomech.2008.03.008

Winter, D.A., Fuglevand, A.J., Archer, S.E., 1994. Crosstalk in surface electromyography: Theoretical and practical estimates. J. Electromyogr. Kinesiol. 4, 15–26. https://doi.org/10.1016/1050-6411(94)90023-X

Wolff, J., 1892. Das Gesetz der Transformation der Knochen. Translated as: The Law of Bone Remodelling, Springer Verlag. Springer: Berlin.

Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., Whittle, M., D'Lima, D.D., Cristofolini, L., Witte, H., Schmid, O., Stokes, I., 2002. ISB Recommendation on Definitions of Joint Coordinate System of Various Joints for the Reporting of Human Joint Motion — Part I: Ankle, Hip, and Spine. J. Biomech. 35, 543–548. https://doi.org/10.1016/S0021-9290(01)00222-6

Zajac, F.E., 1989. Muscle and tendon: properties, models, scaling and application to biomechanics and motor control. Crit. Rev. Biomed. Eng. 17, 359–410.

VI APPENDIX

Ethics

Ottawa Health Science Network Research Ethics Board / *Conseil d'éthique de la recherche du Réseau de science de la santé d'Ottawa*

File Number: H12-10-07

Date (mm/dd/yyyy): 04/17/2014

Université d'Ottawa
Bureau d'éthique et d'intégrité de la recherche

University of Ottawa
Office of Research Ethics and Integrity

Ethics Approval Notice

Health Sciences and Science REB

Principal Investigator / Supervisor / Co-investigator(s) / Student(s)

First Name	Last Name	Affiliation	Role
Mario	Lamontagne	Health Sciences / Human Kinetics	Principal Investigator
Paul	Beaulé	Medicine / Medicine	Co-Principal Investigator

File Number: H12-10-07

Type of Project: Professor

Title: Mobility Assessment of a Dual Mobility Hip Arthroplasty for Osteoarthritic Hip: Feasibility for a Prospective Randomized Controlled Trial

Renewal Date (mm/dd/yyyy)	Expiry Date (mm/dd/yyyy)	Approval Type
04/11/2014	04/10/2015	Ia

(Ia: Approval, Ib: Approval for initial stage only)

Special Conditions / Comments:
N/A

550, rue Cumberland, pièce 154
Ottawa (Ontario) K1N 6N5 Canada
550 Cumberland Street, room 154
Ottawa, Ontario K1N 6N5 Canada
(613) 562-5387 • Téléc./Fax (613) 562-5338
www.recherche.uottawa.ca/deontologie/ www.research.uottawa.ca/ethics/

Ottawa Health Science Network Research Ethics Board / *Conseil d'éthique de la recherche du Réseau de science de la santé d'Ottawa*

File Number: H02-10-07 Bureau d'éthique et 02/09/2015

Université d'Ottawa University of Ottawa

Ethics Approval Notice

Health Sciences and Science REB

Principal Investigator / Supervisor / Co-investigator(s) / Student(s)

First Name	Last Name	Affiliation	Role
Paul	Beaulé	Medicine / Medicine	Principal Investigator
Mario	Lamontagne	Health Sciences / Human Kinetics	Co-investigator
Jae-Jin	Ryu	Others / Others	Co-investigator
Giulia	Mantovani	Health Sciences / Human Kinetics	Student Researcher

File Number: H02-10-07

Type of Project: Professor

Title: FAI: Correlating Hip Morphology to Changes in Cartilage and Subchondral Bone

Renewal Date (mm/dd/yyyy)	Expiry Date (mm/dd/yyyy)	Approval Type
04/07/2015	04/06/2016	Ia

(Ia: Approval, Ib: Approval for initial stage only)

Special Conditions / Comments:
N/A

1

550, rue Cumberland, pièce 154 550 Cumberland Street, room 154
Ottawa (Ontario) K1N 6N5 Canada Ottawa, Ontario K1N 6N5 Canada
(613) 562-5387 • Téléc./Fax (613) 562-5338
www.recherche.uottawa.ca/deontologie/ www.research.uottawa.ca/ethics/

Hip dysfunction and Osteoarthritis Outcome Score (HOOS)

English version LK 2.0

HOOS HIP SURVEY

Today's date: _____/_____/_____ Date of birth: _____/_____/_______

Name: ____

INSTRUCTIONS: This survey asks for your view about your hip. This information will help us keep track of how you feel about your hip and how well you are able to do your usual activities.

Answer every question by ticking the appropriate box, only <u>one</u> box for each question. If you are uncertain about how to answer a question, please give the best answer you can.

Symptoms

These questions should be answered thinking of your hip symptoms and difficulties during the **last week**.

S1. Do you feel grinding, hear clicking or any other type of noise from your hip?

Never	Rarely	Sometimes	Often	Always
☐	☐	☐	☐	☐

S2. Difficulties spreading legs wide apart

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

S3. Difficulties to stride out when walking

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Stiffness

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

S4. How severe is your hip joint stiffness after first wakening in the morning?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

S5. How severe is your hip stiffness after sitting, lying or resting **later in the day**?

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Pain

P1. How often is your hip painful?

Never	Monthly	Weekly	Daily	Always
☐	☐	☐	☐	☐

What amount of hip pain have you experienced the **last week** during the following activities?

P2. Straightening your hip fully

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

What amount of hip pain have you experienced the **last week** during the following activities?

P3. Bending your hip fully

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P4. Walking on a flat surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P5. Going up or down stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P6. At night while in bed

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P7. Sitting or lying

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P8. Standing upright

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P9. Walking on a hard surface (asphalt, concrete, etc.)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

P10. Walking on an uneven surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, daily living

The following questions concern your physical function. By this we mean your ability to move around and to look after yourself. For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your hip.

A1. Descending stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A2. Ascending stairs

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A3. Rising from sitting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A4. Standing

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

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

A5. Bending to the floor/pick up an object

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A6. Walking on a flat surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A7. Getting in/out of car

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A8. Going shopping

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A9. Putting on socks/stockings

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A10. Rising from bed

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A11. Taking off socks/stockings

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A12. Lying in bed (turning over, maintaining hip position)

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A13. Getting in/out of bath

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A14. Sitting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

A15. Getting on/off toilet

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

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

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

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

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Function, sports and recreational activities

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

SP1. Squatting

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP2. Running

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP3. Twisting/pivoting on loaded leg

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

SP4. Walking on uneven surface

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

Quality of Life

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

Never	Monthly	Weekly	Daily	Constantly
☐	☐	☐	☐	☐

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

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

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

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

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

None	Mild	Moderate	Severe	Extreme
☐	☐	☐	☐	☐

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

Extra

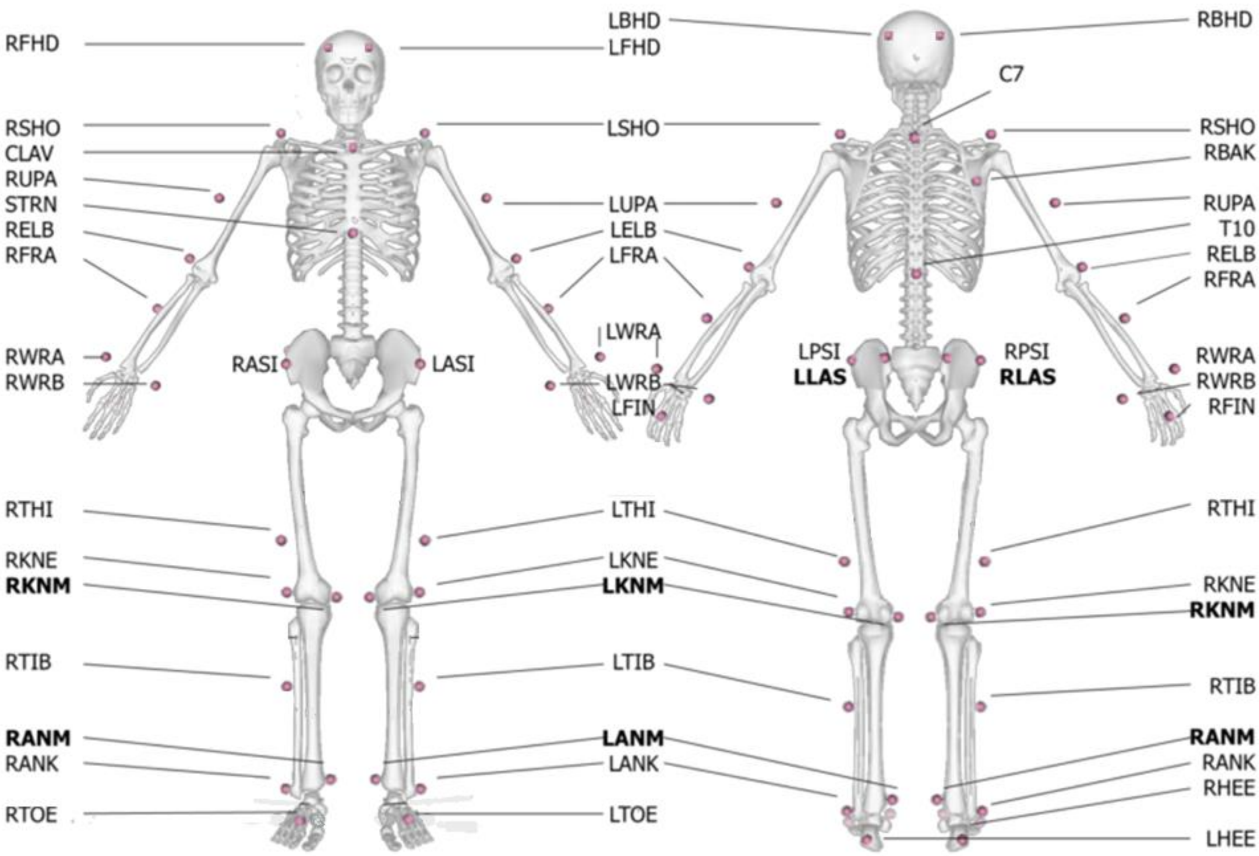


Figure A.1. University of Ottawa Motion Analysis Model (UOMAM) marker set. The Plug-in-Gait marker set acts as the basis with the additions (bold) of medial knee and ankle (LMKN, RMKN, LMAN, RMAN), and lateral pelvis (LLAS, RLAS) markers. Figure adapted from Mantovani (2016).

Table A.1. Coefficient of determination (R^2) of the predicted muscle activation from the squat model and the muscle excitations measured by EMG. The predicted muscle activations were shifted back over the excitations signals in order to account for the electrical mechanical delay.

	R^2 value	RF	TFL	Gmax1	Gmax2	Gmax3	Gmed1	Gmed2	Gmed3	BF	ST	TA	PL	GL	AL	SR	GM
Preoperative	FAI_004	0.77	0.75	0.61	0.64	0.80	0.40	0.50	0.53	0.70	0.42						
Preoperative	FAI_007	0.89	0.82	0.64	0.74	0.81	0.60	0.70	0.81	0.73	0.58						
Preoperative	FAI_017	0.88	0.50	0.80	0.85	0.88	0.64	0.73	0.79	0.75	0.74						
Preoperative	FAI_021	0.76	0.49	0.81	0.77	0.84	0.89	0.91	0.90	0.75	0.80						
Preoperative	FAI_029	0.95	0.83	0.70	0.82	0.80	0.51	0.58	0.60	0.86	0.82						
Preoperative	FAI_032	0.92	0.88	0.80	0.68	0.67	0.53	0.68	0.86	0.82	0.64						
Preoperative	FAI_045	0.86	0.69	0.55	0.92	0.97	0.77	0.78	0.78	0.86	0.80						
Preoperative	FAI_049	0.38	0.84				0.40	0.34	0.50	0.86	0.54						
Preoperative	FAI_065	0.89	0.65	0.72	0.75	0.79	0.42	0.48	0.57	0.82	0.54	0.87	0.68	0.38	0.47	0.85	0.65
Preoperative	FAI_067	0.67	0.58	0.81	0.84	0.83	0.62	0.57	0.60		0.71	0.74	0.83	0.51	0.52	0.92	0.53
Postoperative	FAI_004	0.95	0.69	0.73	0.78	0.85	0.57	0.68	0.73	0.81	0.67	0.86	0.76	0.57	0.42	0.88	0.69
Postoperative	FAI_007	0.87	0.84	0.69	0.75	0.88	0.70	0.69	0.60	0.83	0.79	0.83	0.54	0.56	0.65	0.89	0.45
Postoperative	FAI_017	0.85	0.56	0.78	0.81	0.82	0.55	0.61	0.64	0.79	0.74	0.96	0.68	0.58	0.38	0.80	0.54
Postoperative	FAI_021	0.69	0.59	0.87	0.76	0.85	0.82	0.83	0.84	0.75	0.76	0.81	0.54	0.73	0.72	0.78	0.55
Postoperative	FAI_029	0.88	0.68	0.77	0.76	0.74	0.63	0.68	0.72	0.75	0.72	0.77	0.88	0.69	0.37	0.95	0.47
Postoperative	FAI_032	0.90	0.77	0.71	0.72	0.82	0.88	0.92	0.94	0.76		0.90	0.48	0.53	0.24	0.79	0.63
Postoperative	FAI_045	0.77	0.74	0.85	0.70	0.80	0.74	0.77	0.81	0.91	0.76	0.91	0.54	0.64	0.59	0.75	0.46
Postoperative	FAI_049											0.39	0.42	0.28	0.70	0.50	0.44
Postoperative	FAI_065	0.82	0.80	0.78	0.82	0.67	0.53	0.56	0.54	0.92	0.66	0.42	0.85	0.35	0.33	0.72	0.55
Postoperative	FAI_067	0.76	0.72	0.64	0.66	0.83	0.82	0.80	0.63		0.49	0.93	0.90	0.49	0.21	0.89	0.45
	Pre AVG	0.80	0.70	0.71	0.78	0.82	0.58	0.63	0.69	0.80	0.66	0.80	0.75	0.45	0.50	0.89	0.59
	Post AVG	0.83	0.71	0.76	0.75	0.81	0.70	0.73	0.72	0.82	0.70	0.78	0.66	0.54	0.46	0.79	0.52

RF: rectus femoris; TFL: tensor fascia latea, Gmax1: gluteus maximus (superior), Gmax2: gluteus maximus (middle), Gmax3: gluteus maximus (inferior), Gmax1: gluteus medius (anterior), Gmax2: gluteus medius (middle), Gmax3: gluteus medius (posterior), BF: biceps femoris, ST: semitendinosus, TA: tibialis anteriors, PL: peroneus longus, GL: gracilis, AL: adductor longus, SR: sartorius, GM: gastrocnemius medialis. The blank cases refer to data that were not available due to technical issues.

List of Contributions

Journals

Published

1. Catelli DS, Kowalski E, Beaulé PE, Smit K, Lamontagne M. (2018) Asymptomatic Participants with Femoroacetabular Deformity Demonstrate Stroger Hip Extensors and Greater Pelvis Mobility During the Deep Squat Task. *The Orthopaedic Journal of Sports Medicine.* 6(7):1-10.
2. Catelli DS, Kowalski E, Beaulé PE, Lamontagne M. (2017) Does the Dual-Mobility Hip Prosthesis Produce Better Joint Kinematics during Extreme Hip Flexion Task? *Journal of Arthroplasty*. 32(10):3206-12.
3. Catelli DS, Wesseling M, Jonkers I, Lamontagne M. (2018) A Musculoskeletal Model Customized for Squatting Task. *Computer Methods in Biomechanics and Biomedical Engineering*. 6:1-4.

Submitted

4. Catelli DS, Kowalski E, Beaulé PE, Lamontagne M. Increased Pelvic Mobility and Altered Hip Muscles Contraction Pattern: Two-Year Follow-up Cam FAI Corrective Surgery. Submitted for publication to *Journal of Hip Preservation Surgery, September 2018.*
5. Catelli DS, Ng KCG, Kowalski E, Beaulé PE, Lamontagne M. Femoroacetabular Impingement Surgical Correction Decreases Hip Muscle Forces after 2-Year Follow-Up. Submitted for publication to *Gait & Posture, September 2018*.
6. Catelli DS, Ng KCG, Wesseling M, Kowalski E, Jonkers I, Beaulé PE, Lamontagne M. Femoroacetabular Impingement Surgical Correction Affects Squat Hip Contact Forces 2-Year after Follow-Up. *Prepared for submission to Clinical Biomechanics, December 2018*.

Additional

7. Kowalski E, Catelli DS, Lamontagne M. Side does not matter in healthy young and older individuals - Examining the importance of how we match limbs during gait studies. (2019). *Gait & Posture*. 67:133-136.
8. Kowalski E, Catelli DS, Lamontagne M. Comparing the accuracy of visual and computerized onset detection methods on simulated electromyography signals with varying signal-to-noise ratios. *Prepared for submission to Journal of Biomechanical Engineering, December 2018*.

Book Chapter

9. Lamontagne M, Ng KCG, Mantovani G, Catelli DS. (2015) Biomechanics of Femoroacetabular Impingement. In: Doral MN & Karlsson J. (Org.) Sports Injuries. 2ed. Berlin, Heidelberg: Springer Berlin Heidelberg, v.1, p. 783-795.

Conferences

Podiums

10. Catelli DS, Kowalski E, Carsen S, Antinolfi P, Beaulé PE, Lamontagne M. (2018) Surgical Correction of Femoroacetabular Impingement: An Electromyography Study during Deep Squat. In: 103º *Congresso Nazionale della Società Italiana di Ortopedia e Traumatologia* – SIOT 2018, Bari, Italy.

11. Catelli DS, Ng KCG, Kowalski E, Beaulé PE, Lamontagne M. (2018) Hip Muscle and Contact Forces in Post-Surgical Cam FAI. In: European Orthopaedic Research Society – EORS 2018, 26th Annual Meeting, Galway, Ireland.

12. Lamontagne M, Catelli DS, Kowalski E, Le A, Beaulé PE. (2018) Hip Muscle Strength and Frequency Analysis Following FAI Surgery Correction. In: Canadian Orthopaedic Association - COA 2018, Victoria, Canada.

13. Catelli DS, Wesseling M, Jonkers I, Lamontagne M. (2017) Wrapping Surfaces to Control Moment Arm Lengths during a Squat Task. In: 35th International Conference on Biomechanics in Sport - ISBS 2017 – Cologne, Germany.

14. Beaulé PE, Catelli DS, Kowalski E, Lamontagne M. (2017) La prothèse de la Hanche à Double Mobilité Produit-elle une Meilleure Mécanique Articulaire? In: *92e Congrès de la Société Française de Chirurgie Orthopédique et Traumatologique* (SOFCOT), Paris, France.

15. Lamontagne M, Catelli DS, Kowalski E, Beaulé PE. (2017) Kinematic Analysis of Dual-Mobility vs Single-Bearing Hip Prosthesis during Squatting Task using Statistical Parametric Mapping. In: Canadian Orthopaedic Association - COA 2017, Ottawa, Canada.

16. Catelli DS, Kowalski E, Phillips C, Beaulé PE, Lamontagne M. (2017) Femoroacetabular Impingement Corrective Surgery Does Not Affect Ground Reaction Force Symmetry During Squat Tasks. In: 18th European Federation of National Associations of Orthopaedics and Traumatology Congress - EFORT 2017, Vienna, Austria.

17. Kowalski E, Catelli DS, Ng KCG, Lamontagne M, Beaulé PE. (2016) Comparing Hip Joint Mechanics in Individuals with a Symptomatic or Asymptomatic Cam Deformity during a Stair Ascent Task. In: 17th European Federation of National Associations of Orthopaedics and Traumatology Congress - EFORT 2016, Geneva, CH.

18. Tulchin-Francis K, Backus A, Lamontagne M, Lewis C, Podeszwa D, Edelstein J, Catelli DS, Anderson A. (2016) Functional Outcomes Assessment in Patients with Femoral Acetabular Impingement. In: Gait & Clinical Movement Analysis Society – 2016 Annual Conference, Tutorial #2, Memphis, USA.

19. Catelli DS, Reynolds S, Ng KCG, Lamontagne M, Beaulé PE. (2015) Biomechanical Analysis between Dual Mobility and Conventional THA during Stair Ascent and Descent. In: XXV Congress of the International Society of Biomechanics, Glasgow, UK.

Posters

20. Catelli DS, Kowalski E, Carsen, S, Beaulé PE, Lamontagne M. (2018) Muscle Activity during Deep Squat in Patient with Femoroacetabular Impingement after Corrective Surgery. In: 2018 International Society for Hip Arthroscopy – ISHA Annual Meeting, Melbourne, Australia.

21. Catelli DS, P.E. Beaulé, M. Lamontagne. (2018) Can we develop a biomechanical functional

score to quantify the joint mechanics of THA patients? In: European Orthopaedic Research Society – EORS 2018, 26th Annual Meeting, Galway, Ireland.

22. Lamontagne M, Kowalski E, Catelli DS, Dervin G. (2018) Do Constant Radius Medial Bering Implants Provide Symmetrical Biomechanics During Walking Following On Year of Surgery? In: 19th European Federation of National Associations of Orthopaedics and Traumatology Congress - EFORT 2018, Barcelona, Spain.

23. Catelli DS, Mantovani G, Le A, Beaulé PE, Kowalski E, Lamontagne M. (2018) Effect of Stiff Hip Flexors on High Contact Forces during Gait: In-silico Analysis. In: Orthopaedic Research Society 2018 Annual Meeting, New Orleans, USA.

24. Catelli DS, Le A, Kowalski E, Beaulé PE, Lamontagne M. (2018) EMG Frequency Analysis and Isometric Hip Strength in Femoroacetabular Impingement after Corrective Surgery. In: Orthopaedic Research Society 2018 Annual Meeting, New Orleans, USA.

25. Kowalski E, Catelli DS, Lamontagne M. (2018) Does the Inter-Limb Variability Change with Ages in Healthy Participants during Gait? In: Orthopaedic Research Society 2018 Annual Meeting, New Orleans, USA.

26. Kowalski E, Catelli DS, Lamontagne M. (2018) Side Does Matter: Comparison of Different Limb-Matching Methods during Gait in Healthy Participants. In: Orthopaedic Research Society 2018 Annual Meeting, New Orleans, USA.

27. Kowalski E, Catelli DS, Vu B, Phillips C, Lamontagne M. (2017) EMG Onset Detection Methods for Various Signal to Noise Ratios: Visual or Computadorized Detection. In: 25th Annual and Anniversary Meeting of the European Orthopaedic Research Society (EORS), Munich, Germany.

28. Kowalski E, Vu B, Phillips C, Catelli DS, Lamontagne M. (2017) Using Computer Simulated EMG Signals to Improve Visual Onset Detection. In: XXVI Congress of the International Society of Biomechanics, Brisbane, Australia.

29. Lamontagne M, Catelli DS, Kowalski E, Beaulé PE. (2017) Comparison of Muscle Activity between Symptomatic and Asymptomatic Individuals with Cam Femoroacetabular Deformity during a Squatting Task. In: Orthopaedic Research Society 2017 Annual Meeting, San Diego, USA.

30. Catelli DS, Kowalski E, Lamontagne M, Beaulé PE. (2017) Does Corrective Surgery in Femoroacetabular Impingement Improve Joint Kinematics During Squatting? In: Orthopaedic Research Society 2017 Annual Meeting, San Diego, USA. ORS 2017 Annual Meeting, San Diego, USA.

31. Kowalski E, Catelli DS, Beaulé PE, Lamontagne M. (2017) Going Beyond Patient Reported Outcome Measures (PROMs) In Assessing Surgical Management of Femoroacetabular Impingement. In: Orthopaedic Research Society 2017 Annual Meeting, San Diego, USA.

32. Lamontagne M, Catelli DS, Kowalski E, Dervin G. (2016) Posterior Stabilizer Implants Result in Increased Quadriceps Muscle Activity Compared to Medial Pivot Implants during a Stair Ascent Task. In: 2016 Transatlantic Orthopaedic Congress. New York, USA.

33. Catelli DS, Ng KCG, Lamontagne M, Beaulé PE. (2016) Ground Reaction Force Symmetry during Sitting and Standing Tasks after a Dual Mobility or Conventional Cup Total Hip Arthroplasty. In: 17th European Federation of National Associations of Orthopaedics and Traumatology Congress - EFORT 2016, Geneva, CH.

34. Lamontagne M, Ng KCG, Catelli DS, Beaulé PE. (2016) Do Anatomical Parameters of Cam FAI Influence Hip Joint Mechanics during Level Walking? In: Canadian Orthopaedic Association – COA 2016, Québec City, Canada.

35. Catelli DS, Kowalski E, Lamontagne M, Beaulé PE. (2016) Comparison of Biomechanical Functional Score between Dual Mobility and Conventional Bearings in Total Hip Arthroplasty Patients during Squat Task. In: Orthopaedic Research Society 2016 Annual Meeting, Orlando, USA.

36. Kowalski E, Catelli DS, Lamontagne M, Beaulé PE. (2016) Asymmetry in Patients with a Dual Mobility Total Hip Arthroplasty during Inclined Walking: A Randomized Clinical Trial. In: Orthopaedic Research Society 2016 Annual Meeting, Orlando, USA.

37. Catelli DS, Kowalski E, Lamontagne M, Beaulé PE. (2016) Hip and Pelvis Mechanics and Asymmetry in Cam Femoroacetabular Deformity Patients during Level Gait. In: Orthopaedic Research Society 2016 Annual Meeting, Orlando, USA.

38. Catelli DS, Little R, Lamontagne M, Beaulé PE. (2015) Squat Asymmetry in Patients with Total Hip Arthroplasty: Comparison between Dual Mobility and Standard Prostheses. In: XXV Congress of the International Society of Biomechanics, Glasgow, UK.

39. Catelli DS, Reynolds S, Lamontagne M, Beaulé PE. (2015) Gait Asymmetry in Patients with a Dual Mobility Total Hip Arthroplasty: a Randomized Clinical Trial. In: Orthopaedic Research Society 2015 Annual Meeting, Las Vegas, USA.

40. Catelli DS, Mantovani G, Lamontagne M. (2014) Comparison between EMG Electrodes Placement Using the SENIAM Guidelines and Electrical Stimulation. In: 7th World Congress of Biomechanics, Boston, USA.

41. Mantovani G, Catelli DS, Lamontagne M. (2014) Electromyography Activation Values Measured Using Free Set-up and Isokinetic Device During Maximal Voluntary Isometric Contraction. In: 7th World Congress of Biomechanics, Boston, USA.

Non-Refereed

42. Catelli DS, Lamontagne M. (2018) What the Orthopaedic Surgeon Should Know about Gait and Motion Analysis? In: 31st Annual Tom Smallman Canadian Orthopaedic Association (COA) Basic Science Course. Ottawa Conference and Event Centre, Ottawa, Canada.

43. Catelli DS, Ng KCG, Kowalski E, Beaulé PE, Lamontagne M. (2018) Hip Muscle and Contact Forces in Post-Surgical Cam FAI. In: Proceeding of Hans K. Uhthoff 2018. The Ottawa Hospital, Ottawa, Canada.

44. Catelli DS, Kowalski E, Lamontagne M, Beaulé PE. (2017) In: Proceeding of Medical Devices Innovation Institute and CREATE-BEST 2017 Poster Day. University of Ottawa, Ottawa, Canada.

45. Catelli DS, Kowalski E, Philips C, Beaulé PE, Lamontagne M. (2017) Femoroacetabular Impingement: Does Corrective Surgery Affect Ground Reaction Force Symmetry during Squat Tasks? In: Proceeding of Hans K. Uhthoff 2017. The Ottawa Hospital, Ottawa, Canada.

46. Catelli DS, Ng KCG, Lamontagne M, Beaulé PE. (2016) Biomechanical Analysis between Dual Mobility and Conventional THA during Stairs Tasks. In: Proceeding of HKGSA Conference 2016. The Ottawa Hospital, Ottawa, Canada.

47. Catelli DS, Ng KCG, Lamontagne M, Beaulé PE. (2016) Biomechanical Analysis between Dual Mobility and Conventional THA during Stair Tasks. In: Proceeding of Hans K. Uhthoff 2016. The Ottawa Hospital, Ottawa, Canada.

48. Kowalski E, Catelli DS, Ng KCG, Lamontagne M, Beaulé PE. (2016) Comparing Hip Joint Mechanics in Individuals with a Symptomatic or Asymptomatic Cam Deformity during Stair Ascent Task. In: Proceeding of Hans K. Uhthoff 2016. The Ottawa Hospital, Ottawa, Canada.