Accurate Measurement of Healthy Joint Kinematics to Inform Diagnosis and Treatment

Vasiliki Kefala
Accurate Measurement of Healthy Joint Kinematics to Inform Diagnosis and Treatment

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Vasiliki Kefala

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Advisor: Dr. Kevin B. Shelburne
Abstract

The description of human motion has a primary importance in different scientific areas such as medicine, sports, physical therapy. Kinematics specifically studies pure motion without reference to the causes of motion such as forces. Understanding the kinematics of human movement is of critical importance in medicine and biology. Motion measurement can be used in order to evaluate functional performance of limbs under normal and abnormal conditions. Kinematic knowledge is also important for diagnosis and surgical treatment of joint disease and the design of implants to rehabilitate function. Accurate joint kinematics is essential to protect articular functionality. An alteration may change the transmission of physiological loads, which could lead to degenerative arthrosis from compartmental overload. Thus, accurate measurement of healthy joint motion is needed to establish baseline kinematics and clinical parameters for assessment of natural joint function, diagnosis of pathology, design of treatments, and evaluation of patient outcomes. The main aim of human motion analysis is the description of joint kinematics during daily living activity. Accurate quantification of hip/knee kinematics during activities of daily living and differing demand is essential since joint kinematics during functional tasks are influenced by external forces, joint position and the balance of active and passive contributory forces across the joint of interest. Age range has also a significant impact on joint kinematics. Currently, it is unclear what aspects of the kinematic changes appearing with osteoarthritis are the result of the disease or part of natural aging. To our knowledge, no others have evaluated normal knee function for a cohort age matched to total knee arthro-
plasty (TKA) recipients and during activities that patients with TKA often report to be troublesome, such as descending a step and executing a turn during walking. Most descriptions of knee kinematics have been for younger adults and for a limited span of activities. Additionally, quantitative data of total knee arthroplasty kinematics is crucial for the evaluation of the component failure and for providing guidelines for further advancement of the implant design. TKA is a regular surgical procedure to alleviate pain and restore knee function. Successful functional outcome following TKA is influenced by the geometry and design of the components as well as their interaction with the soft tissue surrounding this articulation. Thus, understanding the effect of design choices on in vivo kinematics and during different dynamic activities of daily living has become more essential since the connection between knee prosthetic kinematics and clinical performance is clearly increasing. Finally, to best of our knowledge no others have investigated and compared the 3D pelvic functional orientation across different populations that include healthy subjects, subjects that have undergone total hip arthroplasty (THA) and spinal-stabilized cohorts and during different static and dynamic activities. Furthermore, most studies have performed their measurements in static settings whereas the pelvic motion is dynamic. The functional orientation of the pelvis varies during different dynamic activities and the pelvis is not a fixed static bone when considering acetabular cup placement. This knowledge will help us to better understand the behavior of all spinopelvic parameters and aid decisions regarding acetabular component alignment. Differences in spinopelvic parameters across different patient populations and across static and dynamic activities are necessary to understand for accurate positioning of the acetabular component during total hip arthroplasty and reduce the likelihood of impingement events.
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Chapter 1

Introduction

"In all things of nature there is
something of the marvellous"

___________________________
Aristotle

Biomechanics, (from Ancient Greek: βίος "life" and μηχανική "mechanics"), is the study of the movement of living things using the science of mechanics that is related to the description of motion and how forces generate motion to living organisms, such as humans, animals, plants and the functional basic units of life, the cells [8]. It is now widely recognized that biomechanics plays an important role in understanding of the fundamental principles of human motion; however, it is an area of study that has a very long history. Biomechanics as a branch of science can be traced back as far as the days of the Greek philosophers Plato and Aristotle in the fourth century B.C. Aristotle wrote the first treatise on movement in animals that analyzed their bodies as mechanical systems and might be considered the first biomechanician. Socrates, who was born 2400 years ago, instructed that we could not begin to comprehend the world around us until we understood our own nature. Who among us has never marveled at the beautiful motions of a dancer or the fast finger movements of a musician? From the time of Aristotle onward there have been
adequate literature on the topic of movement in both animals and humans. Despite the
great minds that have studied the topic, it is only recently that much progress has been
made experimentally.

The description of human motion has a primary importance in different scientific areas
such as medicine, sports, physical therapy. Kinematics specifically studies pure motion
without reference to the causes of motion such as forces. Understanding the kinematics
of human movement is of critical importance in medicine and biology. Motion measure-
ment can be used in order to to evaluate functional performance of limbs under normal
and abnormal conditions. Kinematic knowledge is also important for diagnosis and surgi-
cal treatment of joint disease and the design of implants to rehabilitate function. Accurate
joint kinematics is essential to protect articular functionality. An alteration may change
the transmission of physiological loads, which could lead to degenerative arthrosis from
compartmental overload. Thus, accurate measurement of healthy joint motion is needed to
establish baseline kinematics and clinical parameters for assessment of natural joint func-
tion, diagnosis of pathology, design of treatments, and evaluation of patient outcomes.

Musculoskeletal simulation offers valuable data to researchers and clinicians to better
evaluate pathological conditions and to better understand human movement. Whole-body
movement is regularly obtained through the use of the musculoskeletal modeling software
platforms and is used to estimate joint kinematics and moments, as well as intersegmental
joint loads and muscle forces. Accuracy in the joint kinematic data is crucial since they
are input into those musculoskeletal models and accurate joint kinematics along with the
quadriceps force predictions from musculoskeletal modeling are input into finite element
analyses to simulate activities and predict joint mechanics.

There are three different approaches towards the investigation of human joint kine-
matics. Researchers have assessed human joint motion with in vitro (cadavers) [9, 10],
noninvasive (gait laboratories) [11, 12] and in vivo (roentgen stereophotogrammetry and
fluoroscopy) [13, 14] experiments. The in vitro approach offers the analysis of bone, ligaments, and muscles in cadaveric specimens, however, this approach is costly, and is difficult to apply experimental loading conditions to replicate conditions experienced in vivo and thus can only be used to investigate passive properties of the human body. The most commonly used methods for assessing human joint movement rely upon skin-mounted but they include errors due to soft-tissue artifacts [15]. In vivo measurements of small translations and rotations of the joint is a difficult task that requires advanced radiographic imaging techniques such as dual plane fluoroscopy. Dual plane fluoroscopy allows real-time tracking of joint motion by capturing X-ray videos with precise measurement of six Degree of Freedom (DoF) joint kinematics during different tasks of daily living.

1.1 Motivation and Problem Statement

The knee is the largest joint in the body and having healthy knee joints is needed to carry out most activities of daily living. Additionally, the hip joint is also one of the body’s largest weight-bearing joints, only secondary to the knee joint, that allows motion and gives stability that is required to bear body weight.

Osteoarthritis (OA) is the most regular form of arthritis and a primary cause of disability all over the world, largely due to pain which is the main symptom of the disease. It is the most frequent form of joint disorder in the United States, and it is calculated that more than 27 million Americans are affected [16]. As a degenerative disorder, OA can involve any joint, and it mainly has an effect on the articular cartilage and surrounding tissues. The knee is the most common joint affected from OA with up to 41% of limb arthritis being located in the knee and 19% in hips [17]. The disease development and progression is affected by abnormal joint kinematics under weight bearing conditions. Total Knee Arthroplasty (TKA) and Total Hip Arthroplasty (THA), which are the most common orthopaedic knee and hip
surgeries, are an extremely effective treatment for functional disability. Additionally, these surgeries to overcome pain caused from diseases, like end stage arthritis, from injury or aging. This therapy is employed to replace a deformed joint with a knee or hip prosthesis. The incidence of joint replacement procedures is high, with over 1 million total hip and total knee replacement procedures performed each year in the United States [18]. Thus, precise knowledge of natural knee and hip mechanics provides useful metrics for comparison to knee and hip function following TKA and THA.

The main aim of human motion analysis is the description of joint kinematics during daily living activity. Accurate quantification of hip/knee kinematics during activities of daily living and differing demand is essential since joint kinematics during functional tasks are influenced by external forces, joint position and the balance of active and passive contributory forces across the joint of interest. Furthermore, comparing across higher demand is justified by the need to reveal differences in knee or hip kinematics through activities more challenging to the degrees of freedom of the joint of interest. This investigation of the characteristic joint motion is important for assessment and could also enhance the optimization of implant function that intends to restore joint alignment and to achieve full range of knee flexion. Finally, these data will be useful to compare with and better understand knee kinematic changes in pathologies such as OA, and in repair, such as TKA or THA.

Age range has also a significant impact on joint kinematics. Currently, from studies such as by Hamai et al. [19] it is unclear what aspects of the kinematic changes appearing with OA are the result of the disease or part of natural aging. However, prevalence of OA increases with age. In fact, it is estimated that 1-3% of the older adult population will undergo THA or TKA at some point with an average of 66 years old [20]. Additionally, older adults adapt different movement strategies since they tend to slow their pace and use shorter strides while walking and use unique movement strategies during activities such as stepping-down or turning and pivoting [21, 22, 23]. How normal joint kinematics change
with demand warrants further scrutiny as norms of healthy joint kinematics in older adults may be dominated by intersubject differences. Thus, documenting joint kinematics in older adults is essential for understanding the changes that occur in knee or hip kinematics with OA, TKA and THA.

To conclude, measurement of dynamic joint kinematics in vivo during activities of daily living is important to understand the effects of joint injuries and diseases as well as for evaluating the treatment effectiveness. Documenting normal changes in joint kinematics with age is necessary for understanding the changes in knee or hip kinematics since the motion of younger subjects may not be representative of the age range associated with knee pain, OA, and TKA. Finally, precise knowledge of normal joint mechanics provides useful metrics for comparison to joint function following TKA or THA.

Accurate joint kinematics during dynamic activities are achieved by using fluoroscopy and radiography systems and in our lab we have developed the High Speed Stereo Radiography (HSSR) system which enables the precise measurement of six DoF joint kinematics during different tasks of daily living. HSSR is synchronized with conventional motion capture which is the established standard for the quantification of human movement and allows collecting whole body motion.

### 1.2 Objectives

The objective of this dissertation was to accurately measure joint kinematics during activities of daily living for informing diagnosis, helping design of treatments and improve the design of prosthetic devices. This was achieved by focusing on obtaining healthy data during different activities of daily living, by recruiting people of the appropriate age and finally by assessing data that supports computer models. More specifically the first specific aim was to provide comprehensive patient specific measurements of patellofemoral mo-
tion and patellar tendon mechanics during gait activities in healthy older adults since the patellofemoral joint and quadriceps mechanism are susceptible to many pathologies resulting from acute injury (e.g. dislocation) and chronic disease (e.g. osteoarthritis). For my second specific aim, differences in tibiofemoral and patellofemoral joint kinematics, and patellar tendon mechanics between weight bearing and non-weight bearing activities were investigated due to the fact that patellofemoral dysfunctions have been linked with combined abnormal motion of the tibia and patella relative to the femur. The kinematics of the tibiofemoral and patellofemoral joints can change as subjects change demand on their knee and move from non-weightbearing to weightbearing activities. The amount of kinematic change with weightbearing may vary dramatically between individuals, confounding attempts to generate helpful predictions of weightbearing kinematics. The third specific aim was to discover whether notable differences in mobile and fixed bearing kinematics occur during activities of daily living and compare these results with healthy knee kinematics. Successful functional outcome following TKA is influenced by the geometry and design of the components and the available components for knee arthroplasty are routinely offered in Rotating Platforms (RP) and Fixed-Bearing (FB) alternatives. The choice between RP or FB components remains unclear due to uncertainty in the actual kinematic distinction between the two alternatives in vivo. The final specific aim was to establish functional pelvic orientation that will help improve acetabular cup positioning during total hip arthroplasty (THA). Acetabular component positioning is directly related to the incidences of hip dislocation and wear, the two most common causes of THA failure. One of the factors to consider for optimization of cup orientation on patient specific basis is the position of the pelvis in functional positions.
1.3 Dissertation Overview

Chapter 2 provides a literature review where background information on methods used to measure natural knee and hip joint motion are presented and compared with other existing methods in research and concludes with discussion on how our work adds to the literature.

Chapter 3 describes the Measurement of Patellofemoral Kinematics in Older Adults during Gait Activities whose objective was to describe case series measurements of patellar motion in healthy older adults as they performed three gait activities, determine patellar tendon angle and moment arm, and show if these quantities were activity dependent.

Chapter 4 presents Changes in Natural Tibiofemoral, Patellofemoral, and Patellar Tendon Kinematics with Weightbearing and discusses the combined measurement of the differences in tibiofemoral and patellofemoral kinematics, patellar tendon angle, and patellar tendon moment arm that occur during non-weightbearing and weightbearing activities.

Chapter 5 presents In Vivo Comparison of Rotating Platform and Fixed Bearing Knee Replacements During Lunge and Pivot Activities whose objective was to discover whether notable differences in mobile and fixed-bearing kinematics occur during activity that promotes tibial rotation, and to compare these results with normal healthy kinematics.

Chapter 6 presents Functional Pelvic Orientation for Surgical Planning in Total Hip Arthroplasty whose objective was to establish the functional orientation of the pelvis since it is the most important variable for the surgeon to consider when determining the proper goal for acetabular component orientation.

Finally, Chapter 7 analyzes the contributions to the field of experimental biomechanics made by this dissertation and proposes direction for future work in this area of expertise.
Chapter 2

Background Information and Literature Review

2.1 Natural Joint Motion

Natural joint motion analysis is aimed at providing quantitative knowledge affiliated to the mechanics of the musculoskeletal system while different dynamic activities are being performed, with a particular focus on the relative movement between adjacent bones or joint kinematics. Kinematics is the measurement of motion, that include displacements, velocities, and accelerations during a specific movement. Human motion analysis became one of the most used approaches for evaluating musculoskeletal system physiopathology, and its use keeps growing. Thus, accurate knowledge of natural human joints kinematics, including 3D rigid body and surface kinematics, is important for comprehending of their function and for many clinical applications. Evaluating joint motion characteristics is beneficial for a proper clinical understanding of complex functional movements during activities of daily living. Studying joint motion can assist in the evaluation of the patients’
functionality and progression in their rehabilitation period and for developing strategies to alleviate joint pain. Therefore, an accurate method for studying joint kinematics is needed.

2.2 Methods for Studying Joint Motion

The evolution of several methods for the capture of human joint motion has been driven by the need for new information on the characteristics of normal and pathological human joints. Different lab-based methods are currently available for researchers to quantify joint motion involving in vitro, in vivo, and computational methodologies. In vitro approach allows for the study of tissues on cadaveric specimens while the in vivo investigations are being performed on live subjects. Experimental data obtained from these methods, using computational help, will aid in the development of more complex musculoskeletal models, which may be utilized to evaluate joint function and disease and design surgical interventions and post-operative rehabilitation strategies. Computational modeling can complement experimental studies by enabling prediction of joint loads, contact mechanics, and internal stress/strains, which would otherwise be challenging to measure experimentally.

2.2.1 In vitro joint motion analysis

In vitro joint motion analysis allows for the analysis of bone, ligaments, and muscles in cadaveric specimens. In-vitro studies allow direct measurement of the internal loads and soft tissue forces and mechanical testing can be used to specify the exact response of tissue or joints to precise loading and unloading conditions. The main advantage of the in vitro approach is that any equipment to obtain data that would be invasive on a living subject can be used with no restriction on cadaveric specimen, thus, internal quantities can be acquired during a cadaveric experiment. Additionally, surgical treatments can be investigated without any concern of causing damage or injury to a patient. For example, the implantation
of a TKA on a cadaveric limb can the relationship between the device and the remaining structures such as the ligaments. However, in vitro studies often do not accurately reflect loads that occur during characteristic movements since typically an idealized set of loading and boundary constraints are applied that may not be representative of physiological loading conditions. More specific, even though material properties and forces exerted by passive tissues like ligaments can be calculated with mechanical testing and strain-gauge transducers, active structure’s (muscles) contribution is not available. Testing machines have been developed [24] in order to simulate daily activities such as squatting or gait but still an in vitro experiment can not provide the necessary information regarding the muscle coordination that is mandatory in order to perform a daily activity. Furthermore, in vitro analysis can be expensive and time-consuming.

2.2.2 In vivo joint motion analysis

In-vivo joint motion analysis is a great source of experimental data since they can be obtained from subjects that are alive and can be asked to perform complex motions and tasks of every day life. The most common approach for evaluating joint motion in vivo includes video-based motion capture with reflective markers fixed on the skin. Reconstruction of human movement based on skin mounted reflective markers has become a standard procedure in clinical practice. However, movement of the skin with respect to bone creates uncertainty in the measured joint kinematics, with errors more than 20 mm reported for the actual underlying kinematics and task-displacements of individual skin-mounted markers relative to the underlying bone at the knee [15, 25]. To minimize the inherent inaccuracy of skin-mounted markers, markers have been mounted on pins and inserted directly into the bones [26, 27, 28]. Although this approach can provide high-quality kinematics data,
intraâcortical pins cause pain and impingement and has limited its application in human movement studies.

Skin motion artifact can be avoided by using medical imaging techniques such as fluoroscopy. Newer technologies, such as cine magnetic resonance imaging (MRI) and single- and biplane fluoroscopy or stereo radiography, are capable of accurately recording 3-D joint motion non-invasively. MRI allows assessing movement of the underlying bone directly, however, MRI is not yet capable of achieving high frame rates required for evaluating dynamic function. Furthermore, this method is expensive and the restriction it imposes on the movements (typically a small-diameter cylindrical space) prevents from obtaining full motion kinematics measurement during functional activities like gait [29]. Finally, MRI is not the ideal option to use with subjects that have prosthetic parts in their bodies because of the detrimental effects of large metal objects on both formalities.

Evaluating in vivo dynamic joint kinematics in 6 DOF with accuracy and precision sufficient to detect small changes in position and orientation is technically challenging and advanced radiographic imaging techniques such as dynamic radiography and fluoroscopy is required. Dynamic X-ray imaging (or fluoroscopy) enables accurate, nonâinvasive measurement of bone motion in vivo. Single-plane X-ray fluoroscopy allows direct visualization of the underlying bones and has been used to track bone movements in both healthy and reconstructed joints [30, 31, 32]. However, this method is limited to two dimensional assessment and is imprecise in the out of plane direction [33, 34, 31]. Those limitations may be overcome with stereo radiography. Stereo radiography (or biplane fluoroscopy) enables accurate quantitative assessment of 3-D joint motion with sub millimeter and sub degree accuracy.

However, generally in vivo studies demonstrate some disadvantages that may reduce the potential of in vivo investigations. Even though external body segment kinematics can be measured reliably, internal quantities of interest cannot be quantified in a living subject.
For example, internal loads like ligament and muscle forces are not accessible in vivo and the subject-specific interaction between them and body segment kinematics is still unclear.

2.3 Stereo Radiography

Joint motion analysis using stereo radiography imaging has recently become widespread in the field of biomechanics, yielding significant results. Stereo radiography enables accurate quantitative assessment of three dimensional (3D) joint motion, six degree of freedom (DOF), translation and rotation with sub-millimeter and sub-degree accuracy [35]. It is used to characterize kinematics in healthy normal and symptomatic population and can be adapted to measure high-speed human joint motion in vivo during activities of daily living. In this method, radiographic images are captured in two different planes simultaneously and a sequence of fluoroscopic images of the joint of interest is collected while the subject performs a specific motion task. For each frame, the 3D pose of the joint is obtained by aligning a 3D model of each segment to its two-dimensional (2D) projection in the corresponding image known as 2D-to-3D image registration. The movement is then reconstructed from the sequence of registered poses.

Since stereo radiography can capture six DOF joint motion with high accuracy during different dynamic tasks of daily living it has many applications in orthopaedics, bioengineering and sports medicine. Accurate assessment of joint motion derived from this technique helps evaluating joint function and improving the design of an implant. Moreover the stereo radiography kinematic data could also be used as input to assess deformation of joint structures, to obtain contact stress distributions and for several modeling applications. This knowledge is also crucial for planning surgeries or rehabilitation therapies.

Dynamic kinematic measurements of different joints of the body such as the human knee [36, 37, 38, 39], shoulder [40, 41], spine [42], foot [43] and hip [44] and during
different motor tasks, have been conducted using stereo radiography to capture precise joint kinematics in vivo.

Since the field of view in dual-plane stereo radiography systems is limited, a combination of motion capture systems and stereo radiography are the recommended methods for in-vivo evaluations [45, 15, 46].

2.4 Natural Knee Joint Motion

The knee joint, as the main lower limb motor joint, is the most vulnerable and susceptible joint [47]. The knee impairments can considerably affect the normal daily living routine and mental health of subjects [48]. Thus, deep understanding of the biomechanics of a normal and diseased knee joint is essential for designing knee prosthetic devices, selection of the proper TKA and for optimizing a rehabilitation therapy.

Knee joint anatomy and kinematics are complex. The knee joint is more complicated than a simple hinge joint and moves with a complex set of translations and rotations. It consists of three bones and two joints: the femur and tibia articulate at the tibiofemoral joint (TF) while the femur and patella, at the patellofemoral joint (PF) (Figure 2.1). Both joints must tolerate high contact forces during daily activities such as gait [49]. These two joints work together to form a modified hinge joint that exhibits six degrees of motion during dynamic motor tasks that allows the knee to bend and straighten as well as to rotate slightly from side to side. These six degrees of freedom consist of three rotations (flexion/extension, external/internal and varus/valgus rotation) and three translations (anterior/posterior, medial/lateral, superior/inferior). The three translation DOFs are meaningfully limited by the fibrous capsule, ligaments and muscles. The rotation with the greatest range of motion is around the sagittal plane (flexion and extension), while the varus/valgus and the external/internal rotation around the frontal and transverse are also more restricted and they are
influenced by the position of the joint in the sagittal plane [50]. This phenomenon is called 'screwhome' [51] and rotation depends on the degree of flexion.

Extension is the movement which causes the leg to straighten. Flexion is the movement when the calf touches the posterior thigh. So the range of motion from full extension to full flexion is from 0° to 140° [50]. Extension is the motion which causes the leg to straighten while flexion is the motion when the calf is in contact with the posterior thigh. The range of motion from full extension to full flexion is from 0° to 140° [50] and this range can be changed according to the activity. During gait, the flexion/extension ranges from 0° to 60° while for climbing and descending/ascending stairs and pivoting it ranges from 0° to about 90° [50]. The mechanism of knee flexion indicates a combined movement of rolling, gliding and rotation of the femoral condyles over the tibial plateaus [52] and this combined motion, called rollback, permits a broad rotation on the sagittal plane.

Standard coordinate systems for the knee joint are based on mechanical or anatomical axes. Grood and Suntay, in 1983, created a joint coordinate system providing a geometric description of the three-dimensional rotational and translational movement between two rigid bodies, applied to the knee joint [1] (Figure 2.2). With this approach, the estimated joint displacements became independent of the order in which the components’ translations and rotations occur.

Over the past decade, stereo radiography of in vivo dynamic knee joint motion has arisen and has provided new information on normal and pathological knee movement. Precise measurement of healthy tibiofemoral (TF) and patellofemoral (PF) joint motion is crucial for understanding variations of knee motion in patients with knee pathology and evaluating the result of conservative or surgical treatments such as TKA. Finally, the kine-matics of the tibiofemoral and patellofemoral joints are mightily interdependent, in a way that injury and/or altered motion in the tibiofemoral joint can influence the patellofemoral mechanics and vice versa [53].
Normal knee anatomy. In a healthy knee, these structures work together to ensure smooth, natural function and movement.

Figure 2.1: Normal knee anatomy

Figure 2.2: Coordinate systems used to define the relative positions and orientations of the femur, tibia, and patella of a right knee based on Grood and Suntay approach [1], [2].
2.4.1 Tibiofemoral Joint Kinematics

A complete knowledge of tibiofemoral joint kinematics is crucial for evaluating the function of the normal and pathological joint. It is essential to understand tibiofemoral joint kinematics to diagnose the early onset of osteoarthritis and develop the necessary therapeutic or surgical treatments [54]. Moreover, the results obtained from the kinematic analysis of the TF joint can be utilized in the finite element analysis to comprehend the contact mechanics of the tissues in the knee joint under loading conditions.

Several researchers have reported precision kinematics for the tibiofemoral joint [55, 56, 57, 58, 26] and during different activities of daily living such as a non-weightbearing baseline knee extension [56] and during dynamic weightbearing activities such as normal gait [36, 58], landing [59, 56], step-up activity [60] and lunge [61]. The kinematics of the tibiofemoral joints can change as subjects change demand on their knee and move from non-weightbearing to weightbearing activities. For example, Myers et al [56] measured an increase in anterior displacement of the tibia relative to the femur as demand (defined as increasing knee extensor moment) on the knee increased kinematic changes during weightbearing may alter surgical repairs to the knee made with the knee unloaded. Knowledge of how knee kinematics change with demand can provide kinematic targets for treatments that seek to restore normal function.

Definitions of natural healthy knee motion are subject to the kinematic variability between individuals. While knee kinematics normally change with demand, a description of average motion may still poorly represent the knee motion of an individual. Significantly, treatments that seek to restore normal motion to the knee might be confounded by patient-specific knee motion that may routinely fall outside population norms. Natural knee motion is achieved by different knee motion profiles in different individuals. What’s more, subjects with consistently greater or lesser motion of the TF joint may have corresponding
greater or lesser motion of the PF articulation. While it is important to measure how normal knee kinematics change with demand, these changes warrant further scrutiny as norms of healthy knee kinematics may be dominated by intersubject differences.

**Change in Kinematics with Pathology**

Additionally, several studies have obtained 3D in vivo tibiofemoral kinematics in knees with osteoarthritis under weight-bearing conditions [62, 63, 64, 19]. Acute medial OA knees were demonstrated to have notably less knee extension, less posterior femoral rollback, less tibial internal rotation with flexion, and larger medial shift of the femur when compared with normal knees during knee flexion-extension tasks [63, 19] (Figure 2.3). Hence, kinematic analyses of healthy and OA knees immediately before TKA is crucial to gain a better understanding of pathological and reconstructed knees.

![Figure 2.3: Knee osteoarthritis illustration.](image)

**Change in Kinematics with Age**

Tibiofemoral kinematics may also alter with age. Several prior studies have evaluated young healthy, OA, and TKA subjects, and shown important differences between tibiofemoral kinematics across these cohorts. However, with a few notable exceptions [65], these studies routinely compare older adults with OA and TKA to young healthy controls.
The motion of younger subjects may not be representative of the age range associated with knee pain, OA, and TKA. Older adults tend to use unique movement strategies during activities such as stepping-down or turning and pivoting [21, 22, 23]. While these studies have noted that aging has an impact on knee kinematics measured using marker-based motion capture, few studies have examined the small translations and rotations of the knee in older adults with no history of knee pathology. Documenting knee kinematics in older adults is necessary for understanding the changes in knee kinematics that occur with OA and TKA.

**Change in Kinematics with Joint Replacement**

Quantitative data of total knee arthroplasty (TKA) kinematics is crucial for the evaluation of the component failure and for providing guidelines for further advancement of the implant design. TKA is a regular surgical procedure to alleviate pain and restore knee function. TKA is most regularly executed by substituting damaged bone and cartilage on the articulating surfaces with a combination of metal and polyethylene/ceramic components (Figure 2.4). Successful functional outcome following TKA is influenced by the geometry and design of the components as well as their interaction with the soft tissue surrounding this articulation.

It has been demonstrated that the kinematics of prosthetic knees are unlike the normal knees, and necessitate excessive sliding and rotational motions which may cause high shear stresses at the joint interface. Price et al [66] compared sagittal plane kinematics between normal and TKA patients, and found significantly altered TF kinematics after TKA.

Furthermore, significant conclusions are that there is a broad range of kinematics and that particular implant designs have particular advantages and disadvantages [11, 67]. Wear has been reported to be one of the crucial factors limiting the long-term success of the knee implants in the past [68]. To reduce wear, mobile bearing knees have been introduced as an alternative choice to fixed bearing TKAs in an effort to achieve tibiofemoral kinematics
that better matched healthy function, particularly in tibial rotation. Several studies have investigated the differences between different designs [69, 70, 71]. However, prior work has focused on sagittal plane activities and ignored turning motions likely to test the axial rotation kinematics of a mobile bearing design.

Thus, understanding the effect of design choices on in vivo kinematics and during different dynamic activities of daily living has become more essential since the connection between knee prosthesis kinematics and clinical performance is clearly increasing.

![Total knee replacement illustration.](image)

Figure 2.4: Total knee replacement illustration.

### 2.4.2 Patellofemoral Joint Kinematics

The patellofemoral (PF) joint is susceptible to many pathologies resulting from acute injury (e.g. dislocation) and chronic disease (e.g. osteoarthritis). PF joint motion is clinically significant since many cases with patellofemoral dysfunctions are related to the abnormal motion of the patella relative to the femur, or patellar maltracking [72, 73]. Abnormal patella tracking is assumed to change the mechanical interaction between the patella and femur, and may lead to cartilage degradation and osteoarthritis. In addition, PF maltracking (often consisting of lateral patellar displacement and tilt) may be a key cause of patellofemoral pain [74, 75]. PF related pathologies in the normal knee, like joint laxity, patellar maltracking, cartilage degeneration and anterior knee pain affect approximately
affect approximately 25% of the population [74]. PF pathology and maltracking are frequently addressed with a variety of conservative and surgical treatments. Surgical treatment may consist of tibial tubercle osteotomy to reposition the insertion of the patellar tendon and the patella to improve tracking, stability, and reduce load between the patella and trochlear groove [76]. The PF joint and quadriceps mechanism are also vulnerable to pathology of the tibiofemoral (TF) joint including complications from surgeries such as anterior cruciate ligament (ACL) repair and high tibial osteotomy (HTO) [77]. ACL repair may cause scarring that produces a relative shortening of the patellar tendon and alterations in TF kinematics can change tracking of the patella in femoral groove. Similarly, HTO may alter the TF joint line and create patella baja or pseudo-patella baja [77]. Thus, precise knowledge of healthy patella mechanics can provide useful metrics for comparison to quadriceps function following patella resurfacing in conjunction with partial or total knee arthroplasty (TKA). Complications linked with the PF joint are a common cause for revision of TKA [78]. Moreover, patients after TKA often exhibit long-term adaptations in their movement patterns attributed to quadriceps weakness.

The mechanics of the quadriceps mechanism may be altered by changes in patellar flexion and patellar tendon angle (PTA). PTA, the angle between patellar tendon and the long axis of the tibia, is highly correlated with both PF and TF joint kinematics, patellar tendon length, and the geometry of the articular surfaces of the knee [15]. Assessment of PTA differentiates the biomechanical behavior of the normal knee and knees with implants. In addition, patellar tendon moment arm (PTMA) is an indicator of quadriceps efficiency and sensitive to changes in patellar kinematics [79, 80], movement strategies [81], and muscle contraction [82].

Similarly with the tibiofemoral joints, the patellofemoral joint kinematics can change from non-weightbearing to weightbearing activities since weightbearing can influence strain in repaired ligaments [83, 84] and tracking of the patella in the femoral groove [76]. Re-
searchers using accurate measurement methods such as stereo radiography have reported kinematics for the healthy PF joint [85, 34, 86] in squatting and lunge activities. However, loading conditions during gait are different than squatting and lunge and may lead to differences in patella kinematics [87, 88] and reports of precise measurements of baseline natural patella kinematics during walking, and higher demand tasks in gait such as turning and stairs descent, do not exist, nor do quantities such as ISR, patellar tendon angle and moment arm. What’s more, most studies record knee motion of younger subjects that may not be representative of the age range associated with PF pain, osteoarthritis, and TKA.

2.4.3 Contributions

To our knowledge, no others have evaluated normal knee function for a cohort age matched to TKA recipients and during activities that patients with TKA often report to be troublesome, such as descending a step and executing a turn during walking. Most descriptions of knee kinematics have been for younger adults and for a limited span of activities. Several prior studies have evaluated young healthy, OA, and TKA subjects, and shown important differences between knee kinematics across these three cohorts. However, these studies routinely compare older adults with OA and TKA to young healthy controls. Our study fills a gap in this information by providing a cohort more similar in age to those with OA and TKA. Documenting normal changes in knee kinematics with age is necessary for understanding the changes in knee kinematics that occur with OA and TKA.

Additionally, differences between non-weightbearing and weightbearing are clinically meaningful because kinematic changes may alter surgical repairs to the knee made with the knee unloaded. Part of this work was to investigate the differences in tibiofemoral and patellofemoral kinematics that occur from non-weightbearing to weightbearing activities.
Finally, few have investigated the combined motion of TF and PF joints in healthy living subjects, thus our intent was to determine if changes in tibiofemoral kinematics correspond with changes in patellofemoral kinematics, and examine whether these differences in kinematics are variable between individuals and provide comprehensive patient-specific measurements of TF and PF joint kinematics.

2.5 Natural Hip/Pelvis Joint

The hip joint is unique anatomically, physiologically, and developmentally; thus evaluating the main structure and biomechanics of the hip is crucial for clinicians, physiotherapists and engineers alike. The hip joint is one of the most stable joints, powerful joint in the body due to its heavy musculature, robust architecture, and large range of motion. The hip joint is a ball and socket type of synovial joint between the pelvis and the femur that permits for polyaxial articulation between the body and the lower extremity. The head of the femur is fitted to the acetabulum for a distance extending over nearly half a sphere, and embraced by the acetabular labrum (a ring of cartilage that surrounds the acetabulum) that creates a seal effect to hold in place the joint (Figure 2.6). The hips’ main function is to provide dynamic support from the weight of the body while accommodating force and load transmission from the axial skeleton to the lower extremities, allowing mobility. Thus, the hip joint plays an important role in the human osteoarticular system, both in terms of locomotion and as a load-bearing joint for the torso by transmitting weight to different areas of the pelvis [89].
The normal anatomy of the hip.

Figure 2.5: The normal anatomy of the hip.

The pelvis is responsible for biped ambulation and support of the spinal column and accommodates sexuality, reproduction, storage and elimination of urinary and fecal waste, and indeed represents the human body’s foundation for both form and function [90]. The hip joint connects the pelvic girdle to the lower limb and the head of the femur articulates with the acetabulum of the pelvic (hip) bone. The bony structure of the pelvis is the foundation for the hip joint. The pelvis is oriented in mirror images to the left and right sides of the vertebral column, creating a right hip and a left hip. The pelvis consists of the sacrum, the coccyx, the ischium, the ilium, and the pubis [91, 92](Figure ??). The structure of the pelvis supports the contents of the abdomen while also helping to transfer the weight from the spine to the lower limbs [93]. During gait, the joints within the pelvis work together
to decrease the amount of force transferred from the ground and lower extremities to the spine and upper extremities [93].

Many of the essential movements during daily living such as sitting, standing, walking, running, and lifting, entail architectural contributions at the hip/pelvis joint. Additionally, the pelvis acts in a similar way to a stationary unit with coordinated motion occurring among the lumbar spine and the hip joint as the result of muscle coordination.

![The pelvis helps anchor the muscles and protect the organs in the lower abdomen.](image)

Figure 2.6: The normal anatomy of the pelvis.

**Hip Joint Motion**

Participation in specific activities requires a complex range of hip movements and muscle activity. The hip joint achieves great mobility and stability during different activities of daily living, however, this joint sacrifices mobility for stability as it is designed for weight bearing. The form of the hip joint allows motion of the lower extremity under the control of specific muscles. Researchers are particularly interested in investigating the motion of
the natural hip joint in basic movements. During gait activities, the hip joint plays a crucial role in advancement of the lower extremity.

Hip joint is a multiaxial joint that allows a wide range of motion. The maximum total range of motion of the hip joint is approximately $140^\circ$ of flexion-extension, $75^\circ$ of abduction-adduction, and $90^\circ$ of internal-external rotation. Raising the leg toward the front is termed flexion; pushing the leg toward the back is termed extension. Swinging the leg away from the body laterally (to the side) is termed abduction. Pulling the leg toward or across the body is adduction. Rotation of the femur head in the hip joint permits us to point our toes at any point across an arc that is enabled by the rotational flexibility of the hip. This movement ability lets us to regulate our foot position with respect to the body vis-a-vis pointing the toes in or pointing the toes out. The combination of flexion, extension, abduction, adduction, and rotation into a composite movement is known as circumduction.

Any restrictions in hip function and range of motion could jeopardize each individual’s quality of life and physical activities of daily living. Due to various pathologies such as premature wear and tear of the hip joint and degenerative arthritis or artificial joint replacement, individuals will ultimately adapt their regular functional movements to compensate for joint pain or instability.

Limitations to normal gait, function, and range of motion frequently are the main reason for total hip arthroplasty (THA) in an effort to restore daily functional tasks. Preoperatively, THA patients demonstrate slower ambulatory speeds, decreased cadence, and shortened stride length as a result of reduced hip flexion during contact, and reversal of motion during extension at the end of the stance phase.

**Pelvis Motion**

Broadly, motions of the pelvis are characterized as rotations about one of three cardinal axes, each of which generates motion in one of the planes [93, 92]. Rotation about a
mediolateral axis causes motion within the sagittal plane, and is regularly referred to as anterior or posterior tilt or rotation. With anterior pelvic tilt, the anterior superior iliac spines (ASIS) each move anteriorly and inferiorly while the posterior superior iliac spines (PSIS) each move superiorly [94]. Conversely, posterior pelvic tilt refers to the opposite situation, when the ASIS move posteriorly and superiorly while the PSIS move inferiorly [95, 92]. Rotation about an anteroposterior axis generates motion within the frontal or coronal plane. This movement happens when one side of the pelvis moves lower as the other side moves higher, and is often referred to as pelvic obliquity or list. Generally, this happens while weight-bearing on a single lower extremity and is depicted by the motion of the contralateral side of the pelvis [94, 95, 92]. Rotation about a vertical axis produces movement in the transverse or horizontal plane and in some fields, this is known as forward and backward rotation (Levangie and Norkin 2011), or similarly as anterior and posterior rotation [94, 95, 92]. Others refer to these motions as internal and external rotation [94, 92], and is named similarly to the motions of the lower extremity segments. Just as right femoral internal rotation is counter-clockwise rotation of the femur when viewed from a superior perspective, internal rotation of the pelvis during right stance is counter-clockwise rotation of the pelvis or backward rotation [94, 92].

These same terms for pelvic motion (tilt and rotation) are also utilized to refer to the position of the pelvis in certain postures or activities. When the weight is equally distributed between both lower extremities in standing position, the pelvis is normally level with the left and the right anterior superior iliac spine (ASIS) being at the same height [96], showing no obliquity [92]. In the sagittal plane, the neutral position of pelvis has occasionally been determined as the ASIS being in the same vertical plane as the pubic symphysis [97, 92]. This definition is sometimes extended to denote that a line drawn between the ASIS and PSIS is horizontal, demonstrating no tilt. Use of these bony landmarks gives the clinician the ability to clinician to easily palpate and measure pelvic tilt. Generally, people stand in
11 to 13 degrees of anterior pelvic tilt [98, 92]. However, this pelvic tilt is not the same as the radiological measure of pelvic tilt (pelvic tilt sacral) often used in literature focusing on the spine. In spine literature, the pelvic tilt estimation is the angle between a vertical line passing through the center of the biconfoemoral axis and a line drawn between the axis and the middle of the superior endplate of S1 [99, 92]. Calculating this parameter, pelvic tilt in standing position is usually between 12 and 15 degrees [100]. The motions of the pelvis and hip may also be altered in the presence of abnormal acetabular structure, especially with acetabular dysplasia.

**JCS for the Hip/Pelvis**

For most fields of biomechanical research, the normal human hip joint is considered as a ball and socket joint, with the center of rotation determined at the center of the femoral head. The joint coordinate system (JCS) reported by Grood and Suntay has the benefit of reporting joint motions in clinically relevant terms and is independent of the order in which the rotational transformations are utilized [1] in which the first rotation is depicted as being about an axis fixed in the proximal segment, the final rotation about an axis fixed in the distal segment and the middle rotation about a floating axis mutually perpendicular to the other two axes. The JCS system refers to conventions using Euler angles in the following order: flexion, adduction-abduction and internal-external rotation of the moving segment coordinate system with reference to the fixed segment coordinate system or to fixed global reference frame. A more universally acceptable reference system was sought, to be appropriate for different biomechanical investigations, including gait analysis, radiographic analysis, in vitro studies, and finite element modeling [101]. However, it was accepted that these different areas of research tend to utilize various anatomical landmarks and different reference axes. One reason is that the most reproducible landmarks of the bones are not certainly accessible in vivo in standard practice [101].
The movement of the pelvis is described in equivalent terms with the pelvis considered as the distal segment and the laboratory reference system as the proximal system. The rotations about the axes embedded in the pelvis are described as tilt, about the medio-lateral axis, obliquity, about the anterior-superior axis and rotation about the proximal-distal axis. It is conventional that the value of these angles rely on the order in which they are identified [102]. Based on Wu et al study [101] the pelvic coordinate system is defined with an Euler sequence angle of XYZ and with the anatomical landmarks of ASIS (anterior superior iliac spine), PSIS (posterior superior iliac spine and midpoint of the pubic symphysis). The origin is coincident with the right (or left) hip center of rotation, the Z axis is the line parallel to a line connecting the right and left ASISs, and pointing to the right the X axis is the line parallel to a line lying in the plane defined by the two ASISs and the midpoint of the two PSISs, orthogonal to the Z-axis, and pointing anteriorly and the Y axis is the line perpendicular to both X and Z, pointing cranially [101, 103].

The location of the hip center of rotation has been calculated using either a functional approach [101, 103] or a prediction approach [104] were the functional approach is recommended [101]. This method seems appropriate when it is possible to analyze a sufficient range of motion at the hip [101]. The JCS for the left (or right) hip joint is described with the e1 as the axis fixed to the pelvis and coincident with the Z-axis of the pelvic coordinate system, e3 as the axis fixed to the femur and coincident with the y-axis of the right (or left) femur coordinate system and e2 as the floating axis, the common axis perpendicular to e1 and e3 [101].

2.5.1 Kinematics of Hip/Pelvis

The normal biomechanics of different activities of daily living include complex coordinated movement of the hip, pelvis, and lumbar spine [105]. Kinematic data during activities
of daily living could provide essential information evaluating kinematics of pathological and reconstructed hips. Kinematic analysis of healthy hips during functional dynamic activities is crucial for creating a baseline for evaluating kinematics of those pathological and reconstructed hips. Additionally, pelvis kinematics could be more complex than just a 2D movement in the sagittal plane since it could have a three-dimensional component as a result of a deviation of the rotation axis from the bicoxofemoral axis. Since pelvic posture and kinematics can influence acetabular orientation, an accurate knowledge of the axis for pelvic rotation is important for a more accurate estimation of sagittal pelvic alignment in healthy subjects and possible kinematic disorders in patients with hip or spine impairment [106].

Though imaging processes and diagnostic criteria of hip pathology are often rely on static findings (i.e., radiographs), it is essential to understand that all pathologies of the hip are dynamic in nature and that disease process will have an effect on motion. Furthermore, although the past and current innovation for Total Hip Arthroplasty (THA) has been on improving fixation methods and bearing surfaces, the ongoing challenges with the contemporary THA involve in vivo dynamic occurrence like impingement, and dislocation. Thus, a kinematic evaluation of the hip is crucial for evaluating the cause of hip disease as well as in improving our capacity to reestablish function through THA.

**Natural Kinematics**

Kinematic analysis of healthy hips is essential in order to establish a baseline of ideal mechanics that could be used as a reference for THA evaluation. The hip joint accomplishes great mobility and stability during different activities of daily living. Involvement in particular tasks needs a complex range of hip movements and muscle activation. Thus, kinematic analysis of normal hip/pelvis kinematics during various functional weight-bearing activities is important for estimating kinematics of pathological and reconstructed hips and this
knowledge may assist clinicians in developing more targeted treatment approaches. While the in vivo kinematics of a total hip arthroplasty (THA) are well documented, there is limited information related to the kinematics of healthy, non-arthritic (normal) hips and degenerative hips requiring a THA.

In Hara et al [107] study, healthy hip kinematics were evaluated in four different functional weight-bearing activities over the range of 100° of flexion and 60° of axial rotation by 2D/3D image registration techniques. This study demonstrated that native hip joints showed activity dependent kinematics with coordinated pelvic and femoral dynamic movements. Understanding whether the pelvis or the femur is contributing to altered hip kinematics may assist clinicians in developing better treatments. DeCook et al [108] investigated subjects with a native hip and made a comparison with in vivo kinematics to subjects before and after receiving a THA. They showed that degenerative hips experience more abnormal hip kinematics compared to the normal group that leads to higher articulating surface forces and stresses within the acetabulum. Also, they demonstrated that none of the implanted hips experienced hip separation.

Several studies have reported that abnormal pelvic movement has a significant effect on hip joint kinematics, with probable consequences on hip instability in patients after THA [109, 110, 111]. Several abnormal kinematic patterns have been implied to be factors connected with running associated injury. An increase in anterior pelvic tilt has mainly been implicated, indicating that this motion places the surrounding structures at a higher risk for injury [112, 113]. In particular, increased anterior pelvic tilt has been recognized as a predisposing factor for hamstring strains in runners [114, 113]. Furthermore, a strong correlation has been documented during running between increased anterior pelvic tilt and increased lumbar lordosis [115]. The connection between injury risk and the coordinated movements of the pelvis and lumbar spine has elicited extensive examinations of these
patterns during normal gait and running [112, 113]. In Franz et al [113] the kinematic patterns of the hip and pelvis were highly coordinated during both walking and running.

Since the motion of the spine, pelvis, and hip is coordinated during postural changes, any pathology of this area may influence stand-to-sit pelvis kinematics [116]. The pelvis kinematics in standing to sitting transition is regularly considered as a rotation around the axis that links the center of the two femoral heads (hip axis). This motion originates from anterior pelvic tilt in the standing position (pelvic flexion) and continues to posterior pelvic tilt in the sitting position (pelvic extension) [117]. Lazzanec et al [117] showed that adaptation from standing to sitting positions combines pelvic tilt and anteroposterior pelvic translation. Kinematic variation in the sagittal pelvic tilt has important effects on the functional orientation of the acetabular implant resulting in potential changes on biomechanics of the THA. Furthermore, irregular kinematics could happen for patients with musculoskeletal disorders, thus, affecting the hip-spine complex [111]. Kim et al [111] study showed that hip or spine pathology affected the abnormal stand-to-sit pelvic kinematics and that surgeons should take into account possible abnormal stand-to-sit pelvis kinematics when planning a surgery such as THA.

**Change in Kinematics with Pathology**

Hip pathology can generate kinetic alterations at the hip joint. Regularly structural abnormality related with developmental deformity or degenerative osteoarthritis can have a profound effect on force creation and transmission at the hip joint.

Hip joint osteoarthritis (OA) is a chronic and progressive musculoskeletal condition often leading to pain, functional limitations such as poor kinematics and limiting range of motion and significantly reduced quality of life [118, 119](Figure 2.7). Specifically walking is an essential and common activity of daily living that is frequently affected by hip OA [120]. Evaluating and comprehending whether movement variability during gait
varies according to pain intensity and structural disease severity may help develop better rehabilitation and treatment protocols to prevent disease progression. There is no known cure for hip OA and clinical guidelines suggest that total hip arthroplasty is retained for the end stage of the disease. As patients’ function and ability are directly affected by joint kinematics, there is an interest in estimating the hip kinematics of patients with end-stage hip disorders before THA. One approach to have better outcomes may be to evaluate hip biomechanics and movement strategies during challenging tasks. Previous investigators showed that patients with hip osteoarthritis exhibited abnormal hip kinematics during different daily activities [121, 122]. Stair use, specifically is a relevant task to assess in people with hip OA, since difficulty with stair ascent is one of the main reasons that patients decide to undergo a costly hip joint replacement [123]. Many studies have reported altered biomechanical patterns in people with hip OA during tasks such as gait [124] and stair climbing [125]. Imaging 3D/2D techniques have been used for kinematic analyses of hips affected by OA. In [123], they reported that with respect to squatting, patients with OA were unable to deeply flex their femurs due to limited range of motion (ROM) of the hip joints. Additionally, patients with OA tilted their pelvis more posteriorly to maintain a deeply flexed posture than did healthy subjects [126, 63]. Gait patterns of subjects with osteoarthritis (OA) of the hip are identified by a decreased walking speed and step length, thus, gait analysis is essential for providing information for the functional limitations at various stages of the disorder which could be utilized to inform the development and timely use of targeted therapeutic interventions like exercising.

Other pathological changes that may influence hip kinematics are acetabular dysplasia (also known as developmental dysplasia of the hip or dysplasia) and femoroacetabular impingement (FAI). These two diseases are mainly characterized by altered hip geometry and may be the main causes of early hip OA [127, 128, 129]. FAI is a pathological disorder of the hip that is created from structural deformity of the proximal femoral head or acetabulum
generating altered joint congruency in this articulation [129]. Many studies have assessed hip and pelvis kinematics during gait in persons with cam FAI. The results of these investigations are differing and have documented that patients with cam FAI show decreases in sagittal [130], frontal [131] and transverse plane hip kinematics [130] and diminished frontal plane pelvis motion [132] when compared to healthy subjects. Kennedy et al [132] investigated 18 individuals with cam FAI as well as 19 healthy control individuals and reported that those with FAI demonstrated lower peak abduction angles and smaller frontal and sagittal plane excursions compared to the control group during walking. Information of the biomechanical variations during gait in those with FAI will give further intuition into the functional impairments of the disease as well as information that is needed to inform targeted rehabilitation protocols that target on restoration of physical function.

Developmental dysplasia of the hip (DDH) is an significant disease that causes osteoarthritis (OA) [133], and it is well accepted that high stress concentrations during weight-bearing ambulation could cause OA [134]. Sato et al [135] investigated the contrast femoral translation in DDH and contralateral normal hips during weight-bearing stepping. Dynamic radiographic imaging of dysplastic and contralateral healthy hips was performed and assessed their kinematics using model-image registration process. Their results showed that DDH hips demonstrate higher swing-phase femoral translations during walking-in-place than contralateral normal hips. Assessment of femoral translation in DDH hips provide knowledge that could help us to better understand the pathogenesis of hip OA and how to provide better treatment strategies.
Change in Kinematics with Joint Replacement

Total hip arthroplasty (THA) is the definitive surgical procedure for end-stage osteoarthritis of the hip joint and is highly successful in alleviating pain, improving function, and allowing patients to resume their activities of daily living. Roughly, 2.5% of the population over 40 will undergo THA and the number of primary and revision THA is expected to grow over the next 20 years [136]. Even though THA is effective in accomplishing the main purpose of pain relief, gait deficiencies, such as reduced range of hip and knee flexion/extension and lower range of hip abduction/adduction, may continue years after the implantation [137]. This could lead to an increase risk of falls for THA subjects and restrain their quality of life. Hence, a deep understanding of hip joint biomechanics is
crucial for improving the THA design and performance during different tasks of daily living. Kinematic evaluation of patients following THA have been studied extensively during gait [137, 138]. Fewer studies have evaluated kinematic discrepancies between THA and normal hips during demanding physical activities [139, 140]. Dimitriou et al [136] compared the 3D kinematics of THA and healthy hip during step-up and single-leg stance activities where the THA group showed greater internal rotation than the contralateral healthy hip during the step-up activity but not during single-leg stance and the difference in internal rotation was greatly correlated to the difference in femoral anteversion and anterior translation of hip joint center between implanted and healthy hip joint. In implanted hips, Koyanagi et al [141] showed that the mean maximum hip flexion ROM was 86.2°, which is smaller than the mean maximum hip flexion angle of 95.4° in normal healthy hips. Komiyama et al [126] revealed that THA range of hip joint motion increased and the pelvis tilted anteriorly more after THA than before the surgery during the squatting activity.

Appropriate acetabular cup positioning plays a crucial role in accomplishing both stability and mobility of an artificial hip joint and is significant for avoiding complications such as dislocation [142], impingement [143] and accelerated bearing wear [144]. Implant positioning during surgery has been shown to impact the incident of dislocation. Improper implant positioning, such as retroversion of an acetabular cup, could cause implant impingement, leading to dislocation. The wear of the components has been well documented, and many factors have been shown to contribute to premature component wear, like material quality, acetabular cup position, and implant misalignment during the implant surgery. Joint instability and hip separation, established as the femoral head center displacement relative to the acetabular cup center, has also been reported to increase the wear volume. The risk of hip instability and separation increases postoperatively after THA. Additionally, variations in THA component placement can cause unwelcome postoperative mechanics.
Several studies have been performed to evaluate the in vivo hip kinematics and specifically hip separation in the implanted hip [145]. DeCook et al [108] revealed that even very small hip separation can cause edge loading in THA and the mayor differences between the preoperative degenerative joints and the postoperative prosthetic joints definitely designate the harmful consequences of hip joint separation and sliding.

2.5.2 Acetabular Alignment and Functional Orientation of the Pelvis

Acetabular component placement is directly associated to the occurrences of hip dislocation and wear, the two most common causes of THA [146, 142]. Factors to pay attention to during the optimization of cup orientation on subject specific basis involve femoral anteversion, hip biomechanics and the orientation of the pelvis in functional positions [147]. Functional orientation of the pelvis after surgery is one of the most crucial variables for the surgeon to review when deciding the proper goal for acetabular component orientation.

Surgical navigation has been reported to provide better accuracy and precision of component positioning compared to traditional freehand techniques [148, 3]. Surgical navigation gives the ability to the surgeon to make fine adjustments to component position based on factors like pelvic tilt (PT) [149]. For example, a navigation system can estimate the tilt-adjusted retroversion of an acetabular component. Until recently, methods utilized range from adjustment based on calculation of PT in a supine position by digitizing the floor plane and anterior pelvic plane (APP) to modification without estimation of the APP in the lateral position [3]. APP is regularly used as a superficial anatomical landmark during THA surgery, with most of the subjects being in the supine position. APP is determined as the plane through the right anterior superior iliac spine (right-ASIS), the left anterior superior iliac spine (left-ASIS) and the pubic symphysis (SP). The APP and the coronal (frontal) plane is determined as pelvic tilt angle (PT) (Figure 2.8).
Acetabular component alignment is considered to have a significant role in THA dislocations [150]. For many decades, hip surgeons have depended on traditional safe acetabular zones to minimize implant instability. Lewinnek et al [142] documented a safety zone for acetabular cup placement and based on their results, an inclination of $40^\circ \pm 10^\circ$ and anteversion of $15^\circ \pm 10^\circ$ should allow the greatest feasible range of motion of the hip with the minimum dislocation risk. Nevertheless, the validity of Lewinnek safe zones has been challenged by recent studies [151, 152, 153, 154]. Abdel et al [151] demonstrated that approximately 60% of 206 dislocated THA surgeries had a cup positioned within the so-called safe zone. Additionally, acetabular orientation is a dynamic parameter that can be affected by forces contributing on the pelvis by changes in pelvic tilt, obliquity, and rotation [6, 155, 156, 157]. These forces may derive from above the hip joint such as change in spinal alignment, at the hip, such as muscle weakness or advanced hip deterioration, or below the hip, like limb-length discrepancy [157]. These parameters can be adapted to
alter the acetabular socket orientation secondary to pelvic motion. Additionally, even with navigation systems [158], the evaluation of the safe zone still relies on static measurements that do not take into account the pelvis dynamic changes. Thus, planning and evaluation of the intended position of the acetabular component in the supine position, maybe is not ideal for predicting clinically important changes in its orientation during functional activities, resulting from individual pelvic kinematics. Ideal orientation is subject-specific and requires assessment of functional pelvic tilt pre-operatively.

Lazennec et al used EOS Imaging (EOS, formerly biospace med) [6, 109, 106, 117] to improve our knowledge of the coordinated motion of the spine-pelvis-hip joint during postural change such as from lying to standing and from standing to sitting. Their work demonstrated that the pelvis tilts during postural changes, and since the acetabulum is part of the pelvis, pelvic motion will change the anteversion and inclination of the acetabulum, thus, it is not maintained in the position achieved at the time of surgery. This alteration of the acetabular angles is the main cause that Lazennec et al [6] reported affecting the sagittal 'functional' cup placement in THA versus the coronal inclination and anteversion accomplished at surgery. Lazennec et al [159] noted that the kinematics of the lumbar spine has an effect on the functional tilt of the pelvis in the supine, standing and seated positions. The variation in sagittal pelvic tilt between supine and more functional positions was reported first in 2003 and was documented to be very patient-specific [159, 160, 161]. DiGioia et al [160] reported an unexpected arc of sagittal pelvic movement as big as 70° in their series of 84 patients.

Any kinematic variation in sagittal PT will have a significant consequence on the functional anteversion and inclination of the acetabulum [155]. As the pelvis rotates posteriorly, with the iliac spines moving posteriorly with respect to the pubic tubercles, the functional anteversion of the acetabular components will rise. This rotation is helpful against dislocation and edge-loading in flexion of the hip, but could cause wear in extension and anterior
instability. Thus, assessing the orientation of the acetabular component seems more beneficial during functional flexion and extension rather than when the patient is in the supine position, since the problems associated with malorientation of the acetabular component are more likely to occur during those movements. Anteversion of the normal acetabulum is a variable that is defined by pelvic orientation and flexion of the pelvis increases anteversion, while pelvic extension decreases anteversion of the acetabulum cup.

Finally, it is important to consider whether PT changes postoperatively when utilizing preoperative PT for tilt-modified acetabular component placement. The effect of PT on acetabular component anteversion has been documented and assessed as approximately 0.7° increase in anteversion for each degree of posterior PT. In [3], was reported that the mean change in functional anteversion was 0.74° increase in anteversion per degree increase in posterior tilt (range: −0.75° to −0.72°) and that the functional inclination of the acetabular component showed a nonlinear response to change in pelvic tilt. They reported that while the mean change in inclination is 0.29° per degree of posterior tilt, the impact on inclination increased substantially with increasing posterior tilt. For example, if the acetabular component is positioned at 30° of APP anteversion and 40° of APP inclination, the functional inclination will increase to 47° when a patient has 15° of posterior PT, which is a change in inclination of 0.47° per degree of posterior tilt. Figures 2.9, 2.10 from Marrat et al [3] demonstrate the change of cup anteversion and inclination, and how these values can be influenced by pelvic tilt and the functional acetabular component position for given values of anterior pelvic plane anteversion and sagittal pelvic tilt.
Figure 2.9: Figure demonstrating position of acetabular shell with 30° APP anteversion and 40° APP inclination with A) 20° of anterior pelvic tilt and B) 20° of posterior pelvic tilt changing the functional anteversion to 44.2° and the functional inclination to 50.9° [3].

Figure 2.10: Functional Acetabular Component Position for Given Values of APP anteversion and Sagittal Pelvic Tilt [3].
2.5.3 Hip-Spine Relations and Sagittal Balance

Connections between sagittal spinal alignment and acetabular orientation garner substantial research effort with the aim of optimising prosthetic cup implantation. The sagittal balance is a postural strategy that permits the femoral heads and pelvis to support the center of gravity of the trunk and is determined by the spinal curves and the orientation of the pelvis [6]. Recent investigations evaluating the connection between the spine and the pelvis have improved our understanding of pelvic and spinal orientation. Spino-pelvic radiographic parameters have been utilized to assess balance in patients with a spinal deformity or spondylolisthesis [162]. It has been revealed that compensatory spine and pelvic dynamics are important to keep optimal balance and a functional range of movement (ROM) in both the healthy and implanted hip. Thus, loss of compensation at the spino-pelvic junction may decrease the effectiveness and increase the risk of complications following THA. The importance of the pelvic area in sagittal balance is apparent for spinal surgeons, who take into account sacral slope, pelvic incidence, and pelvic tilt in their planning and analyses [6, 162, 147].

Pelvic tilt, or the connection between the spine and the pelvis in the sagittal plane [155] directly has an effect on the functional orientation of the acetabular cup and that influences the biomechanics, impingement-free motion, and the stabilization of the joint. Variations in the spinal balance can change pelvic rotation through alterations in pelvic tilt and obliquity [6]. Pelvic tilt may have various definitions between hip and spine surgeons [6, 5]. In hip reconstruction, pelvic tilt is commonly determined as the angle between the global coronal axis of the body and the anterior pelvic plane, while for spine surgeons, it is an angle estimated by a line dropped from the midpoint of the sacral endplate to the center of the bicoxo-femoral axis and a vertical line extending this axis in the standing position [157]. When the pelvic tilt is neutral, using anterior referencing points in THA is indicative of
the patient’s functional anatomy. Nevertheless, any situation that changes the pelvic tilt, whether it was pre-existing at the time of THA procedure or following surgery, may accordingly translate into functional malposition of the acetabular cup [155]. Studies have documented a higher value in acetabular anteversion (AA) of 0.7° for every 1° increase in posterior pelvic tilt [155].

Each discrete patient is distinguished by the morphological parameter of the pelvic incidence (PI). This angle (55° ±10.6°) is estimated on lateral views as the angle between the line perpendicular to the middle of the cranial sacral endplate and the line joining the middle of this endplate to the center of the bicoxofemoral axis [6]. Sacral slope (SS) is the angle formed by the horizontal and the superior S1 endplate and pelvic tilt (PT) is the angle between the vertical and the line joining the middle of the sacral endplate and the center of the bicoxofemoral axis, which is the line between the geometric center of both femoral heads. Finally, Pelvic incidence is geometrically correlated to sacral slope and pelvic tilt with the following relationship, PI = SS + PT (Figure 2.11). Raised pelvic incidence relates with raised sacral slope and lordosis, and low pelvic incidence with low sacral slope and lordosis.

Figure 2.11: Geometrical construction showing that SS and PT are complementary angles, since the horizontal line of SS and the vertical line of PT are perpendicular. The red triangle comprises the pelvic tilt angle on its left side and the sacral slope angle on its right side [4].

42
The standing position includes a forward tilt of the pelvis. The superior S1 endplate on lateral view underlies a sacral slope angle with the horizontal of about (40.6° ± 8.5) [163] (Figure 2.12). The degree of posterior pelvic tilt able to compensate for the anterior sagittal imbalance relies on the pelvic incidence. The higher the PI, the higher the theoretic flexibility of posterior pelvic tilt. Oppositely, patients with small PI have less capacity for compensation. During the sitting position the opposite is happening. The pelvis tilts backwards and SS declines, to a mean of 20° to 25° and sometimes it becomes very low (5° to 10°) or even negative [6]. The distinction in sacral slope between the standing and sitting positions is related to the available flexion of the lumbar-sacral junction, as distinct from possible hip-joint flexion [6].

The sitting down process substantially alters the orientation of the anterior pelvic plane, which is presently the common reference for modifying acetabular cups positioning [164]. Lewinnek’s plane (APP) is retrieved when the subject is lying down from the morphologic data collected for THA planning. However, Lewinnek’s plane is not certainly vertical in the standing position, and tilt is very changeable in the sitting position [165] (Figure 2.12).

In the supine position when the lower limbs are in extension, SS is frequently higher than in the standing position [159, 6] and this additional pelvic tilt may be badly tolerated if the spine is stiff or deformed.

Finally, it has been reported that compensatory spine and pelvic dynamics are significant to keep optimal standing balance and a functional range of movement (ROM) in both the healthy and implanted hip. Studies have revealed that spinal fusion may alter the adaptation of the spinopelvic junction [116], therefore, loss of compensation at the spino-pelvic junction may decline the effectiveness and enlarge the risk of complications after THA surgery [109].
Figure 2.12: Progression from standing to sitting positions causes considerable modification in the orientation of the anterior pelvic plane (A.P.P. or Lewinnek plane) [5].

2.5.4 Contributions

To best of our knowledge no others have investigated and compared the 3D pelvic functional orientation across different populations that include healthy subjects, subjects that have undergone THA and spinal-stabilized cohorts and during different static and dynamic activities. Most studies have focused on comparing healthy cohorts with preoperatively or postoperatively THA subjects or with patients with lumbosacral fusions but do not compare all different groups of patients concurrently. Furthermore, most studies have performed their measurements in static settings whereas the pelvic motion is dynamic. All radiographic investigations are limited since the radiographic view is of a single position at
a single moment in time, however, the connection between the spine and pelvis is dynamic. The functional orientation of the pelvis varies during different dynamic activities and the pelvis is not a fixed static bone when considering acetabular cup placement.

Thus, we have assessed the global 3D aspects of pelvic motion using a high speed stereo radiography system performing a 2D/3D imaging technique since any coronal or axial rotation of the pelvis in functional positions would also affect acetabular orientation. Furthermore, calculating specific parameters at the spino-pelvic junction assesses spinal balance and pelvic compensation. This will help us to clearly estimate the consequences of the pelvic motion for more proper acetabular cup positioning. Additionally, the position of the pelvis in the sagittal plane may differ notably between functional positions and taking into account the interindividual variation is crucial. Based on these changes, the angles of orientation of the acetabular component during dislocation and edge-loading will be not be the same from those estimated from standard CT and radiographs. PT may change significantly during different activities of daily living like walking or rising from a chair that could be considered in a dynamic analysis of component alignment. As such, further investigation needs to be conducted among different populations and different tasks to investigate which subject may present a substantial dynamic change.

This knowledge will help us to better understand the behavior of all spinopelvic parameters and investigate which functional parameter correlates the most with patient-specific pelvic orientation, and aid decisions regarding acetabular component alignment. Differences in PI and spinopelvic parameters (APPA, SS, PTS) across different patient populations and across static and dynamic activities are necessary to understand for accurate positioning of the acetabular component during THA and reduce the likelihood of impingement events.
Chapter 3

Measurement of Patellofemoral Kinematics in Older Adults during Gait Activities

3.1 Abstract

The patellofemoral (PF) joint is susceptible to many pathologies resulting from acute injury, chronic disease and complications following surgical treatment of the knee. The objectives of this study were to describe case series measurements of patellar motion in healthy older adults as they performed a non-weightbearing seated knee extension and three gait activities, determine patellar tendon angle and moment arm, and show if these quantities were activity dependent. A stereo radiography system was utilized to obtain the 3D PF kinematics of seventeen healthy people over 55 years of age (8F/9M, 66 ± 7.9 years old, 75.7 ± 20.5 kg) as they performed level walking, a step down, and a pivot turn. For a similar portion of the gait cycle, patellar flexion (6.2° ± 5.8°) and average range of motion (ROM) (11° ± 5.9°) for walking with a step down was greater compared to
the other gait activities (gait ROM $6.9^\circ \pm 4.3^\circ$, pivot ROM $5.7^\circ \pm 3.3^\circ$), while the average range of motion for patella tilt was greater during walking with a pivot turn ($8.6^\circ \pm 3.9^\circ$). However, each subject displayed distinct PF kinematic trends during all activities with a few notable exceptions. Patella medial-lateral translation was significantly different between weightbearing and non-weightbearing activities. Importantly, the knee extensor mechanism characteristics of patellar tendon angle and moment arm showed considerable variation across subjects but were largely unaltered by changing activities. The variation between subjects and the different behavior of the patella during the step down and pivot emphasized the need for analysis of a range of activities to reveal individual response to pathology and treatment in patellar maltracking and osteoarthritis.

### 3.2 Introduction

The patellofemoral (PF) joint is susceptible to many pathologies resulting from acute injury, chronic disease and complications following surgical treatment of the knee. PF pathologies are frequently associated with abnormal motion of the patella relative to the femur, or patellar maltracking (often consisting of lateral patellar displacement and tilt) [72, 73]. Patellar maltracking can change the loads and stresses in the soft tissues between the patella and femur, which may lead to chondromalacia and osteoarthritis [166]. In addition, PF maltracking may be a key cause of patellofemoral pain [73, 75, 74], which affects 25% of the population [74, 167]. PF pathology and maltracking may be addressed with a variety of conservative and surgical treatments, including physical therapy designed to strengthen the quadriceps via weightbearing exercise [168], and surgical treatment such as tibial tubercle osteotomy to reposition the insertion of the patellar tendon to improve patella tracking and stability [73]. Normal PF kinematics are also affected by pathology of the tibiofemoral (TF) joint including changes caused by surgeries such as anterior cruciate ligament repair.
and high tibial osteotomy. ACL repair may cause scarring that produces a relative shortening of the patellar tendon and alterations in TF kinematics that can change tracking of the patella in the femoral groove [77, 169]. Similarly, HTO may create patella baja or pseudopatella baja [77]. Likewise, complications such as PF pain or poor tracking are a common cause for revision of total knee arthroplasty (TKA) [78]. Moreover, even in successful TKA surgery, patients often exhibit long-term adaptations in their movement patterns attributed to quadriceps deficits) [170]. Whether the result of pathology or treatment, changes in patellar tracking influence the mechanics of the quadriceps mechanism through altered patellar tendon angle (PTA) [15] and moment arm (PTMA) [79, 80]. PTMA is one measure of quadriceps efficiency and sensitive to changes in patellar kinematics. Precise knowledge of healthy patella kinematics can provide useful metrics for comparison to quadriceps function following patella resurfacing in conjunction with partial or TKA. Even so, there is a lack of precise knowledge of healthy patella kinematics in older adults, patellar tendon angle and moment arm, which can provide benchmarks for comparison to patellar function in pathology and treatment. What’s more there are few reports of patella motion during walking, and reports of precise measurements of normal patella kinematics during higher demand gait tasks such as turning and stair descent do not exist, nor do corresponding patellar tendon angle and moment arm measurements in vivo. Studies using accurate measurement methods such as fluoroscopy have reported kinematics for the healthy PF joint [85, 34, 86] in squatting and lunge activities. However, loading conditions during gait are different than squatting and lunge and may lead to differences in patella kinematics [87, 88]. The patellofemoral joint kinematics can change from non-weightbearing to weightbearing activities since weightbearing can influence strain in repaired ligaments and tracking of the patella in the femoral groove [76]. Furthermore, the prior studies of patella kinematics did not include comparison with other activities, including non-weightbearing, to determine if the measurements were activity dependent. Finally, prior studies recorded patella motion
of younger subjects that may not be representative of the older age range most associated with PF pain, osteoarthritis, and TKA. Older adults tend to slow their pace and use shorter strides while walking [171] and use unique movement strategies during activities such as stepping-down or turning and pivoting [172, 22, 23]. While these studies and others [173] have noted that aging has an impact on knee kinematics measured using marker-based motion capture, no studies have examined the small translations and rotations of the patella in older adults with no history of knee pathology. Documenting normal patella kinematics during different activities in older adults is crucial for better understanding the changes in knee kinematics that occur with pathology and treatment. The objectives of this study were to provide patient-specific measurements of patella motion relative to the femur, patellar tendon angle, and patellar tendon moment arm in healthy older adults as they performed three gait activities and a non-weightbearing knee extension. To our knowledge, no other study has investigated native patella kinematics during activities reported to be altered with age, namely descending a step and executing a turn during walking. We hypothesized that patella flexion, tilt, and medial-lateral translation, and patellar tendon angle and moment arm would be similar across subjects, yet activity dependent and sensitive to the changing movement demands of level gait, step down, and turning.

### 3.3 Methods

#### 3.3.1 Participants

Seventeen healthy subjects (8F/9M, age = 66 ± 7.9 years, body mass = 75.7 ± 20.5 kg, body mass index = 26.4 ± 5.1 kg/m², height = 166.8 ± 10.9 cm) (Table 3.1) participated in the study. This study was approved by the University of Denver Institutional Review Board and all participants provided informed consent. Subjects had no history of
injuries or surgeries to the lower limbs and could perform activities of daily living without pain or discomfort.

Table 3.1: Subjects specifications including Insall-Salvati ratio (ISR) which is used to determine presence/absence of patella alta and patellar tendon length (PTL) with the highlighted values indicating patella alta.

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### 3.3.2 Procedures

The participants performed four activities (Figure 3.1a): 1) unloaded knee extension in which seated individuals slowly extended one knee starting from relaxed flexion to full extension (seated knee extension (SKE)); 2) level walking at a self-selected pace (gait); 3) walking and executing a 90° medial change in direction on the planted foot (pivot); 4) and step down during gait from a seven-inch platform (step down). One trial was collected for each activity. High speed stereo radiography (HSSR) was used to measure the kinematics of the PF joint of the dominant limb. HSSR captures two radiographic views to
enable three-dimensional tracking of the bones in the knee [174]. The HSSR system is composed of two matching custom radiography systems with 40 cm diameter image intensifiers integrated with high-speed digital cameras [174]. Image distortion created by the image intensifiers was subtracted by imaging a radio opaque mesh of known dimension, and then forming a transformation to correct distortion from subsequent images of the subjects (XROMM Undistorter, Brown University, RI). The capture volume was calibrated from imaging a custom-calibration cube enclosing 52 steel beads of precisely-known position and size [174, 46]. The accuracy of the HSSR system for tracking the bones of the knee was assessed using a human knee phantom fixed to translational and rotational stages [174]. A human knee phantom was fixed to translational and rotational stages with accuracy to 0.025 mm and 0.01°, respectively. The knee phantom consists of human femur, tibia, fibula, and patella bones encased in thermoplastic in an anatomic full extension position. The knee phantom was translated in 0.254 mm increments in a square pattern of 21 points that traveled in and out of plane with the radiography system. Similarly, the knee phantom was axially rotated by 1.00° increments for 5.00° in both directions, for a total of 11 points. Static stereo radiographs were captured at each translational and rotational position. Three-dimensional models of the distal femur, proximal tibia, and patella were reconstructed from the CT data using ScanIP (Simpleware Inc., Mountain View, USA). The position and orientation of the patella and femur was found using Autoscoper (Brown University), which optimized the positions of the two bones to the two-dimensional stereo radiography images. The average translational tracking error of the patella was 0.1 ±0.05 mm and the average rotational tracking error was 0.4° ±0.2°. Similarly, for the femur the average translational tracking error was 0.1 ± 0.12 mm and the average rotational tracking error was 0.1° ± 0.13°. Other studies using similar dual-plane radiography equipment and methods recorded patella tracking accuracy of 0.46-1.14 mm in translation and 0.99°-1.04° in rotational [175, 176]. The improved accuracy obtained from our results could be based
on the fact that a phantom has been used instead of a cadaver with intact soft tissue and
the reported errors are those that might be obtained under ideal operating conditions. Im-
ages during all activities were captured with pulsed radiography (pulse width 750 μs, 60
kV, and 63 mA) at 100 frames/second for the gait activities and 50 frames/second for the
SKE. Following the laboratory data collection, a static computed tomography (CT) scan
with 1.0 mm slice thickness was obtained of each subject's knee (GE Optima 660,120kV
bone technique).

![Knee Extension, Gait, Pivot, Step Down](image)

![Femoral Coordinate System](image)

![Patellar Tendon](image)

**Figure 3.1:** a) Activities performed by the participants, b) 3D image registration to the 2D stereo images, and c) the femoral coordinate system that was applied to the patella d) patellar tendon moment arm and patellar tendon angle illustration.

### 3.3.3 Data Processing

Three-dimensional models of the distal femur, proximal tibia, and patella were recon-
structed from the CT data using ScanIP (Simpleware Inc., Mountain View, USA). The
position and orientation of the patella and femur was found using Autoscooper (Brown Uni-
versity [177]), which optimized the positions of the three-dimensional bone models to the
two-dimensional stereo radiography images (Figure 3.1b). HSSR was synchronized with a conventional motion caption system that was used to define the start and end for each activity (heel strike, opposite toe off and heel off for each subject). The femoral coordinate system for each subject was defined by fitting a cylinder to the medial and lateral posterior femoral condyles, with the center placed at the trochlea. The femoral coordinate system was applied to the patella at full extension with the medial-lateral (ML) axis defined as the line through the long axis of the cylinder, the superior-inferior (SI) axis aligned to the posterior aspect of the femur, and the anterior-posterior (AP) axis defined as the cross product of the ML and SI axes (Figure 3.1c) [46]. The coordinate system of the tibia and patella was assigned coincident with the femoral coordinate system with the knee at full extension [46, 178]. The motion of the patella relative to the femur was calculated using methods described by Grood and Suntay [1]. The kinematics of the patella were filtered using a 4th order low-pass Butterworth filter with a cutoff frequency of 6 Hz for all activities. TF kinematics for the same subjects during the same activities were described in prior work by Kefala et al [46].

3.3.4 Data Analyses

PF kinematics were calculated and described as a function of TF flexion. Comparisons of the average, standard deviation and total range of motion (ROM) of PF flexion-extension (FE), medial-lateral tilt (ML), and ML translation within each task were made across subjects and activities for a similar portion of the stance phase among the activities. ROM was calculated as the difference between the maximum and the minimum for each degree of freedom (DOF). PF flexion was also reported relative to the long axis of the tibia (PTFE) to enable comparison to previous reports, mainly in the TKA literature, that used this alternative standard [87]. Patellar baja was recorded using Insall-Salvati ratio (ISR) estimated as
a ratio between the length of the patellar tendon and vertical height of the patella. Values greater than 1.2 are considered to be patella alta and values less than 0.8 are considered to be patella baja [75]. Patellar tendon angle (PTA) was taken as the line between the inferior pole of the patella, the tibial tuberosity, and the long axis of the tibia (Figure 3.1d). Moment arm of the patellar tendon (PTMA) was calculated as the perpendicular distance from the patellar tendon to the instantaneous axis of rotation (IAR) of the tibiofemoral joint (Figure 3.1d). IAR was calculated using the equations described by Spoor and Veldpaus et al [179].

Two way without replication analysis of variance (ANOVA) was used to compare ROM and averages of patellar flexion, tilt, ML translation, tendon angle, and moment arm between activities. Post-hoc tests of paired t-Tests with Bonferroni correction were performed when a significant ANOVA test was identified. Differences were considered statistically significant when p<0.05 for the ANOVA test and p<0.017 was used for the adjusted p value in the post-hoc analyses. Finally, the data were analyzed at 10% increments from 20-60% for a similar portion of the gait activities.

3.4 Results

Complete tabular patella kinematics for subjects 1 through 8 are provided at digitalcommons.du.edu/orthopaedic_biomechanics.

3.4.1 Seated Knee Extension

Patella kinematics during SKE were reported over a similar flexion range as the gait activities, between 0° and 20° knee flexion. PF flexion during unloaded SKE was similar across subjects (Figure 3.2) (avg 3.9° ±2.8° between 0° and 20° flexion). The ML translation of some subjects remained near neutral during the SKE (Figure 3.2) while others were
consistently medial. Flexion of the patella measured with respect the tibia, PTFE, showed a similar pattern for all subjects (Figure 3.2).

![Graphs showing patellofemoral and patellotibia kinematics](image)

**Figure 3.2:** Patellofemoral and patellotibia kinematics for the seated knee extension showing flexion(+)/extension(-) (FE), medial(-)/lateral(+) (ML) tilt, medial(-)/lateral(+) translation, and patellotibia flexion/extension (PTFE) for all seventeen subjects for a matched range with the gait percent cycle (0-20 deg of flexion).

### 3.4.2 Level Walking

To compare kinematics over a common phase of the gait cycle across the three activities, results are presented for the single-limb stance phase of gait between opposite toe off (OTO) and heel-off (HO) (Fig.2). During level walking, average flexion of the patella was 1.4° ±2.8°, with a trend towards higher flexion at OTO (Figures 3.3, 3.4, Table 3.2). Average ML tilt was medial (avg −2.2° ± 5.5° between OTO and HO) (3.4, Table 3.2), however five of the seventeen subjects walked with lateral patella tilt (Figure 3.3). On average the ML position of the patella was near neutral (avg −0.2° ± 4.5° between OTO and HO) though

55
most subjects walked with their patellae positioned either medially or laterally (Figure 3.3). PTFE was similar in all subjects during the stance phase of gait (avg 7.3° ±1.7°).

**Figure 3.3:** Gait, pivot, and step down trials showing PF and PTFE and ML tilt and ML translation for all seventeen subjects. In all trials 0% represents heel strike (HS) (or first impact (FI) for step down) and 100% is toe off (TO). In gait, the average opposite toe off (OTO) phase occurred at 22% and heel off (HO) at 76%. For pivoting, OTO occurred at 24% of the stance phase. For step down, full weight acceptance (FWA) was between 10-22% for all subjects, except from one participant whose FWA occurred at 40% and average HO occurred at 62%. Subject #5 did not perform the pivot activity and subject #1 did not perform the step down activity.

### 3.4.3 Pivot

During pivot, most subjects began with a slightly flexed patella (OTO avg 4.2°±3.6°) and completed the turn in a more extended angle (Figures 3.3, 3.4). At OTO the patella
of most of the subjects was medially tilted (avg $-3.0^\circ \pm 2.4^\circ$), followed by a movement toward lateral tilt as the subjects executed the pivot and approached HO (Figure 3.4). The average ML tilt was slightly lateral ($1.7^\circ \pm 4.7^\circ$) throughout the activity (Table 3.2). For most subjects, the patella was medially positioned throughout the activity (avg $-4.3 \pm 3.4$ mm). Average PTFE during pivot was $6.2^\circ \pm 2.5^\circ$ (OTO avg $-9.1^\circ \pm 3.5^\circ$).

![Figure 3.4: Average (solid line) and one standard deviation (shaded region) for FE, ML tilt, ML translation, and PTFE for all seventeen subjects comparing unloaded knee extension (black dashed line) with gait and higher demand activities of pivot and step down.](image)

### 3.4.4 Step Down

During step down, most subjects began the activity with the patella of their leading knee in flexed position (PF FE avg $8.6^\circ \pm 6.3^\circ$ at OTO), while towards HO most subjects demonstrated an extended patella angle (except Subject 16) (Figures 3.3, 3.4). The subjectsâ€™
patellae were predominantly medially tilted (−2.9° ± 5.9°) except for two subjects whose patellae were laterally tilted throughout the activity (Subject 8 and Subject 17) (Figure 3.3). Greater medial translation was observed for most of the subjects (avg -1.6 ± 3.6 mm at OTO).

Table 3.2: Average and average range of motion for PF kinematics and average for PTA and PTMA during seated knee extension (for a similar flexion range with the gait activities), gait pivot and step down. Data in parenthesis indicate the values for a similar flexion range or portion of the stance phase among the activities.

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3.4.5 Kinematic Comparisons

Significant difference was found across the three activities at 30, 40 and 50% of the gait cycle for ML tilt (e.g. F (2,28) = 6.7, p<0.005 at 30%) and ML translation (e.g. F (2,28) = 4.4, p=0.02 at 30%). Post-hoc analyses using the t-Test with Bonferroni correction indicated that PF FE was significantly higher for the step down, compare to pivot (p=0.006) and level gait (p<0.001). No other significant differences were found among the activities. Significant intrasubject variability was found across the three gait activities at 30, 40, and 50% of the gait cycle ((Figure 3.4) in ML tilt (e.g. F (14,28) = 6.6, p<0.001 at 30%) and ML translation (e.g. F (14,28) = 2.6, p=0.01 at 30%). Furthermore, during step down, most subjects began the activity with the patella of their leading knee almost 5° more flexed compared to level gait or pivot and at OTO, the PTFE of most subjects was more flexed compared with level gait (Figure 3.4, but nearly the same by HO.

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Additionally, significant difference was found across the four activities at 30, 40, and 50% of the gait cycle for ML tilt (e.g. F (3,42) = 7.3, p ≤ 0.001 at 30%) and ML translation (e.g. F (3,42) = 3.9, p = 0.013 at 30%). (SKE was compared at similar knee flexion angles to the gait activities). In the post-hoc analyses, the non-weightbearing SKE was found to be significantly different in ML tilt (p=0.01) and ML (p=0.02) translation with the pivot, but not with step down and level gait. Finally, significant intrasubject variability was found across the four activities at 30, 40, and 50% of the gait cycle (Figure 3.3) in ML tilt (e.g. F (14,42) = 9.6, p ≤ 0.001 at 30%) and ML translation (e.g. F (14,42) = 3.5, p<0.001 at 30%)

3.4.6 Patella Range of Motion

Over a common portion of the stance phase of each activity (20%-60% in Figs. 2 and 3), patella flexion, tilt, and lateral translation was different in the three activities (Figure 3.5). FE ROM of the patella was significantly different among the activities (F (2,28) =4.5, p=0.007) and was the highest (11° ±5.9°) for step down compared to gait (6.9° ±4.3°) (p=0.05), pivot (5.7°±3.3°) (p=0.006) and SKE (7.0° ±3.0°) (p=0.03). Post hoc analysis indicated that ML tilt ROM was significantly higher during pivot (8.6° ±3.9°) compared to gait (3.7° ±3.2°, p<0.001), step down (4.5°±3.4°, p<0.001) and SKE (2.9° ±2.2°) (Figure 3.5).
Figure 3.5: Average range of motion and standard deviation for all activities and subjects, FE (top left), ML tilt (top right), and ML translation (bottom) for a similar flexion range and percent cycle. Range of motion was calculated by as the maximum minus the minimum during each activity. Star indicates significant difference among the activities.

3.4.7 Patellar Tendon Angle and Moment Arm

PTA during SKE averaged $14.8^\circ \pm 0.2^\circ$ (Table 3.2) and spanned from $21.9^\circ$ in Subject 4 to $7.3^\circ$ in Subject 6 (Figure 3.6). Average PTA during the three gait activities was similar in magnitude ($15.8^\circ \pm 4.8^\circ$, $15.3^\circ \pm 4.2^\circ$ and $14.4^\circ \pm 4.8^\circ$ for gait, pivot and step down, respectively) (Table 3.2, Figure 3.7a), yet varied widely between subjects and spanned from $26.2^\circ$ in Subject 15 to $9.2^\circ$ in Subject 2 in level walking (Figure 3.7b), from $7.5^\circ$ in Subject 2 to $21.7^\circ$ in subject 4 for pivot (Figure 3.7c), and from $4.1^\circ$ in subject 2 to $24.3^\circ$ in subject 15 for step down (Figure 3.7d). PTMA for SKE averaged $42.4 \pm 8.5$ mm, spanning between $27.1$ mm in Subject 12 (at $100^\circ$ knee flexion) and $58.6$ mm in Subject 6 ($0^\circ$ knee flexion) ((Figure 3.6)). The average moment arm was also similar between activities with little deviation between OTO and HO, $47.0 \pm 0.6$ mm, $51.5 \pm 0.7$ mm and $48.7 \pm 1.7$ mm
Figure 3.6: Patellar tendon angle during seated knee extension for all subjects for a similar flexion range with the gait activities (top left), average PTMA (blue) and PTA (green) for all subjects during gait with average PTMA and PTA across subjects shown by dashed lines (top right), PTMA during the seated knee extension for all subjects for a similar flexion range with the gait activities (bottom left).

for gait, pivot and step down, respectively (Table 2). Variation in PTMA spanned between 38.4 mm in Subject 8 and 61.1 mm in Subject 13 in level walking (Fig. 5b), between 41.8 mm in subject 1 and 64.5 mm in subject 15 for pivot (Figure 3.7c), and finally between 36.4 mm in subject 2 and 63.7 mm in subject 16 for step down (Figure 3.7d).
Figure 3.7: Comparison of the PTA during the three gait activities for a similar flexion range (a), average PTMA (blue) and PTA (green) for all subjects during gait (b) average PTMA (blue) and PTA (green) for all subjects during pivot (c) and step down (d), average PTMA and PTA across subjects shown by dashed lines.

3.5 Discussion

The objectives of this study were to provide comprehensive patient-specific measurements of patella motion, patellar tendon angle, and patellar tendon moment arm in healthy older adults as they performed three gait activities and a non-weightbearing SKE. To our knowledge, no other study has investigated normal patella kinematics during activities reported to be altered with age, namely descending a step and executing a turn during walking. We hypothesized that patella flexion, tilt, and medial-lateral translation, and patellar tendon angle and moment arm would be similar across subjects, yet activity dependent and
sensitive to the changing movement demands of level gait, step down, and pivoting. In support of our hypotheses, weightbearing resulted in changes in patellar kinematics compared to the non-weightbearing SKE marked by a significant change in ML tilt and translation. Differences in patellar kinematics were noted between the three gait movements. Peak patellar flexion was found to be significantly greater during the step down while greater patellar tilt ROM was observed for the pivot. PTA and PTMA for SKE were similar to the values found for gait activities and to results from previous studies. Opposing our hypotheses, patellar tendon angle and moment arm for the gait activities were insensitive to gait demands. In addition, subjects walked with unique kinematic features showing significant variability from 30-50% of the cycle for ML tilt and ML translation. For example, the patella of some subjects remained laterally or medially tilted throughout the gait activities. PTA and PTMA were also shown to be unique to each subject. Additionally, only patellar flexion during level gait and pivoting supported our hypothesis of similar kinematics across subjects. Most subjects showed distinct trends in kinematics characterized by medial or lateral positioning of the patella that was consistent through the weightbearing activities. While on average the patella moved medially throughout the activities in our results contrary to the in vitro results of others [88, 180, 181] there was also considerable variability between subjects [180, 181]. For example, ML translation of Subject 11 followed a medial trend (Figure 3.3, while Subject 16 followed an opposite lateral trend (Figure 3.3, while still others remained mostly neutral (e.g. Subject 14 in Figure 3.3 throughout the activities. Similar differences between subjects were also described though not reported in prior studies. Koh et al [85] investigated knee extension of one male using intracortical pins and found that the patella shifted laterally with knee flexion (9 mm from 0 to 60° flexion); whereas, more recent weightbearing in vivo measurements described a medial shift during gait on average [182], and in the early portion of a lunge [86], that proceeded to a lateral
patellar shift as knee flexion increased [86]. Notably, these prior studies did not compare
kinematics between individuals or report individual results.

Level walking resulted in changes to patellar kinematics relative to the unweighted
SKE, most remarkably in a significant increase in ML translation ROM. For instance, av-
erage patella ROM from 2.0 ± 1.4 mm during SKE to 4.2 ± 2.2 mm in ML translation
during level walking (also, compare gait with SKE in (Figures 3.5). While, on average,
the patella moved medially from OTO to TO, there were considerable differences between
subjects (compare individual and average data in Figures 3.3, 3.4). In general, the subjects
maintained the same medial or lateral translation trend in patellar movement they displayed
in the SKE. While there was no prior data for comparison to our measurements of patella
kinematics in gait, Nha et al [86] reported medial translation similar to our results, but
with lateral tilt over a similar range of knee motion during a weightbearing lunge. The
more demanding pivot and step down produced changes in patella kinematics relative to
level walking since significant differences in average ROM for PF flexion and lateral tilt
were shown for step down and pivot, respectively, relative to gait (3.5). Over a similar
portion of stance phase in all three gait activities (20-60% of stance phase, Fig. 3), step
down had the highest average patellar FE (Table 3.2), and highest FE ROM (Table 3.2,
3.5). The higher patellar flexion in step down may be explained by the greater knee flexion
angle with weight acceptance (8.6° ± 6.3° compared with 5.6° ± 4.1° at OTO). In addition,
Shalhoub et al [183] showed that PF kinematics were more sensitive to change in extensor
muscle loading with the knee near full extension compared with deep flexion, which may
explain the sensitivity of patella kinematics to higher demand gait activities. Over the same
portion of the stance phase, the average ML tilt was lateral during the pivot, while all other
activities were medial (Figures 3.3). At the completion of the pivot, external rotation of the
tibia (avg 6.6° in Kefala et al. [46]) corresponded with an increased lateral tilt of the patella
(Figures 3.3, 3.4). Similarly, Li et al [61] recorded a correlation between patellar tilt and
tibiofemoral rotation during a lunge activity. In addition, ML tilt ROM for the pivot was significantly higher compared with the other activities and shifted toward lateral tilt (Figure 3.5). These results are supported by reported measurements in cadavers in which enforced tibial IE was followed by patellar ML tilt, especially near full extension [88].

### 3.5.1 Patellar Tendon Moment Arm and Angle

Patellar tendon moment arm was similar in magnitude and variability to in vivo measurements reported in previous studies. In Im et al [79], peak moment arm was 46.5 mm in females and 50.6 mm in males, which was close to the average of 47.0 ± 0.6 mm between 0 and 20° found in this study level walking (Table 3.2). (In our study, average moment arm in level walking was 46.6 in females and 47.6 in males.) These results also agree with the in vitro study of Ali et al [181], which found a similar range of PTMA between 0 and 20° of knee flexion. Across activities, our results for PTMA ranged from 47 to 51.5 mm (Table 3.2) with the highest value occurring during the pivot. Although PTMA was consistent for each individual, there was a range of this important measurement (compare subjects in Figure 3.7b) that could impact the estimation of muscle and knee joint forces in silico [184]. Furthermore, measurement of individual differences is becoming more essential to support the drive towards personalized medicine in treatment for orthopaedic pathologies [185].

On average our results for patellar tendon angle were consistent with previous reports (e.g. Miller et al that reported values of 15° ± 2.9° at 20° of flexion similar to values in Table 2) [66, 181, 186, 187], however, there was considerable variation between subjects. The angle of the patellar tendon is important because patellar tendon force partly determines loading in the PF and TF joints, including the AP shear force applied to the tibia [188]. Patellar tendon angle was positive and remained nearly constant in each individual over the knee range of motion utilized during gait (Figures 3.7a,b), yet this angle in some subjects
was small (9.2° in Subject 2) and large in others (26.2° in Subject 15). For everyone, the magnitude of their PTA across activities was similar, except for a notable increase during the pivot as the subjects approached TO (Figures 3.7a). This increase in PTA at TO might be explained by the high external tibial rotation prior to TO described in a previous study of these subjects [46].

The data collected in this study provide insight into how the native kinematics of the patellofemoral joint of aging individuals can vary from simple activities of daily living to higher demand tasks such as step descending or pivot by defining a general trend of patellar tracking when comparing tracking during different functional activities. These data can help clinicians and scientists that seek to understand PF pathology, to facilitate future improvements in the patella design and to provide better knowledge when prescribing rehabilitation intervention.

3.5.2 Limitations

A limitation of this study was the measurement of one trial for each activity. Repeated measurements may have provided assessment of intra-subject variability; however, this was deemed unjustifiable considering the additional x-ray exposure. In addition, the number of subjects might be considered relatively small. However, our results showed a clear distinction between activities as have prior studies with similarly sized cohorts [34, 86, 178]. Unfortunately, the number of subjects was insufficient to test possible differences between males and females as others have noted [189]. (In our study, females trended toward lower PTMA during gait activities and greater lateral shift and lesser tilt of the patella during gait while for the step down an opposite trend was observed where the females had lesser shift and greater tilt. Kinematics during the pivot were similar in males and females.) Finally, our analysis did not include the heel-strike portion of the gait cycle.
3.5.3 Conclusions

This study investigated 6DOF natural kinematics of the patella during three gait activities for a cohort of older adults to provide benchmarks for evaluation of knee pathology and treatment. The data collected in this study provide insight into how the kinematics of the patellofemoral joint of aging individuals can vary between activities of daily living, and between individuals. Weightbearing in walking resulted in changes to patellar kinematics relative to the unweighted SKE, most notably in increasing ML translation ROM among subjects. Most notably, there was significantly greater ML tilt ROM during the pivot and greater patellar flexion during the step down, when compared to level gait. Importantly, the extensor characteristics of patellar tendon moment arm and angle ranged widely across subjects but were mostly unaltered by changing activities. These data can help clinicians and scientists to evaluate PF maltracking, facilitate future improvements in patella design in TKA, and provide targets for rehabilitation. With some notable exceptions, each subject displayed distinct PF kinematics that followed consistent trends through the three gait activities. The variation between subjects, the difference between weightbearing and non-weightbearing activities and the different behavior of the patella during the step down and pivot emphasized the need for subject-specific analysis with a range of activities.
Chapter 4

Changes in Natural Tibiofemoral, Patellofemoral, and Patellar Tendon Kinematics with Weightbearing

4.1 Abstract

**Background**: Few studies have investigated the combined motion of the tibiofemoral (TF) and patellofemoral (PF) joints in healthy older adults. Combined measurements of TF and PF joint kinematics are necessary because knee pathologies and treatments frequently concern both articulations. In addition, the kinematics of the TF and PF joints can vary as the demand on the knee changes from non-weightbearing to weightbearing activities. The goals of this research were combined measurement of the differences in TF and PF kinematics, patellar tendon angle (PTA), and patellar tendon moment arm (PTMA) that occur during non-weightbearing and weightbearing activities.

**Methods**: High speed stereo radiography (HSSR) was used to measure the kinematics of the TF and PF joints in subjects as they performed one seated non-weightbearing (knee
extension) and two weightbearing (lunge and chair rise) activities. Patellar tendon angle and moment arm were extracted from the subject’s PF and TF kinematics. Kinematics and the root mean square difference between non-weightbearing and weightbearing activities were compared across subjects and activities.

**Results:** Although tibia motion relative to the femur was consistent in trend, internal rotation and anterior translation increased with the weightbearing lunge (avg 2.7°, 1.6 mm) and chair rise (avg 2.7°, 2.0 mm). Patellar flexion was consistent with knee flexion across activities, however, patellar tilt and medial-lateral translation changed from non-weightbearing to weightbearing (avg RMSD 3.7°, 3.0 mm). Changes in PTA and PTMA from non-weightbearing to weightbearing were not significant.

**Conclusions:** While weightbearing elicited changes in knee kinematics, in most DOFs these differences were exceeded by intersubject differences. These results provide comparative kinematics for the evaluation of knee pathology and treatment in older adults, while emphasizing consideration of subject-specific kinematics.

### 4.2 Introduction

Reported measurements of natural kinematics of both the tibiofemoral (TF) and patellofemoral (PF) articulations of the healthy knee during activities of daily living are rare. Studies have reported precision kinematics for the tibiofemoral joint [57, 190, 191] and the patellofemoral joint [85, 192], yet only a few have investigated the combined motion of TF and PF joints in healthy living subjects [61, 193]. Combined measurements of TF and PF joint kinematics are necessary because knee pathologies and treatments frequently concern both articulations. In vitro studies have documented how alteration in TF joint translation and rotation can change PF joint function [194, 195, 88]. Likewise, PF dysfunctions, such as patellar maltracking, have been linked with combined abnormal motion of the tibia and
patella relative to the femur [196, 197]. Furthermore, complications following treatments at the tibiofemoral joint such as anterior cruciate ligament (ACL) repair, may cause scarring that produces a relative shortening of the patellar tendon [1], and procedures such as high tibial osteotomy and total knee arthroplasty may alter the joint line and create patella baja or pseudo-patella baja [169, 198]. Combined measurement of TF and PF kinematics also enables calculation of important quantities such as patellar tendon angle (PTA), which partly determines the loading vector at the tibiofemoral joint [199, 187, 200, 201], and patellar tendon moment arm (PTMA), which has been used as a surrogate measurement for the efficiency of the quadriceps [202]. The kinematics of the tibiofemoral and patellofemoral joints can change as subjects change demand on their knee and move from non-weightbearing to weightbearing. For example, Myers et al [56] measured an increase in anterior displacement of the tibia relative to the femur as demand (defined as increasing knee extensor moment) on the knee increased. This is significant because kinematic changes during weightbearing may alter surgical repairs to the knee made with the knee unloaded. For example, weightbearing can influence strain in repaired ligaments [83, 84] and tracking of the patella in the femoral groove [76]. Knowledge of how knee kinematics change with demand can provide kinematic targets for treatments that seek to restore normal function. However, the amount of kinematic change with weightbearing may vary dramatically between individuals, confounding attempts to create useful predictions of weightbearing kinematics [184]. Hume et al demonstrated that spline representations of knee kinematics may adequately predict knee motion of a group, but poorly predict individual kinematics. Significantly, treatments that seek to restore normal weightbearing motion to the knee might be confounded by patient-specific knee motion that may routinely fall outside population norms. Knee kinematics may also change with age. Several prior studies have evaluated young healthy, OA, and TKA subjects, and shown important differences between knee kinematics across these cohorts. However, with a few notable
exceptions [65], these studies routinely compare older adults with OA and TKA to young healthy controls. The motion of younger subjects may not be representative of the age range associated with knee pain, OA, and TKA. Older adults tend to use unique movement strategies during activities such as stepping-down or turning and pivoting [21, 22, 23]. While these studies and others [173] have noted that aging has an impact on knee kinematics measured using marker-based motion capture, few studies have examined the small translations and rotations of the knee in older adults with no history of knee pathology. Documenting knee kinematics in older adults is necessary for understanding the changes in knee kinematics that occur with OA and TKA. How normal knee kinematics change with demand warrants further scrutiny as norms of healthy knee kinematics in older adults may be dominated by intersubject differences. The goals of this research were (1) perform combined measurement of the differences in tibiofemoral and patellofemoral kinematics, PTA, and PTMA that occur from non-weightbearing to weightbearing, (2) determine if changes in tibiofemoral kinematics correspond with changes in patellofemoral kinematics, and (3) examine whether these differences in kinematics were variable between individuals.

4.3 Methods

4.3.1 Participants

Eleven healthy individuals with no history of injuries or surgeries to the lower limbs (5F/6M, age = 64 ± 7.9 years, body mass = 74.3 ± 19.7 kg, body mass index = 26.0 ± 4.7 Kg/m2, height = 165.7 ± 10.9 cm) provided informed consent and participated in the University of Denver Institutional Review Board approved study (Table 4.1).
Table 4.1: Subjects specifications including Insall-Salvati ratio (ISR) which is used to determine presence/absence of patella alta and patellar tendon length (PTL) with the highlighted values indicating patella alta.

<table>
<thead>
<tr>
<th>Subjects</th>
<th>Gender</th>
<th>Height (cm)</th>
<th>Weight(Kg)</th>
<th>Age</th>
<th>BMI</th>
<th>ISR</th>
<th>PTL</th>
</tr>
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<td>65.8</td>
<td>56</td>
<td>24.8</td>
<td>1.04</td>
<td>46.17</td>
</tr>
<tr>
<td>3</td>
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<tr>
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<td>1.05</td>
<td>45.35</td>
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<td>1.06</td>
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<td>73</td>
<td>23.3</td>
<td>1.34</td>
<td>50.30</td>
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</tbody>
</table>

4.3.2 Protocol

High speed stereo radiography (HSSR) was used to measure the kinematics of the TF and PF joints in each subject’s dominant knee [35, 86]. HSSR captures two radiographic views to enable three-dimensional tracking of the bones in the knee. The HSSR system is composed of two matching custom radiography systems with 40 cm (16”) diameter image intensifiers integrated with high-speed, high-definition (1080x1080) digital cameras [35]. The accuracy of the HSSR system for tracking the tibia and femur was confirmed by comparison to measurements made in bones with implanted tantalum beads [35]. Studies using similar dual-plane radiography equipment and methods recorded patella tracking accuracy of 0.46-1.14 mm in translation and rotational accuracy of 0.99°-1.04° [203, 176]. All activ-
ities were captured at 50 frames/second and obtained with pulsed radiography (pulse width 750 $\mu$s, 60 kV, and 63 mA). The effective dose of this experiment as a whole was 0.28 mSv, which was the combined effective dose from CT (0.16 mSv [204]) and from fluoroscopy (0.12 mSv, PCXMC, STUK, Helsinki, Finland).

4.3.3 Data Collection

Participants performed three activities: 1) knee extension in which the individuals were seated and slowly extended their knee from high flexion to full extension (seated knee extension); 2) weightbearing deep knee bend (lunge); 3) and standing up from a chair (chair rise) (Figure 4.1a). One subject did not complete the lunge, and so data from ten subjects were reported for that activity. Following the laboratory data collection, a static bone CT with a slice thickness of 1.0 mm was obtained of each subject’s dominant knee.
Figure 4.1: a) Participants performed seated knee extension, rising from a chair, and lunge with their dominant knee in view of the radiography system, b) 2D/3D image registration and c) the origin of the femoral coordinate system for each subject was defined by fitting a cylinder to the medial and lateral posterior condyles, with the center placed at the trochlea. The coordinate system of the tibia and patella was assigned coincident with the femoral coordinate system at full extension, d) patellar tendon angle (PTA) was taken as the line between the inferior pole of the patella, the tibial tuberosity, and the long axis of the tibia, and moment arm (PTMA) was calculated as the perpendicular distance from the patellar tendon to the instantaneous axis of rotation of the tibiofemoral joint.

4.3.4 Data Processing

Three-dimensional models of the distal femur, proximal tibia, and patella bones were reconstructed from the CT data using ScanIP (Simpleware Inc.). Positions of the three-dimensional bone models were matched to the two-dimensional stereo radiography images to quantify the translational and rotational pose of the tibia and patella relative to the femur (Autoscooper, Brown University, [177]) (Figure 4.1b). The origin of the femoral coordinate system for each subject was defined by fitting a cylinder to the medial and lateral posterior condyles, with the center placed at the trochlea (Figure 4.1c) [205, 46]. The medial-lateral (ML) axis was defined as the line through the long axis of the cylinder while the superior-
inferior (SI) axis was aligned to the posterior aspect of the femur. The anterior-posterior (AP) axis was defined as the cross product of the ML and SI axes. The coordinate system of the tibia and patella was assigned coincident with the femoral coordinate system during non-weightbearing full extension [46, 206] to reveal changes from each participant’s neutral pose [207]. The motion of tibia and patella was described relative to the femur [1], and filtered using a 4th order low-pass Butterworth filter with a cutoff frequency of 2 Hz for all three activities. The contact points of the femur on the tibial plateau were estimated with tibiofemoral low-point kinematics [30, 31, 191], calculated by finding the most distal points of the femoral condyles relative to the tibia as a function of flexion. Low-point kinematics of the femur relative to the medial/lateral tibial plateau were used to illustrate notable individual differences.

4.3.5 Data Analyses

Six kinematic quantities were reported with respect to tibial flexion angle: tibial internal rotation (TIE), varus rotation (TVV), anterior translation (TAP), and patellar flexion (PFE), tilt (PTilt), and medial translation (PML). In addition, patellar tendon angle (PTA) was taken as the line between the inferior pole of the patella, the tibial tuberosity, and the long axis of the tibia. Patellar tendon moment arm (PTMA) was calculated as the perpendicular distance from the patellar tendon to the instantaneous axis of rotation (IAR) of the tibiofemoral joint. The patella tendon was defined as a straight line from the patella apex to the tibial tuberosity, and the IAR was calculated using the equations described by Spoor and Veldpaus et al [179](Figure 4.1d). Root mean square deviation (RMSD) was used to quantify the differences between the non-weightbearing (NWB) seated knee extension and weightbearing (WB) lunge and chair rise for each individual and averaged across subjects (average RMSD). RMSD of TIE, TVV, TAP, PFE, PTilt, PML, PTA, and PTMA was cal-
culated in 10-degree increments from 40° to 80° of knee flexion. Comparisons were made within the common range of knee angles (40° to 80°) because the minimum and maximum knee angles differed across subjects and activities. In addition, RMSD values were scaled by the total excursion of each DOF during the task. Scaled RMSD provided context by revealing differences that exceed the range of motion of that DOF [25, 184, 208]. Accordingly, a scaled value of RMSD greater than 1.0 indicated a substantial change in RMSD that exceeded the range of motion during the activity. Paired Student’s t-tests were used to compare kinematic data between activities, and differences were considered statistically significant when p<0.05.

4.4 Results

Complete tabular patella kinematics for subjects 1 through 8 are provided at digital-commons.du.edu/orthopaedic_biomechanics.

4.4.1 Tibiofemoral Kinematics

On average, tibial internal rotation (TIE) increased with weightbearing (Figure 4.2A). Average TIE for the common flexion range was 13.3° ±5.3° for seated knee extension, 16° ±4.7° for lunge, and 16° ±7.4° for chair rise (Table 4.2 summarizes the results for TF kinematics for the entire range of motion for each activity and for the common flexion range), yielding an average RMSD from WB to NWB of 3.9° for lunge and 4.0° for chair rise (scaled RMSD 0.5 and 0.6, respectively, Table 4.3). Individual RMSD was lowest in Subject 5 (1.4 and 1.7, respectively), and highest in Subject 2 (3.7 and 6.2, respectively). Substantial variation in TIE was observed that remained consistent across activities (Figure 4.2C, Figure 4.3). Most of the subjects remained in tibial varus (TVV, Figure 4.2A, Figure 4.3) throughout the seated knee extension (avg. 1.9° ±3.7°), and during lunge and chair
rise (avg. 1.3° ±3.6°, 1.2° ±2.8°, respectively). The change in TVV with weightbearing was small (avg. RMSD 1.5 and 1.8 in lunge and chair rise, respectively). Even so, scaled RMSD was greater than 0.5 across almost all subjects due to the overall small excursion of TVV (Table 4.3). Tibial anterior translation (TAP, Figure 4.2A) during the lunge and chair rise was more anterior (avg 1.9 ± 2.4 mm during seated knee extension compared with 3.5 ± 1.7 mm for lunge and 3.9 ± 2.2 mm for chair rise, Table 4.2 resulting in RMSD values of 2.1 and 2.5 mm and scaled RMSD of 0.6 and 0.7 (Table 4.3). Low point kinematics indicated a more medial pivot for nine of the subjects for all three activities (e.g. Subject 10, Figure 4.4). However, two subjects displayed a medial pivot during seated knee extension that was not observed during the two weight bearing activities (e.g. Subject 3, Figure 4.4).
Figure 4.2: (A) Comparison of tibia kinematics relative to the femur showing average and one standard deviation of the three activities, (B) Comparison of patella kinematics relative to the femur showing average and one standard deviation of the three activities, (C) TIE range of motion for each subject for all three activities (the lunge was not performed by Subject 1).
Figure 4.3: Knee extension, lunge, and chair rise trials showing TIE and TVV rotation and TAP translation for all seventeen subjects.
Figure 4.4: Illustrative low-point kinematics for two subjects during the three activities showing consistent tibial rotation across activities with distinct subject differences.
Table 4.2: Maximum, minimum, total range of motion, and average kinematics for the tibia relative to the femur during the three activities over the total range of knee flexion and over the common knee flexion (40° to 80°) in parentheses.

<table>
<thead>
<tr>
<th>Knee Extension</th>
<th>TF External Rotation [deg]</th>
<th>TF Valgus Rotation [deg]</th>
<th>TF Anterior Translation [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum</td>
<td>-21.8±6.9(−16.9±6.1)</td>
<td>-3.3±2.8(−2.6±3.7)</td>
<td>9.0±4.7(4.1±3.2)</td>
</tr>
<tr>
<td>Minimum</td>
<td>0.2±0.9(−9.9±5.0)</td>
<td>2.2±2.1(−0.9±3.6)</td>
<td>-0.6±0.9(0.3±1.7)</td>
</tr>
<tr>
<td>Average</td>
<td>-11.4±4.4(−13.3±5.3)</td>
<td>-0.7±2.5(−1.9±3.7)</td>
<td>3.2±2.3(1.9±2.4)</td>
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<tr>
<td>ROM</td>
<td>21.9±6.4(7.1±2.9)</td>
<td>5.5±2.3(1.7±0.8)</td>
<td>9.7±4.1(3.8±1.8)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Lunge</th>
<th>TF External Rotation [deg]</th>
<th>TF Valgus Rotation [deg]</th>
<th>TF Anterior Translation [mm]</th>
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<tbody>
<tr>
<td>Maximum</td>
<td>-24.8±6.5(−19.3±5.8)</td>
<td>2.7±4.4(−2.4±3.6)</td>
<td>8.7±5.2(5.0±1.9)</td>
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<td>Minimum</td>
<td>-6.7±8.4(−12.7±4.4)</td>
<td>-2.5±3.6(−0.1±3.7)</td>
<td>1.5±1.8(2.1±1.7)</td>
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<td>Average</td>
<td>-17.9±6.0(−16.0±4.7)</td>
<td>0.2±4.4(−1.3±3.6)</td>
<td>5.5±3.4(3.5±1.7)</td>
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<tr>
<td>ROM</td>
<td>18.0±8.0(6.6±3.3)</td>
<td>5.2±1.9(2.4±0.9)</td>
<td>7.2±4.1(2.9±1.2)</td>
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<thead>
<tr>
<th>Chair Rise</th>
<th>TF External Rotation [deg]</th>
<th>TF Valgus Rotation [deg]</th>
<th>TF Anterior Translation [mm]</th>
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<tr>
<td>Maximum</td>
<td>-19.7±8.0(−19.0±7.4)</td>
<td>-2.4±2.8(−2.2±2.8)</td>
<td>6.1±2.6(5.3±2.5)</td>
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<td>Minimum</td>
<td>-3.3±4.9(−11.7±7.3)</td>
<td>0.7±2.4(−0.2±2.8)</td>
<td>-0.1±1.1(2.6±1.7)</td>
</tr>
<tr>
<td>Average</td>
<td>-11.7±6.1(−16.0±7.4)</td>
<td>-1.0±2.6(−1.2±2.8)</td>
<td>2.8±1.1(3.9±2.2)</td>
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<tr>
<td>ROM</td>
<td>16.4±7.9(7.3±3.6)</td>
<td>3.1±1.4(2.0±1.1)</td>
<td>6.2±3.2(2.7±1.3)</td>
</tr>
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</table>

Table 4.3: Root mean square difference (RMSD) between the non-weightbearing and weightbearing activities averaged across all subjects. Scaled RMSD was calculated as the RMSD divided by the ROM in each degree of freedom. Green highlights scaled RMSD values of 0.5 or greater, yellow highlights values of scaled RMSD of 1 or greater.

<table>
<thead>
<tr>
<th>RMSD</th>
<th>TF VV</th>
<th>TF IE</th>
<th>TF AP</th>
<th>PF FE</th>
<th>PF Tilt</th>
<th>PF ML</th>
<th>PFMA</th>
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<tr>
<td>Scaled Value</td>
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<td><strong>0.5</strong></td>
<td><strong>0.6</strong></td>
<td><strong>0.1</strong></td>
<td><strong>1.0</strong></td>
<td><strong>1.2</strong></td>
<td><strong>0.3</strong></td>
<td><strong>0.1</strong></td>
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<tr>
<td>Chair Rise</td>
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<td></td>
<td></td>
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<td></td>
<td></td>
<td></td>
<td></td>
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<tr>
<td>Scaled Value</td>
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<td><strong>0.6</strong></td>
<td><strong>0.7</strong></td>
<td><strong>0.1</strong></td>
<td><strong>0.9</strong></td>
<td><strong>1.3</strong></td>
<td><strong>0.1</strong></td>
<td><strong>0.1</strong></td>
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81
4.4.2 Patellofemoral Kinematics

All three activities demonstrated similar trends in PF flexion (PFF, Figure 4.2B) with an average of 39.4° ± 6.6°, 38.7° ± 5.9° and 37.1° ± 5.5° for seated knee extension, lunge and chair rise, respectively (Table 4.4 summarizes the results for PF kinematics for the entire range of motion for each activity and for the common flexion range). Average RMSD from WB to NWB was 3.7° for lunge and 4.7° for chair rise (scaled RMSD 0.1 and 0.1, respectively, Table 4.3). Average ML tilt (PTilt, Figure 4.2B) was medial (avg -4.7° ± 6.6°, -5.1° ± 6.7°, -3.8° ± 6.1° for seated knee extension, lunge and chair rise, respectively), while individually most subjects consistently maintained a medial or lateral tilt throughout the activities (Figure 4.5). Average RMSD was 3.7 for lunge and 3.6 for chair rise, which was similar to the ROM (average scaled RMSD was 1.0 in lunge and 0.9 in chair rise). Patellar ML translation (PML, Figure 4.2B) was medial in most subjects for seated knee extension and lunge (avg -1.9 ± 4.3 mm and -1.7 ± 4.3 mm, respectively), with increased medial position during chair rise (-4.1 ± 2.2 mm, Table 4.4). Individually, some of the subjects’ patellae remained in a more lateral position (Figure 4.5). Average RMSD was 2.8 for lunge and 3.2 for chair rise, which were greater than the ROM (avg scaled RMSD was 1.2 in lunge and 1.3 in chair rise, Table 4.3).
Figure 4.5: Knee extension, lunge, and chair rise trials showing PFFE and PTilt rotation and PML translation for all eleven subjects.
Table 4.4: Maximum, minimum, total range of motion, and average kinematics for the patella relative to the femur during the three activities over the total range of knee flexion and over the common knee flexion (40° to 80°) in parentheses.

<table>
<thead>
<tr>
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<th>PF Flexion [deg]</th>
<th>PF Lateral Tilt [deg]</th>
<th>PF Lateral Translation [mm]</th>
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<td></td>
<td></td>
</tr>
<tr>
<td>Maximum</td>
<td>82.1±5.7(56.3±7.4)</td>
<td>-11.8±6.7(−6.5±8.1)</td>
<td>-5.6±4.8(−3.1±4.3)</td>
</tr>
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<td>Minimum</td>
<td>-0.6±0.4(23.1±5.2)</td>
<td>3.8±5.2(−2.7±8.3)</td>
<td>4.1±3.4(−0.7±4.1)</td>
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<td>Average</td>
<td>34.9±12.1(39.4±6.6)</td>
<td>-4.6±6.7(−4.7±8.3)</td>
<td>-1.2±3.5(−1.9±4.3)</td>
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<tr>
<td>ROM</td>
<td>82.7±5.7(33.2±3.9)</td>
<td>15.6±4.3(3.8±1.7)</td>
<td>9.7±4.0(2.4±1.2)</td>
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<td><strong>Lunge</strong></td>
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<td>Maximum</td>
<td>72.8±11.5(52.8±6.1)</td>
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<td>Minimum</td>
<td>12.75±14.2(23.2±3.1)</td>
<td>1.7±10.9(−3.1±6.0)</td>
<td>1.8±5.2(0.2±4.9)</td>
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<td>Average</td>
<td>51.8±12.6(38.7±5.9)</td>
<td>-5.3±6.7(−5.1±6.7)</td>
<td>-1.6±4.7(−1.7±4.3)</td>
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<tr>
<td>ROM</td>
<td>60.1±14.5(29.5±4.1)</td>
<td>10.8±8.1(4.1±2.1)</td>
<td>7.4±4.7(3.7±2.8)</td>
</tr>
<tr>
<td><strong>Chair Rise</strong></td>
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</tr>
<tr>
<td>Maximum</td>
<td>53.8±6.7(50.1±5.0)</td>
<td>-7.4±5.7(−5.3±6.1)</td>
<td>-6.3±2.6(−5.8±2.9)</td>
</tr>
<tr>
<td>Minimum</td>
<td>-1.9±4.5(20.3±5.2)</td>
<td>-0.8±3.9(−2.2±6.1)</td>
<td>-0.3±2.8(−2.4±2.9)</td>
</tr>
<tr>
<td>Average</td>
<td>22.1±4.8(37.1±5.5)</td>
<td>-3.5±5.2(−3.8±6.1)</td>
<td>-2.8±2.1(−4.1±2.2)</td>
</tr>
<tr>
<td>ROM</td>
<td>55.8±9.4(29.8±4.3)</td>
<td>8.2±3.7(3.1±1.4)</td>
<td>6.1±3.7(3.4±3.9)</td>
</tr>
</tbody>
</table>

### 4.4.3 TF/PF Correlation

Strong correlation was found between tibial and patellar flexion for all three activities ($R^2 = 0.96$ for knee extension, $R^2 = 0.94$ for lunge and $R^2 = 0.92$ for chair rise) (Figure 4.6) (Table 4.5) with similar rate of change (0.69, 0.72, 0.70 for seated knee extension, lunge and chair rise respectively). The majority of subjects had increased patella ML tilt with an increase in TF IE rotation for all activities (Table 4.5). Additionally, strong correlation was found for most of the subjects for the TF IE rotation and PF FE rotation for the
seated knee extension while for chair rise and lunge less subjects were found to be strongly correlated for those DOFs. For seated knee extension strong correlation was found for most of the subjects for PF FE and TF AP.

Figure 4.6: Knee extension, lunge, and chair rise trials showing PFFE and TFFE rotation correlation for all eleven subjects.

Table 4.5: Correlation Values for TF and PF kinematics for different DOFs for seated knee extension, lunge and chair rise.

| # Sub | Knee Extension | | | Lunge | | | Chair Rise Up | |
|-------|----------------|---|---|---|---|---|---|---|---|
|       | PF AP / TF AP  | PF AP / TF IE | PF IE / TF IE | PF AP / TF AP  | PF AP / TF IE | PF IE / TF IE | PF AP / TF AP  | PF AP / TF IE | PF IE / TF IE |
| 1     | 0.853          | 0.456         | 0.983         | —              | —              | 0.716         | 0.855         | 0.934         |
| 2     | 0.582          | 0.886         | 0.489         | 0.469          | 0.923          | 0.958         | 0.718         | 0.858         | 0.617         |
| 3     | 0.777          | 0.963         | 0.839         | 0.461          | 0.368          | 0.383         | 0.631         | 0.990         | 0.279         |
| 4     | 0.884          | 0.938         | 0.682         | 0.753          | 0.905          | 0.983         | 0.952         | 0.949         | 0.950         |
| 5     | 0.066          | 0.782         | 0.941         | 0.265          | 0.963          | 0.889         | 0.109         | 0.989         | 0.809         |
| 6     | 0.911          | 0.864         | 0.012         | 0.997          | 0.976          | 0.848         | 0.878         | 0.887         | 0.473         |
| 7     | 0.960          | 0.854         | 0.833         | 0.983          | 0.612          | 0.371          | 0.949         | 0.957         | 0.512         |
| 8     | 0.968          | 0.914         | 0.056         | 0.978          | 0.815          | 0.470          | 0.008         | 0.888         | 0.972         |
| 9     | 0.886          | 0.957         | 0.663         | 0.949          | 0.108          | 0.009          | 0.163         | 0.924         | 0.868         |
| 10    | 0.956          | 0.777         | 0.600         | 0.965          | 0.913          | 0.765          | 0.812         | 0.960         | 0.702         |
| 11    | 0.952          | 0.829         | 0.048         | 0.014          | 0.178          | 0.504          | 0.029         | 0.430         | 1.000         |

4.4.4 Patellar Tendon Angle and Moment Arm

All subjects demonstrated consistent PTAs across the activities, with the highest PTAs at low knee flexion angles (e.g. avg. 11.7° ± 4.4° at full extension during seated knee extension) and the lowest PTAs at maximum knee flexion (e.g. avg. 1.0° ± 5.3° at 80° during
seated knee extension, Figure 4.7). Average PTA was 6.0° ± 8.4°, 1.7° ± 5.2° and 8.9° ± 5.8° for seated knee extension, lunge, and chair rise, respectively. Significant differences were found from 40° to 50° between knee extension and lunge and chair rise (p=0.002 and p=0.004, respectively). Excepting these differences, the individual PTA of each subject remained consistent during activities as demonstrated by low RMSD and scaled RMSD (Table 4.3). Across individuals, the differences in PTA were large compared with the differences between activities (Figure 4.7). Across activities, patellar tendon moment arm (PTMA) was highest at full extension (e.g. 51.3 ± 1.4 mm during seated knee extension) and lowest at high knee flexion angles (e.g. 38.2 ± 1.9 mm at 80° during seated knee extension, (Figure 4.8)). Average PTMA was 41.8 ± 8.3, 39.8 ± 5.9, 47.1 ± 5.5 mm, for knee extension, chair rise, and lunge, respectively (Figure 4.8). Although there were large differences between individuals (Figure 4.8), no significant differences were found between activities as further demonstrated by the low RMSD and scaled RMSD (Table 4.3).

4.5 Discussion

Quantification of natural knee kinematics is essential for assessment of joint function in the diagnosis of pathologies, such as progression of osteoarthritis and patellar instability, and for evaluation of outcomes following conservative or surgical treatment. In this study, a high speed stereo radiography system was used to investigate multiple DOFs of TF and PF kinematics in healthy older adults during a non-weight bearing seated knee extension and two weight-bearing activities, lunge and chair rise. Tibia motion relative to the femur was consistent in trend during the seated knee extension, lunge and chair rise, with the exceptions of increased internal rotation and anterior translation during weightbearing. The axis of tibial rotation was on the medial side of the knee (medial pivot) for all three activities, with notable intersubject differences in the amount of rotation (Figure 4.2C). Average
patella ROM during the lunge and chair rise activities was within the ROM of the knee extension, demonstrating consistent tracking of the patella, however, moving from non-weightbearing to weightbearing created changes in patellar tilt and medial-lateral translation. Although changes in PTA and PTMA from non-weightbearing to weightbearing were small, substantial individual differences were measured. Like prior studies, general trends in tibia motion relative to the femur were similar for all subjects across activities. Tibial kinematics during seated knee extension were in agreement with Myers et al [56] where peak internal rotation for unweighted knee extension was $14.5^\circ \pm 7.7^\circ$ at $90^\circ$ of knee flexion (cf. $16^\circ$ in Figure 4.2A) and peak anterior translation was $2.6 \pm 2.1$ mm (cf. 4 mm in Figure 4.2A). In further agreement with their findings, weightbearing caused changes in
Figure 4.8: Patellar tendon moment arm (PTMA) across subjects during the seated knee extension (up left) during lunge (up right), during Chair Rise (bottom left) and comparison of average and one standard deviation across activities (bottom right).

kinematics. In the current study, internal rotation and anterior translation increased during lunge and chair rise. However, the average RMSD values for TIE (3.9 and 4.0 for lunge and chair rise) were less than the TIE standard deviation of the subjects (Table 4.2 and Figure 4.2A), suggesting tibial rotation was more sensitive to subject differences than weightbearing. Furthermore, on average, the axis of tibial rotation was located on the medial side of the knee lending support to the concept of medial pivot in total knee arthroplasty [209], however the amount of internal rotation varied widely between subjects (Figure 4.2C) with some showing only moderate evidence of medial pivot (Figure 4.4). TAP increased with weightbearing for most subjects as expected from prior reports (Figure 4.2A) [56], however the impact was moderate relative to the ROM of TAP (e.g. scaled RMSD was 0.7 for
chair rise). Conversely, TVV rotation was consistent for all three activities as demonstrated by the small average standard deviation (Table 4.2 and Figure 4.2A) and the low RMSD values compared to the other DOFs (Table 4.3). These results agreed with several studies that have measured small amounts of TVV during activities [46, 210]. However, TVV too was affected by weightbearing as demonstrated by the scaled RMSD values that showed these changes were on the order of the natural TVV range of motion (e.g. scaled RMSD of 1.1 for chair rise). As shown in prior studies [194, 34], there was a strong correlation between patellar flexion and tibiofemoral flexion both on average and across individuals (e.g. \( \text{avg } R^2 = 0.96 \) for knee extension). Moreover, weightbearing had the smallest effect on patellar flexion with the lowest RMSD and scaled RMSD of all the kinematics measured (Table 4.3). In contrast, weightbearing changed both Ptilt and PML. The change in Ptilt with weightbearing was medial in some subjects and lateral in others and as great as the unweighted ROM leading to scaled RMSD of 0.9 and 1.0 for lunge and chair rise. Greater PML was observed for the chair rise compared to seated knee extension and lunge. In the results of Koh et al [85], the patella shifted laterally with knee flexion (9 mm from 0 to 60° of flexion) for a seated knee extension, while Nha et al [86] found a medial translation of the patella during a weightbearing lunge exercise at low flexion angles (up to 30° of knee flexion) followed by lateral translation up to 90° of flexion. This medial to lateral movement of the patella was similar to the trend found in our results (Figure 4.2B). The large variability (Table 4.4) and scaled RMSD of 1.2 and 1.3 for lunge and chair rise, respectively (Table 4.3), indicated the dependence of PML on subject and activity differences. While several studies have presented the tibiofemoral or patellofemoral kinematics separately, there are only a few studies that have focused on investigating the kinematic relationship between joint motions [194, 195, 34, 193]. In agreement with in vitro studies, both tibial internal rotation and patellar medial tilt increased with knee flexion [194]. Some in vitro studies have shown the impact of increased tibial rotation on patellar tracking. For
example, Li et al [58] demonstrated that increased external tibial rotation and increased posterior tibial translation raised patellofemoral contact pressures on the lateral facet of the patella. In vivo, Li et al [61] described a strong correlation between external rotation of the femur relative to the tibia (or tibial internal rotation) and patellar lateral tilt and medial translation (relative to the tibia). As both increase with knee flexion (Figures 4.2A, 4.2B), this correlation was also present in most of our subjects (avg $R^2 = 0.6$, $R^2 = 0.6$, $R^2 = 0.7$ for knee extension, lunge and chair rise respectively), however, those who utilized greater external/ internal tibial rotation during their movements did not show greater patellar tilt or translation. In other words, inducing high amounts of tibial rotation may influence patellar tracking in vitro [58], but this did not necessarily describe our subjects who naturally utilize greater tibial rotation in movement. For instance, Subjects 3 and 10 who displayed very different tibial rotation and low point kinematics in Figure 4.4 had similar patellar kinematics. This was also true for subjects with greater anterior translation during weight-bearing who did not demonstrate greater changes in patellar kinematics. Scaled RMSD of the tibiofemoral kinematics did not correlate with the scaled RMSD of the patellofemoral kinematics, suggesting that high variability in one DOF did not indicate high variability in another.

### 4.5.1 Patellar Tendon Angle and Moment Arm

Our results for PTA were consistent with the magnitude and trend of previous studies that measured PTA during knee extension and deep knee flexion [181, 187, 87, 188]. In agreement with prior reports, the patellar tendon angle was positive (angled forward relative to the long axis of the tibia) with the knee extended and decreased with knee flexion (Figure 4.7). Differences in PTA were small between NWB and WB (e.g. scaled RMSD was 0.1 for lunge and chair rise). Even so, there were large differences in PTA between subjects,
spanning from $7^\circ$ to $21^\circ$ at full extension (Figure 4.7), which is important due to the notable influence of PTA on patellofemoral [211], tibiofemoral [212], and ligament loads [213]. Our results for PTMA were supported by values reported in previous studies. In Krevolin et al [214] the peak moment arm ranged from 40-60 mm and occurred at about $45^\circ$ of knee flexion. Our results for PTMA also ranged from 40-60 mm over the same range of knee flexion but peaked between 0 and $30^\circ$. These results were more in line with the study of Ali et al [181] where similar trends in PTMA were observed for a simulated deep knee bend. The effect of weightbearing on PTMA was small, however, like PTA, there were large differences between individuals. For instance, PTMA ranged between 45 to 60 mm at full extension (Figure 4.8) across subjects. Substantial variations in PTMA can have implications for repairs that involve preservation of the quadriceps mechanism, such as total knee arthroplasty [215] and the calculation of muscle forces using musculoskeletal models that frequently assume the same moment arm across subjects [216, 217].

4.5.2 Limitations

There were several notable limitations of this study. The results were based on the measurement of one trial for each activity and subject. Repeated measurements may have given better results by enabling assessment of intra-subject variability, however this was deemed unjustifiable for the additional x-ray exposure. In addition, compared to traditional motion capture, the number of subjects might be considered relatively small; however, subject numbers were comparable to other studies of knee kinematics using similar imaging technology [61, 210]. The number of subjects did not allow statistical examination of differences between male and female subjects. Even so, TF kinematics between the male and female subjects were observed to be similar, although there was a trend toward greater medial Ptilt and lateral translation in the female subjects in weightbearing. Our study
was focused on measurements in older adults, which may differ from results in younger cohorts. We chose to measure older adults because the motion of younger subjects may not be representative of the age range associated with knee pain, OA, and TKA. Finally, using the femoral shaft proximal to the knee rather than the hip center to define the sagittal orientation of the vertical axis of the femoral coordinate system might have created error in the definition of zero flexion angle. We chose not to record the hip joint center during CT to reduce x-ray exposure.

4.5.3 Conclusions

This study investigated the natural kinematics of the tibia and patella relative to the femur in healthy older adults, using a high speed stereo radiography system. Measurements were made during a non-weightbearing activity and two weightbearing activities for a cohort of older adults. While weightbearing elicited changes in knee kinematics, in most DOFs intersubject differences exceeded the differences observed due to weightbearing. Similarly, patellar tendon angle and moment arm were consistent for the three activities but varied substantially among subjects. These results provide comparative kinematics for the evaluation of knee pathology and treatment in older adults, and emphasize the need for considering subject-specific kinematics.
Chapter 5

In vivo comparison of rotating platform and fixed bearing knee replacements during lunge and pivot activities

5.1 Abstract

Aims: The objective of this study was to discover whether notable differences in mobile and fixed-bearing kinematics occur during activity that promotes tibial rotation, and to compare these results with normal healthy kinematics. We hypothesized that rotating-platform knee replacements would exhibit greater rotation of the tibia relative to the fixed-bearing knee replacements.

Materials and Methods: The in vivo motion of the tibia relative to the femur was measured in subjects with posterior stabilized fixed-bearing (FB) and rotating-platform (RP) total knee arthroplasties using a high-speed stereo radiography system during a lunge and gait with a change in direction (pivot).
**Results and Conclusion:** The in vivo tibiofemoral kinematics of mobile and fixed-bearing total knee prostheses were similar during two activities of daily living that included an activity that challenged tibial rotation. Measurements of internal/external (IE) rotation had higher variability and significantly greater IE range of motion with the RP compared with the FB, and this greater amount of variability of RP was not unlike the healthy knee.

**Clinical relevance:** The patterns in kinematics for both designs were similar to healthy data but with less IE rotation, yet the RP implants more closely replicated the asymmetrical posterior condylar translation and range of motion of the healthy knee during activity that challenges tibial rotation.

### 5.2 Introduction

The main purpose of total knee arthroplasty (TKA) is to alleviate pain and to restore function in subjects with severe knee osteoarthritis. To maintain the necessary range of motion without overloading the surrounding soft tissue structures, replication of tibiofemoral kinematics of the native knee is considered to be essential. Successful functional outcome following TKA is influenced by the geometry and design of the components as well as their interaction with the soft tissue surrounding this articulation. A big variety of different prosthetic designs exist in TKA, for example based on the condylar geometry design, posterior cruciate retention versus substitution, bearing mobility, bone versus ligament referenced placement, cemented versus uncemented placement, with or without patella resurfacing etc. Fixed-bearings (FB) are most commonly used with respect the bearing mobility feature. Mobile-bearing TKAs were introduced in the 1970s by Goodfellow and O’Connor [52] as an alternative choice to fixed bearing TKAs in an effort to accommodate greater internal/external (IE) rotation, more like the native knee. In the normal knee, internal rotation of the tibia occurs with knee flexion as the lateral condyle of the femur experiences greater
posterior motion on average than the medial condyle [31, 218]. In addition, activities such as turning while walking can require greater relative rotation [46]. Mobile-bearing prosthesis may better enable tibial rotation by allowing asymmetrical posterior translation of the femoral condyles [216]. Today, available components for knee arthroplasty are routinely offered in rotating and fixed-bearing alternatives. However, the choice between rotating-bearing or fixed-bearing components remains unclear due to uncertainty in the actual kinematic distinction between the two alternatives in vivo. The rotating platform (RP) in TKA is based on a low contact stress (LCS) theory, in which a bearing with higher upper-surface congruency is compensated by undersurface mobility which reduces articular contact stress without enhancing bone-implant interface stress. This could make the implant to last more since polyethylene wear and aseptic loosening will be reduced. Wear is less of a problem with the continuing development of polyethylene quality, however loosening remains a problem in current TKA. Even so the primary RP knees or its derived implant designs are accepted as an effective implant design [219, 220], the clinical results demonstrate no significant differences than those for standard fixed bearing (FB) knees [221]. Namba et al [222] reported mobile bearing non-posterior-stabilized variants demonstrated a higher risk of failure than FB non-posterior-stabilized designs. Several in vitro studies [16, 223] have demonstrated that mobile-bearing prostheses have higher articular conformity, less contact stress, and lower polyethylene wear rates relative to fixed-bearing prostheses, although wear performance in vivo are similar [52, 224]. Also, no differences have been observed between the two designs in patient reported outcomes [225, 226] and knee flexion range of motion [226, 227, 111]. In vivo measurements of knee kinematics using fluoroscopy have noted only slightly greater tibial rotation on average in knees with rotating bearings during stance, kneeling and deep knee bend [228, 229] or in some cases no difference in tibial rotation during step up [230, 71]. Ranawat et al [222] reported an average tibial internal rotation of 4.1 degrees for a FB TKA design and 7.3 degrees for
a mobile-bearing TKA design during a deep knee bend. Dennis et al [231] showed, that maximum rotation angles in 76 posterior-cruciate sacrificed mobile-bearing TKAs and 212 posterior-stabilized FB TKAs were absolutely similar (5.5 degrees of internal rotation and 2.1 degrees of external rotation, during a deep knee bend). As the amount of tibial rotation also depends strongly on condylar surface design [232], it is important to assess the kinematics of commonly available contemporary designs. In addition, prior work has focused on sagittal plane activities and ignored turning motions likely to test the axial rotation kinematics of a rotating platform design. Measurement of kinematics during turning motions are challenging using single-plane fluoroscopy due to poor out of plane accuracy, but can be achieved using bi-plane or stereo radiography setups. The objective of this study was to discover whether notable differences in mobile and fixed-bearing kinematics occur during activity that promotes tibial rotation, and to compare these results with normal healthy kinematics. To achieve this, we measured the in vivo tibiofemoral kinematics of posterior stabilized fixed-bearing and rotating-platform total knee arthroplasties using a high-speed stereo radiography system during a lunge and gait with a change in direction (pivot). We hypothesized that rotating-platform knee replacements would exhibit greater rotation of the tibia relative to the fixed-bearing knee replacements.

5.3 Methods

Twenty-nine subjects (12M/17F, 62.9±10.4 years, 2.9±4.9 years post-surgery, BMI: 26.4±3.2 kg/m2) with posterior-stabilized rotating-platform (RP) total knee replacements and nine subjects (3M/6F, 62.8±10.6 years, 2.9±4.9 years post-surgery, BMI: 26.5±3.2 kg/m2) with posterior-stabilized fixed-bearing (FB) total knee replacements. Twelve subjects had bi-lateral TKAs of which both knees were included in the study, resulting in a total of fifty knees tested, 38 RP and 12 FB. Both designs are from the same knee system.
All subjects provided informed consent and were recruited for this study after approval by the Institutional Review Board of the University of Denver. All subjects were at least nine months post-operation, received Attune® (DePuy-Synthes, Warsaw, IN) components, were free of limiting lower extremity pathology, and able to perform the activities required by the study. All subjects scored greater than 69 on Part 2 - Function of the Knee Society Score (KSS) questionnaire. Each subject performed two activities: a single-leg weight-bearing lunge and a pivot turn during gait (consisting of a 90° internal change in direction with the planted foot of the examined knee) (Figure 5.1a). High-speed stereo radiography (HSSR) was used to capture accurate 3D measurement of bone and implant motion [174]. The HSSR system is composed of two 40 cm diameter image intensifiers with high-speed, high-definition (1080x1080) digital cameras positioned at a relative 70° angle [174]. The accuracy of the HSSR system in implant tracking is 0.2 ± 0.1 mm in translation and 0.4° ± 0.3° in rotation [174], comparable to other stereo radiography systems [195]. Collection frequency of radiography images was 50 Hz for lunge and 100 Hz for pivot. Femur and tibial CAD geometries for each subject were obtained from the manufacturer. The three-dimensional CAD geometries were used to track the motion of the femoral and tibial components from the two-dimensional stereo radiography images (Autoscooper, Brown University, www.xromm.org [233]), (Figure 5.1b). Local coordinate systems were defined for each implant component as follows: the origin of the femoral coordinate system was located along the flexion-extension axis of femoral condylar geometry between the most medial and lateral points. The normal direction of the distal plane on the fixation surface was used to define the superior-inferior (SI) axis orientation, and the edge points of that plane to define the medial-lateral (ML) axis. Anterior-posterior axis was defined using the cross product of the SI and ML axis (Figure 5.1c). The coordinate system of the tibia was assigned coincident with the femoral coordinate system at full extension [46, 206]. Motion of tibia relative to the femur was calculated using methods described by Grood and
Suntay [234]. The kinematics of tibia were filtered using a 4th order low-pass Butterworth filter with a cutoff frequency of 2 Hz for lunge and 6 Hz for pivot. In addition, rollback of the medial and lateral femoral condyles relative to the tibial baseplate were estimated with tibiofemoral low point kinematics [30, 31, 218]. Low point measurements were calculated by finding the most distal points of the femoral geometries relative to the tibia as a function of flexion. Tibial rotation relative to the femur in FB and RP TKA was quantified by comparison of the average and standard deviation of the internal/external range of motion (IEROM) during the two tasks, and by comparison of the femoral condyle low-points of the medial and lateral femoral condyles. In addition, comparisons were made with prior results for the healthy knee from the same laboratory [46]. Average IEROM was calculated as the difference between the maximum and the minimum for each subject and then averaged across subjects for each activity. To present a more complete picture of kinematic differences, VV rotation and AP translation were also compared for the two activities. Unpaired student t-tests were used for matched age, gender, BMI and number of subjects and differences were considered statistically significant when p<0.05. To accommodate for the uneven number of subjects between the two different populations, a resampling-bootstrap procedure [235] was performed to estimate the difference between sample means, medians and correlation coefficients for the average IEROM.
Figure 5.1: a) Lunge and pivot activities performed by the participants. b) 3D image registration to the 2D stereo images and c) The femoral coordinate system fixed in the femoral component.

5.4 Results

Single Leg Lunge

During lunge, the average IE angle (Figure 5.2a) was greater for the RP (-6.1° ± 2.8°) than for the FB (-3.1° ± 1.4°), and subjects with the RP also utilized significantly greater IEROM (avg 9.7° ± 4.4°, 5.7° ± 2.6° for the RP and FB, respectively) (Table 5.1). The mean difference between FB and RP IEROM was 4.0° and the true mean fell between 1.9° (minimum of 95% CI) and 5.9° (maximum of 95% CI) while the median (3.6° ± 1.2°) was between 1.2° and 5.9° and since the mean and median values do not cross zero the two groups were significantly different. The standard deviation of the mean-difference distribution was 1.0° with 95% confidence. Additionally, no significant correlation was found for
the two groups, where the mean correlation coefficient was 0.03 ± 0.4 with minimum -0.6 and maximum 0.7 of 95% confidence interval. The RP design also had greater variability (4.3°, 2.9° for RP and FB, respectively) but it was not statistically significant (p=0.1). Increased internal tibial rotation with knee flexion followed the same trend as prior results for the healthy knee, but the amount of rotation was significantly less in both FB and RP subjects (Figures 5.3a, 5.4). The true mean (8.4° ± 1.2°) lies between the minimum 6.0° and maximum 10.5° of the 95% confidence interval for the RP and healthy knee comparison showing significant difference between the two groups. Similarly, significant difference was found between the FB and healthy knee data where the true mean (12.3° ± 0.3°) is between 11.6° (minimum) and 13.0° (maximum) of 95% confidence interval. Location of the lowest points of the femoral component with respect to the tibial baseplate were calculated from 0 to 80° of knee flexion for lunge (Figure 5.5). On average, the medial lowest point of the RP translated 1.5 ± 0.8 mm which was significantly different from the lowest point of the subjects with FB that translated 5.2 ± 0.4 mm on the medial side (p<0.001). The lateral side of the RP design translated 7.3 ± 1.4 mm (Figures 5.5, 5.6, 5.7) and was found significantly different when compared with the lowest points of the subjects with the FB design that translated 7.9 ± 0.8 mm on the lateral side (p<0.001). The average rotation of the lowest points was found significantly different among the two groups (p<0.001) (9.0° ± 2.1° and 3.8° ± 0.9° for the RP and FB subjects, respectively) (Figure 5.5). All subjects demonstrated similarly increasing anterior translation with flexion (Figure 5.2a), with no significant difference between RP and FB. RP subjects commonly demonstrated distinct kinematics relative to the FB subjects (representative subject, Figure 5.6). On average, AP translation was 0.5 ± 3.0 mm for RP, while for FB the average was 0.1 ± 1.9 mm (Fig 2a). Varus Valgus (VV) rotation was also not significantly different among subjects (Figure 5.2a); average VV angle was -0.5° ± 0.2° for the RP subjects and -0.4° ± 0.1° for the FB.
Table 5.1: Average and average range of motion for TF kinematics in the lunge and pivot.

<table>
<thead>
<tr>
<th>RELATIVE</th>
<th>TF External Rotation [deg]</th>
<th>TF Valgus Rotation [deg]</th>
<th>TF Anterior Translation [mm]</th>
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</thead>
<tbody>
<tr>
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<td></td>
<td></td>
</tr>
<tr>
<td>Average</td>
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<td>0.5±3.0</td>
</tr>
<tr>
<td>ROM</td>
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<td>1.6±0.9</td>
<td>9.9±4.0</td>
</tr>
<tr>
<td><strong>Pivot RP</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Average</td>
<td>2.5±1.8</td>
<td>0.2±0.2</td>
<td>-0.02±0.2</td>
</tr>
<tr>
<td>ROM</td>
<td>10.9±5.0</td>
<td>2.1±1.3</td>
<td>3.0±1.8</td>
</tr>
<tr>
<td><strong>Lunge FB</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Average</td>
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<td>-0.4±0.1</td>
<td>0.1±1.9</td>
</tr>
<tr>
<td>ROM</td>
<td>5.7±2.6</td>
<td>1.4±0.6</td>
<td>7.2±3.8</td>
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<tr>
<td><strong>Pivot FB</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Average</td>
<td>1.6±1.3</td>
<td>-0.01±0.1</td>
<td>0.5±0.2</td>
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<td>ROM</td>
<td>6.3±3.1</td>
<td>1.9±1.1</td>
<td>2.8±1.1</td>
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</table>
Figure 5.2: a) Kinematics of the Lunge activity for all subjects shown in dashed lines over median and one standard deviation b) Kinematics of the Pivot activity for all subjects shown in dashed lines over median (bold lines) and one standard deviation (shaded regions).
Figure 5.3: Kinematics of the Lunge (a) and Pivot (b) compared with healthy data from Kefala et al (2017), bold lines are the median and shaded regions are one standard deviation.

**Pivot Turn During Gait**

Most of the RP and FB subjects began the pivot with tibial internal rotation during weight acceptance, then completed the turn with tibial external rotation (Figure 5.2b) before toe-off. Average IE angle for the subjects was similar (RP $2.5^\circ \pm 1.8^\circ$ and FB $1.6^\circ \pm 1.3^\circ$) (Table 5.1). The variability as quantified by average standard deviation was found to be $4.5^\circ$ for the RP design and $2.3^\circ$ for the FB. Average IEROM was significantly greater for the RP subjects ($10.9^\circ \pm 5.0^\circ$) than for the FB subjects ($6.3^\circ \pm 3.1^\circ$) (Figure 5.4) where the true mean ($4.6^\circ \pm 1.2^\circ$) lies between the minimum $2.2^\circ$ and maximum $6.6^\circ$ of the 95% confidence interval and the median ($6.6^\circ \pm 1.5^\circ$) lies between $3.0^\circ$ (minimum) and $8.6^\circ$ (maximum). Moreover, no significant correlation was found among the two groups where the mean correlation coefficient was found to be $0.1 \pm 0.2$ with a minimum of -0.3 and a maximum of 0.6 of 95% confidence interval. External rotation increased during the pivot.
following the same trend as prior results for the healthy knee, but the amount of rotation was less for FB and RP TKA subjects (Figure 5.3b). The IEROM of the FB design was statistically different when compared to the healthy knee where the mean difference (8.8° ± 0.3°) lies between 8.4° (minimum) and 9.4° (maximum) for the 95% confidence interval (Figure 5.4). No significant difference was found for the IEROM of the RP design when compared to healthy data where the mean difference (4.2° ± 1.2°) lies between 1.7° (minimum) and 6.5° (maximum) for the 95% confidence interval. During pivot, the position of the femoral low points with respect to the tibial baseplate translated posteriorly on the medial side of both designs (3.6 ± 0.6 mm for RP, 5.8 ± 0.5 mm for FB) and anteriorly on the lateral side (3.6 ± 0.9 mm for RP, 4.4 ± 0.6 mm for FB) (Figures 5.5, 5.6, 5.7). The average rotation between low points was significantly different between the two designs (p<0.001) and found to be 5.1° ± 0.7° for the RP subjects and 2.9° ± 0.8° for the FB. RP subjects commonly demonstrated distinct kinematics relative to the FB subjects (representative subject, Figure 5.6). Although similar values of AP translation were observed for both RP and FB for both activities, average motion range in AP was higher for RP (9.9 ± 4.0 mm) compared to FB (7.2 ± 3.8 mm) (Table 5.1) for the pivot turn. The amount of VV of the subjects during pivot was small like that during the lunge (Figure 5.2b).

![Graphs showing comparison of IEROM between FB, RP and healthy data from Kefala et al (2017) for lunge (left) and pivot (right).]

Figure 5.4: Comparison of IEROM between FB, RP and healthy data from Kefala et al (2017) for lunge (left) and pivot (right).
Figure 5.5: Comparison of Medial and Lateral low point kinematics for the Lunge and Pivot. All subjects are shown in dashed lines over median (bold lines) and one standard deviation (shaded regions).
Figure 5.6: a) Low point kinematics of a representative RP subject and FB subject during lunge, b) low point kinematics of a representative RP subject and FB subject during pivot.
Figure 5.7: Average medial and lateral AP low point kinematics for both RP and FB groups for all subjects during a) lunge and b) pivot. Blue and red solid lines are the average RP and FB lateral AP low point kinematics respectively while the blue and red dash lines are the average RP and FB medial AP low point kinematics respectively.

5.5 Discussion

The objective of this study was to discover whether notable differences in mobile and fixed-bearing kinematics occur during activities that promote tibial rotation, and to compare these results with normal healthy kinematics. A high-speed stereo radiography system was used to investigate 6DOF TF kinematics during lunge and pivot. During the lunge and pivot, TF kinematics for both designs demonstrated similar trends but with significantly greater IEROM and greater IE rotation intrasubject variability for the RP design. The trend in kinematics of FB and RP was similar to healthy data, however, the amount of IE rotation was less than healthy for both activities, with the RP design demonstrating less difference in the amount of IE rotation from the healthy knee. The IE rotation between the FB and RP groups was similar during both activities, with the RP demonstrating significantly greater IEROM and greater variability. During the lunge, which required higher knee flexion angles than the pivot, IE rotation angle increased with flexion angle with both implant designs (Figure 5.3). Patients with the RP design had significantly greater average IEROM (9.7°)
than the FB group (5.7°) which was similar to the results of Shi et al [236] and Tamaki et al [69] during a deep knee bend activity with a posterior-stabilized fixed bearing design. Ranawat et al [229] also found that a mobile bearing, posterior stabilized implant had greater axial rotation than the fixed bearing implant replacements in their study during a deep knee bend activity. In contrast, Liu et al [54] found no significant difference in tibiofemoral axial rotation between fixed bearing and mobile-bearing posterior stabilized knee prosthesis from squatting to standing. Likewise, Wolterbeek et al [71] found no difference between a fixed bearing design and posterior stabilized RP during a step-up motion, and Shi et al [236] found no significant difference during a weight bearing deep knee flexion exceeding 120°. The pivot, an activity that challenges the axial rotation of the lower limb, revealed a significantly greater average IEROM of 10.9° for the RP, and 6.3° axial rotation for the FB, supporting the results found during the lunge. While the allowed rotation of the RP design had great effect on IE rotation of the RP subjects, there was not a significant effect on AP or VV. The FB and RP posterior stabilized knee replacements had similar AP translation and VV rotation for both activities. Liu et al [237] also found no significant difference in AP translation between fixed bearing and mobile-bearing posterior stabilized knee prosthesis from squatting to standing. These results were further supported by the step-up motion measured by Wolterbeek et al [71] and weight-bearing deep flexion measured by Shi et al [70]. Notably, these results were also consistent for the pivot activity that induced greater excursion of tibial rotation. The two implant designs demonstrated kinematic trends that were similar with normal knee kinematics, but with significant differences in IE rotation (Figure 5.3). During the lunge, increasing flexion angle corresponded with increased internal tibial rotation with both implants but less than that utilized by subjects with normal knees (Figure 5.3). The amount of rotation measured in the RP design during pivot was higher compare to FB design similar to healthy knee IEROM. These results supported our hypothesis that RP knee replacements would exhibit greater rotation of
the tibia relative to the FB knee replacements, more like healthy healthy motion. In addition, the RP displayed greater intersubject variability (Figure 5.3b) and IEROM (Figure 5.4) for both activities, more like the individual differences observed during the pivot of healthy subjects. Even so, the variability and ROM of IE and VV in healthy subjects performing the lunge was much greater than in the implanted subjects. AP translation for both designs only partially restored the anterior tibial translation observed in the normal knee (Figure 5.3). The progression of low points across the medial and lateral sides of the proximal tibia was unique between the lunge and pivot, demonstrating the need for activities that challenge rotation when comparing RP and FB designs. For the lunge activity, and in both the RP and FB designs, the medial condyle moved anterior and the lateral condyle moved posteriorly on average as the knee was flexed (Figure 5.5). In their measurements of deep knee bend, Ranawat et al [229] and Shi et al [70] also showed some anterior motion of the medial condyle and posterior translation of the lateral condyle of a mobile-bearing prosthesis. In contrast, Delport et al [228] compared contact position during stance with kneeling and found a mostly steady contact position on the medial and lateral sides of the FB and posterior translation of the medial and lateral sides of the RP with flexion. Unlike some prior reports [228], the variability of the FB was less than RP (Figure 5.5). The greater intersubject differences of the RP can be explained by rotation of the insert as shown in Figure 6 with rotation occurring near the center of the insert, however, the greater variability of the RP knees did not approach that described in the healthy knees (Figure 5.3). For the pivot activity, movement of the medial and lateral low points reversed direction from the lunge. The FB and RP lateral low points moved anterior (opposite the lunge) due to external rotation of the tibia (whereas the tibia rotated internally on average during the lunge). For some subjects, insert rotation occurred in the RP in the opposite direction of the lunge (compare RP in Figure 5.6). This study was limited by the smaller number of subjects with a fixed bearing prosthesis. However, this is justified by the low variability
in kinematics of the fixed bearing subjects. In addition, only a single trial of each activity was used for measurement of implant kinematics, removing the possibility of examining intrasubject variability. Processing only one trial of each activity was deemed justifiable to limit the total x-ray exposure of the subjects. This study showed that the in vivo TF kinematics of mobile and fixed-bearing total knee prostheses were similar during two activities of daily living including an activity that challenged rotation. Measurements of IE rotation had higher variability with the RP compared with the FB, and this greater amount of variability of RP was not unlike the healthy knee. The patterns in kinematics for both designs were similar to healthy data but with less IE rotation, yet the RP implants more closely replicated the asymmetrical posterior condylar translation of the healthy knee.
Chapter 6

Differences in 3D Functional Pelvic Orientation between Static and Dynamic Activities in THA

6.1 Abstract

Acetabular component placement is directly associated with the incidences of the two most common causes of total hip arthroplasty (THA) failure, hip dislocation and wear. The pelvis tilts during postural changes and since the acetabulum is part of the pelvis, pelvic motion changes the anteversion and inclination of the acetabulum. Pelvic orientation between supine, sitting and standing is unique to each patient and can vary dramatically within each position. Furthermore, certain pelvic positions in patients with spinal disorders can affect acetabular alignment after THA, increasing the risk of dislocation in certain patient populations. High-speed stereo radiography (HSSR) was used to estimate PI and spinopelvic functional parameters (PTS, SS, APPA) during dynamic activities and across different populations. Four subjects (5M, 68.6±5years, BMI: 26.1±2.9kg/m2) that
include two healthy subjects, one THA, and one spinal-stabilized cohort, were enrolled for the purpose of this study. The highest pelvic arc of motion was observed for one of the healthy subjects (38.7°) while the minimum pelvic ROM was observed for the THA subject (32.9°). Notable differences in APPA were observed between the supine position and neutral standing and between the supine position and neutral seated for all subjects. The greatest difference was found for the second healthy subject between the supine position and neutral sit activity (25.3°). PI changed little across activities for all subjects and the highest SS was assessed for the first healthy subject with an average of 36.1° ±10.9° with 44° being the highest value for the neutral standing position. The highest average value for PTS was found for the second healthy subject (31.7° ±12.6°). Significant correlation was found between APPA, SS and PTS for both dynamic activities and for all subjects. Finally, our results demonstrated that the position of the pelvis in the sagittal plane may vary substantially between functional positions and across different populations and the extent of change is specific to each individual.

6.2 Introduction

Acetabular component placement is directly associated with the incidences of hip dislocation and wear, the most common causes of total hip arthroplasty (THA) failure [146, 142]. For primary THA 0.5% to 10% dislocation rates have been documented [238] and 10% to 25% after revision surgery [239]. The alignment of the acetabular cup plays an important role in the mechanical stability of a THA and a CT scan has been the gold standard for evaluating the position of the acetabular component in relation to pelvic orientation to best restore native acetabular geometry while accounting for dislocation risk. [240]. The CT scan is done with the patient in the supine position, however, the sitting position is where most posterior dislocations occur. Lewinnek et al [142] reported a safety zone for
acetabular cup positioning and based on their assessments, an inclination of $40^\circ \pm 10^\circ$ and anteversion of $15^\circ \pm 10^\circ$ should enable the greatest possible range of motion of the hip with the minimum dislocation risk with respect the pelvic tilt motion. However, the validity of Lewinnek safe zones has been challenged by recent studies [151, 152, 153, 154].

Abdel et al [151] demonstrated that approximately 60% of 206 dislocated THA surgeries had a cup aligned within the so-called safe zone. Additionally, the evaluation of the safe zone relies on static measurements that do not take into account the pelvis dynamic changes. Acetabular orientation is a dynamic parameter that can be affected by forces contributing on the pelvis by changes in pelvic tilt, obliquity, and rotation [6, 155, 156, 157]. The true spatial orientation of the acetabulum requires an assessment of the anterior plane pelvic angle (APPA). Methods utilized range from adjustment based on calculation of APPA in a supine position by digitizing the floor plane and anterior pelvic plane (APP) to modification without estimation of the APP in the lateral position [3]. APP is regularly used as a superficial anatomical landmark during THA surgery, with most of the subjects being in the supine position. APPA changes are related with a change in the spatial orientation of the pelvis and thus influence the alignment of the acetabular component. Therefore, APPA differences between the supine and the upright position may be responsible for THA dislocations, if cup positioning is only relied on APPA values assessed in the supine position. Additionally, APPA may change postoperatively. The effect of APPA on acetabular component anteversion has been documented and assessed as approximately $0.7^\circ$ increase in anteversion for each degree of posterior APPA but the impact of APPA on acetabular component inclination may be equally important [155]. Thus, planning and evaluation of the intended position of the acetabular component only in the supine position may not account for clinically important changes in its orientation during functional activities, resulting from individual pelvic kinematics. Optimal orientation is subject-specific and needs an assessment of functional pelvic tilt pre-operatively. Factors to pay attention to during
the optimization of cup orientation involve femoral anteversion, hip kinematics and the orientation of the pelvis in functional positions [147]. Functional orientation of the pelvis after surgery is one of the most crucial variables for the surgeon to review when deciding the proper goal for acetabular component orientation.

Surgical navigation has been reported to provide better accuracy and precision of component positioning compared to traditional freehand techniques [148, 3]. Surgical navigation also gives the ability to the surgeon to make fine adjustments to component position based on individual’s factors like APPA [149]. However, even with navigation systems [158], the evaluation of the safe zone still relies on static measurements that do not take into account the pelvis dynamic orientation. Pelvic orientation between supine, sitting and standing is unique to each patient and can vary dramatically within each position.

Furthermore, certain pelvic positions in patients with spinal disorders can affect acetabular alignment after THA, increasing the risk of dislocation in certain patient populations [6]. Recent studies show that the highly variable biomechanical relationship between the lumbar spine and hip joint determines the functional orientation of the cup [159]. Lazennec et al [6, 109, 106, 117] demonstrated that the pelvis tilts during postural changes, and since the acetabulum is part of the pelvis, pelvic motion will change the anteversion and inclination of the acetabulum, thus, it is not maintained in the position achieved at the time of surgery. This alteration of the acetabular angles is the main cause that Lazennec et al [6] reported the sagittal ‘functional’ cup placement in THA versus to the coronal inclination and anteversion accomplished at surgery. Recent investigations evaluating the connection between the spine and the pelvis have improved our understanding of pelvic and spinal orientation. Spino-pelvic radiographic parameters have been utilized to assess balance in patients with a spinal deformity or spondylolisthesis [162]. It has been revealed that compensatory spine and pelvic dynamics are important to keep optimal balance and a functional range of movement (ROM) in both the healthy and implanted hip. Thus, loss of
compensation at the spino-pelvic junction may decrease the effectiveness and increase the risk of complications following THA. The importance of the pelvic morphology in sagittal balance is apparent for spinal surgeons, who take into account spinopelvic parameters, such as the morphologic pelvic incidence (PI) angle and functional parameters such as APPA, sacral slope (SS) angle, and pelvic tilt sacral (PTS) angle, [6, 162, 147].

The relationship between the lumbosacral spine and the pelvis is dynamically related to positional change, and may become more complex by co-existing pathology. The pelvic shape in the sitting position is different from that in the standing position consequently, analysis of lumbar alignment in the sitting position must include consideration of pelvic alignment. Specifically, both the sitting position and the standing position should be evenly considered in defining the ideal position for casting treatment or operative spinal fusion. Studies have revealed that spinal fusion may alter the adaptation of the spinopelvic junction [116], therefore, loss of compensation at the spino-pelvic junction may decline the effectiveness and enlarge the risk of complications after THA surgery [109]. While the PI is a static measurement unaffected by a pelvic change in position, the most reproducible measurement used to assess dynamic motion of the pelvis is sacral slope [165]. The sacral slope is normally 40° while standing and decreases to 20° when sitting, representing 20° of pelvic motion between standing and sitting.

Although previous studies have reported the aforementioned parameters during static conditions, further investigation needs to be conducted under activities of daily living to identify three-dimensional dynamic pelvic movement that may differ from what is currently captured in clinical x-rays and may be valuable in making surgical decisions. To the best of our knowledge no others have investigated and compared the 3D pelvic functional orientation across different populations that include healthy subjects, subjects that have undergone THA and spinal-stabilized cohorts and during different static and dynamic activities. Most studies have focused on comparing healthy cohorts with preoperatively or
postoperatively THA subjects or with patients with lumbosacral fusions but not compare all different groups of patients concurrently. Furthermore, most studies have performed their measurements in static settings whereas the pelvic motion is dynamic.

For the purpose of this study we have assessed the global 3D aspects of pelvic motion using a high speed stereo radiography system performing a 2D/3D imaging technique. Three dimensional evaluation of pelvic orientation is important because any coronal or axial rotation of the pelvis in functional positions would also affect acetabular orientation. This will help us to clearly estimate the consequences of the pelvic motion for more proper acetabular cup positioning. Additionally, the position of the pelvis in the sagittal plane may differ notably between functional positions and taking into account the interindividual variation is crucial. Based on these changes, the angles of orientation of the acetabular component during dislocation and edge-loading will not be the same from those estimated from standard CT and radiographs.

Thus, the objectives of this work is to use high-speed stereo radiography (HSSR) to 1) estimate PI and spinopelvic functional parameters (PTS, SS, APPA) during dynamic activities and 2) investigate the range of these parameters with activity and 3) to form a more reliable evaluation of patient-specific pelvic orientation to inform acetabular cup placement. We hypothesized that substantial differences will be observed across different subject populations during the same activity and in the same subject during different activities.
6.3 Methods

6.3.1 Participants

Four subjects (4M, 68.6±5 years, BMI: 26.1±2.9 kg/m2) participated in the study. This study was approved by the University of Denver Institutional Review Board and all participants provided informed consent. Data were obtained from two healthy subjects that had no history of injuries or surgeries to the lower limbs and could perform activities of daily living without pain or discomfort, from one total hip replacement patient with greater than six months post-operation and with no other lower extremity surgeries and one subject with lumbar spine fusion. Further exclusion criteria included no primitive or metastatic tumors of hip joint, no injury to ligaments or muscles in the joint being imaged.

6.3.2 Procedures

The participants performed seven activities (Figure 6.1a), two dynamic ones: 1) Normal treadmill walking at a self selected pace (gait); 2) and standing up from a chair (chair rise); and five static activities that were meant to capture the extents of pelvic motion. Three were during standing position: 3) Neutral standing; 4) back extended and one leg at 90° (Leg90); and 5) Slouching and two during sitting position: 6) Neutral seated and 7) seated and fully bent over at the waist (Bent over seated). Supine position was retrieved from CT data. One trial was collected for each activity.

High speed stereo radiography (HSSR) was used to capture the 3D position of the pelvis. HSSR captures two radiographic views to enable three-dimensional tracking of the bones in the knee [25]. The HSSR system is composed of two matching custom radiography systems with 40 cm diameter image intensifiers integrated with high-speed digital cameras [35]. Image distortion created by the image intensifiers was subtracted by imaging
a radio opaque mesh of known dimension, and then forming a transformation to correct dis-\n\textit{tortion from subsequent images of the subjects (XROMM Undistorter, Brown University,\nRI). The capture volume was calibrated from imaging a custom-calibration cube enclosing\n52 steel beads of precisely-known position and size [35, 46]. Images during all activities\nwere captured with pulsed radiography (pulse width 750 $\mu$s, 60 kV, and 63 mA) at 50\nframes/second. Following the laboratory data collection, a static computed tomography\n(CT) scan with 1.0 mm slice thickness, or MRI was obtained of each subject’s pelvis (GE\nOptima 660,120kV bone technique).

![Figure 6.1](image)

\textbf{Figure 6.1:} a) Activities performed by the participants. b) Local pelvis coordinate system and c)\n3D image registration to the 2D stereo images. d) Spinopelvic parameters of APPA, SS, PTS, PI e)\nAPPA and its functional orientation of the pelvis during different positions.
6.3.3 Data Processing

Three-dimensional models of the pelvis were reconstructed from the CT data using ScanIP (Simpleware Inc., Mountain View, USA). The position and orientation of the pelvis was found using Autoscopec (Brown University), which optimized the positions of the pelvis bone to the two-dimensional stereo radiography images (Figure 6.1c). HSSR was synchronized with a conventional motion capture system that was used to define the global coordinate system.

Motion of the pelvis was described with respect to a reference global ground coordinate system. The local pelvis coordinate system for each subject was established based on three bony landmarks on the pelvis: the bilateral anterior superior iliac spines (RASIS and LASIS), the midpoint of the bilateral pubic tubercles (pubic symphysis). The center was taken at the origin of the right hip joint center by fitting a sphere in the acetabular cup. The positive X axis points from the origin to the right ASIS. The Y axis lies in the plane defined by the left and right ASISs and the midpoint of the pubic symphysis and points anterior orthogonal to the X axis. The Z axis is perpendicular to both X and Y, positive cranially (superiorly in the erect standing position) (Figure 6.1b). Rotation is the angle of rotation of the pelvis about Z axis. Obliquity is the angle of rotation of the anterior-posterior axis of the pelvis out of the horizontal plane and tilt is the angle of rotation about the medio-lateral X axis of the pelvis.

Spinopelvic Parameters

The anterior pelvic plane angle (APPA) was defined as the angle between the anterior pelvic plane and a vertical reference line (Figure 6.1d). A positive value indicates pelvic anteversion (anterior tilt) (Figure 6.1e). The Pelvic incidence (PI) was measured as the angle formed by the perpendicular line to the tangent line to the centre of the sacral plateau
(S1) and the line connecting this centre to the centre of the axis of the femoral heads (the midpoint of bicoxofemoral axis) (Figure 6.1d). The sacral slope (SS) was defined as the angle between a line tangent to the upper plate of S1 and a horizontal reference line (Figure 6.1d) while the PTS was estimated between the line connecting the midpoint of the upper plate of S1 to the midpoint of bicoxofemoral axis and a vertical reference line (Figure 6.1d). A simple geometrical construction based on the properties of right-angled triangles demonstrates that the PI is the sum of the PTS and the SS. The latter two parameters are positional angles and vary inversely to one another: PI = PTS + SS. Thus, since the value of the PI is fixed for a given patient, the summation of the PTS and the SS is invariable, and as one increases, the other necessarily decreases. Finally, when the pelvis rotates backward (retroversion), PTS (pelvic tilt sacral) increases while when the pelvis rotates forward (anteversion), PTS decreases.

6.3.4 Data Analysis

Comparisons were performed for each subject across different activities and across subjects for each activity. Differences were calculated between static and selected points from dynamic activities that correspond to a similar position of the pelvis between the activities. Range of motion (ROM) was calculated as the difference between the maximum and the minimum for APPA for each subject for all activities to assess the range of APPA for each individual. The Pearson r correlation was used to assess the relationship between the spinopelvic parameters in both dynamic activities, ranging between +1.0 and -1.0. Finally, the coefficient of determination was calculated (R²).
6.4 Results

Anterior Pelvic Plane Angle

The highest pelvic arc of motion was observed for one of the healthy subjects (38.7°) while the lowest ROM was observed for the second healthy subject (34.9°) (Figure 6.2b,c). Intersubject variability across the same or different populations was observed during standing and seated positions (Figures 6.3,6.4 respectively). Progression to a more posterior tilt of the pelvis was demonstrated from standing to seated position and the highest APPA change was found for the THA subject (23.1°) (Figure 6.5). A more anterior position of the pelvis was observed as a whole for the first healthy subject where the highest anterior tilt was shown for the bent over seated activity (30°). The second healthy subject demonstrated a more posterior position of the pelvis during the different tasks with the highest posterior value being observed for the neutral seated activity (−27.2°). Regarding the spine subject, the pelvic motion was not fixed in either anterior or posterior direction among the activities, resulting in 37.1° ROM. The highest APPA value was observed during the bent over seated activity (24.5°). Finally, during sitting positions an increased posterior tilt of the pelvis was demonstrated for the THA subject which decreased when bending forward, achieving the most anterior tilt of the pelvis (Figure 6.7a,d, Table 6.1) (3.5°). Table 6.1 summarizes all the APPA values for all subjects and activities.
Figure 6.2: a) Comparison of the three subjects during the gait activity b) First healthy subject during gait and static standing activities and c) Second healthy subject during gait and static standing activities d) THA subject during gait and static standing activities
Figure 6.3: Comparison of the four subjects during the neutral standing activity. Our data were compared with literature [3, 6, 7]

Figure 6.4: Comparison of the four subjects during the neutral seated activity. Our data were compared with literature [3, 6, 7]
Figure 6.5: APPA change from neutral standing to neutral seated position (on the left) and illustration of APPA change for the THA subject (on the right)

During gait a more anterior position of the pelvis was shown throughout the entire activity for the THA subject when compared to the two healthy subjects (Figure 6.2a). The first healthy subject remained in a more neutral pelvic position for the whole trial while subject two demonstrated a posterior tilt of the pelvis during the entire activity (Figure 6.2). During the chair rise activity, for all three populations, a similar trend was observed, where the pelvis was in a more posterior position in the beginning of the trial which slowly increased to a more anterior position of the pelvis towards the end of the activity as the subject approached the standing position (Figure 6.7a).

Notable differences in APPA were observed between the supine position and neutral standing and between the supine position and neutral seated for all subjects (Figure 6.6). The greatest difference was found for the second healthy subject between the supine position and neutral sitting activity (25.3°). For the same subject a difference of 12.5° was observed between supine position and neutral standing, a difference of 6° and 0.9° with the most anterior point and the most posterior point of the gait activity with neutral standing respectively. The difference between neutral sit and chair rise was 15.6° and 7.6° when compared to bent over activity. For the first healthy subject the greatest difference was assessed between the supine position and the neutral sit (20.8°) task while 8.5° was the dif-
ference between neutral stand and supine position. When the most anterior point of the gait activity was compared with the neutral stand a difference of 2.8° was found and 5.3° with the most posterior point of the same task. For the spinal fusion subject a 12.6° and a 10.6° difference was estimated between the supine position and neutral stand and between supine and neutral sit respectively. When compared chair rise activity with the static neutral sitting position a difference of 4° was found and a difference of 12° when the chair rise activity was compared with the bent over trial. Finally, notable differences in APPA were observed for the THA subject between the supine position and neutral standing of 7.9°, and between the supine position and neutral seated of 15.2°. Differences were observed for the chair rise activity when compared with the neutral seat (5.5°) and bend forward during sitting (5.3°). Additionally, differences of 4.1° were shown for the APPA between the neutral stand and the most posterior point of gait activity and 9.8° with the most anterior point.

Figure 6.6: APPA change from supine position to all static positions for all subjects. The lowest and highest APPA change is demonstrated on the plots for each subject.
Figure 6.7: a) Comparison of the three subjects during the chair rise activity b) First healthy subject during chair rise and static sitting activities and c) Second healthy subject during chair rise and static sitting d) THA subject during chair rise and static sitting activities
Table 6.1: APPA values for all subjects during all static and dynamic activities.

<table>
<thead>
<tr>
<th>APPA (°)</th>
<th>Healthy 1</th>
<th>Healthy 2</th>
<th>Spine</th>
<th>THA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Supine</td>
<td>12.1</td>
<td>-1.9</td>
<td>0.7</td>
<td>11.5</td>
</tr>
<tr>
<td>Neutral Stand</td>
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<td>-14.4</td>
<td>-11.9</td>
<td>-3.6</td>
</tr>
<tr>
<td>Leg90 Stand</td>
<td>6.3</td>
<td>-7.9</td>
<td>3.7</td>
<td>2.6</td>
</tr>
<tr>
<td>Slouch Stand</td>
<td>14.4</td>
<td>-25.9</td>
<td>-12.6</td>
<td>-16.6</td>
</tr>
<tr>
<td>Neutral Seated</td>
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<td>-27.2</td>
<td>-9.9</td>
<td>-26.7</td>
</tr>
<tr>
<td>Bent Over-Seated</td>
<td>30</td>
<td>7.7</td>
<td>24.5</td>
<td>3.5</td>
</tr>
<tr>
<td>Gait Range</td>
<td>-1.7 to 0.8</td>
<td>-8.4 to -13.5</td>
<td>–</td>
<td>0.5 to 6.2</td>
</tr>
<tr>
<td>Chair Rise Range</td>
<td>–</td>
<td>0.1 to -11.6</td>
<td>-5.9 to 12.5</td>
<td>-1.8 to -21.2</td>
</tr>
<tr>
<td><strong>ROM</strong></td>
<td><strong>38.7</strong></td>
<td><strong>34.9</strong></td>
<td><strong>37.1</strong></td>
<td><strong>38.2</strong></td>
</tr>
</tbody>
</table>

**Pelvic Incidence**

PI changed little across activities for all subjects. The greatest value for the PI was found for one of the healthy subjects with an average value of 60.6° ± 0.9° while the minimum PI was found for the THA subject with an average of 46.4° ± 0.8°. Table 6.2 summarizes all the PI values for all subjects and all activities. In Figure 6.8 PI is illustrated during gait and chair rise activities for all subjects.
Figure 6.8: a) Comparison of the three subjects during the chair rise activity b) First healthy subject during chair rise and static sitting activities and c) Second healthy subject during chair rise and static sitting d) THA subject during chair rise and static sitting activities

Table 6.2: PI values for all subjects during all static and dynamic activities.

<table>
<thead>
<tr>
<th>PI (°)</th>
<th>Healthy 1</th>
<th>Healthy 2</th>
<th>Spine</th>
<th>THA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Supine</td>
<td>49.0</td>
<td>60.2</td>
<td>43.8</td>
<td>47.2</td>
</tr>
<tr>
<td>Neutral Stand</td>
<td>48.4</td>
<td>62.5</td>
<td>41.3</td>
<td>45.6</td>
</tr>
<tr>
<td>Leg90 Stand</td>
<td>46.1</td>
<td>60.0</td>
<td>43.5</td>
<td>45.7</td>
</tr>
<tr>
<td>Slouch Stand</td>
<td>48.4</td>
<td>61.1</td>
<td>43.0</td>
<td>46.6</td>
</tr>
<tr>
<td>Neutral Seated</td>
<td>48.0</td>
<td>59.3</td>
<td>41.0</td>
<td>48.0</td>
</tr>
<tr>
<td>Bent Over-Seated</td>
<td>50.0</td>
<td>61.2</td>
<td>42.0</td>
<td>46.0</td>
</tr>
<tr>
<td>Gait Range</td>
<td>48.3</td>
<td>60.5</td>
<td>–</td>
<td>45.8</td>
</tr>
<tr>
<td>Chair Rise Range</td>
<td>–</td>
<td>60.2</td>
<td>40.4</td>
<td>46.0</td>
</tr>
</tbody>
</table>

Range: 46.1 to 50, 59.3 to 62.5, 41 to 43.8, 45.6 to 48
Sacral Slope

The highest SS was assessed for the first healthy subject with an average of 36.1° ±10.9° with 44° being the highest value for the neutral stand position (Figure 6.9a). The smallest SS was observed for the subject with the spinal fusion and the average was 22.8° ±5.9° (Figure 6.11a). A similar value was estimated for the second healthy subject and the THA subject with averages of 28.9° ±12.4° and 28.2° ±12.9° respectively (Figures 6.10a, 6.12a). The most stiff pelvis was found for the subject with the spinal fusion, where the difference between neutral stand and neutral sit activities was found to be 4.1°. For the rest of the subjects the difference between those activities was the highest for the first healthy subject, 30.5°; and 11.8° and 23.6° for the other healthy and THA subject, respectively (Figure 6.13). Table 6.3 summarizes all the SS values for all the subjects and all activities. Figure 6.8 shows a similar trend for the SS during the gait activity, demonstrating small changes throughout the entire trial. That was not the case for the chair rise activity (Figure 6.8), where the THA subject demonstrated a small SS in the beginning of the trial which was slowly increasing toward the end. A similar behaviour was observed for the subject with the spinal fusion. However, the healthy subject demonstrated less change for the SS throughout the activity (Table 6.3).

![Figure 6.9: a) Comparison of the SS values for the first healthy subject for all activities b) Comparison of the PTS values for the first healthy subject for all activities](image)

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Figure 6.10: a) Comparison of the SS values for the second healthy subject for all activities b) Comparison of the PTS values for the second healthy subject for all activities

Figure 6.11: a) Comparison of the SS values for the subject with the spinal fusion for all activities b) Comparison of the PTS values for the subject with the spinal fusion for all activities
Figure 6.12: a) Comparison of the SS values for the THA subject for all activities b) Comparison of the PTS values for the THA subject for all activities

Figure 6.13: SS values for neutral standing, neutral sitting and the SS change from standing to seated position
Table 6.3: SS values for all subjects during all static and dynamic activities.

<table>
<thead>
<tr>
<th>SS (°)</th>
<th>Healthy 1</th>
<th>Healthy 2</th>
<th>Spine</th>
<th>THA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Supine</td>
<td>44.0</td>
<td>39.1</td>
<td>31.7</td>
<td>43.0</td>
</tr>
<tr>
<td>Neutral Stand</td>
<td>44.0</td>
<td>25.8</td>
<td>21.0</td>
<td>29.6</td>
</tr>
<tr>
<td>Leg90 Stand</td>
<td>40.7</td>
<td>31.9</td>
<td>30.1</td>
<td>37.5</td>
</tr>
<tr>
<td>Slouch Stand</td>
<td>43.0</td>
<td>14.5</td>
<td>17.3</td>
<td>16.7</td>
</tr>
<tr>
<td>Neutral Seated</td>
<td>13.5</td>
<td>14.0</td>
<td>16.9</td>
<td>6.0</td>
</tr>
<tr>
<td>Bent Over-Seated</td>
<td>31.5</td>
<td>48.0</td>
<td>19.5</td>
<td>36.3</td>
</tr>
<tr>
<td>Gait Range</td>
<td>39.8 to 41.1</td>
<td>27.5 to 33.1</td>
<td>–</td>
<td>34.9 to 41.3</td>
</tr>
<tr>
<td>Chair Rise Range</td>
<td>–</td>
<td>29.1 to 41.2</td>
<td>21.2 to 39.4</td>
<td>11.5 to 30.9</td>
</tr>
<tr>
<td>Range</td>
<td>13.5 to 44</td>
<td>14 to 48</td>
<td>16.9 to 39.4</td>
<td>6 to 43</td>
</tr>
</tbody>
</table>

**Pelvic Tilt Sacral**

The highest average value for PTS was found for the second healthy subject (31.7° ±12.6°). The average values for the first healthy subject, the spinal fusion subject and the THA were 13.4° ±10.8°, 20.8° ±5.7° and 18.7° ±13.2° respectively (Figures 6.11b, 6.12b). The smallest difference between neutral stand and neutral sit activity was 0.3° for the spine subject which was much smaller compare to the rest subjects (28.6°, 10°, 26° for the two healthy and THA subjects respectively). Table 6.4 summarizes all the PTS values for all the subjects and all activities. PTS demonstrated the expected behaviour during the gait and chair rise activities following the expected increase or decrease resulting in the summation of PI, along with the SS (Figure 6.8).
Table 6.4: PTS values for all subjects during all static and dynamic activities.

<table>
<thead>
<tr>
<th>PTS (°)</th>
<th>Healthy 1</th>
<th>Healthy 2</th>
<th>Spine</th>
<th>THA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Supine</td>
<td>4</td>
<td>21.1</td>
<td>12.4</td>
<td>3.5</td>
</tr>
<tr>
<td>Neutral Stand</td>
<td>6.4</td>
<td>36.5</td>
<td>24.5</td>
<td>16</td>
</tr>
<tr>
<td>Leg90 Stand</td>
<td>7.6</td>
<td>27.3</td>
<td>13.4</td>
<td>11.7</td>
</tr>
<tr>
<td>Slouch Stand</td>
<td>8.2</td>
<td>46.7</td>
<td>27.5</td>
<td>29.9</td>
</tr>
<tr>
<td>Neutral Seated</td>
<td>35</td>
<td>46.5</td>
<td>24.2</td>
<td>42</td>
</tr>
<tr>
<td>Bent Over-Seated</td>
<td>19.5</td>
<td>12.6</td>
<td>22.9</td>
<td>9.6</td>
</tr>
<tr>
<td>Gait Range</td>
<td>7.3 to 9.3</td>
<td>28.2 to 33.2</td>
<td>–</td>
<td>6.3 to 11.9</td>
</tr>
<tr>
<td>Chair Rise Range</td>
<td>–</td>
<td>19.5 to 30.1</td>
<td>4.9 to 19.9</td>
<td>15.8 to 35.2</td>
</tr>
<tr>
<td><strong>Range</strong></td>
<td><strong>4 to 19.5</strong></td>
<td><strong>12.6 to 46.7</strong></td>
<td><strong>4.9 to 27.5</strong></td>
<td><strong>3.5 to 35.2</strong></td>
</tr>
</tbody>
</table>

**Relationships**

In gait activity, the relationship analysis showed a significant correlation between APPA, SS and PTS for all subjects (Figure 6.14). Similar results were found for the chair rise activity (Figure 6.15). The relationship analysis results between all three parameters are reported in Tables 6.5 to 6.6 and 6.7 to 6.9 for gait and chair rise respectively for all subjects.
Figure 6.14: a) Correlation between APPA vs SS, APPA vs PTS and SS vs PTS for the gait activity
Figure 6.15: a) Correlation between APPA vs SS, APPA vs PTS and SS vs PTS for the chair rise activity

Table 6.5: Correlation among the spinopelvic parameters for the two healthy subjects during gait activity.

<table>
<thead>
<tr>
<th>Spinopelvic Parameters</th>
<th>APPA</th>
<th>SS</th>
</tr>
</thead>
<tbody>
<tr>
<td>SS</td>
<td>r = 0.98</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>P&lt;.0001</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>R² = 0.97</td>
<td>-</td>
</tr>
<tr>
<td>PTS</td>
<td>r = -0.99</td>
<td>r = -0.98</td>
</tr>
<tr>
<td></td>
<td>P&lt;.0001</td>
<td>P&lt;.0001</td>
</tr>
<tr>
<td></td>
<td>R² = 0.98</td>
<td>R² = 0.96</td>
</tr>
</tbody>
</table>
Table 6.6: Correlation among the spinopelvic parameters for the THA subject during gait activity.

<table>
<thead>
<tr>
<th>Spinopelvic Parameters</th>
<th>APPA</th>
<th>SS</th>
</tr>
</thead>
<tbody>
<tr>
<td>SS</td>
<td>$r = 0.80$</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>$P &lt; .0001$</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>$R^2 = 0.64$</td>
<td>-</td>
</tr>
<tr>
<td>PTS</td>
<td>$r = -0.99$</td>
<td>$r = -0.8$</td>
</tr>
<tr>
<td></td>
<td>$P &lt; .0001$</td>
<td>$P &lt; .0001$</td>
</tr>
<tr>
<td></td>
<td>$R^2 = 0.99$</td>
<td>$R^2 = 0.64$</td>
</tr>
</tbody>
</table>

Table 6.7: Correlation among the spinopelvic parameters for the healthy subject during chair rise activity.

<table>
<thead>
<tr>
<th>Spinopelvic Parameters</th>
<th>APPA</th>
<th>SS</th>
</tr>
</thead>
<tbody>
<tr>
<td>SS</td>
<td>$r = 1$</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>$P &lt; .0001$</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>$R^2 = 0.99$</td>
<td>-</td>
</tr>
<tr>
<td>PTS</td>
<td>$r = -0.98$</td>
<td>$r = -0.98$</td>
</tr>
<tr>
<td></td>
<td>$P &lt; .0001$</td>
<td>$P &lt; .0001$</td>
</tr>
<tr>
<td></td>
<td>$R^2 = 0.97$</td>
<td>$R^2 = 0.96$</td>
</tr>
</tbody>
</table>
Table 6.8: Correlation among the spinopelvic parameters for the THA subject during chair rise activity.

<table>
<thead>
<tr>
<th>Spinopelvic Parameters</th>
<th>APPA</th>
<th>SS</th>
</tr>
</thead>
<tbody>
<tr>
<td>SS</td>
<td>$r = 1$</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>$P &lt; .0001$</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>$R^2 = 0.99$</td>
<td>-</td>
</tr>
<tr>
<td>PTS</td>
<td>$r = -1$</td>
<td>$r = -1$</td>
</tr>
<tr>
<td></td>
<td>$P &lt; .0001$</td>
<td>$P &lt; .0001$</td>
</tr>
<tr>
<td></td>
<td>$R^2 = 1$</td>
<td>$R^2 = 0.99$</td>
</tr>
</tbody>
</table>

Table 6.9: Correlation among the spinopelvic parameters for the subject with spinal fusion during chair rise activity.

<table>
<thead>
<tr>
<th>Spinopelvic Parameters</th>
<th>APPA</th>
<th>SS</th>
</tr>
</thead>
<tbody>
<tr>
<td>SS</td>
<td>$r = 1$</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>$P &lt; .0001$</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>$R^2 = 0.99$</td>
<td>-</td>
</tr>
<tr>
<td>PTS</td>
<td>$r = -1$</td>
<td>$r = -0.99$</td>
</tr>
<tr>
<td></td>
<td>$P &lt; .0001$</td>
<td>$P &lt; .0001$</td>
</tr>
<tr>
<td></td>
<td>$R^2 = 0.99$</td>
<td>$R^2 = 0.98$</td>
</tr>
</tbody>
</table>

6.5 Discussion

For proper alignment of the acetabular cup, knowledge of the pelvic orientation is essential. The objectives of this study were to establish the functional orientation of the pelvis by using HSSR for higher accuracy and estimate the differences in PI and spinopelvic parameters (APPA, SS, PTS) across different patient populations and across static and dynamic
activities of daily living. This knowledge is necessary to enable acetabular component placement that reduces the likelihood of impingement events. To our knowledge, no other study has investigated and compared the 3D pelvic functional orientation across different populations that include healthy subjects, subjects that have undergone THA and spinal-stabilized cohorts and during different static and dynamic activities such as gait and stair rise. We hypothesized that substantial differences will be observed across different subject populations during the same activity and in the same subject during during different activities. In support of our hypotheses, we found substantial individual variation between functional positions of the pelvis across the different populations. Notable differences were observed when comparing APPA between different static and dynamic conditions for the same subject as well as when compared to the rest of the subjects. The APPA was not neutral for all the subjects and varied substantially among the different activities. Spinopelvic metrics that demonstrated different degrees of PI among the subjects affects the pelvic motion. A more stiff pelvis was found for the spinal fusion subject with respect the SS value when compared to the rest of the subjects.

APPA in all of the functional positions which we studied demonstrated variations from the supine position. The inter-subject variation in all positions was considerable, highlighting the significance of individual pre-operative evaluation. The changes in position were substantial with the greatest change being a posterior tilt of 25.3° from supine to neutral seated for the second healthy subject.

The greatest absolute change in APPA moving from supine to standing was 12.6° for the spinal fusion subject. Large changes were shown between the supine and bent over seated positions, with a mean absolute rotation of 16.6° for all subjects, and a maximum posterior tilt of −23.8° for the subject with the spinal fusion. All subjects moved posteriorly from supine to neutral stand positions that could lead to an excessive functional antervation in extension. Conversely, the pelvis was tilted anteriorly greater than 16° for all of the sub-
jects from supine to bent over seated positions that leads to a less acetabular component antversion in flexion which put the subjects at risk of posterior instability. Furthermore, our results demonstrate that the APPA when the subject is in supine position is not neutral for all the subjects including one healthy with anterior tilt of the pelvis and one THA subject with posterior tilt of the pelvis indicating high intersubject variability. This knowledge could be used to further improve the optimal cup alignment on an individual patient basis. As the pelvis rotates posteriorly, the functional anteversion of the acetabular cup will increase. This rotation is helpful against dislocation and edge-loading in flexion of the hip, but could lead to wear in extension and anterior instability. Therefore, estimating the orientation of the acetabular cup seems more helpful during functional flexion and extension rather than when the subject is in supine position, since with malorientation of the acetabular cup are more likely to happen during those instances.

Previous studies have reported a mean seated APPA in the range of $-25^\circ$ to $-36^\circ$, where all patients were seated in a relaxed posture, upright in a chair [161, 160]. Similar values were observed for two of our subjects during neutral sit position while for the other two subjects much less posterior tilt was observed.

The posterior APPA from standing to sitting is normally $20^\circ$ [159, 6]. The two healthy subjects moved posteriorly between the two static activities with an average of $12.6^\circ$ while the minimum posterior tilt was reported for the spinal fusion subject. The highest result was shown for the THA subject that demonstrated a posterior tilt of $23.1^\circ$ from standing to sitting. Finally, the APPA ROM, was quite large, spanning greater than $30^\circ$ and reflecting a considerable variation between patients. The ROM of both dynamic activities was within the range that was observed from the static activities indicating that static activities can sufficiently provide the pelvic arc of motion.

Prior work on pelvic positioning by DiGioia et al [160], Nishihara et al [161], and Parratte et al [241] all showed high intersubject variability, whether supine or standing.
Similarly, our results also demonstrated a high intersubject variability of APPA with a range of \(-27.2^\circ\) to \(30^\circ\) for all subjects during all activities. These results further enhance the suggestions reported from prior studies in literature [149, 242] that information about patient-specific preoperative pelvic tilt should be a factor to consider when defining the ideal acetabular cup alignment for each subject.

Pelvic tilt differences emphasize the essential individual variation between various functional positions of the pelvis. The greatest difference between the instant of the most anterior point of the gait activity and the neutral stand was \(6^\circ\) for the second healthy subject and \(9.8^\circ\) with the instant of the most posterior point of the gait activity for the THA subject. Additionally, \(15.6^\circ\) was the highest difference from chair rise to neutral sit activity while \(12^\circ\) was the highest difference from chair rise to bent over activity for the spinal fusion subject. Additionally, one subject might have a \(-14.4^\circ\) posterior pelvic tilt during neutral standing and another could have a \(3.6^\circ\) anterior pelvic tilt in exactly the same neutral standing position.

PI is fixed a morphological measurement of the anterior to posterior dimension of the pelvis that defines the position of the femoral heads relative to the spine. PI regulates sagittal lumbar lordosis [243] and determines the relative position of the sacral plate in relation to the femoral heads. The greater the PI, the greater the SS and the higher the degree of lumbar lordosis [243]. Prior work has shown the lower value of PI in the population is approximately \(35^\circ\), the higher around \(85^\circ\) and the average being \(51.9^\circ\) [5, 244]. Our results demonstrated the highest PI value for one of our healthy subjects (60.6\(^\circ\), Table 6.2) and the lowest for our subject with the spinal fusion (42.1\(^\circ\)). Those two subjects demonstrated the highest and the lowest SS values throughout the activities (Figures 6.10a, 6.11a ). Generally, PI remained unchanged throughout the activities for each subject as expected and our values are similar to those that have been reported from previous studies [5, 6, 159].
Our results are in agreement with literature since our subject with the highest PI demonstrated the highest posterior tilt during standing (−14.4°, Tables 6.1, 6.2). Oppositely, patients with small PI have less capacity for compensation. Based on the literature, during the sitting position the pelvis tilts backwards and SS declines, to a mean of 20° to 25° and sometimes it becomes very low (5° to 10°) or even negative [6]. Our lowest SS value during neutral sitting position was found for the THA subject (6°) and the highest for the spinal fusion subject (16.9°) (Table 6.3). The distinction in sacral slope between the standing and sitting positions is related to the available flexion of the lumbar-sacral junction, as distinct from possible hip-joint flexion [6]. In the supine position when the lower limbs are in extension, SS was found higher than the standing position [159, 6] and this additional pelvic tilt may be badly tolerated if the spine is stiff or deformed.

That was the case for our subject with the spinal fusion since the most stiff pelvis was demonstrated for that subject where much less than 10° of motion was observed (4°) at the spine-pelvis junction between the neutral standing and neutral seated positions (Figures 6.8a, 6.11a). When the spine stiffens the pelvis will compensate by tilting posteriorly in an attempt to maintain sagittal balance, a trend that was observed for our subject with spinal fusion for some of the activities.

Our results for PTS are in agreement with what exists in literature [159, 6]. We also observed from our results that as the pelvis tilts posteriorly the PTS increases and inversely as the pelvis tilts anteriorly (Figures 6.14, 6.15, Tables 6.5 to 6.9).

During the dynamic activity of gait small changes were observed for both SS and PTS angles throughout the activity for all subjects, however, intersubject variability was observed between the two healthy subjects (Figure 6.8) but that was not the case for the THA subject that followed a similar trend with one of the healthy subjects (Figure 6.8).

During the chair rise activity SS was found minimum in the beginning of the trial (seated position) for all subjects (Figure 6.8) which was slowly increased towards the end
of the trial as expected from what has been reported in previous studies [159, 6] for static sitting and standing positions.

Finally, a strong correlation was observed between the values of the APPA changes during the gait activity and chair rise activity (Figures 6.14, 6.15, Tables 6.5 to 6.9) with the values of SS and PTS for all subjects indicating that the pelvic arc movement affects similarly the change of all spinopelvic parameters during the dynamic activities. Those results are not in agreement with what has been found for the static activities of standing and sitting where no correlation was found between APPA and the values of SS and PTS [245, 164] indicating the need for dynamic measurements. Additionally, the substantial differences observed between the static and the dynamic activities during both standing and sitting emphasizes the need to better understand the dynamic positioning of the pelvis during clinical examination of spinopelvic mobility. However, analyzing spinopelvic parameters in the standing and sitting positions, including the stressful poses of bent over during sitting or the Leg90 activity during standing, provides an overall estimate of the pelvic arc of motion during activities of daily living.

Limitations

This study has limitations. First, the number of subjects was small, however it is an ongoing study and 26 more subjects will be added for the purpose of this analysis. Secondly, we did not perform a reliability and repeatability study for the technique of measurement of all the spinopelvic parameters for all subjects, however consistency on the results was observed when repetition of the spinopelvic parameters was performed for one of our subjects. Finally, no correlation of the functional parameters (APPA, SS, PTS) with the morphological parameter of PI was estimated due to the limited number of subjects.
Conclusions

The objective of this work was to use high-speed stereo radiography (HSSR) to estimate PI and spinopelvic functional parameters (PTS, SS, APPA) during static and dynamic activities and investigate the range of these parameters with activity to form a more reliable evaluation of patient-specific pelvic orientation to inform acetabular cup placement. The data collected in this study provide insight to the hip surgeons for intraoperative decisions that aid to further improvement of the acetabular component orientation.

Specifically, our results demonstrated that the position of the pelvis in the sagittal plane may vary substantially between functional positions and across different populations and the extent of change is specific to each individual. As a result of these changes, the angles of orientation of the acetabular component during dislocation will be different from those obtained from standard CT and radiographs. Differences were also observed between the static and the dynamic activities during both standing and sitting indicating the need to better understand the dynamic positioning of the pelvis during clinical examination of spinopelvic mobility. APPA is not neutral during supine or standing positions and differences were observed between the supine and neutral stand and seated position that are important to consider during surgical approaches that place the cup with the patient in the supine position. Preoperative assessment of the 3D functional pelvic orientation on a patient-specific basis is recommended during different static and dynamic activities of daily living of the cup to prevent prosthetic dislocation or impingement. Finally, improved understanding of spinopelvic parameters may eventually allow for more comprehensive patient-specific cup placement, potentially minimizing the risk of dislocation and wear-associated loosening.

In our opinion investigation of the 3D functional orientation of the pelvis is essential since any coronal or axial rotation of the pelvis in functional positions would also affect ac-
etabular orientation and those rotations cannot be seen with standard imaging. We also observed from our results that APPA differences between dynamic and static activities change substantially indicating the need for future studies to consider a dynamic analysis for the component position. Additionally, activities that are commonly related with posterior dislocation, such as rising from a chair or bent over for shoe tying where the hip is flexed further, seems more relevant to investigate the functional orientation of the pelvis rather than when the patient is in the supine position. Bending forward during sitting brings the pelvis into a more anteriorly tilted position where the acetabular component can become less anteverted and risk posterior dislocation. Finally, we suggest that hip surgeons take into account both APPA and SS for each patient so that the planning would be based on a precise anatomic and functional alignment of the cup.
Chapter 7

Conclusions and Recommendations

The objective of this dissertation was to accurately measure joint kinematics during activities of daily living for informing diagnosis, helping design of treatments and improving the design of prosthetic devices. Accurate measurement of healthy joint motion is crucial for establishing baseline kinematics and clinical parameters for evaluation of natural joint function that could be used as a reference for treatment evaluation. The goal was to obtain healthy data from the hip and knee joint during different activities of daily living, by recruiting people of the appropriate age since the motion of younger subjects may not be representative of the age range associated with joint pain, disease or treatment, to use those data as a reference for comparing the effectiveness of different prosthetic devices and by assessing data that support computer models.

Chapter 3 provides comprehensive patient specific measurements of patellofemoral motion and patellar tendon mechanics during gait activities in healthy older adults since the patellofemoral joint and quadriceps mechanism are susceptible to many pathologies resulting from acute injury (e.g. dislocation) and chronic disease (e.g. osteoarthritis). The objective of this study was to describe case series measurements of patellar motion in healthy older adults as they performed three gait activities and a non-weightbearing knee extension,
determine patellar tendon angle and moment arm, and show if these quantities were activity dependent. To accomplish this, dual-plane radiography and CT imaging were employed, which enables accurate measurement of bone motion and thus the ability to identify subtle differences between individuals. To our knowledge, no other study has investigated native patella kinematics during activities reported to be altered with age, namely descending a step and executing a turn during walking. These data can help clinicians and scientists that seek to understand PF pathology, to facilitate future improvements in the patella design and to provide better knowledge when prescribing rehabilitation intervention. Our results demonstrated, with some notable exceptions, that each subject displayed distinct PF kinematics that followed consistent trends through the three gait activities. The variation between subjects, the difference between weightbearing and non-weightbearing activities and the different behavior of the patella during the step down and pivot emphasized the need for subject-specific analysis with a range of activities. Future work should also include investigation of differences between males and females since notable differences exist in PF alignment and incidence of TF and PF disorders in males and females. Additionally, providing static alignment (Q angle) would help to better inform the clinicians. However, adding examination of differences between males and females would require a substantial increase in the number of subjects. More specifically, based on power analysis (G*Power, HHU, Duseldor), finding differences between males and females in one of the primary variables across the three different activities suggests a total of 64 participants would be required. Thus, adding more subjects for the purpose of this analysis would further improve our knowledge for future improvements of the patella design and investigating the differences that exist between genders during different activities of daily living.

Chapter 4 discusses the combined measurement of the differences in tibiofemoral and patellofemoral kinematics, patellar tendon angle, and patellar tendon moment arm that occur during non-weightbearing and weightbearing activities. Differences in tibiofemoral and
patellofemoral joint kinematics, and patellar tendon mechanics between weight bearing and non-weight bearing activities were investigated due to the fact that patellofemoral dysfunctions have been linked with combined abnormal motion of the tibia and patella relative to the femur. The kinematics of the tibiofemoral and patellofemoral joints can change as subjects change demand on their knee and move from non-weightbearing to weightbearing activities. The amount of kinematic change with weightbearing may vary dramatically between individuals, confounding attempts to generate helpful predictions of weightbearing kinematics. Our results demonstrated that while weightbearing elicited changes in knee kinematics, in most DOFs these differences were exceeded by intersubject differences. These results provide comparative kinematics for the evaluation of knee pathology and treatment in older adults, while emphasizing consideration of subject-specific kinematics. Several prior studies have evaluated young healthy, OA, and TKA subjects, and shown important differences between knee kinematics across these three cohorts. However, these studies routinely compare older adults with OA and TKA to young healthy controls. This study fills a gap in this information by providing a cohort more similar in age to those with OA and TKA. Documenting normal changes in knee kinematics with age is necessary for understanding the changes in knee kinematics that occur with OA and TKA while motion of younger subjects may not be representative of the age range associated with knee pain, OA, and TKA. Finally, while prior studies have noted that aging has an impact on knee kinematics measured using marker-based motion capture, few studies have examined the small translations and rotations of the knee in older adults with no history of knee pathology. For these reasons our research describes TF and PF kinematics for a cohort similar in age to TKA recipients. Measurement of healthy subjects that are age matched to populations most associated with knee OA is crucial since countless treatments are designed to restore healthy function to the knee and body without clear comparative mechanics for healthy function in this age group. However, future work could consider assessment of TF
and PF kinematics during different activities of daily living for diverse individuals which may differ from our results in older cohorts and include younger subjects, subjects that have knee OA, and following TKA, with equal numbers who are male, female since this will help to better identify the differences among the different populations and discover what features of the disease persist after treatment.

The aim of chapter 5 was to discover whether notable differences in mobile and fixed bearing kinematics occur during activities of daily living and compare these results with healthy knee kinematics. It has been demonstrated that the kinematics of prosthetic knees are unlike the normal knees, and necessitate excessive sliding and rotational motions which may cause high shear stresses at the joint interface. Successful functional outcome following TKA is influenced by the geometry and design of the components and the available components for knee arthroplasty are routinely offered in Rotating Platforms (RP) and Fixed-Bearing (FB) alternatives. Thus, understanding the effect of design choices on in vivo kinematics and during different dynamic activities of daily living has become more essential since the connection between knee prosthesis kinematics and clinical performance is clearly increasing. Furthermore, prior work has focused on sagittal plane activities and ignored turning motions likely to test the axial rotation kinematics of a rotating platform design. Measurement of kinematics during turning motions are challenging using single-plane fluoroscopy due to poor out of plane accuracy, but can be achieved using bi-plane or stereo radiography setups. Thus, this study investigated the in vivo tibiofemoral kinematics differences in mobile and fixed-bearing kinematics that occur during activities that also promote tibial rotation, lunge and gait with a change in direction (pivot), with a high-speed stereo radiography system for sub-millimeter and sub-degree accuracy. Our results showed that during lunge and pivot, TF kinematics for both designs demonstrated similar trends but with significantly greater IE ROM and greater IE rotation intrasubject variability for the RP design. The trend in kinematics of FB and RP was similar to healthy data, however, the
amount of IE rotation was less than healthy for both activities, with the RP design demonstrating less difference in the amount of IE rotation from the healthy knee. Future work could include a higher number of FB prosthesis design since this study was limited by the smaller number of subjects with a fixed bearing prosthesis. However, this is justified by the low variability in kinematics that was observed for the fixed bearing subjects.

The goal of chapter 6 was to establish functional pelvic orientation that will help improve acetabular cup positioning during total hip arthroplasty (THA). For proper alignment of the acetabular cup, knowledge of the pelvic orientation is essential. The objectives of this study were to establish the functional orientation of the pelvis by using HSSR for higher accuracy and estimate the differences in PI and spinopelvic parameters (APPA, SS, PTS) across different patient populations and across static and dynamic activities of the daily living. This knowledge is necessary to inform acetabular cup placement during THA and reduce the likelihood of impingement events. To our knowledge, no other study has investigated and compared the 3D pelvic functional orientation across different populations that include healthy subjects, subjects that have undergone THA and spinal-stabilized cohorts and during different static and dynamic activities such as gait and stair rise. Results from this study demonstrated that the position of the pelvis in the sagittal plane may vary substantially between functional positions and across different populations and the extent of change is specific to each individual. As a result of these changes, the angles of orientation of the acetabular component during dislocation will be different from those obtained from standard CT and radiographs. Differences were also observed between the static and the dynamic activities during both standing and sitting indicating the need to better understand the dynamic positioning of the pelvis during clinical examination of spinopelvic mobility. Preoperative assessment of the 3D functional pelvic orientation on a patient-specific basis is recommended during different static and dynamic activities of daily living so the hip surgeon can adjust the alignment of the cup to prevent prosthetic dislocation or im-
pingement. In our opinion investigation of the 3D functional orientation of the pelvis is essential since any coronal or axial rotation of the pelvis in functional positions would also affect acetabular orientation and those rotations cannot be seen with standard imaging. We also observed from our results that APPA differences between dynamic and static activities change substantially indicating the need for future studies to consider a dynamic analysis for the component position. Additionally, activities that are commonly related with posterior dislocation, such as rising from a chair or bent over for shoe tying where the hip is flexed further, seems more relevant to investigate the functional orientation of the pelvis rather than when the patient is in the supine position. Benting over activity during sitting brings the pelvis into a more anteriorly tilted position where the acetabular component can become less antverted and risk posterior dislocation. Finally, we suggest that hip surgeons take into account both APPA and SS for each patient so that the planning would be based on a precise anatomic and functional alignment of the cup.
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