

**THE APPLICATION OF BIPLANAR VIDEORADIOGRAPHY TO
THE STUDY OF KINEMATIC AND KINETIC FACTORS
ASSOCIATED WITH NON-CONTACT DECELERATION ACL
INJURY**

By

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BS, Boston University, 2007

Thesis

Submitted in partial fulfillment of the requirements for the
degree of Doctor of Philosophy in Biomedical Engineering at Brown University

Providence, Rhode Island

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This dissertation by Daniel Leo Miranda is accepted in its present form by the Department of Biomedical Engineering as satisfying the dissertation requirement for the degree of Doctor of Philosophy.

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Curriculum vitae

Daniel (Danny) Leo Miranda was born on December 31st, 1984 in Fort Collins, Colorado to Drs. Henry (Rick) and Jeanne Miranda. Danny was raised in Fort Collins, Colorado with his younger sister Maria and younger brother Joseph (Joey). Danny graduated from the International Baccalaureate program at Poudre High School in 2003. Following his high school graduation, Danny attended Boston University where he pursued a degree in Biomedical Engineering. Danny recieved his Bachelor's of Science degree in Biomedical Engineering from Boston University's College of Engineering in May of 2007. After graduating from Boston University, Danny enrolled in Brown University's Graduate School and began his Ph.D. work in the Bioengineering Lab within the Department of Orthopaedics at the Alpert Medical School and Rhode Island Hospital. Working with Dr. Braden C. Fleming, his dissertation advisor, he has been a part of an incredibly productive research team focused on the anterior cruciate ligament. As part of this research team, he has been an author on several peer-reviewed publications, conference proceedings, podium presentations, and poster presentations.

Personal information

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Education and training

Brown University
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Thesis: The application of biplanar videoradiography to the study of kinematic and kinetic factors associated with non-contact deceleration ACL injury

Boston University
Department of Biomedical Engineering, College of Engineering
Boston, MA
Bachelor of Science, Biomedical Engineering, May 2007

Advisor: Dr. Kenneth Lutchen

Thesis: An *in situ* study of deep inspirations during bronchoprovocation in excised lung lobes

Summary of current research

Goals

- Investigate the kinematic, kinetic, and neuromuscular factors associated with the gender bias observed in non-contact ACL injury using advanced medical imaging, force plates, and electromyography
- Explore neuromuscular and biomechanical knee function in ACL reconstructed and ACL deficient men and women before and after surgery
- Impact novel gender specific injury prevention, reconstruction, and rehabilitation strategies
- Reduce the economic burden and long term consequences of ACL injuries

Methods

- Collaborated with research scientists at Brown University's Departments of Orthopaedics, Ecology and Evolutionary Biology, and Computer Science to develop software for non-invasively tracking in vivo human joint motion while minimizing hardware costs and maximizing computational efficiency
- Developed and optimized algorithms in MATLAB for creating accurate, reproducible, and clinically relevant outcome measures for describing knee joint kinematics using computed tomography (CT)
- Designed instrumentation for assessing systematic error of custom data collection systems using SolidWorks
- Integrated hardware and software data collection systems for non-invasively studying in vivo human joint motion

Human subject research experience

- Created, edited, and defended human subject research protocols and consent forms according to Institutional Review Board (IRB) standards
- Organized and implemented recruitment and consent of human subjects
- Conducted human subject research

Research and development interests

Biomechanics, Sports Injury, Anterior Cruciate Ligament (ACL), Kinematics, Kinetics, Osteoarthritis, Injury Prevention, Injury Rehabilitation, Disease Progression, Medical Imaging, Image Processing, High-Speed Motion Capture, Biomechanical Modeling, Muscle Activity, Tissue Engineering, Mechanical Testing, Orthopaedic Medical Devices, Prosthetics, Product Design, Product Development, Data Collection / Processing / Analysis

Skills

Software MATLAB, SolidWorks, C-Motion Visual3D, Materialise Mimics, Autodesk Maya, Geomagic Studio

Hardware Biplanar Videoradiography, Optical Motion Capture, Electromyography (EMG), Computed Tomography (CT), Magnetic Resonance Imaging (MRI), Force Plates, Data Acquisition

Additional Human Subject Research, Large Animal Model (porcine and equine) Research, In Situ Cadaveric Model Studies, Surgical Preparation

Honors and awards

Orthopaedic Research Society poster recognition, January 2011

Top scored Orthopaedic Research Society poster in adult knee poster category; invited to be displayed at the 2011 American Academy of Orthopaedic Surgeons Annual Meeting

Brown University travel award, March 2010, January 2011, February 2012

Granted through the Department of Biology and Medicine and the Center for Biomedical Engineering

Boston University research fellowship, May 2006 - August 2006

Industry Sponsored Research and Design Fellowship in Biomedical Engineering award

Teaching experience

Brown University, Providence, RI

Graduate research mentor, May 2009 - present

- Mentored undergraduate, medical students, and orthopaedic residents during independent research

Teaching assistant, biotechnology in medicine, September 2009 - December 2009

- Presented lectures on orthopaedic medical devices and synthetic biology
- Instructed undergraduate students during regular office hours
- Graded assignments and exams
- Built and maintained an interactive course website

Additional professional experience

Brown University, Providence, RI

Conference coordinating staff, American Society of Biomechanics, August 2010

- Aided in planning and hosting the 2010 annual meeting of the American Society of Biomechanics
- Provided computer and presentation support for poster and podium presenters
- Participated in lab and campus tours
- Coordinated group activities

Leadership

Graduate Biomedical Engineering Society, Brown University Chapter, September 2010 - present

Engineering Student Advisor, Boston University College of Engineering, May 2005 – August 2005

Societal affiliations

Orthopaedic Research Society (ORS), 2010 - present

American Society of Biomechanics (ASB), 2008 - present

Peer-reviewed publications

1. **Miranda DL**, Fadale PD, Hulstyn MJ, Shalvoy RM, Machan JT, Fleming BC. Knee biomechanics during a jump-cut maneuver: Effects of gender & ACL surgery. *Med Sci Sports Exerc.* Under Review (submitted July 2012, in review).
2. **Miranda DL**, Rainbow MJ, Crisco JJ, Fleming BC. Kinematic differences between optical motion capture and biplanar videoradiography during a jump-cut maneuver. *J Biomech.* Under Review (submitted July 2012, in review).
3. **Miranda DL**, Schwartz JB, Loomis AC, Brainerd EL, Fleming BC, Crisco JJ. Static and dynamic error of a biplanar videoradiography system using marker-based and markerless tracking techniques. *J Biomech Eng.* 2011; 133(12):121002. PMID: 22206419
4. **Miranda DL**, Rainbow MJ, Leventhal EL, Crisco JJ, Fleming BC. Automatic determination of anatomical coordinate systems for three-dimensional bone models of the isolated human knee. *J Biomech.* 2010; 43(8):1623-6. PMID: 20167324

Peer-reviewed conference abstracts

1. **Miranda DL**, Rainbow MJ, Crisco JJ, Fleming BC. Kinematic differences between optical motion capture and biplanar videoradiography during a jump-cut maneuver. The Gordon Research Conference on Musculoskeletal Biology & Bioengineering. Proctor Academy, August 5-10, 2012. *Poster Presentation.*
2. Rainbow MJ, Cheung R, **Miranda DL**, Schwartz JB, Crisco JJ, Davis IS, Fleming BC. Tracking high-speed patella motion using biplanar videoradiography: an accuracy study. The 36th Annual Meeting of the American Society of Biomechanics. Hilton University of Florida Conference Center, Gainesville, FL. August 15-18, 2012. *Podium Presentation.*
3. Rainbow MJ, **Miranda DL**, Cheung R, Fleming BC, Davis IS. Automatic Determination Of A Morphology-Based Anatomical Coordinate System Of The Human Patella. The 59th Annual Meeting and 3rd World Congress on Exercise is Medicine of the the American College of Sports Medicine, Moscone West Convention Center, San Francisco, CA. May 29-June 2, 2012. *Poster Presentation.*
4. **Miranda DL**, Rainbow MJ, Crisco JJ, Fleming BC. Kinematic Errors due to Soft Tissue Artifact during a Jump-Cut Maneuver: A Comparison between Optical Motion Capture and Bi-planar X-ray Technologies. The 58th Annual Meeting of the Orthopaedic Research Society, Moscone West Convention Center, San Francisco, CA. February 4-7, 2012. *Poster Presentation.*

5. Biercevicz AM, **Miranda DL**, Murray MM, Machan JT, Fleming BC. T2* Weighted MRI Derived Morphology and Signal Intensity to Determine Structural Properties of an ACL Reconstruction Graft in a Porcine Model. The 58th Annual Meeting of the Orthopaedic Research Society, Moscone West Convention Center, San Francisco, CA. February 4-7, 2012. *Podium Presentation*.
6. **Miranda DL**, Rainbow MJ, Crisco JJ, Fleming BC. Biplanar X-ray Derived ACL Excursions during a Jump-Cut Maneuver Associated with ACL Injury. International Symposium on Ligaments & Tendons - XII, Hilton Financial District, San Francisco, CA. February 3, 2012. *Podium Presentation*.
7. Biercevicz AM, **Miranda DL**, Murray MM, Machan JT, Fleming BC. MRI Derived Morphology and Signal Intensity to Determine Structural Properties of an ACL Reconstruction Graft or ACL Primary Repair at One Year in a Porcine Model. International Symposium on Ligaments & Tendons - XII, Hilton Financial District, San Francisco, CA. February 3, 2012. *Poster Presentation*.
8. Childs B, Brainerd EL, **Miranda DL**, Crisco JJ, Kirker-Head CA. Novel Imaging Modality for In Vivo 3D Tracking of Equine Metacarpophalangeal Joint Kinematics: A Preliminary Study. The American College of Veterinary Surgeons Veterinary Symposium, Chicago, IL. November 3-5, 2011. *Poster Presentation*.
9. **Miranda DL**, Schwartz JB, Dawson MM, Fleming BC, Crisco JJ. Markerless Tracking Error of a Bi-Planar X-Ray Motion Capture System. The 35th Annual Meeting of the American Society of Biomechanics, The Westin Long Beach, Long Beach, CA. August 10-13, 2011. *Poster Presentation*.
10. **Miranda DL**, Gosselin MM, Brainerd EL, Crisco JJ, Fleming BC. Gender Differences in Human Knee Function During Maneuvers Associated with Non-Contact ACL Injury. The 57th Annual Meeting of the Orthopaedic Research Society, Long Beach Convention Center, Long Beach, CA. January 13-16, 2011. *Poster Presentation*.
11. **Miranda DL**, Brainerd EL, Fleming BC, Crisco JJ. Accuracy and Precision of 3-D Skeletal Motion Capture Technology. The 56th Annual Meeting of the Orthopaedic Research Society, Morial Convention Center, New Orleans, LA. March 6-9, 2010. *Short-Talk Podium Presentation & Poster Presentation*.
12. Hillstrom HJ, Syrkin G, Lenhoff M, **Miranda DL**, Backus SI, Wolfe S, Crisco JJ. Three-Dimensional Wrist Kinematic Accuracy: A Comparison of Markerless Bone Registration and Surface Markers in a Cadaver Model. The 55th Annual Meeting of the Orthopaedic Research Society, Venetian Hotel Resort and Casino, Las Vegas, NV. February 22-25, 2009. *Poster Presentation*.
13. **Miranda DL**, Rainbow MJ, Crisco JJ, Fleming BC. Standardized Anatomical Coordinate Systems for 3-D Bone Models of the Human Knee. The 55th Annual Meeting of the Orthopaedic Research Society, Venetian Hotel Resort and

Casino, Las Vegas, NV. February 22-25, 2009. *Short-Talk Podium Presentation
& Poster Presentation.*

14. **Miranda DL**, Rainbow MJ, Brainerd EL, Fleming BC. A Preliminary Study: Tracking 3D Kinematics of the Goat Knee Joint In Vivo. North American Congress on Biomechanics, University of Michigan, Ann Arbor, MI. August 5-9, 2008. *Poster Presentation.*
15. **Miranda DL**, Rainbow MJ, Brainerd EL, Fleming BC. Tracking 3D Kinematics of Healthy and ACL-Transected Goat Knee Joints In Vivo: A Preliminary study. The 34th Annual Northeast Bioengineering Conference, Brown University, Providence, RI. April 4-6, 2008. *Podium Presentation.*

Invited lectures

1. Colorado State University, Orthopaedic Research Center. Using biplanar x-ray to quantify the neuromechanics of jumping & cutting in ACL-I, ACL-D, and ACL-R athletes. June 2011.
2. Rhode Island Hospital, Department of Orthopaedics. Using XROMM to quantify the neuromechanics of jumping & cutting in ACL-I, ACL-D, and ACL-R athletes. May 2011.
3. Brown University, Department of Ecology and Evolutionary Biology. Exploring the gender differences in the neuromechanics of jumping and landing in ACL-I and ACL-R athletes. December 2010.

Preface and acknowledgements

The past five years at Brown University have been an incredible journey. I came here in the fall of 2007 and joined the Bioengineering Laboratory, which has become a second home to me. At Brown, I have had the opportunity to work with incredibly talented faculty, staff, and students. Being a part of this academic environment has truly been a privilege. I not only consider them mentors and colleagues, but friends as well. I hope to continue these friendships and collaborations throughout my life and career. Thank you all.

I am extremely grateful to my advisor, Dr. Braden Fleming. He has been a great role model throughout my time at Brown, and I have certainly learned an incredible amount working with him these past five years. Having his input and guidance on my thesis work was invaluable. The freedom with which he let me pursue my interests has also been a blessing and something that I am very grateful for. Most notably, his availability and willingness to participate in every aspect of my research has been made working for him easy. I can honestly say that I have become a better engineer, researcher, and scientist under his guidance. For that I am truly thankful.

My thesis committee has been fantastic, and I appreciate their guidance and feedback throughout this process. Dr. Trey Crisco has been an integral part of my research in the Bioengineering Lab. His mentorship on biomechanics, validation, and problem solving has been an essential part of my education. Dr. Beth Brainerd has been

someone whom I have had the pleasure of learning from and working with since the day I got to Brown. She was instrumental in much of my research moving from an idea to an outcome. Without her leadership and ability to bring a community of scientists and engineers together, XROMM would not be what it has become. Dr. David Laidlaw has given his insights and expertise in medical imaging and computational problem solving to provide an incredibly unique perspective on the biomechanical problems that were faced in my thesis work. Having Dr. Mike Bey as a part of my committee has been a joy, and he has provided an insightful outside viewpoint on the use of biplanar videoradiography. Thanks to all of you for helping make this research possible.

I have been blessed with an incredible family who have been supportive throughout my entire dissertation experience. Thank you Mom and Dad; without everything you have given me throughout my life I would not be where I am today. Maria and Joey, thank you for being such incredible siblings. Despite being the oldest, I look up to both of you. Maria, you have always been such a strong, smart, and beautiful person, and I have certainly tried to live up to your example. Joey, I admire your courage, and your enthusiasm with which you pursue art is inspirational.

I am so grateful to have become a part of the Hehl family during my time at Brown. My mother and father-in-law have been a second set of parents for much longer than I have been married to Christina. They are some of the most generous people that I have met, and I am so proud to call them Mom and Dad. My brother-in-law, Steve, has become a good friend and someone whose company I very much enjoy.

Finally, I need to thank my friends. I am lucky to have so many people from so many different aspects of my life that I can count on. I am unsure how I have been blessed with all of your friendships, but I am thankful for them each day.

Dedication

I lovingly dedicate this to my wonderful wife, Christina. Her love and friendship have been and will perptually be my inspiration.

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Nomenclature

ACS	Anatomical Coordinate System
ADT	Angular Displacement Transducer
ACL	Anterior Cruciate Ligament
PLB	ACL Posterolateral Bundle
AMB	ACL Anteromedial Bundle
ACL _{INT}	ACL-Intact
ACL _{REC}	ACL-Reconstructed
ACL _{DEF}	ACL-Deficient
AUC	Area Under the Curve
BF	Biceps Femoris
BMI	Body Mass Index
COBRE	Centers of Biomedical Research Excellence
CPU	Central Processing Unit
CT	Computed Tomography

CI	Confidence Interval
DOF	Degree(s) Of Freedom
DRR	Digitally Reconstructed Radiograph
DLT	Direct Linear Transform
EMG	Electromyography
FOV	Field Of View
FPS	Frames Per Second
G	Gastrocnemius
GPU	Graphics Processing Unit
H	Hamstring
HAM	Helical Axis of Motion
II	Image Intensifier
JCS	Joint Coordinate System
FE or FL/EX	Kinematic Flexion/Extension Rotation
AA or AD/AB	Kinematic Adduction/Abduction Rotation
IE or IN/EX	Kinematic Internal/External Rotation
ML or ME/LA	Kinematic Medial/Lateral Translation
AP or AN/PO	Kinematic Anterior/Posterior Translation
CD or CO/DI	Kinematic Compression/Distrraction Translation
LG	Lateral Gastrocnemius

LOAD	Loading Phase
LWM	Local Weighted Mean
MG	Medial Gastrocnemius
NIH	National Institutes of Health
OMC	Optical Motion Capture
PTOA	Post-Traumatic Osteoarthritis
PREP	Preparatory Phase
Q	Quadriceps
RF	Rectus Femoris
ST	Semitendinosus
STA	Soft Tissue Artifact
SID	Source to Image Distance
SD	Standard Deviation
3D or 3-D	Three-Dimensional
VM	Vastus Medialis
XROMM	X-Ray Reconstruction of Moving Morphology

Chapter 1

Introduction

Overview

The knee is a diarthrodial joint composed of the patellofemoral joint and the tibiofemoral joint. The patellofemoral joint is the articulation between the anterior aspect of the distal femur with the posterior aspect of the patella. The tibiofemoral joint, which is the focus of this work, is the articulation between the distal femur and the proximal tibia. Tibiofemoral joint motion is three-dimensional (3-D) and exists in the sagittal, coronal, and axial anatomic planes. The six-degree-of-freedom (DOF) motion is quantified as three rotational motions and three translational motions (Figure 1.1).

The articulating surfaces of the femur and tibia are covered with articular cartilage and interact with the menisci and synovial fluid. In addition, the knee is comprised of four major ligaments (Figure 1.2): the medial collateral ligament, the lateral collateral ligament, the anterior cruciate ligament (ACL), and the posterior cruciate ligament. The ACL is the primary ligament and focus of the work herein. The ACL originates on the medial aspect of the lateral femoral condyle and inserts on the anterior aspect of the tibial plateau. It functions to prevent excess anterior tibial translation, restrain internal tibial rotation, and stabilize knee abduction and adduction [1–8].

ACL injuries commonly occur during sport activities and result from the sudden deceleration of the tibia with respect to the femur. This often occurs under non-contact conditions where kinematic and kinetic factors produce excessive stress on the ligament. Furthermore, women suffer non-contact deceleration ACL injuries significantly more than men when participating in the same sport activities. Functional healing of the ACL is rare in both males and females and often results in an unstable ACL deficient (ACL_{DEF}) knee. Presently, arthroscopically-assisted ligament reconstruction (Figure 1.3) is typically required to treat the resulting functional instability. However, while ACL reconstruction is effective at restoring stability to the joint, the risk of developing premature post-traumatic osteoarthritis (PTOA) is not mitigated. While

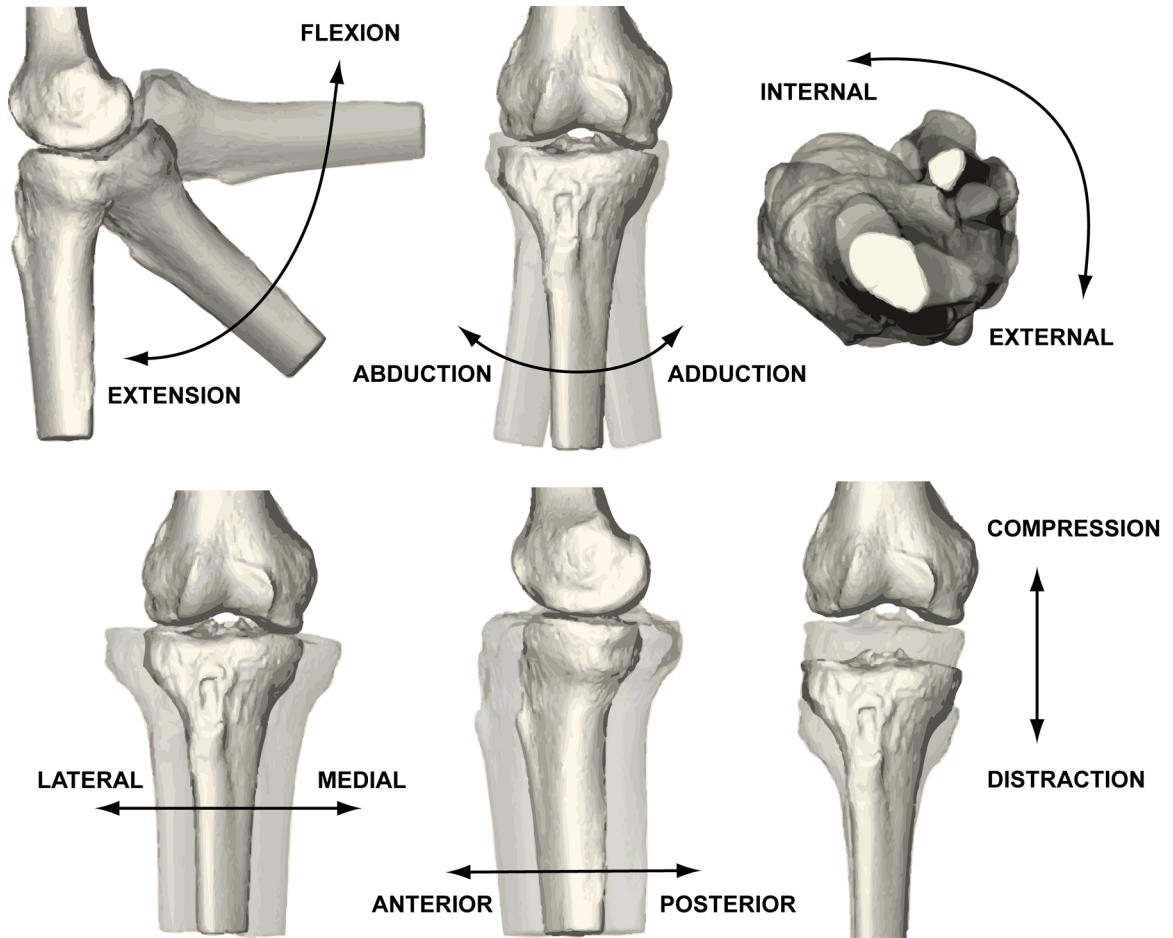


Figure 1.1: Illustration showing the six-DOF motion of the tibiofemoral joint. The top row represents three rotations: flexion/extension (left), abduction/adduction (middle), and internal/external (right). The bottom row represents the three translations: medial/lateral (left), anterior/posterior (middle), and compression/distraction (right).

much research has focused on the biomechanical factors contributing to the gender bias observed for non-contact deceleration ACL injury and the premature joint degradation seen despite ACL reconstruction, it remains unclear what kinematic and kinetic differences influence the injury mechanism and set the stage for premature onset and progression of PTOA.

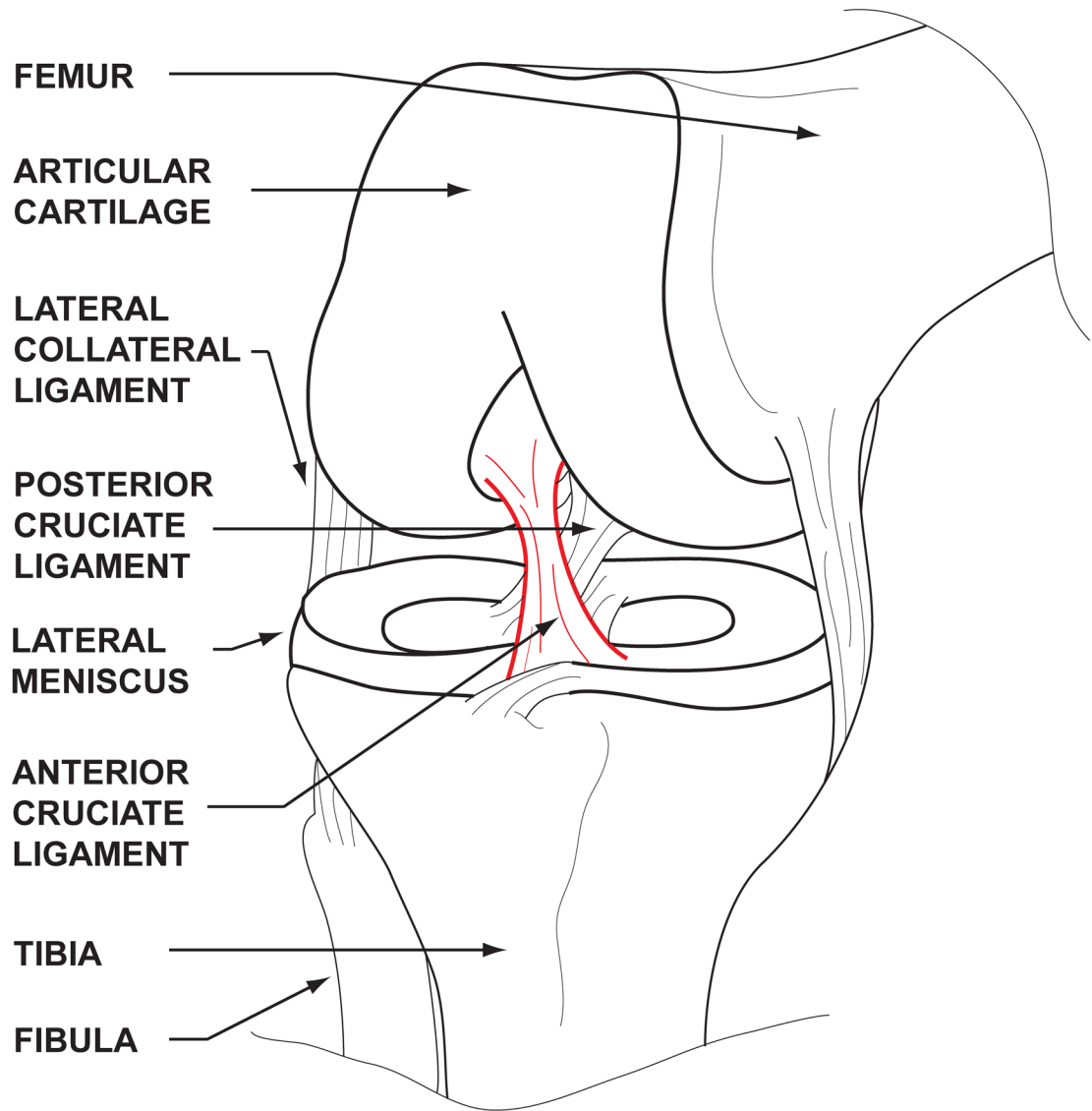


Figure 1.2: Illustration representing the primary structures of the tibiofemoral joint. The ACL is located at the center of the joint and is colored in red. The medial collateral ligament and medial meniscus (not labeled) are located on the right, opposite the lateral collateral ligament and lateral meniscus, respectively.

The overall objective of this work is to compare the biomechanics of male and female

ACL-intact (ACL_{INT}) and ACL-reconstructed (ACL_{REC}) recreational athletes with the hope of elucidating both gender and ACL reconstruction status differences. This work makes significant use of custom biplanar videoradiography technology to accurately quantify the motion of the knee (Figure 1.4), and as a result, techniques need to be developed and validated to safely measure *in vivo* knee kinematics and kinetics using this technology. Therefore, this work is a collection of method development, validation, and hypothesis driven research focused on identifying differences between gender and knee status in recreational athletes performing a maneuver associated with non-contact deceleration ACL injury. It is our long-term hope that the results obtained from this work will inform novel injury prevention and rehabilitation strategies to reduce injury rate and mitigate the long term consequences observed after ACL reconstruction.

1.1 Background

1.1.1 Anatomy and function of the ACL

The ACL is one of four major ligaments comprising the knee. It is made of two bundles: the anteromedial bundle and the posterolateral bundle, which are taut in flexion and extension respectively [1]. The ACL originates from within the notch of the distal femur, where the ligament fibers are attached to the medial wall of the lateral femoral condyle. The ACL inserts anterior to the intercondyloid eminence of the tibia, blending with the anterior horn of the lateral meniscus. This anatomy allows the ACL to prevent excess anterior translation of the tibia relative to the femur [2]. In fact, studies have shown that more than 80% of the force restraining excess anterior translation is provided by the ACL between 30° and 90° of knee flexion [3]. The ACL also induces internal rotation of the tibia during anterior tibial

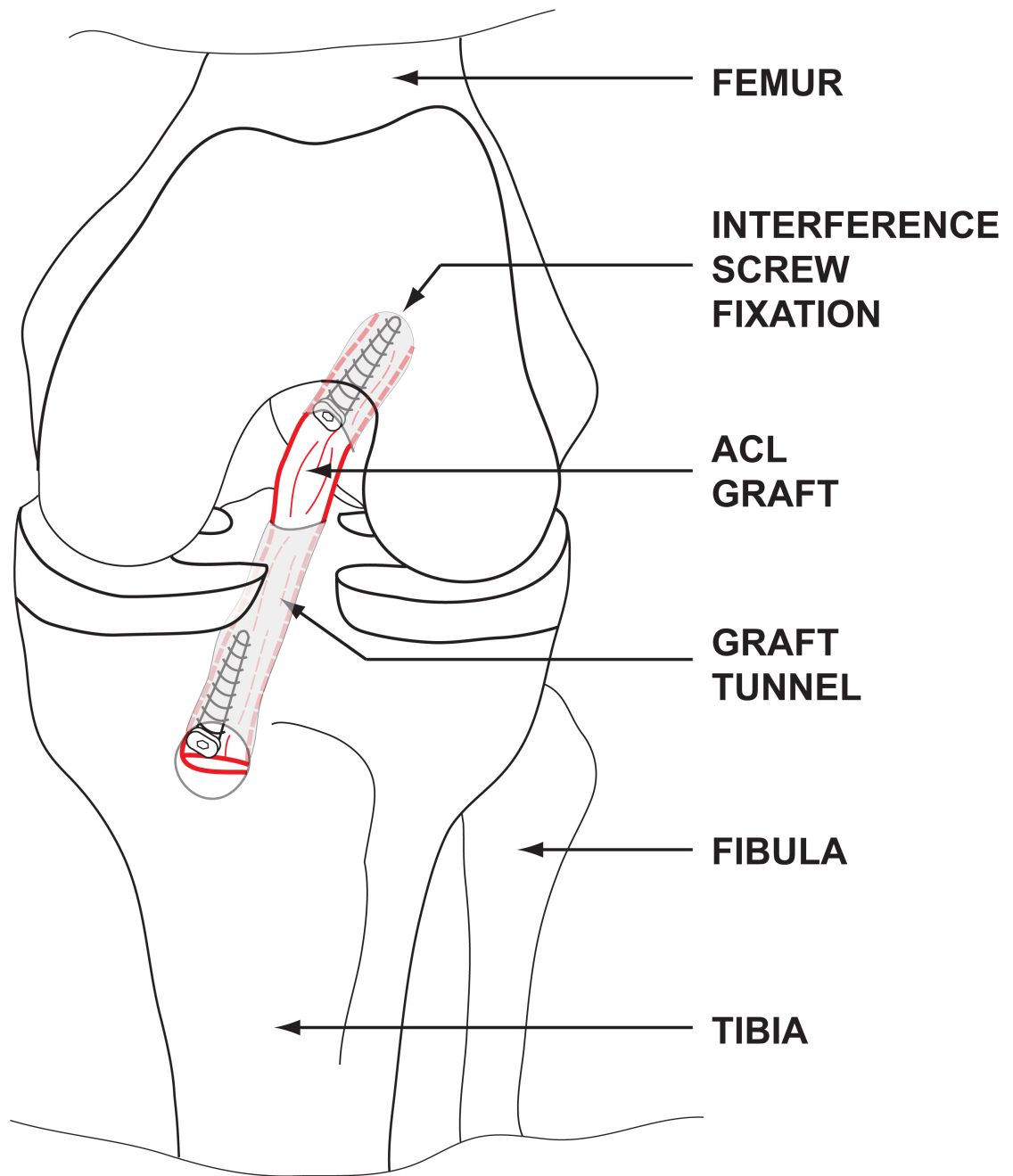


Figure 1.3: Illustration representing an arthroscopic ACL reconstruction. The ACL graft is highlighted in red and is secured within the bone by two interference screws. The drill tunnel, which is highlighted in gray, attempts to restore the natural geometry of the ACL.

translation. This suggests that the ACL restrains internal rotational moments during anterior-posterior tibial translation. Additionally, the ACL is a secondary stabilizer to abduction and adduction forces about the knee [3–7]. Ultimately, the intact ACL contributes significantly to the natural motion of the knee: a delicate balance of ‘rolling’ and ‘gliding’ of the femur on the tibia during flexion and extension [1,9,8].

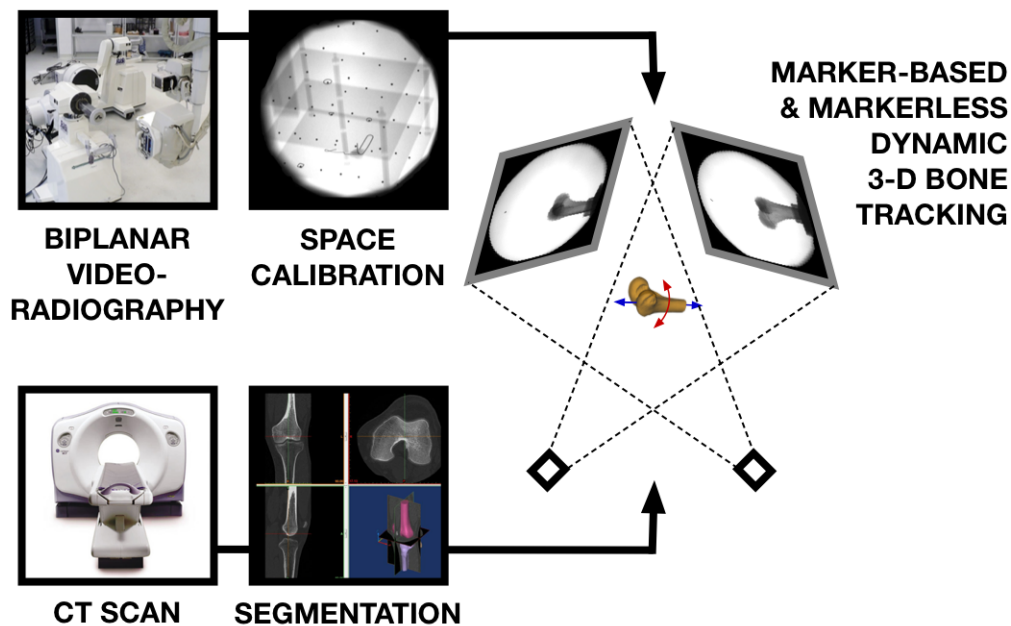


Figure 1.4: Schematic overview of biplanar videoradiography technology. There are two essential pieces of data that are required. First is the actual biplanar videoradiography motion capture sequences. This system allows for videoradiography to be captured at up to 1,000 frames-per-second. The 3-D position of any point within its field of view is known once the system is calibrated. The second piece of data is the CT scan, which provides the volumetric images of the bones or objects of interest. The bone models are isolated using standard segmentation techniques and then combined with the biplanar videoradiography sequences. Finally, the 3-D motion of the bones or objects of interest can be extracted after performing either marker-based or markerless tracking in a virtual 3-D environment.

1.1.2 Non-contact deceleration ACL injury

ACL injuries commonly result from the sudden deceleration that occurs prior to a rapid direction change or leg landing [10]. These types of non-contact maneuvers produce abduction and internal rotational moments about the extended knee joint [11–14]. Excessive quadriceps contraction combined with hamstring weakness during extension is also thought to contribute to the non-contact deceleration ACL injury mechanism. The quadriceps muscles are active during knee extension while the hamstring muscles act as antagonists to the quadriceps during deceleration of knee extension [14–20]. The combination of these complex and violent biomechanical factors rarely results in an isolated disruption of the ACL. It is reported that approximately 80% of ACL injuries are associated with either traumatic tears of the medial meniscus, lateral meniscus, medial collateral ligament, or a combination of all three [1,21,22]. Additionally, chondral and subchondral bruising from bony impaction is likely to occur [23–25]. After ACL injury, the normal ‘roll-glide’ kinematics of the knee are disrupted, resulting in increased anterior tibial translation, increased internal tibial rotation, and altered joint contact mechanics [1].

1.1.3 Surgical ACL reconstruction

Research presently supports the use of arthroscopically-assisted reconstruction of the ACL to improve the sagittal plane stability and function of the knee for most patients after injury [1]. Reconstruction begins with a surgical removal of the torn ACL tissue. Once the original ACL tissue is removed, it is replaced with a tendon graft harvested from either the central third of the patellar tendon, the medial hamstrings, or an allograft [26]. Despite the general success of ACL reconstruction in restoring the stability of the knee in the sagittal plane, both immediate and long-term problems

remain. In the instances where autograft tissue is harvested, biomechanical and structural changes at the area of the graft are sacrificed along with the procedure related morbidities. When allograft tissue is used, cost, availability, immune response, and disease transmission present problems for the patient. In either case, large prospective studies have shown that the risk of tearing the reconstructed autograft is up to 5.7% and is even greater in younger patients [27]. Ultimately, no matter how precisely a reconstruction restores the natural geometric anatomy of the ACL, native sensory nerve fibers within the joint capsule and the complex fan-shaped bony integration is not retained. Thus, a failure to fully restore both the neuromuscular and rotational kinematics of the knee is likely. These failures may influence the premature degradation of the knee joint, which have been shown to occur with or without ACL reconstruction [1,26,28].

1.1.4 ACL injury related demographics

Failure of the ACL has been observed as far back as 1845 when Amédée Bonnet described three essential signs indicative of acute ACL injury for patients who have not fractured a bone: “a snapping noise, haemarthrosis, and loss of function are characteristic of ligamentous injury in the knee.” Currently, significant problems still remain despite the advancements in treatment options for ligamentous injury [26]. Currently, over 400,000 ACL injuries occur in the United States each year [29]. It is estimated that approximately 70% of these injuries are non-contact, resulting from the sudden deceleration that occurs prior to a rapid direction change or leg landing [30]. It has been shown that when participating in the same high-risk activities involving jumping and cutting maneuvers, females tend to suffer ACL injury at a rate two to ten times greater than males [31–33]. The total surgical cost of an ACL reconstruction is estimated to fall between \$5,300 and \$7,000 based on data published in 2012 [34].

Furthermore, data published in 2011 estimates the cost of a quality adjusted life-year to be \$10,000 per injury [35]. These surgical and rehabilitation costs can amount to millions of dollars annually for women alone. This is in addition to the long term costs associated with post-traumatic knee OA, which is thought to prematurely present in 80% to 90% of patients as early as seven years after injury [29].

The increased risk of ACL injury to female athletes in conjunction with the growing numbers of female collegiate sport participants has heightened public awareness of the gender bias. This has prompted researchers to probe anatomical (ACL size, femoral notch size, Q-angle, etc...) and biomechanical (kinematics and kinetics of the leg and trunk, quadriceps activation, etc...) factors that may lead to the increased rate of ACL injuries in female athletes. Although the links between gender and ACL injury are not fully understood, differing anatomy and joint mechanics likely play a significant role. Understanding these gender differences in both ACL_{INT} and ACL_{REC} knees may lead to novel preventative and rehabilitation strategies [14].

1.1.5 Optical motion capture techniques

Because the non-contact injury mechanism is associated with highly dynamic maneuvers, studies have focused on measuring motion of and forces across the knee joint during laboratory controlled jumping, landing, cutting, and pivoting maneuvers [13,14,36–41]. These data suggest that women, compared with men, appear to cut, pivot, and land from a jump with less knee flexion, increased knee abduction, increased external rotation of the hip, and low hamstring activity relative to quadriceps activity (quadriceps-dominant contraction). While muscle activation data and ground reaction force (GRF) data collected from standard electromyography (EMG) and force plate systems are acceptably reliable, kinematic measurements performed using traditional optical motion capture (OMC) techniques are associated with signif-

icant errors. These techniques employ reflective surface markers to track underlying bone motion [11,13,16,37,40,42–47]. The surface markers are generally fixed externally to the skin and have been shown to produce large errors due to the relative motion between the markers and the bones. This relative motion is a consequence of the elasticity of the skin and muscle tissue surrounding the bone. Specifically, the quadriceps and hamstrings are large muscles that surround the femur, and the gastrocnemius, soleus and tibialis anterior muscles surround the tibia. During highly dynamic jumping, pivoting, and cutting maneuvers, the muscles, skin, and bone may not move as a single rigid body; rather, they likely move with their own distinct motion patterns.

Studies using implanted bone pins have shown that markers attached to the surface of the skin move as much as 30mm relative to the bone during rapid movements [48,49]. In an attempt to reduce these inaccuracies, Andriacchi et al. has developed innovative techniques that track a cluster of markers uniformly distributed on a limb segment. These techniques employ algorithms that model soft tissue properties and detect violations in rigid body mechanics, yet *in vivo* validation is limited [48,49]. Using the point cluster method, Alexander and Andriacchi were able to show that error was significantly reduced (0.8 mm – average error, 2.6 mm – max error) during low speed stair stepping activities [50]. However, this study validated the point cluster method by tracking only the tibia of a single subject using bone pins that could significantly stiffen the soft tissue surrounding the bone. These errors would likely increase under more dynamic maneuvers associated with non-contact deceleration ACL injury, especially for the more muscular thigh segment. Not only are these errors problematic when evaluating dynamic knee kinematics, they are unacceptable when attempting to estimate stresses to the meniscus, cartilage, and ligaments. Changes in these tissues, especially during dynamic maneuvers, can be small enough that errors of even a few millimeters will produce enormous uncertainties in the data [51,52].

1.1.6 Bone pin, CT, and MRI based tracking techniques

Measuring dynamic joint motion with bone pins has been used in a handful of studies involving human subjects [48,53–55] and large animal models [56]. While this technique eliminates the errors caused by relative skin and muscle motion in relation to the underlying bone, it is highly invasive and inevitably alters normal movement. The invasive nature of this technique would certainly constrain the number of willing participants, hindering an investigator’s ability to conduct adequate studies. Alternative methods employing magnetic resonance imaging (MRI) and computed tomography (CT) to track joint motion *in vivo* have been successful; however, they are limited to static positions or low frame rates [57–59]. Additionally, highly dynamic jumping, pivoting, and cutting maneuvers would be impossible to carry out within a confined CT or MRI environment. The use of conventional fluoroscopy, which allows *in vivo* visualization of the bones, is limited to a single plane and motion can only be visualized and tracked in two-dimensions (2-D); any out of plane motion or object displacement is not quantifiable and prone to errors [60]. Furthermore, the frame rates and shutter speeds associated with most conventional fluoroscopes are too low to eliminate errors from motion blur during dynamic maneuvers [61].

1.1.7 Biplanar videoradiography motion capture techniques

High-speed biplanar videoradiography has been employed in a number of studies that take advantage of the technology’s ability to capture 3-D skeletal motion *in vivo* [62,63]. The advancements in the techniques have allowed investigators to progress from marker-based bone tracking techniques to markerless tracking techniques. Traditionally, a minimum of three radio-opaque spherical beads are implanted into bones of interest, and their 3-D locations are digitized using standard direct linear transform

(DLT) methodology [63]. The combination of these 3-D marker location data with CT or MRI based 3-D bone models allow quantitative and qualitative skeletal motion to be assessed in 3-D. This marker-based tracking technique is capable of accuracy and precision below 0.1 mm and 0.1 degrees, making it ideal for studying the affect of skeletal motion on ligamentous and cartilagenous tissues.

These marker-based tracking techniques still require invasive bead implantation procedures that limit the number of willing human subject volunteers. Recently, markerless bone registration algorithms have been developed to eliminate the need for embedding radio-opaque beads into the bones of interest [64,65]. The simplest of these techniques, labeled as scientific roscoping, requires an experienced investigator to manually manipulate CT or MRI based bone models to match the biplanar radiographs in a 3-D environment [66,67]. These techniques require significant user interactions and can be laborious and time consuming for large data sets with significant bone motion over many frames. This has lead to the development of 3-D volumetric algorithms to track bone motion [65]. These techniques are based on the assumption that a properly oriented projection through a 3-D volumetric model will produce an image similar to the captured bi-planar videoradiographs. Because these techniques rely heavily on computational power they significantly reduce investigator interactions. Validation studies conducted by Anderst et al. and Bey et al. have shown these markerless tracking techniques to be capable of accuracy and precision similar to marker-based tracking techniques ($\sim 0.68 - 0.32$ mm bias) [68–70].

1.2 Significance

The combination of high accuracy and precision with non-invasive model based bone tracking algorithms provide an extremely powerful method for tracking human joint

motion *in vivo*. The application of this technique to study gender and ACL reconstruction status differences during a jump-cut maneuver associated with non-contact deceleration ACL injury has not been done. The qualitative and quantitative data collected using high-speed biplanar videoradiography, EMG, and force plates from ACL_{INT} and ACL_{REC} subjects performing dynamic jumping, cutting, and pivoting maneuvers will provide reliable, six-DOF joint kinematics and kinetics directly before and after ground contact. Furthermore, the tools and techniques necessary for capturing joint kinematics and kinetics during dynamic maneuvers will contribute to a better understanding of the gender specific mechanisms responsible for non-contact deceleration ACL injury. The results from this study will help engineers, physicians, and therapists develop more effective injury prevention and rehabilitation techniques. Furthermore, establishing these techniques using the large bones comprising the knee (femur, tibia, and patella) is an important step in translating this technology to study more complex joints like the ankle and wrist.

1.3 Specific aims

1.3.1 Specific Aim 1

Develop and validate methods for accurately quantifying in vivo knee bone motion using biplanar videoradiography

The first aim was designed to validate software techniques for tracking and quantifying *in vivo* bone motion using biplanar videoradiography. The first objective was to accurately validate new markerless tracking software that implemented previously described markerless tracking algorithms on a graphics processor. We used independent, gold standard instrumentation and cadaveric bone models to determine the

static and dynamic marker-based and markerless tracking error associated with the W.M. Keck Foundation’s biplanar videoradiography system.

The second objective was to develop an algorithm for defining robust coordinate systems based on the 3-D geometry of the distal femur and proximal tibia in order to extract clinically relevant joint kinematics. We used CT-based cadaveric 3-D bone models as inputs to an algorithm; this takes advantage of the inertial properties and morphology of the femur and tibia to build ACSs based on their articulating surfaces.

1.3.2 Specific Aim 2

Collect and quantify the kinematics of ACL_{INT} individuals performing an activity associated with non-contact deceleration ACL injury

The objective of the second aim was to determine the feasibility of collecting and quantifying the leg motion of ACL_{INT} recreational athletes performing an activity associated with non-contact deceleration ACL injury. Additionally, we wanted to optimize methods for combining motion capture modalities to expand the field of view (FOV) of the biplanar videoradiography system. We also wanted to understand how soft tissue affects kinematic measurements obtained from traditional OMC techniques during the proposed activity.

To address this aim, we designed an experimental set-up capable of synchronously measuring leg motion using biplanar videoradiography and OMC as well as EMG and GRF data. We recruited ten ACL_{INT} subjects and had them perform a jump-cut maneuver which is designed to mimic sport activities that are associated with non-contact deceleration ACL injury. We hypothesized that the soft tissue surrounding the femur would influence the kinematic measurements more than the soft tissue surrounding the tibia. Additionally, we hypothesized that soft tissue artifact would

significantly influence the OMC joint kinematic measurements after ground contact when compared to the biplanar videoradiography measurements.

1.3.3 Specific Aim 3

Investigate gender and ACL reconstruction status differences in kinematics and kinetics of ACL_{INT} and ACL_{REC} subjects performing an activity associated with non-contact deceleration ACL injury

The objective of the third aim was to investigate the kinematic and kinetic gender and ACL reconstruction status differences using the methods developed and implemented in Specific Aim 1 and Specific Aim 2. We recruited ten ACL_{REC} subjects and collected kinematic, kinetic, EMG, and GRF data while they performed the jump-cut maneuver. We hypothesized that kinetic and kinematic differences would be dependent on both gender and ACL reconstruction status. Specifically, we hypothesized that ACL_{INT} subjects would have larger GRF after contact as compared to ACL_{REC} subjects. We also hypothesized that ACL_{REC} subjects would have similar anterior tibial translation values but altered secondary rotational kinematics when compared to the ACL_{INT} subjects. Additionally, we hypothesized that female subjects would land with less knee flexion and larger peak vertical GRF as compared to the male subjects. Finally, we hypothesized that females would abduct and internally rotate their tibia more following contact and they would have increased anterior tibial translation after contact as compared to their male counterparts.

1.4 References

- [1] Dargel J., Gotter M., Mader K., Pennig D., Koebke J., and Schmidt-Wiethoff R., 2007, “Biomechanics of the anterior cruciate ligament and implications for surgical reconstruction,” *Strategies Trauma Limb Reconstr*, 2(1), pp. 1–12.
- [2] Fleming B. C., Renstrom P. A., Beynnon B. D., Engstrom B., Peura G. D., Badger G. J., and Johnson R. J., 2001, “The effect of weightbearing and external loading on anterior cruciate ligament strain,” *J Biomech*, 34(2), pp. 163–170.
- [3] Butler D. L., Noyes F. R., and Grood E. S., 1980, “Ligamentous restraints to anterior-posterior drawer in the human knee. A biomechanical study,” *J Bone Joint Surg Am*, 62(2), pp. 259–270.
- [4] Fukubayashi T., Torzilli P. A., Sherman M. F., and Warren R. F., 1982, “An in vitro biomechanical evaluation of anterior-posterior motion of the knee. Tibial displacement, rotation, and torque,” *J Bone Joint Surg Am*, 64(2), pp. 258–264.
- [5] Grood E. S., Stowers S. F., and Noyes F. R., 1988, “Limits of movement in the human knee. Effect of sectioning the posterior cruciate ligament and posterolateral structures,” *J Bone Joint Surg Am*, 70(1), pp. 88–97.
- [6] Lipke J. M., Janecki C. J., Nelson C. L., McLeod P., Thompson C., Thompson J., and Haynes D. W., 1981, “The role of incompetence of the anterior cruciate and lateral ligaments in anterolateral and anteromedial instability. A biomechanical study of cadaver knees,” *J Bone Joint Surg Am*, 63(6), pp. 954–960.
- [7] Markolf K. L., Kochan A., and Amstutz H. C., 1984, “Measurement of knee stiffness and laxity in patients with documented absence of the anterior cruciate ligament,” *J Bone Joint Surg Am*, 66(2), pp. 242–252.
- [8] Smith P. N., Refshauge K. M., and Scarvell J. M., 2003, “Development of the concepts of knee kinematics,” *Arch Phys Med Rehabil*, 84(12), pp. 1895–1902.
- [9] Freeman M. A. R., and Pinskerova V., 2005, “The movement of the normal tibio-femoral joint,” *J Biomech*, 38(2), pp. 197–208.
- [10] Gianotti S. M., Marshall S. W., Hume P. A., and Bunt L., 2009, “Incidence of anterior cruciate ligament injury and other knee ligament injuries: a national population-based study,” *J Sci Med Sport*, 12(6), pp. 622–627.
- [11] Besier T. F., Lloyd D. G., Cochrane J. L., and Ackland T. R., 2001, “External loading of the knee joint during running and cutting maneuvers,” *Med Sci Sports Exerc*, 33(7), pp. 1168–1175.
- [12] Boden B. P., Dean G. S., Feagin J. A., and Garrett W. E., 2000, “Mechanisms of anterior cruciate ligament injury,” *Orthopedics*, 23(6), pp. 573–578.

- [13] Chappell J. D., Yu B., Kirkendall D. T., and Garrett W. E., 2002, "A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks," *Am J Sports Med*, 30(2), pp. 261–267.
- [14] Griffin L. Y., Albohm M. J., Arendt E. A., Bahr R., Beynnon B. D., Demaio M., Dick R. W., Engebretsen L., Garrett W. E., Hannafin J. A., Hewett T. E., Huston L. J., Ireland M. L., Johnson R. J., Lephart S., Mandelbaum B. R., Mann B. J., Marks P. H., Marshall S. W., Myklebust G., Noyes F. R., Powers C., Shields C., Shultz S. J., Silvers H., Slauterbeck J., Taylor D. C., Teitz C. C., Wojtys E. M., and Yu B., 2006, "Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting, January 2005," *Am J Sports Med*, 34(9), pp. 1512–1532.
- [15] Aune A. K., Cawley P. W., and Ekeland A., 1997, "Quadriceps muscle contraction protects the anterior cruciate ligament during anterior tibial translation," *Am J Sports Med*, 25(2), pp. 187–190.
- [16] Colby S., Francisco A., Yu B., Kirkendall D., Finch M., and Garrett W., 2000, "Electromyographic and kinematic analysis of cutting maneuvers. Implications for anterior cruciate ligament injury," *Am J Sports Med*, 28(2), pp. 234–240.
- [17] DeMorat G., Weinhold P., Blackburn T., Chudik S., and Garrett W., 2004, "Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury," *Am J Sports Med*, 32(2), pp. 477–483.
- [18] MacWilliams B. A., Wilson D. R., DesJardins J. D., Romero J., and Chao E. Y., 1999, "Hamstrings cocontraction reduces internal rotation, anterior translation, and anterior cruciate ligament load in weight-bearing flexion," *J. Orthop. Res*, 17(6), pp. 817–822.
- [19] More R. C., Karras B. T., Neiman R., Fritschy D., Woo S. L., and Daniel D. M., 1993, "Hamstrings—an anterior cruciate ligament protagonist. An in vitro study," *Am J Sports Med*, 21(2), pp. 231–237.
- [20] Torzilli P. A., Deng X., and Warren R. F., 1994, "The effect of joint-compressive load and quadriceps muscle force on knee motion in the intact and anterior cruciate ligament-sectioned knee," *Am J Sports Med*, 22(1), pp. 105–112.
- [21] Beynnon B. D., Bernstein I. M., Belisle A., Brattbakk B., Devanny P., Risinger R., and Durant D., 2005, "The effect of estradiol and progesterone on knee and ankle joint laxity," *Am J Sports Med*, 33(9), pp. 1298–1304.
- [22] Gillquist J., and Messner K., 1999, "Anterior cruciate ligament reconstruction and the long-term incidence of gonarthrosis," *Sports Med*, 27(3), pp. 143–156.
- [23] Costa-Paz M., Muscolo D. L., Ayerza M., Makino A., and Aponte-Tinao L., 2001, "Magnetic resonance imaging follow-up study of bone bruises associated with anterior cruciate ligament ruptures," *Arthroscopy*, 17(5), pp. 445–449.

- [24] Johnson D. L., Urban W. P., Caborn D. N., Vanarthos W. J., and Carlson C. S., 1998, "Articular cartilage changes seen with magnetic resonance imaging-detected bone bruises associated with acute anterior cruciate ligament rupture," *Am J Sports Med*, 26(3), pp. 409–414.
- [25] Mathis C. E., Noonan K., and Kayes K., 1998, "'Bone bruises' of the knee: a review," *Iowa Orthop J*, 18, pp. 112–117.
- [26] Murray M. M., 2009, "Current status and potential of primary ACL repair," *Clin Sports Med*, 28(1), pp. 51–61.
- [27] Wright R. W., Dunn W. R., Amendola A., Andrish J. T., Bergfeld J., Kaeding C. C., Marx R. G., McCarty E. C., Parker R. D., Wolcott M., Wolf B. R., and Spindler K. P., 2007, "Risk of tearing the intact anterior cruciate ligament in the contralateral knee and rupturing the anterior cruciate ligament graft during the first 2 years after anterior cruciate ligament reconstruction: a prospective MOON cohort study," *Am J Sports Med*, 35(7), pp. 1131–1134.
- [28] von Porat A., Roos E. M., and Roos H., 2004, "High prevalence of osteoarthritis 14 years after an anterior cruciate ligament tear in male soccer players: a study of radiographic and patient relevant outcomes," *Ann. Rheum. Dis*, 63(3), pp. 269–273.
- [29] Junkin D. M., Johnson D. J., Fu F. H., Mark M. D., Willenborn M., Fanelli G. C., and Wascher D. W., 2009, "Knee Ligament Injuries," *Orthopaedic Knowledge Update: Sports Medicine 4*, Amer Academy of Orthopaedic, pp. 135–153.
- [30] McNair P. J., Marshall R. N., and Matheson J. A., 1990, "Important features associated with acute anterior cruciate ligament injury," *N. Z. Med. J*, 103(901), pp. 537–539.
- [31] Griffin L. Y., Agel J., Albohm M. J., Arendt E. A., Dick R. W., Garrett W. E., Garrick J. G., Hewett T. E., Huston L., Ireland M. L., Johnson R. J., Kibler W. B., Lephart S., Lewis J. L., Lindenfeld T. N., Mandelbaum B. R., Marchak P., Teitz C. C., and Wojtyk E. M., 2000, "Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies," *J Am Acad Orthop Surg*, 8(3), pp. 141–150.
- [32] Arendt E., and Dick R., 1995, "Knee injury patterns among men and women in collegiate basketball and soccer. NCAA data and review of literature," *Am J Sports Med*, 23(6), pp. 694–701.
- [33] Gwinn D. E., Wilckens J. H., McDevitt E. R., Ross G., and Kao T. C., 2000, "The relative incidence of anterior cruciate ligament injury in men and women at the United States Naval Academy," *Am J Sports Med*, 28(1), pp. 98–102.
- [34] Genuario J. W., Faucett S. C., Boublik M., and Schlegel T. F., 2012, "A cost-effectiveness analysis comparing 3 anterior cruciate ligament graft types: bone-patellar

tendon-bone autograft, hamstring autograft, and allograft,” *Am J Sports Med*, 40(2), pp. 307–314.

[35] Lubowitz J. H., and Appleby D., 2011, “Cost-effectiveness analysis of the most common orthopaedic surgery procedures: knee arthroscopy and knee anterior cruciate ligament reconstruction,” *Arthroscopy*, 27(10), pp. 1317–1322.

[36] Cerulli G., Benoit D. L., Lamontagne M., Caraffa A., and Liti A., 2003, “In vivo anterior cruciate ligament strain behaviour during a rapid deceleration movement: case report,” *Knee Surg Sports Traumatol Arthrosc*, 11(5), pp. 307–311.

[37] Decker M. J., Torry M. R., Wyland D. J., Sterett W. I., and Richard Steadman J., 2003, “Gender differences in lower extremity kinematics, kinetics and energy absorption during landing,” *Clin Biomech (Bristol, Avon)*, 18(7), pp. 662–669.

[38] Huston L. J., Vibert B., Ashton-Miller J. A., and Wojtys E. M., 2001, “Gender differences in knee angle when landing from a drop-jump,” *Am J Knee Surg*, 14(4), pp. 215–219; discussion 219–220.

[39] Lephart S. M., Ferris C. M., Riemann B. L., Myers J. B., and Fu F. H., 2002, “Gender differences in strength and lower extremity kinematics during landing,” *Clin. Orthop. Relat. Res*, (401), pp. 162–169.

[40] Pollard C. D., Davis I. M., and Hamill J., 2004, “Influence of gender on hip and knee mechanics during a randomly cued cutting maneuver,” *Clin Biomech (Bristol, Avon)*, 19(10), pp. 1022–1031.

[41] Weinhold P. S., Stewart J.-D. N., Liu H.-Y., Lin C.-F., Garrett W. E., and Yu B., 2007, “The influence of gender-specific loading patterns of the stop-jump task on anterior cruciate ligament strain,” *Injury*, 38(8), pp. 973–978.

[42] Chappell J. D., Herman D. C., Knight B. S., Kirkendall D. T., Garrett W. E., and Yu B., 2005, “Effect of fatigue on knee kinetics and kinematics in stop-jump tasks,” *Am J Sports Med*, 33(7), pp. 1022–1029.

[43] Imwalle L. E., Myer G. D., Ford K. R., and Hewett T. E., 2009, “Relationship between hip and knee kinematics in athletic women during cutting maneuvers: a possible link to noncontact anterior cruciate ligament injury and prevention,” *J Strength Cond Res*, 23(8), pp. 2223–2230.

[44] Yu B., Herman D., Preston J., Lu W., Kirkendall D. T., and Garrett W. E., 2004, “Immediate effects of a knee brace with a constraint to knee extension on knee kinematics and ground reaction forces in a stop-jump task,” *Am J Sports Med*, 32(5), pp. 1136–1143.

[45] Ford K. R., Myer G. D., Smith R. L., Vianello R. M., Seiwert S. L., and Hewett T. E., 2006, “A comparison of dynamic coronal plane excursion between matched male

and female athletes when performing single leg landings,” *Clin Biomech* (Bristol, Avon), 21(1), pp. 33–40.

[46] McLean S. G., Lipfert S. W., and van den Bogert A. J., 2004, “Effect of gender and defensive opponent on the biomechanics of sidestep cutting,” *Med Sci Sports Exerc*, 36(6), pp. 1008–1016.

[47] Ford K. R., Myer G. D., Toms H. E., and Hewett T. E., 2005, “Gender differences in the kinematics of unanticipated cutting in young athletes,” *Med Sci Sports Exerc*, 37(1), pp. 124–129.

[48] Holden J. P., Orsini J. A., Siegel K. L., Kepple T. M., Gerber L. H., and Stanhope S. J., 1997, “Surface movement errors in shank kinematics and knee kinetics during gait,” *Gait & Posture*, 5(3), pp. 217–227.

[49] Reinschmidt C., van den Bogert A. J., Nigg B. M., Lundberg A., and Murphy N., 1997, “Effect of skin movement on the analysis of skeletal knee joint motion during running,” *J Biomech*, 30(7), pp. 729–732.

[50] Alexander E. J., and Andriacchi T. P., 2001, “Correcting for deformation in skin-based marker systems,” *J Biomech*, 34(3), pp. 355–361.

[51] Fleming B. C., Beynnon B. D., Renstrom P. A., Johnson R. J., Nichols C. E., Peura G. D., and Uh B. S., 1999, “The strain behavior of the anterior cruciate ligament during stair climbing: an in vivo study,” *Arthroscopy*, 15(2), pp. 185–191.

[52] Shepherd D. E., and Seedhom B. B., 1999, “Thickness of human articular cartilage in joints of the lower limb,” *Ann. Rheum. Dis*, 58(1), pp. 27–34.

[53] Eberhart H. D., and Inman V. T., 1951, “An evaluation of experimental procedures used in a fundamental study of human locomotion,” *Ann. N. Y. Acad. Sci*, 51(7), pp. 1213–1228.

[54] Lafortune M. A., Cavanagh P. R., Sommer H. J., and Kalenak A., 1994, “Foot inversion-eversion and knee kinematics during walking,” *J. Orthop. Res*, 12(3), pp. 412–420.

[55] Neptune R. R., and Hull M. L., 1995, “Accuracy assessment of methods for determining hip movement in seated cycling,” *J Biomech*, 28(4), pp. 423–437.

[56] Tapper J. E., Funakoshi Y., Hariu M., Marchuk L., Thornton G. M., Ronsky J. L., Zernicke R., Shrive N. G., and Frank C. B., 2009, “ACL/MCL transection affects knee ligament insertion distance of healing and intact ligaments during gait in the Ovine model,” *J Biomech*, 42(12), pp. 1825–1833.

[57] Moore D. C., Crisco J. J., Trafton T. G., and Leventhal E. L., 2007, “A digital database of wrist bone anatomy and carpal kinematics,” *J Biomech*, 40(11), pp. 2537–2542.

- [58] Sheehan F. T., Zajac F. E., and Drace J. E., 1998, “Using cine phase contrast magnetic resonance imaging to non-invasively study in vivo knee dynamics,” *J Biomech*, 31(1), pp. 21–26.
- [59] Wolfe S. W., Neu C., and Crisco J. J., 2000, “In vivo scaphoid, lunate, and capitate kinematics in flexion and in extension,” *J Hand Surg Am*, 25(5), pp. 860–869.
- [60] Banks S. A., and Hodge W. A., 1996, “Accurate measurement of three-dimensional knee replacement kinematics using single-plane fluoroscopy,” *IEEE Trans Biomed Eng*, 43(6), pp. 638–649.
- [61] Tashman S., 2008, “Comments on ‘validation of a non-invasive fluoroscopic imaging technique for the measurement of dynamic knee joint motion’,” *J Biomech*, 41(15), pp. 3290–3291: author reply 3292–3293.
- [62] Bey M. J., Peltz C. D., Ciarelli K., Kline S. K., Divine G. W., van Holsbeeck M., Muh S., Kolowich P. A., Lock T. R., and Moutzouros V., 2011, “In vivo shoulder function after surgical repair of a torn rotator cuff: glenohumeral joint mechanics, shoulder strength, clinical outcomes, and their interaction,” *Am J Sports Med*, 39(10), pp. 2117–2129.
- [63] Tashman S., and Anderst W., 2003, “In-vivo measurement of dynamic joint motion using high speed biplane radiography and CT: application to canine ACL deficiency,” *J Biomech Eng*, 125(2), pp. 238–245.
- [64] DeFrate L. E., Gill T. J., and Li G., 2004, “In vivo function of the posterior cruciate ligament during weightbearing knee flexion,” *Am J Sports Med*, 32(8), pp. 1923–1928.
- [65] You B. M., Siy P., Anderst W., and Tashman S., 2001, “In vivo measurement of 3-D skeletal kinematics from sequences of biplane radiographs: application to knee kinematics,” *IEEE Trans Med Imaging*, 20(6), pp. 514–525.
- [66] Abebe E. S., Utturkar G. M., Taylor D. C., Spritzer C. E., Kim J. P., Moorman C. T. 3rd, Garrett W. E., and DeFrate L. E., 2011, “The effects of femoral graft placement on in vivo knee kinematics after anterior cruciate ligament reconstruction,” *J Biomech*, 44(5), pp. 924–929.
- [67] Li G., Van de Velde S. K., and Bingham J. T., 2008, “Validation of a non-invasive fluoroscopic imaging technique for the measurement of dynamic knee joint motion,” *J Biomech*, 41(7), pp. 1616–1622.
- [68] Bey M. J., Kline S. K., Tashman S., and Zauel R., 2008, “Accuracy of biplane x-ray imaging combined with model-based tracking for measuring in-vivo patellofemoral joint motion,” *J Orthop Surg Res*, 3, p. 38.

- [69] Anderst W., Zauel R., Bishop J., Demps E., and Tashman S., 2009, “Validation of three-dimensional model-based tibio-femoral tracking during running,” *Med Eng Phys*, 31(1), pp. 10–16.
- [70] Bey M. J., Zauel R., Brock S. K., and Tashman S., 2006, “Validation of a new model-based tracking technique for measuring three-dimensional, in vivo glenohumeral joint kinematics,” *J Biomech Eng*, 128(4), pp. 604–609.

Chapter 2

Static and dynamic error of a biplanar videoradiography system using marker-based and markerless tracking techniques

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Abstract

The use of biplanar videoradiography technology has become increasingly popular for evaluating joint function *in vivo*. Two fundamentally different methods are currently employed to reconstruct 3-D bone motions captured using this technology. Marker-based tracking requires at least three radio-opaque markers to be implanted in the bone of interest. Markerless tracking makes use of algorithms designed to match 3-D bone shapes to biplanar videoradiography data. In order to reliably quantify *in vivo* bone motion, the systematic error of these tracking techniques should be evaluated. Herein, we present new markerless tracking software that makes use of modern GPU technology, describe a versatile method for quantifying the systematic error of a biplanar videoradiography motion capture system using independent gold standard instrumentation, and evaluate the systematic error of the W.M. Keck XROMM Facility’s biplanar videoradiography system using both marker-based and markerless tracking algorithms under static and dynamic motion conditions. A polycarbonate flag embedded with 12 radio-opaque markers was used to evaluate the systematic error of the marker-based tracking algorithm. Three human cadaveric bones (distal femur, distal radius, and distal ulna) were used to evaluate the systematic error of the markerless tracking algorithm. The systematic error was evaluated by comparing motions to independent gold standard instrumentation. Static motions were compared to high accuracy linear and rotary stages while dynamic motions were compared to a high accuracy angular displacement transducer. Marker-based tracking was shown to effectively track motion to within 0.1 mm and 0.1° under static and dynamic conditions. Furthermore, the presented results indicate that markerless tracking can be used to effectively track rapid bone motions to within 0.15° for the distal aspects of the femur, radius, and ulna. Both marker-based and markerless tracking techniques were in excellent agreement with the gold standard instrumentation for both static

and dynamic testing protocols. Future research will employ these techniques to quantify *in vivo* joint motion for high-speed upper and lower extremity impacts such as jumping, landing, and hammering.

Keywords: *biplanar, X-ray, videoradiography, motion capture, systematic error, validation, accuracy, skeletal, biomechanics, digitally reconstructed radiograph, marker-based tracking, markerless tracking, software, hardware*

2.1 Introduction

Biplanar videoradiography systems that directly measure three-dimensional (3-D) *in vivo* skeletal motion have been developed in part to address the inherent limitations of optical motion capture systems that utilize skin-based marker sets [1–9]. Current skeletal tracking algorithms are classified as marker-based tracking [4,5,7,10,11] or markerless tracking [1,3,6,8,9,12]. Marker-based tracking methods require implantation of at least three radio-opaque spherical markers within each bone. The invasiveness of this technique severely limits its applicability for studying *in vivo* human joint motion. Markerless tracking makes use of algorithms designed to match 3-D bone shapes to biplanar videoradiography data. Employing this technology to obtain quantitative data on human joint motion first requires an understanding of its systematic error, particularly in the dynamic setting.

The systematic error of specific biplanar videoradiography systems has been previously documented using markerless tracking algorithms [13–18]. These studies provide valuable information on existing biplanar videoradiography technologies and stress the importance of establishing error measures for specific joints and system configurations. However, these studies typically involve protocols where cadaveric bones undergo static, semi-static, or gravitational based pendulum-like motion that may not simulate high-speed lower and upper extremity impacts associated with activities like jumping, landing, or hammering. Additionally, previous methods for evaluating systematic error have compared markerless tracking to marker-based tracking using the same data set (intra-specimen). While this provides a convenient and acceptable measure of markerless tracking error, it is not ideal for independently assessing the overall systematic error of a biplanar videoradiography system for both marker-based and markerless tracking techniques.

Work from You et al. [13] and Bey et al. [15] has made significant contributions

to 3-D skeletal motion capture technology by describing and implementing robust markerless tracking algorithms. However, these algorithms require time consuming data processing [13] or expensive computational clusters [3]. Recent advancements in graphics processing unit (GPU) technology make it an ideal candidate for processing computer vision algorithms such as those developed by You et al. [13] and Bey et al. [15]. These markerless tracking algorithms can be efficiently implemented on the GPU of a single workstation, substantially reducing computational time [19] and equipment expenses.

The goals of this study are to: (1) provide an outline of new markerless tracking software that makes use of modern GPU technology; (2) describe a versatile method for quantifying the systematic error of a biplanar videoradiography system using independent gold standard instrumentation; and (3) evaluate the systematic error of a biplanar videoradiography system using both marker-based and markerless tracking algorithms under static and dynamic motion conditions.

2.2 Methods

2.2.1 Description of XROMM Facilities and Resources

2.2.1.1 Hardware

The biplanar videoradiography system in the W. M. Keck Foundation XROMM Facility at Brown University (Providence, RI, USA) consists of two Varian Medical Systems model G-1086 X-ray tubes (Palo Alto, CA, USA), two EMD Technologies model EPS 45-80 pulsed X-ray generators (Saint-Eustache, Quebec, Canada), two 16 inch Dunlee (Aurora, IL, USA) model TH9447QXH590 image intensifiers (IIs), and

two Phantom v10 high-speed digital video cameras (Vision Research; Wayne, NJ, USA). The X-ray tubes are suspended from the ceiling by overhead tube cranes, and the IIs are mounted on mobile gantries (Figure 2.1). The system can deliver pulsed X-ray generation up to 150 Hz and can record in continuous X-ray generation at up to 1,000 frames-per-second (FPS). The system’s pixel resolution is 1800x1800, and the overall resolution of the imaging chain is approximately 2 line pairs/mm.

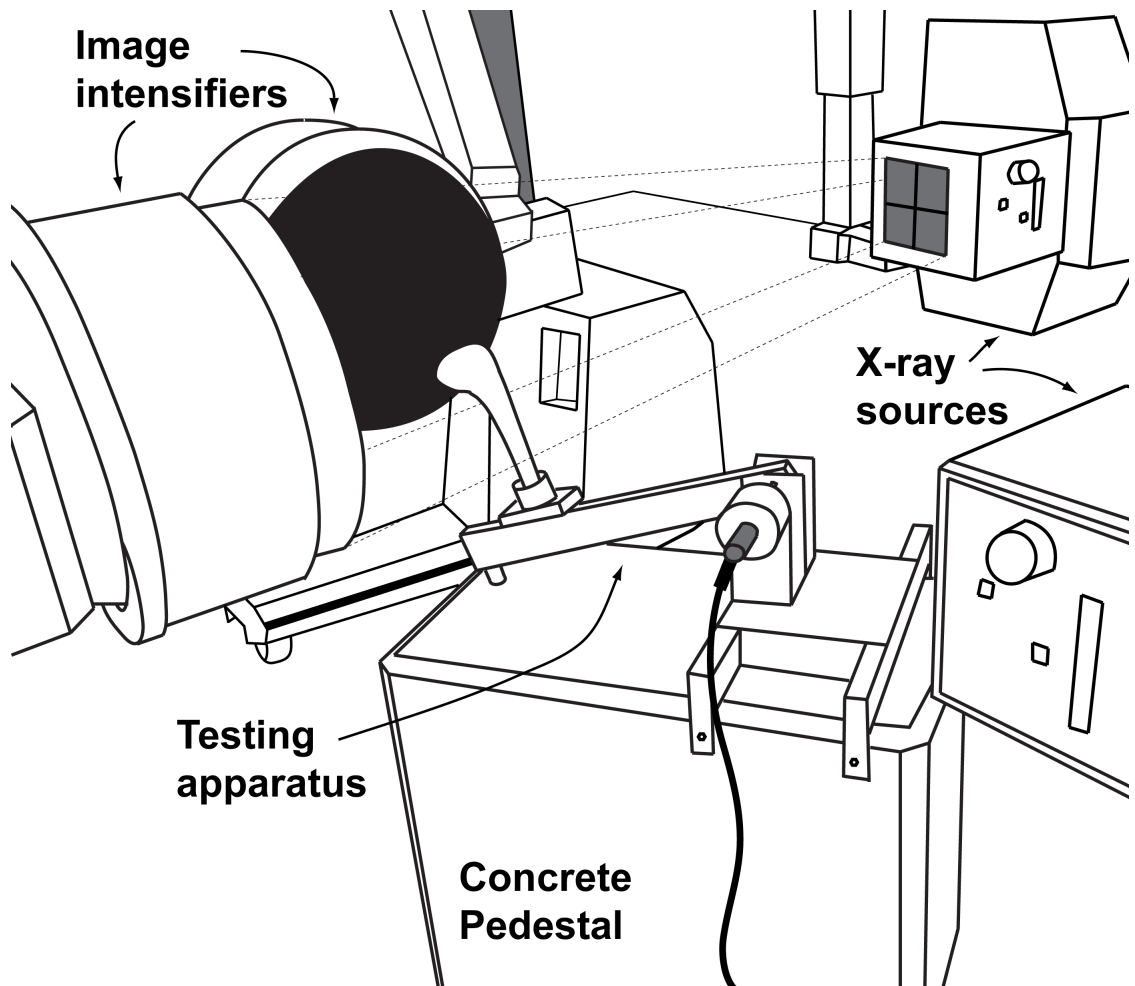


Figure 2.1: Illustrated representation of the experimental testing environment within the W. M. Keck Foundation XROMM Facility. The dynamic testing apparatus is positioned within the field of view determined by the overlapping X-ray beams. A representative X-ray beam is illustrated with dotted lines projecting from one of the X-ray sources.

Image de-distortion and 3-D space calibration for the biplanar XROMM facility have been described in detail previously and are available to the public [10]. Briefly, im-

age distortion is addressed by imaging a patterned sheet of perforated metal (Part 9255T641; McMaster-Carr, Robinson, NJ, USA) and using a local weighted mean (LWM) distortion correction algorithm implemented in MATLAB (XrayProject; Brown University, Providence, RI, USA). To calibrate the 3-D space, an acrylic calibration cube containing 64 steel beads is imaged. The markers are tracked in both planes using custom MATLAB software (XrayProject), and the calibration parameters are calculated using standard direct linear transform (DLT) techniques [20].

2.2.1.2 Software

X-ray reconstruction of moving morphology (XROMM) is a computational process that combines motion data from X-ray video and shape data from 3-D CT-based bone scans [10]. There are two fundamentally different ways to combine these data, marker-based XROMM and markerless XROMM.

Marker-based XROMM, also referred to as Dynamic Radiostereophotogrammetric Analysis [5], requires marker sets of three or more radio-opaque beads to be implanted into each rigid body. The 3-D positions of the radio-opaque beads are reconstructed from the biplanar videoradiography data and then used to calculate frame-by-frame motion for each marker set and, therefore, the respective rigid body in which they are implanted. The development of marker-based XROMM has been built on previous work on canines [4,21] and humans [22]. Custom MATLAB software (XrayProject) has been developed to process all marker-based XROMM data using standard DLT techniques [20,23], as previously described by Brainerd et al. [10].

Markerless XROMM can be performed by auto-registration of a CT volume to biplanar videoradiography data. The following description of the auto-registration algorithm is built on previous work described by You et al. [13] and Bey et al. [15]. The

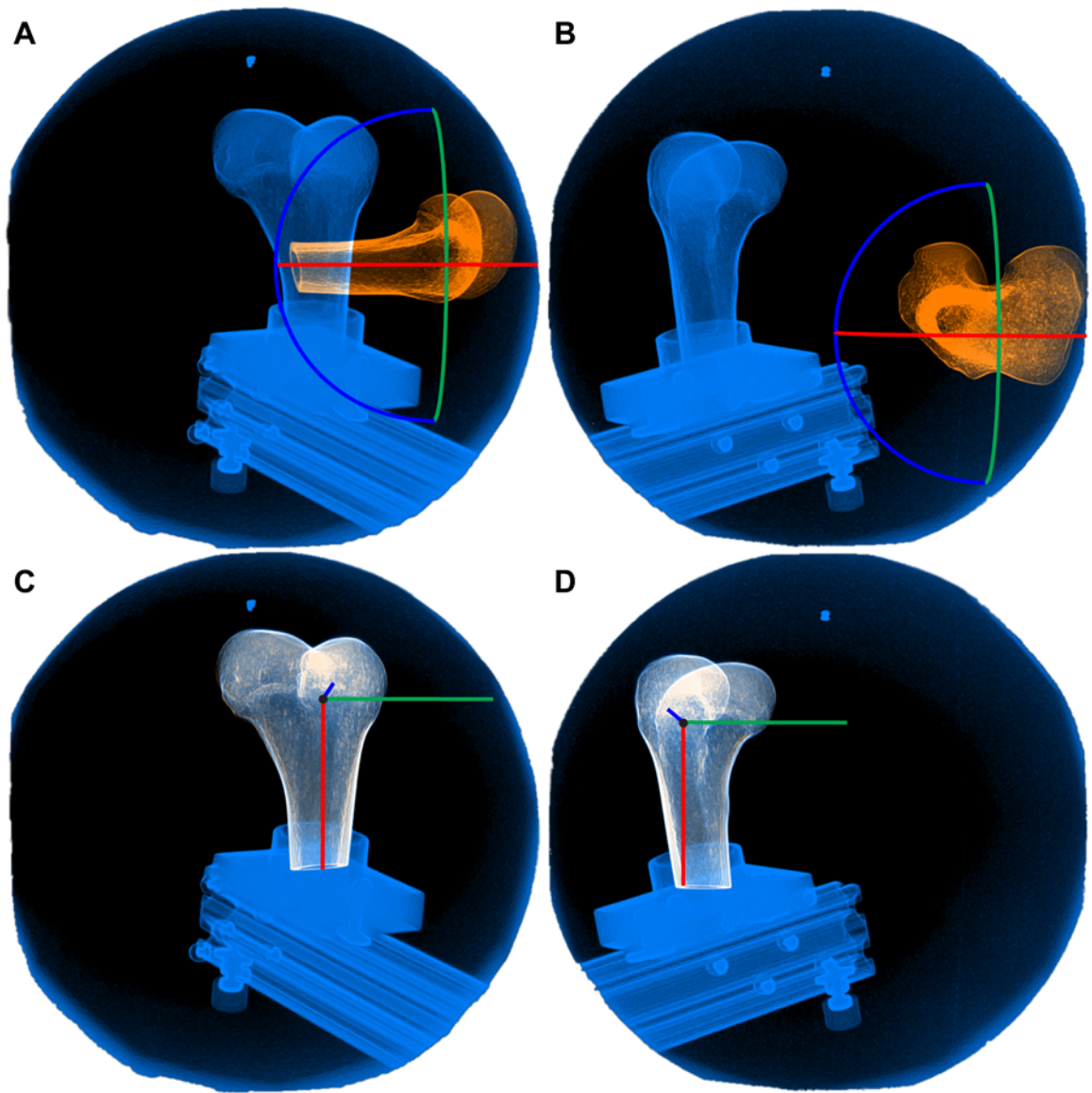


Figure 2.2: XROMM Autoscooper 3-D software environment. This figure illustrates the before (A and B) and after (C and D) results obtained from the auto-registration algorithm using an initial guess that was extrapolated from the previous frames. Additionally, the constrained-axis rotation and translation manipulators are shown in A and B and C and D, respectively. The Autoscooper software (executable and source) is publically available.

auto-registration algorithm consists of four major components. First, a volume visualization technique is used to generate digitally reconstructed radiographs (DRRs) from a CT volume using a standard ray casting approach and the known orientation of the biplanar videoradiography system [24–26]. Second, both the radiographs and DRRs are processed to enhance features and detect edges [27]. Third, a normalized cross correlation is used to measure the similarity between the radiographs and DRRs [28]. Finally, a downhill simplex optimization algorithm iterates over the six-degree-of-freedom (DOF) motion parameters until the desired correlation has been reached [29]. The entire auto-registration algorithm requires an initial guess of bone position and orientation. This can be manually assigned by a user or extrapolated from previously tracked frames.

There are several user defined parameters that control the creation of the DRRs. These include the sampling frequency of the ray casting, the intensity of the virtual X-rays, and a threshold density, which determines if a particular sample will contribute to the accumulated value along each ray. Once the DRRs have been generated, there are additional filters available, such as contrast enhancement and Sobel edge detection. Contrast and edge detection filters can also be applied to the biplanar videoradiography sequence. These settings can enhance the algorithm’s ability to find a solution. For the purposes of this study, these parameters were selected by the user to provide the best visual match between the DRR and the biplanar videoradiography sequence (Figure 2.2).

Once the user has fixed the above parameters, the objective function takes in the x-y-z position and x-y-z orientation of the CT volume as free parameters. It outputs a single scalar value representing the normalized cross correlation between the grayscale values of the DRR and radiograph. The correlation equation, adapted from You et al. [13], used in this study is

$$C_i(\hat{p}) = \frac{\sum_{x,y} [r_i(x, y) - \overline{r_i(x, y)}] [d_i(x, y, \hat{p}) - \overline{d_i(x, y, \hat{p})}]}{\left\{ \sum_{x,y} [r_i(x, y) - \overline{r_i(x, y)}]^2 \sum_{x,y} [d_i(x, y, \hat{p}) - \overline{d_i(x, y, \hat{p})}]^2 \right\}^{1/2}}$$

where

C_i is the correlation between the DRR and radiograph from camera i ;

p is the vector containing the six-DOF motion parameters of the CT bone model: xyz-position and xyz-rotation;

$r_i(x, y)$ is the radiograph from camera i after filtering with edges added;

$\overline{r_i(x, y)}$ is the mean of the radiography from camera i after filtering with edges added;

$d_i(x, y, p)$ is the DRR from camera i generated from the CT bone model in position p after filtering with edges added;

and $\overline{d_i(x, y, p)}$ is the mean of the DRR from camera i generated from the CT bone model in position p .

The final similarity measure for a biplanar videoradiography system is the product of the correlations from each camera,

$$C(\hat{p}) = C_1(\hat{p})C_2(\hat{p})$$

The optimization routine searches the six-DOF parameter space for the position and orientation values that maximize this correlation.

The solution space for the algorithm is noisy and has many local minima. To avoid local minima, the simplex is reset and run multiple times using the result of the

previous optimization as the initial guess for the next. In general, this technique will not converge if the initial guess is too far away; however, the user supplied initial guesses are generally close enough for quick convergence. Within dynamic data sets, initial guesses extrapolated from previous frames also converge quickly due to the high video frame rate and good frame coherence.

The described auto-registration algorithm for extraction of skeletal kinematics from markerless XROMM data has been implemented in an open source software package that is available to the public (Autoscooper; Brown University, Providence, RI, USA). The software provides a graphical interface for markerless bone tracking over a video sequence (Figure 2.2). The images displayed in Figures 2.2A and 2.2B show the DRRs and video sequences for the two X-ray sources before auto-registration. The DRR is displayed in orange and the videoradiography sequences are displayed in blue. The images displayed in Figures 2.2C and 2.2D show the DRR and biplanar videoradiography sequence for the two X-ray sources after auto-registration is completed. The overlay of the DRR with the videoradiography sequence is represented in white. In addition to implementing the auto-registration algorithm, the software allows the user to rotate and translate the 3-D CT volume to best match the DRRs to the biplanar videoradiography sequence. This can be done by using the constrained-axis rotation and translation manipulators shown in Figures 2.2A and 2.2B and 2.2C and 2.2D, respectively. All six position and orientation parameters can be plotted within the software to assist the user in identifying registration inconsistencies between video frames.

The implementation of the auto-registration algorithm makes significant use of general purpose computation on a GPU. Highly parallelized GPU implementations of this algorithm have shown significant speed improvements over the traditional central processing unit (CPU) implementations [19]. Specifically, the DRR generation, image

processing, and similarity measure all take place on a single GPU with NVIDIA’s Fermi architecture (GeForce GTX 480; EVGA, Brea, CA, USA). To generate the DRR, each ray was processed independently by a separate thread. The image processing was done in a similar manner, with each pixel processed independently by a separate thread. The normalized cross correlation similarity measure was also parallelized using a tree-based parallel sum, and the resulting output is read off the GPU. It was not necessary to parallelize the downhill simplex optimization algorithm as the bottleneck exists in DRR generation, image processing, and similarity measure calculation. As a result, the described auto-registration algorithm converges in less than one second for a single frame on a GPU equipped workstation (Precision T7400; Dell, Round Rock, TX, USA).

2.2.2 Systematic Error Testing Protocols

2.2.2.1 Static Testing Protocol

Static error was evaluated by translating and rotating a set of human cadaver bones and a polycarbonate marker flag by known increments with high precision linear (NB4 Series; Newmark Systems, Mission Viejo, CA, USA) and rotary (RT-3 Series; Newmark Systems) positioning stages with accuracies of 0.001 mm and 0.002°, respectively (Figure 2.3A). The bones used in this study include a distal femur, distal radius, and distal ulna (Figure 2.4). The bones were manually stripped of soft tissue, cleaned using tergazyme in a hot water bath, disinfected with hydrogen peroxide, and dried at room temperature. The proximal end of each bone was rigidly fixed in a polyvinyl chloride pot using urethane resin (Smooth-Cast 300Q; Smooth-On, Easton, PA, USA). The marker flag was created by embedding 12 one millimeter diameter tantalum markers into two cylindrical polycarbonate posts (Figure 2.3B).

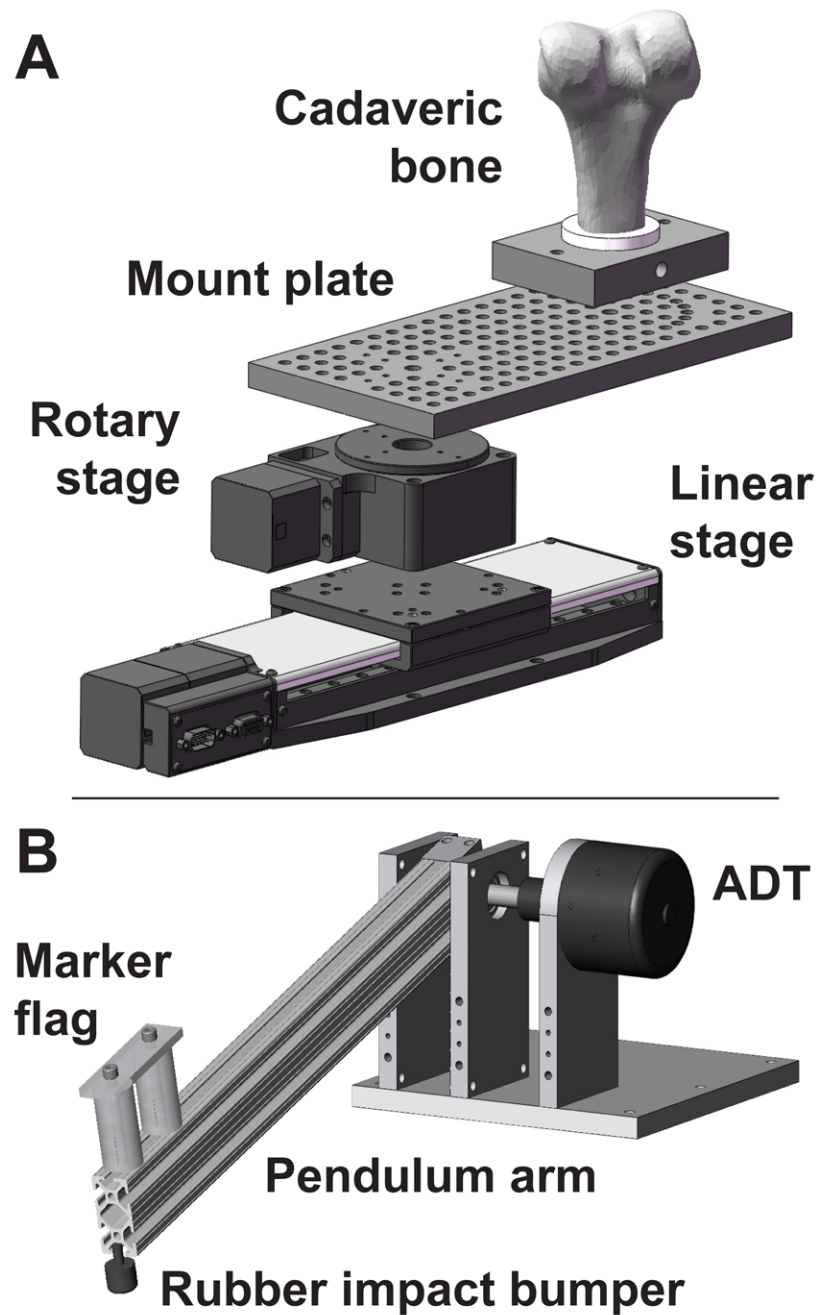


Figure 2.3: Static (A) and dynamic (B) testing apparatus. Both apparatuses were rigidly fixed to a concrete pedestal (Figure 2.1) for all static and dynamic testing.

For each bone and the marker flag, twenty trials of 15 translational motion steps (0.000, 0.001, 0.010, 0.100, 1.000, 10.000, 10.100, 25.000, 25.100, 50.000, 50.100, 75.000, 75.100, 100.000, and 100.100 mm) and 15 rotational motion steps (0.000, 0.002, 0.010, 0.100, 1.000, 10.000, 10.100, 25.000, 25.100, 50.000, 50.100, 75.000, 75.100, 100.000, and 100.100°) were performed using the static testing apparatus shown in Figure 2.3A. An average reference position for the translational and rotational tests were determined from an additional twenty trials at 0.000 mm and 0.000°, respectively. All marker tracking was performed using the marker-based software discussed earlier.

Translational and rotational errors were determined as the difference between the computed rigid body translation or rotation and the true linear or rotary stage value.

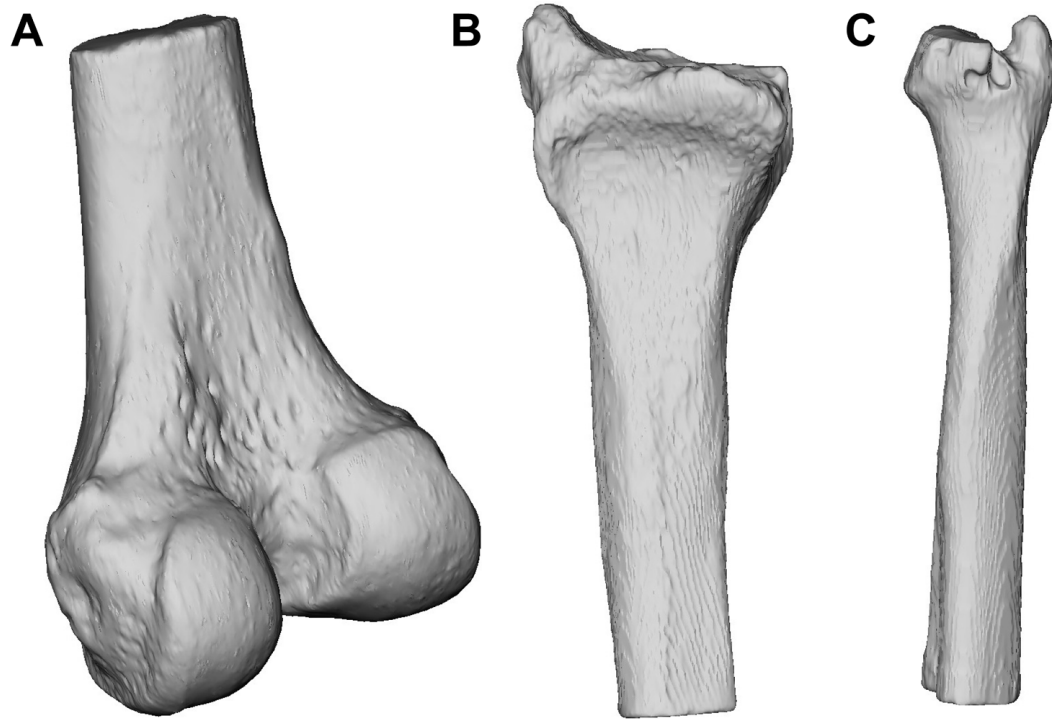


Figure 2.4: Images displaying the morphology of the three bones used in this study. Panels A, B, and C are the 3-D CT models of the distal femur, distal radius, and distal ulna, respectively.

2.2.2.2 Dynamic Testing Protocol

Dynamic error was evaluated using the same bones and the marker flag employed in the static testing protocol. A specifically designed impact pendulum was fabricated (Figure 2.3B). A computer-aided design of the impact pendulum is publically available. The fulcrum of the pendulum was attached to a high precision ($\pm 0.06^\circ$) angular displacement transducer (ADT: Series 600; Trans-Tek Inc., Ellington, CT, USA). This allowed the pendulum arm to spin along the same axis as the mechanical axis of the ADT. Angular displacement transducer data were collected at 5,000 Hz and synchronized with the biplanar videoradiography system.

For each bone and the marker flag, five pendulum drop-impact trials were performed using the apparatus shown in Figure 2.3B. For each trial, the pendulum arm was dropped from a position held outside the field of view (FOV) of the XROMM system. As the arm fell, it entered the FOV and impacted a concrete pedestal. The motion of the arm and the attached bone or marker flag was recorded for all pendulum impacts until a stationary, steady state was achieved. Additionally, an average reference position for each bone and the marker flag were determined by collecting a stationary trial at the impact position. All markerless tracking was performed using the auto-registration software discussed earlier.

Rotational error was determined for each bone and the marker flag by calculating the difference between the computed rigid body rotation and true ADT value.

2.2.3 Imaging Parameters

The X-ray tube voltage and current were set at 70 kVp and 200 mA, and the source to image distance (SID) was set to approximately 140 cm for each testing protocol. For

all static testing, the biplanar videoradiography system recorded in pulsed (4 ms) X-ray generation mode at 60 fps. During dynamic testing, the biplanar videoradiography system recorded in continuous X-ray generation mode at 250 fps. To optimize image quality and eliminate motion blur (dynamic protocol), each high-speed video camera was shuttered between 1/1,300 and 1/2,000 s depending on the bone or marker flag being imaged.

Clinical CT scans for each bone were acquired in the axial plane (Lightspeed; GE, Piscataway, NJ, USA) at 80 kVp, using GE's SMART mA and Bone Plus reconstruction algorithm. The table speed and pitch were set at 0.562° and 5.62 mm/rotation for each scan. The image volumes contained 243 image slices for the distal femur and 218 image slices for the distal radius and distal ulna. In-plane image resolution was set at 512x512 pixels for a voxel size of 0.217x0.217x0.625 mm³. Each bone was isolated from its entire volume using thresholding and segmentation tools implemented in commercially available software (Mimics 14; Materialise, Ann Arbor, MI, USA). These procedures follow well established techniques [30,31].

2.2.4 Data Analysis

All analyses were performed on unfiltered data. Helical axis of motion (HAM) rotation and translation variables were used to describe all rigid body kinematics obtained from the XROMM system. This permits a direct comparison of values without the need to define a common identical coordinate system in both the XROMM and testing apparatuses space. Systematic error, defined as the difference between the measured rigid body motion (XROMM) and the true value of the parameter being measured (stages or ADT), was determined for every static and dynamic data point. These data are summarized using sample median, 25–75 percentile, and range statistics.

Absolute error data are summarized using sample mean and standard deviation (SD) statistics.

The distributions of angular velocities and angular accelerations were determined for the marker flag and each bone in order to highlight the motions being imaged during the dynamic protocol. Angular velocity was defined as the discrete derivative of the measured position data. Angular acceleration was defined as the discrete second derivative of the measured position data. Correlations between error, angular velocity, and angular acceleration were determined for all dynamic trials using standard linear regression techniques.

2.3 Results

The distributions of error values (minimum, 25th percentile, median, 75th percentile, and maximum) for the static rotational (Figure 2.5A) and translational (Figure 2.5B) movements were lower and more tightly clustered for the marker flag than for each bone tracked using the described markerless tracking software. For marker-based tracking of the marker flag, the mean static rotational and translational absolute errors were estimated to be $0.09 \pm 0.08^\circ$ (Figure 2.5C) and 0.12 ± 0.08 mm (Figure 2.5D), respectively. For markerless tracking, the mean static rotational absolute errors were $0.30 \pm 0.18^\circ$, $0.39 \pm 0.18^\circ$, and $0.44 \pm 0.26^\circ$ for the distal femur, distal radius, and distal ulna, respectively (Figure 2.5C). The markerless tracking mean static translational absolute errors were 0.25 ± 0.16 mm, 0.33 ± 0.27 mm, and 0.30 ± 0.30 mm for the distal femur, distal radius, and distal ulna, respectively (Figure 2.5D).

For the dynamic protocol, the minimum, 25th percentile, median, 75th percentile, and maximum angular velocity was 0.0, 47.2, 95.8, 172.7, and 6.2×10^3 degrees/s,

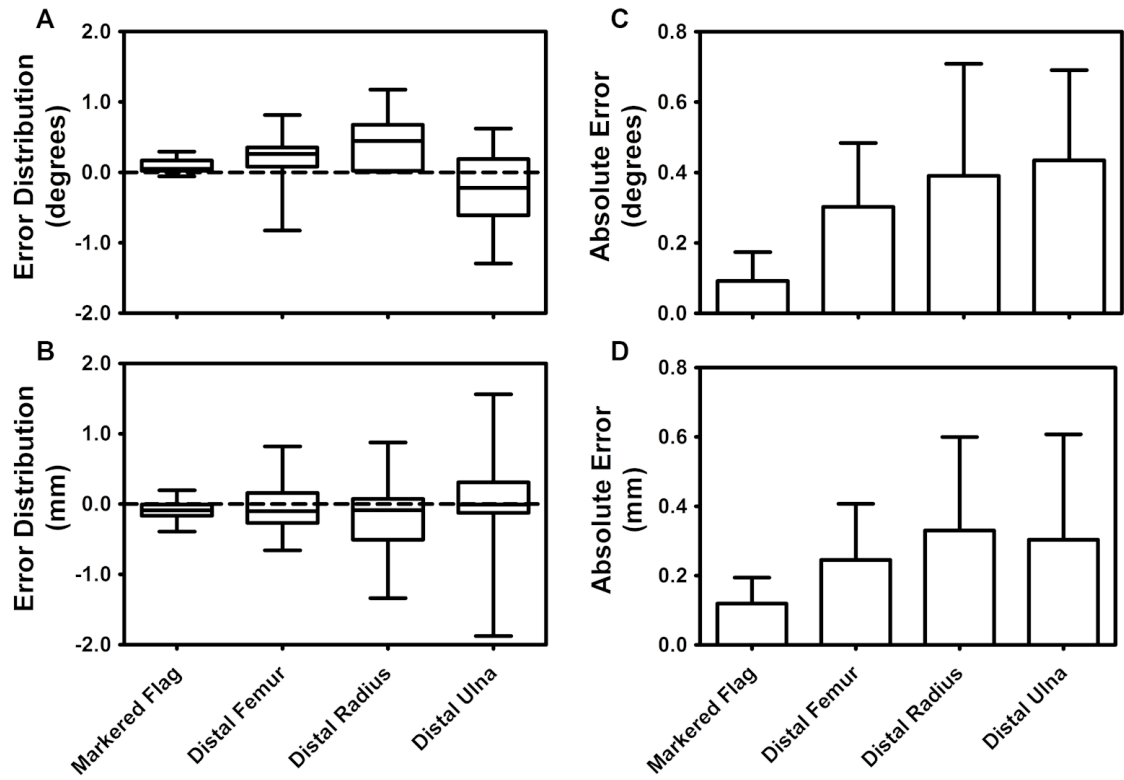


Figure 2.5: Static error results. Box and whiskers rotational (A) and translational (B) plot displaying range, 25–75 percentile, and median static error for each specimen. Mean (+ SD) rotational (C) and translational (D) absolute static error for each specimen.

respectively (Figures 2.6A and 2.6B). The minimum, 25th percentile, median, 75th percentile, and maximum angular accelerations was 0.0, 1.3×10^3 , 2.9×10^3 , 5.1×10^3 , and 1.6×10^6 degrees/s [2], respectively (Figures 2.6C and 2.6D). Approximately 50% of the velocities and accelerations were above 95 degrees/s and 2,900 degrees/s [2], respectively.

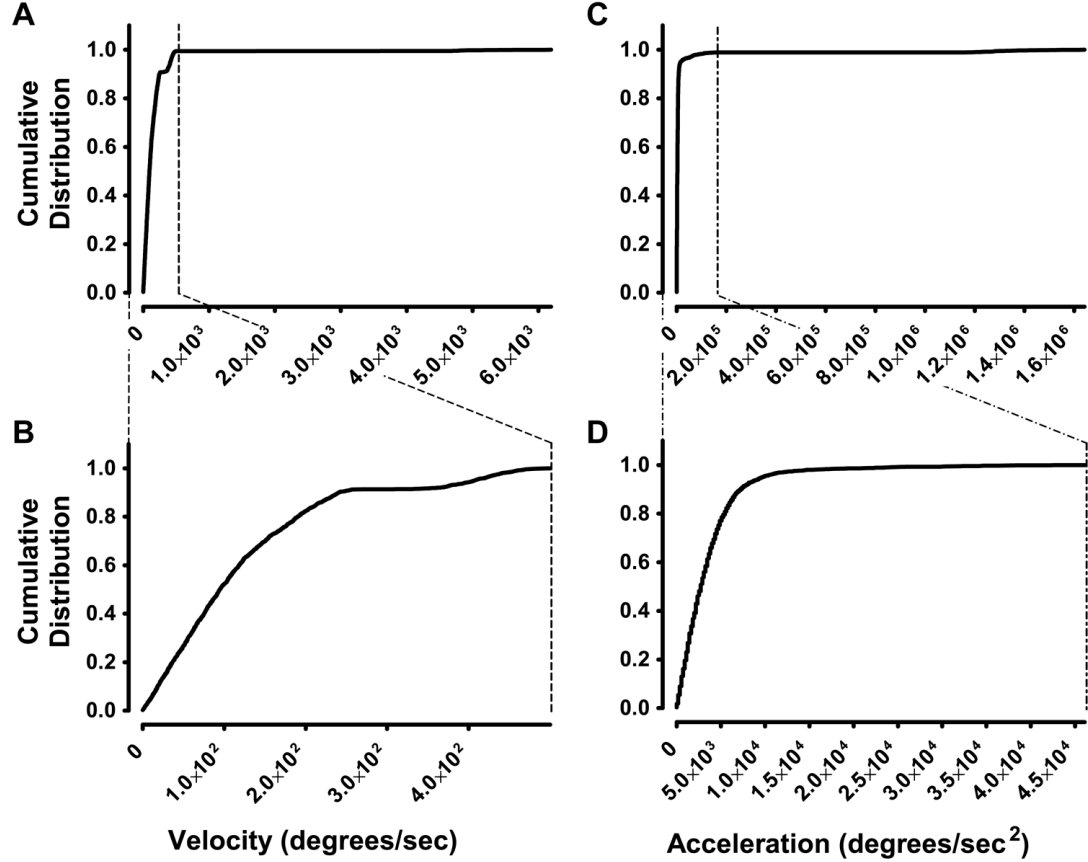


Figure 2.6: Cumulative distributions of all velocities and accelerations tested during the dynamic protocols. The full cumulative distributions of velocities and accelerations are shown in A and C, respectively. The majority of velocities and accelerations are shown in B and D, respectively. These data are taken from the data windowed by the vertical dotted lines present in A and C.

The distribution of dynamic rotational error values (minimum, 25th percentile, median, 75th percentile, and maximum) were similar for the marker flag and each bone tracked using the described markerless tracking software (Figure 2.7A). However, the total range of error values was lower for the marker flag. For marker-based tracking,

the mean dynamic absolute error was estimated to be $0.10 \pm 0.06^\circ$ (Figure 2.7B). For markerless tracking, the mean dynamic absolute error was estimated to be $0.14 \pm 0.09^\circ$, $0.10 \pm 0.09^\circ$, $0.14 \pm 0.12^\circ$ for the distal femur, distal radius, and distal ulna, respectively (Figure 2.7B). In addition, angular velocity and acceleration were found to be very poor predictors of systematic error for the marker flag and all bones ($R^2 \leq 0.01$). Furthermore, the change in error for a unit change in either angular velocity or acceleration was below 0.036 for the marker flag and all bones.

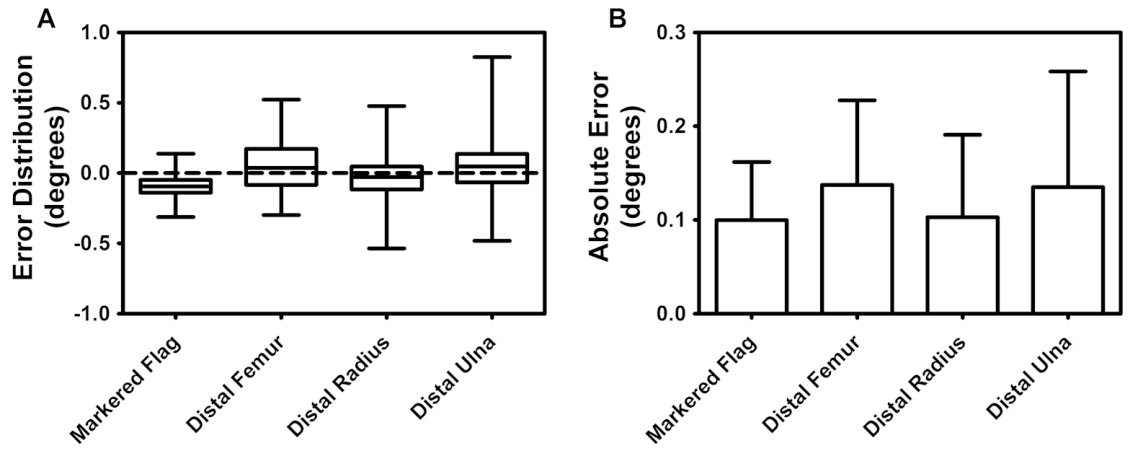


Figure 2.7: Dynamic error results. (A) Box and whiskers plot displaying range, 25–75 percentile, and median dynamic error for each specimen. (B) Mean (+1 SD) absolute dynamic error for each specimen.

2.4 Discussion

The goals addressed by this study were threefold. First, an outline of new markerless tracking software that makes significant use of modern GPU technology has been provided. Second, a versatile method for quantifying the systematic error of a biplanar videoradiography system using independent gold standard instrumentation was described. Finally, these methods were applied to evaluate the systematic error of a biplanar videoradiography system using marker-based and markerless tracking

algorithms under static and dynamic motion conditions.

The results presented in this article indicate that the biplanar videoradiography hardware and Autoscooper software described here are capable of measuring sub-millimeter and sub-degree bone motion under the given testing conditions. The results were obtained using independent instrumentation as a gold standard, rather than an intra-specimen marker-based comparison. This provides two advantages: first, the systematic error of both marker-based and markerless tracking can be assessed independently; and second, eliminating the need to remove all marker signatures from the biplanar videoradiography data saves considerable preprocessing time and effort. The presented results are consistent with similar studies investigating errors associated with markerless bone tracking. A study by Bey et al. reports markerless tracking of the glenohumeral joint [15] to be within 0.5 mm of marker-based tracking. In another study, Bey et al. reports markerless tracking of the patellofemoral joint [16] to be within 0.455 mm and 0.987° of marker-based tracking. A study from Martin et al. [3] reports markerless tracking errors on a separate system to be within 0.25 mm of marker-based tracking for the distal femur and pelvis. Additional studies have shown similar markerless tracking results for the knee [6,18].

As anticipated, the systematic error of the biplanar videoradiography hardware and Autoscooper software system fell below 1 mm and 1° for both static and dynamic marker-based and markerless tracking. It was expected that the marker-based tracking would be equivalent for both the static and dynamic protocols because the W. M. Keck XROMM Facility is equipped with digital video cameras capable of capturing at high shutter speeds that effectively stop the motion at a given frame. Additionally, because spherical markers were being tracked, changes in marker flag orientation and location throughout the imaging FOV were not expected to affect tracking. The absolute error for the marker-based static and dynamic error were not statistically

significant ($p = 0.10$) from each other.

Conversely, a surprising finding was the larger error observed for the static markerless tracking protocol compared to the dynamic markerless tracking protocol. The dynamic markerless tracking protocol showed lower error values for all bones tested ($p < 0.01$ for the distal femur, distal radius, and distal ulna). These findings may be associated with the initial guess required as input to the markerless tracking algorithm. During static testing, each frame is tracked independent of the previous and succeeding trial. During dynamic testing, each frame uses position and orientation information from the previous (or succeeding) frame or frames to determine the bone position and orientation at the current frame. The variation in error observed between different bones is thought to be a consequence of factors reported in previous studies [16]. Particularly, 3-D bone shape, bone density, edge definition, and orientation within the FOV contribute to the observed measurement error. As an example, the relatively large errors observed for the distal ulna are thought to arise from a lack of bony features compared to the distal radius and distal femur. As a result of this deficiency, the 2D projection (radiograph and DRR) of the distal ulna appears similar for multiple 3-D orientations. This highlights the need to conduct study specific error measurements on the system configuration being used and the bones being tracked. Establishing a method for testing the systematic error of a biplanar videoradiography system allows these measurements to be made consistently and relatively easily.

The described method for quantifying the systematic error of a biplanar videoradiography system allows for the independent assessment of marker-based and markerless tracking techniques under static and dynamic motion conditions. Specifically, the ability to assess errors at high speeds and during large accelerations such as highly dynamic impacts and direction changes is important for studying complex activities like jump-landing or hammering. Few studies have assessed markerless tracking er-

ror using independent “gold standards”. Li and colleagues [14] have evaluated the systematic error of a biplanar fluoroscope system using a high accuracy materials testing machine with favorable results; however, the low video capture frame rates (≤ 30 fps) limit the system’s ability to measure highly dynamic joint motions [32]. The method described herein was successful in measuring the systematic error of a biplanar videoradiography system tracking bone motion during high-speed impact conditions.

Moreover, the versatility of this method allows it to be applied to diverse experiments and testing conditions that require different system configurations and bones. It has become increasingly evident that quantifying errors associated with different bones and system configurations is an important and recommended part of any biplanar videoradiography study [10,11,14–18,33,34]. The variation in error observed for the different bones reported herein, as well as those reported in the literature, further highlight the importance of quantifying study specific errors. Additional static measurements were performed where each bone was rotated about its long axis (axis through the diaphysis), and the results were consistent with those reported. The axis of rotation would most likely affect a bone with few distinguishing features. The bones tested in this study provided enough features for the algorithm to successfully converge for multiple orientations. For studies investigating specific rotations for a uniformly featured bone, the axis of rotation associated with the outcome measure would need to be validated. Both testing apparatuses allow for the attachment of specimens in any orientation and could be tailored to suit these study specific validation questions. These methods can be used in conjunction with the described auto-registration software to efficiently evaluate systematic error for the majority of biplanar videoradiography studies.

The XROMM Autoscooper software provides the user with an elegant interface (Fig-

ure 2.2) for markerless bone tracking. Its use of GPU hardware significantly improves the speed over standard CPU implementations. You and colleagues [13] report total processing time to be 100 seconds per frame for two radiographic images. This can translate to hours of processing time for a single data set. Specifically, highly dynamic motions and impacts that occur during high-speed lower and upper extremity impacts require high video capture frame rates (> 150 fps) [35–40]. These types of data sets produce hundreds of frames for a single motion capture trial. Bey et al. [15] and Martin et al. [3] have taken steps to reduce processing times by making use of multi-workstation computational clusters that allow the algorithm to parallelize its computations over many processors; however, this amount of computational power can be prohibitively expensive and may be impractical for many laboratories attempting to process biplanar videoradiography data. Making use of modern GPU technology significantly speeds up computation time without the need for large computational clusters. The current implementation of XROMM Autoscooper on one desktop workstation with a single GPU (480 processing cores) provides a substantial cost advantage. This particular configuration requires between 0.5 and 1.5 seconds per frame depending on the size of the CT volume and visualization parameters. These time estimates will vary with different GPU hardware. Overall, implementing auto-registration algorithms on GPUs successfully reduces processing time, financial commitments, and saves valuable laboratory space.

Previous studies have thoroughly documented the limitations of biplanar videoradiography technology. Compared to traditional optical motion capture technology, biplanar videoradiography technology is not as readily available. Additionally, the X-ray exposure increases risk to subjects, and the relatively small imaging volume (approximately 2,000 in³) limits the range of activities that can be studied when compared to traditional motion capture technology [37]. Furthermore, the registration algorithm presented herein does require at least one user-supplied initial guess

that is visually close to the actual bone position and orientation. In practice, user input is typically required throughout data processing to assist the auto-registration algorithm converge on a final solution. The Autoscooper software requires different degrees of user input based on the quality of the collected data and the complexity of the joints and motions being processed. The results presented in this study are limited to idealized tracking situations. Only one bone was tracked during each test and no soft tissues were present. This eliminates bone and soft tissue overlap, which may increase processing time and measurement errors *in vivo*.

In summary, the results presented herein indicate that the biplanar videoradiography hardware and the publically available GPU-based markerless tracking software can be used to effectively track rapid bone motions to within 0.15° for the distal aspects of the femur, radius, and ulna. Future research will employ these techniques to quantify *in vivo* joint motion for high-speed upper and lower extremity impacts such as jumping, landing, and hammering. Furthermore, additional algorithms and image processing techniques will continue to be explored in hopes of further improving markerless tracking performance and efficiency.

2.5 Conflicts of interest

None.

2.6 Acknowledgements

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2.7 References

- [1] Bey, M. J., Brock, S. K., Beierwaltes, W. N., Zauel, R., Kolowich, P. A., and Lock, T. R., 2007, “In Vivo Measurement of Subacromial Space Width During Shoulder Elevation: Technique and Preliminary Results in Patients Following Unilateral Rotator Cuff Repair,” *Clin. Biomech. (Bristol, Avon)*, 22(7), pp. 767–773.
- [2] Bey, M. J., Kline, S. K., Zauel, R., Lock, T. R., and Kolowich, P. A., 2008, “Measuring Dynamic In-Vivo Glenohumeral Joint Kinematics: Technique and Preliminary Results,” *J. Biomech.*, 41(3), pp. 711–714.
- [3] Martin, D. E., Greco, N. J., Klatt, B. A., Wright, V. J., Anderst, W. J., and Tashman, S., 2011, “Model-Based Tracking of the Hip: Implications for Novel Analyses of Hip Pathology,” *J. Arthroplasty*, 26(1), pp. 88–97.
- [4] Tashman, S., and Anderst, W., 2003, “In-Vivo Measurement of Dynamic Joint Motion Using High Speed Biplane Radiography and CT: Application to Canine ACL Deficiency,” *J. Biomech. Eng.*, 125(2), pp. 238–245.
- [5] Tashman, S., Kolowich, P., Collon, D., Anderson, K., and Anderst, W., 2007, “Dynamic Function of the ACL-Reconstructed Knee During Running,” *Clin. Orthop. Relat. Res.*, 454, pp. 66–73.
- [6] Torry, M. R., Myers, C., Pennington, W. W., Shelburne, K. B., Krong, J. P., Giphart, J. E., Steadman, J. R., and Woo, S. L.-Y., 2010, “Relationship of Anterior

Knee Laxity to Knee Translations During Drop Landings: A Bi-Plane Fluoroscopy Study,” *Knee Surg. Sports Traumatol. Arthrosc.*, 19(4), pp. 653–662.

[7] Deneweth, J. M., Bey, M. J., McLean, S. G., Lock, T. R., Kolowich, P. A., and Tashman, S., 2010, “Tibiofemoral Joint Kinematics of the Anterior Cruciate Ligament-Reconstructed Knee During a Single-Legged Hop Landing,” *Am. J. Sports Med.*, 38(9), pp. 1820–1828.

[8] Li, G., DeFrate, L. E., Rubash, H. E., and Gill, T. J., 2005, “In Vivo Kinematics of the ACL During Weight-Bearing Knee Flexion,” *J. Orthop. Res.*, 23(2), pp. 340–344.

[9] Li, G., DeFrate, L. E., Sun, H., and Gill, T. J., 2004, “In Vivo Elongation of the Anterior Cruciate Ligament and Posterior Cruciate Ligament During Knee Flexion,” *Am. J. Sports Med.*, 32(6), pp. 1415–1420.

[10] Brainerd, E. L., Baier, D. B., Gatesy, S. M., Hedrick, T. L., Metzger, K. A., Gilbert, S. L., and Crisco, J. J., 2010, “X-Ray Reconstruction of Moving Morphology (XROMM): Precision, Accuracy and Applications in Comparative Biomechanics Research,” *J. Exp. Zool. A Ecol. Genet. Physiol.*, 313(5), pp. 262–279.

[11] Anderst, W. J., Vaidya, R., and Tashman, S., 2008, “A Technique to Measure Three-Dimensional In Vivo Rotation of Fused and Adjacent Lumbar Vertebrae,” *Spine J.*, 8(6), pp. 991–997.

[12] Gatesy, S. M., Baier, D. B., Jenkins, F. A., and Dial, K. P., 2010, “Scientific Rotoscopy: A Morphology-Based Method of 3-D Motion Analysis and Visualization,” *J. Exp. Zool. A Ecol. Genet. Physiol.*, 313(5), pp. 244–261.

[13] You, B. M., Siy, P., Anderst, W., and Tashman, S., 2001, “In Vivo Measurement of 3-D Skeletal Kinematics From Sequences of Biplane Radiographs: Application to Knee Kinematics,” *IEEE Trans. Med. Imaging*, 20(6), pp. 514–525.

[14] Li, G., Van de Velde, S. K., and Bingham, J. T., 2008, “Validation of a Non-Invasive Fluoroscopic Imaging Technique for the Measurement of Dynamic Knee Joint Motion,” *J. Biomech.*, 41(7), pp. 1616–1622.

[15] Bey, M. J., Zauel, R., Brock, S. K., and Tashman, S., 2006, “Validation of a New Model-Based Tracking Technique for Measuring Three-Dimensional, In Vivo Glenohumeral Joint Kinematics,” *J. Biomech. Eng.*, 128(4), pp. 604–609.

[16] Bey, M. J., Kline, S. K., Tashman, S., and Zauel, R., 2008, “Accuracy of Bi-plane X-Ray Imaging Combined With Model-Based Tracking for Measuring In-Vivo Patellofemoral Joint Motion,” *J. Orthop. Surg. Res.*, 3, p. 38.

[17] Anderst, W., Zauel, R., Bishop, J., Demps, E., and Tashman, S., 2009, “Validation of Three-Dimensional Model-Based Tibio-Femoral Tracking During Running,” *Med. Eng. Phys.*, 31(1), pp. 10–16. [18] Torry, M. R., Shelburne, K. B., Peterson, D.

- S., Giphart, J. E., Krong, J. P., Myers, C., Steadman, J. R., and Woo, S. L.-Y., 2011, "Knee Kinematic Profiles During Drop Landings: A Biplane Fluoroscopy Study," *Med. Sci. Sports Exercise*, 43(3), pp. 533–541.
- [19] Babenko, P., and Shah, M., 2008, "MinGPU: A Minimum GPU Library for Computer Vision," *J Real-Time Image Proc.*, 3(4), pp. 255–268.
- [20] Chen, L., Armstrong, C. W., and Raftopoulos, D. D., 1994, "An Investigation on the Accuracy of Three-Dimensional Space Reconstruction Using the Direct Linear Transformation Technique," *J. Biomech.*, 27(4), pp. 493–500.
- [21] Tashman, S., Anderst, W., Kolowich, P., Havstad, S., and Arnoczky, S., 2004, "Kinematics of the ACL-Deficient Canine Knee During Gait: Serial Changes Over Two Years," *J. Orthop. Res.*, 22(5), pp. 931–941.
- [22] Tashman, S., Collon, D., Anderson, K., Kolowich, P., and Anderst, W., 2004, "Abnormal Rotational Knee Motion During Running After Anterior Cruciate Ligament Reconstruction," *Am. J. Sports Med.*, 32(4), pp. 975–983.
- [23] Hedrick, T. L., 2008, "Software Techniques for Two- And Three-Dimensional Kinematic Measurements of Biological and Biomimetic Systems," *Bioinspir. Biomim.*, 3(3), p. 034001.
- [24] Marsalek, L., Hauber, A., and Slusallek P., 2008, "High-Speed Volume Ray Casting With CUDA," *Proceedings of the IEEE Symposium on Interactive Ray Tracing*.
- [25] Luebke, D., and Parker, S., 2008, "Interactive Ray Tracing With CUDA," Technical presentation, http://www.nvidia.com/.../tech_presentations.html. PDF URL: http://www.nvidia.com/.../NVISION08-Interactive_Ray_Tracing.pdf.
- [26] Kaufman, A., 1991, *Volume Visualization*, IEEE Computer Society Press, Los Alamitos, CA.
- [27] Russ, J., 2007, *The Image Processing Handbook*, CRC/Taylor and Francis, Boca Raton, FL.
- [28] Sonka, M., 1999, *Image Processing, Analysis, and Machine Vision*, PWS Pub., Pacific Grove, CA.
- [29] Nelder, J. A., and Mead, R., 1965, "A Simplex Method for Function Minimization," *Comput. J.*, 7(4), pp. 308–313.
- [30] Miranda, D. L., Rainbow, M. J., Leventhal, E. L., Crisco, J. J., and Fleming, B. C., 2010, "Automatic Determination of Anatomical Coordinate Systems for Three-Dimensional Bone Models of the Isolated Human Knee," *J. Biomech.*, 43(8), pp. 1623–1626.

- [31] Crisco, J. J., Coburn, J. C., Moore, D. C., Akelman, E., Weiss, A.-P. C., and Wolfe, S. W., 2005, “In Vivo Radiocarpal Kinematics and the Dart Thrower’s Motion,” *J. Bone Jt. Surg., Am.*, 87(12), pp. 2729–2740.
- [32] Tashman, S., 2008, “Comments on ‘Validation of a Non-Invasive Fluoroscopic Imaging Technique for the Measurement of Dynamic Knee Joint Motion’,” *J. Biomech.*, 41(15), pp. 3290–3293.
- [33] Hurschler, C., Seehaus, F., Emmerich, J., Kaptein, B. L., and Windhagen, H., 2008, “Accuracy of Model-Based RSA Contour Reduction in a Typical Clinical Application,” *Clin. Orthop. Relat. Res.*, 466(8), pp. 1978–1986.
- [34] Seehaus, F., Emmerich, J., Kaptein, B. L., Windhagen, H., and Hurschler, C., 2009, “Experimental Analysis of Model-Based Roentgen Stereophotogrammetric Analysis (MBRSA) on Four Typical Prosthesis Components,” *J. Biomech. Eng.*, 131(4), p. 041004.
- [35] Ford, K. R., Myer, G. D., Toms, H. E., and Hewett, T. E., 2005, “Gender Differences in the Kinematics of Unanticipated Cutting in Young Athletes,” *Med. Sci. Sports Exercise*, 37(1), pp. 124–129.
- [36] Ford, K. R., Myer, G. D., Smith, R. L., Vianello, R. M., Seiwert, S. L., and Hewett, T. E., 2006, “A Comparison of Dynamic Coronal Plane Excursion Between Matched Male and Female Athletes When Performing Single Leg Landings,” *Clin. Biomech. (Bristol, Avon)*, 21(1), pp. 33–40.
- [37] Taylor, K. A., Terry, M. E., Utturkar, G. M., Spritzer, C. E., Queen, R. M., Irribarra, L. A., Garrett, W. E., and DeFrate, L. E., 2011, “Measurement of In Vivo Anterior Cruciate Ligament Strain During Dynamic Jump Landing,” *J. Biomech.*, 44(3), pp. 365–371.
- [38] Hewett, T. E., Myer, G. D., Ford, K. R., Heidt, R. S., Colosimo, A. J., McLean, S. G., van den Bogert, A. J., Paterno, M. V., and Succop, P., 2005, “Biomechanical Measures of Neuromuscular Control and Valgus Loading of the Knee Predict Anterior Cruciate Ligament Injury Risk in Female Athletes: A Prospective Study,” *Am. J. Sports Med.*, 33(4), pp. 492–501.
- [39] Leventhal, E. L., Moore, D. C., Akelman, E., Wolfe, S. W., and Crisco, J. J., 2010, “Carpal and Forearm Kinematics During a Simulated Hammering Task,” *J. Hand Surg. [Am]*, 35(7), pp. 1097–1104.
- [40] Coˆte´, J. N., Raymond, D., Mathieu, P. A., Feldman, A. G., and Levin, M. F., 2005, “Differences in Multi-Joint Kinematic Patterns of Repetitive Hammering in Healthy, Fatigued and Shoulder-Injured Individuals,” *Clin. Biomech. (Bristol, Avon)*, 20(6), pp. 581–590.

Chapter 3

Automatic determination of anatomical coordinate systems for three-dimensional bone models of the isolated human knee

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Abstract

The combination of three-dimensional (3-D) models with biplanar videoradiography is increasingly popular for evaluating joint function *in vivo*. Applying these modalities to study knee motion with high accuracy requires reliable anatomical coordinate systems (ACSs) for the femur and tibia. Therefore, a robust method for creating ACSs from 3-D models of the femur and tibia is required. We present and evaluate an automated method for constructing ACSs for the distal femur and proximal tibia based solely on 3-D bone models. The algorithm requires no observer interactions and uses model cross-sectional area, center of mass, principal axes of inertia, and cylindrical surface fitting to construct the ACSs. The algorithm was applied to the femur and tibia of 10 (unpaired) human cadaveric knees. Due to the automated nature of the algorithm, the within specimen variability is zero for a given bone model. The algorithm's repeatability was evaluated by calculating variability in ACS location and orientation across specimens. Differences in ACS location and orientation between specimens were low (< 1.5 mm and $< 2.5^\circ$). Variability arose primarily from natural anatomical and morphological differences between specimens. The presented algorithm provides an alternative method for automatically determining subject-specific ACSs from the distal femur and proximal tibia.

Keywords: *knee, kinematics, anatomical coordinate system, three-dimensional model, surface fitting*

3.1 Introduction

Accurate bone-based coordinate systems are critical for studying the effect of kinematics on ligament and articular cartilage deformation [1–4]. X-ray-based three-dimensional (3-D) skeletal motion-capture technologies require reliable methods for establishing femoral and tibial anatomical coordinate systems (ACSs) to measure knee kinematics.

Standard methods for defining femoral and tibial ACSs use the knee, hip, and ankle joints. Typically, the knee’s flexion/extension (FE) axis is defined as the vector through a cylinder fitted to the femoral condyles. Additional axes are built using the center of the femoral head. The tibial ACS is traditionally defined using medial and lateral points on the tibial plateau combined with the ankle’s center [5,6].

An alternative method is necessary for *ex vivo* biomechanical studies using isolated knee preparations that do not include the proximal femur and distal tibia. Ideally, the method could be used both *in vivo* and *ex vivo*. Herein, we present and evaluate an automated method for constructing subject-specific ACSs for the distal femur and proximal tibia based on bony geometry derived from 3-D images (Figure 3.1).

3.2 Methods

3.2.1 Bone models

CT images of the distal femur and proximal tibia of 10 fresh frozen cadaver knees (7 right, 3 left; 7 male, 3 female, aged 58.3 ± 11.1 years) were acquired (80 kVp, 400 mA, $0.22 \times 0.22 \times 0.625$ mm³: LightSpeed; GE, Piscataway, NJ, USA). CT-based 3-D

bone models were then generated using Materialise Mimics 12.01 (Ann Arbor, MI, USA).

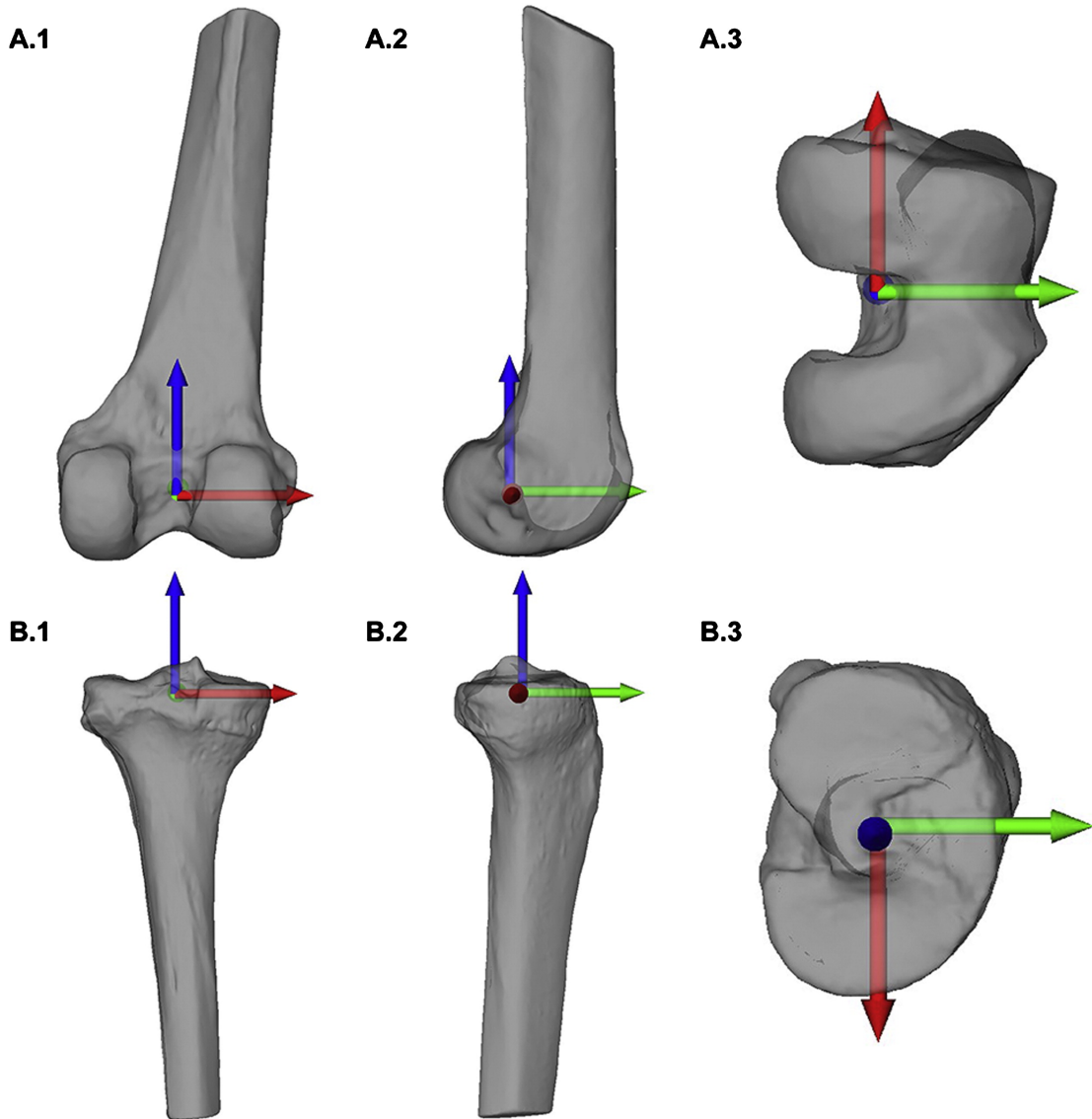


Figure 3.1: The algorithm begins with 3-D models of the distal femur and proximal tibia generated from a CT scan of the knee. Initial inertial axes are used to isolate the femoral diaphysis, femoral condyles, and tibial plateau. (A) The ML axis of the femoral ACS is determined by fitting a cylinder to the condyles. The AP axis of the femoral ACS is calculated by crossing the ML axis with the inertial axis aligned along the femoral diaphyseal's length. The long axis of the femoral ACS is calculated by re-crossing the AP axis with the ML axis. (B) The three axes of the tibial ACS were defined using the isolated plateau's inertial axes.

3.2.2 Femoral ACS

The primary axis of the femoral ACS (medial/lateral, ML) was established from its articulating surfaces. The femoral condyles were isolated using a plane oriented over two iterations. This was done to align the plane's ML axis with the distal femur's anatomy to evenly capture the articulating surfaces (Figure 3.2A.1). A cylinder was fit to the condyles using a MATLAB-based Gauss–Newton algorithm (Figure 3.2B). The cylinder's center vector formed the ML axis of the femoral ACS because it approximates the knee's FE axis [5,7,8].

The normal vector defining the plane's first iteration was determined by crossing a vector connecting a point on the distal femur (pt_1) and a point on the posterior femur (pt_2) with a vector pointing in the ML direction. This vector was defined using the femur's inertial axes [9]. The points isolated from the plane's first iteration were fit with a cylinder. The cylinder's center vector was used to define a new ML vector. The normal vector defining the plane's final iteration was determined by crossing the new ML vector with the vector connecting pt_1 and pt_2 .

Pt_1 was determined by extending a vector distally through the femoral diaphysis until it intersected with the femur (Figure 3.2A.1 and 3.2A.2). The smallest inertial axis of only the femoral diaphysis defined this vector. The diaphysis was isolated by calculating the femur's inertial axes. Axial cross-sections oriented using these axes were determined along the femur's length (Figure 3.3). Three locations were determined along the smallest inertial axis of the femur: (1) the maximum cross-sectional area; (2) half the maximum cross-sectional area; and (3) 30% farther from the distal end of the femur than the second location. Finally, a plane oriented using the femur's inertial axes and positioned at the third location was used to isolate the diaphysis.

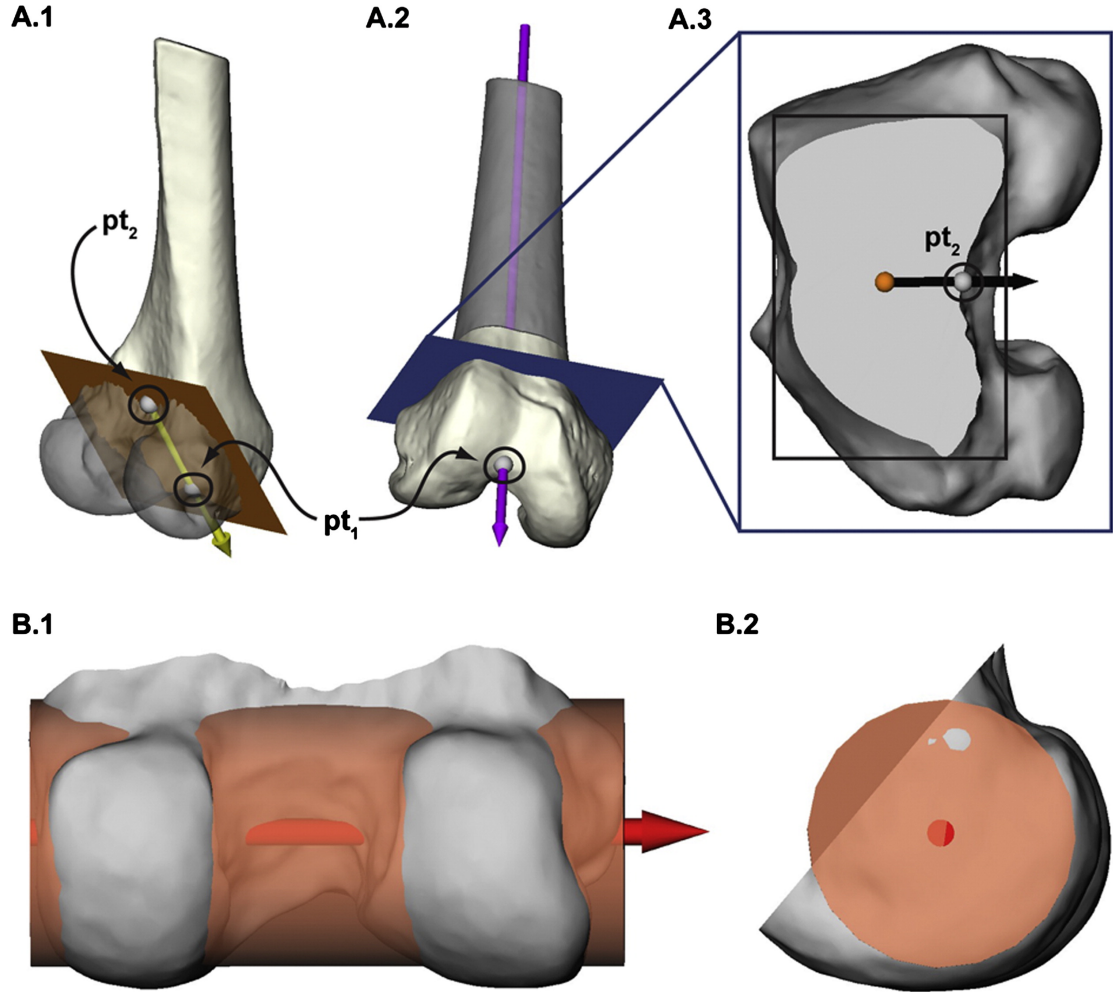


Figure 3.2: (A.1) Posterior view of the plane used to isolate to the condyles. The yellow arrow represents the vector connecting pt_2 to pt_1 . (A.2) Anterior view of the vector through the diaphysis (purple arrow). The intersection of this vector with the distal femur defined pt_1 . The axial plane placed within the metaphysis was used to determine pt_2 . (A.3) top view of the axial cross-section of the femoral metaphysis. The orange sphere is the center of the box bounding the femur. The black arrow represents the femurs inertial axis pointing in the AP direction placed at the center of the femoral bounding box. The intersection of this vector with the posterior femur defined pt_2 . (B.1) Distal view of the condyles. (B.2) Lateral view of the condyles. The cylinder is fit to the condyles using a least-squares algorithm. The condyles were isolated using the second iteration of the plane shown in A.1. The red arrow is the vector through the center of the cylinder, denoted as the ML axis of the femoral ACS.

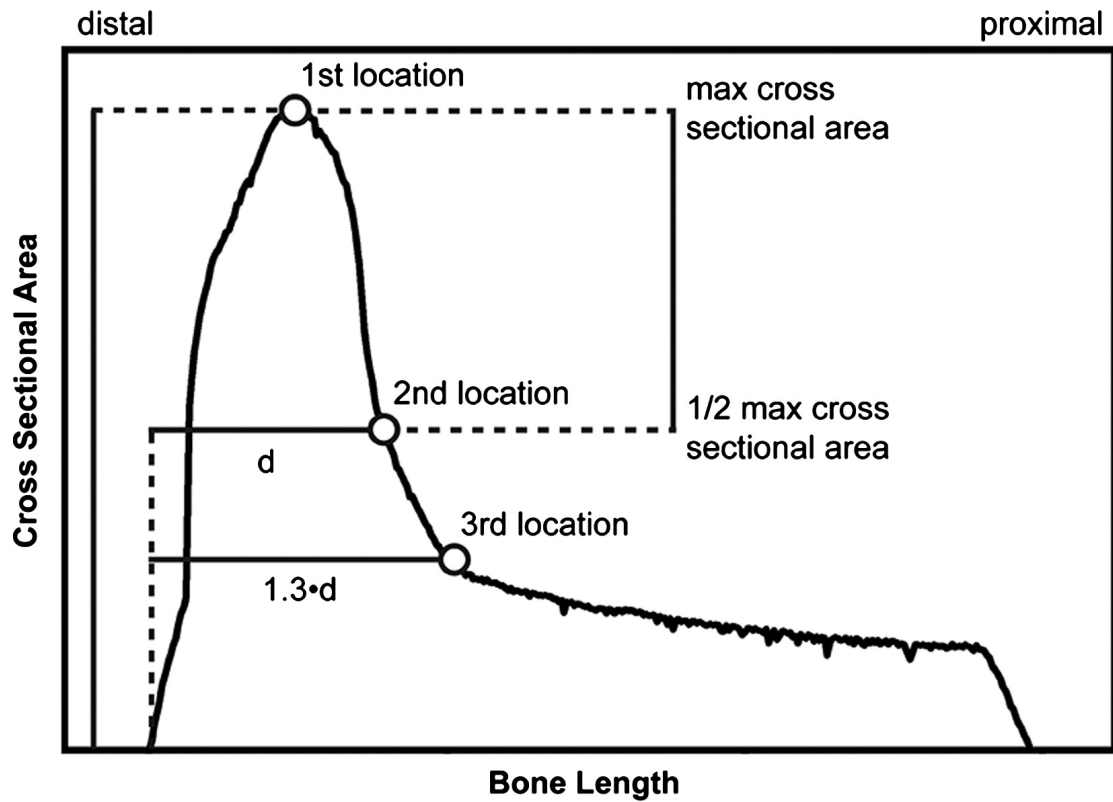


Figure 3.3: Representative graph showing cross-sectional area versus bone length. Cross-sectional area was used to define three locations along the length of the bone that were used to isolate the femoral condyles, diaphysis, and tibial plateau. The first location was positioned at the maximum cross-sectional area. The second location was positioned a distance (d) from the distal end of the femur. The distance (d) was determined by half the maximum cross-sectional area. The third location was set 30% farther from the distal end of the femur than the second location. The placement of these locations was based on preliminary evaluations of consistency across specimens. The locations were appropriate for all tested femurs and their placement did not vary. Small variations did not significantly alter the ACS definitions.

Pt₂ was established using the previously described axial cross-sections. A plane at the second location, oriented using the femur's inertial axes, was used to determine the center of a box bounding the femur (Figure 3.2A.3). Finally, pt₂ was determined by extending the femur's inertial axis pointing in the anterior/posterior (AP) direction through the posterior femur.

The femoral ACSs secondary axis (AP) was calculated by crossing the vector through the diaphysis with the ML axis. The femoral ACS's third axis was calculated by re-crossing the AP and ML axes. The origin of the femoral ACS was positioned at the centroid of the cylinder fit to the condyles.

3.2.3 Tibial ACS

The articulating surface of the tibial plateau was used to define the ML, AP, and third axes of the tibial ACS. The plateau was isolated by calculating the tibia's inertial axes. A plane oriented using the tibia's inertial axes and positioned at the largest cross-sectional area was used to isolate the plateau. The inertial axes of only the plateau were designated as the ML, AP, and third axes of the tibial ACS. The origin of the tibial ACS was positioned at the plateau's center of mass.

3.2.4 Data analysis

The algorithm's repeatability was evaluated by computing the differences in location and orientation of each ACS after aligning each bone to a template bone using a surface registration protocol. Prior to registration, the bone models were scaled to similar volumes according to a scale factor determined by balancing the width of each specimen's condyles or tibial plateau. Femoral and tibial surface registrations

were then performed using Geomagic Studio’s (v10; Morrisville, NC, USA) best-fit alignment algorithm.

3.3 Results

The described algorithm was successful in automatically constructing ACSs for the femur and tibia from the CT-based bone models. The algorithm’s repeatability was evaluated by computing the differences in location and orientation of each ACS compared to the mean ACS, which was determined for both by averaging each specimen’s axes and origins. Location differences were evaluated as 2-D component (x, y, z) distances and absolute 3-D distances. Orientation differences were evaluated as angular ACS x-, y-, and z-axis distances. Mean femoral and tibial ACS x-, y-, and z-location and orientation differences were consistent, with a bias of less than 1.5 mm and 2.5° (Figure 3.4). Additionally, the femoral ACS’s average absolute 3-D location difference was slightly larger (1.7 mm) with a more dispersed range (CI: 1.3 – 2.2 mm) than the tibial ACSs (1.1 mm, CI: 0.7 – 1.5 mm).

3.4 Discussion

We have presented an automated method for constructing subject-specific ACSs for the distal femur and proximal tibia based solely on their 3-D bony geometry. The algorithm was automated to eliminate time-consuming user interactions that may introduce theoretical bias from point or region selection. The ACSs were designed to define a knee joint coordinate system (JCS) based on a geometrical model of the femur’s cylindrical surface rolling on top of the tibia’s planar surface [7,8]. Describing

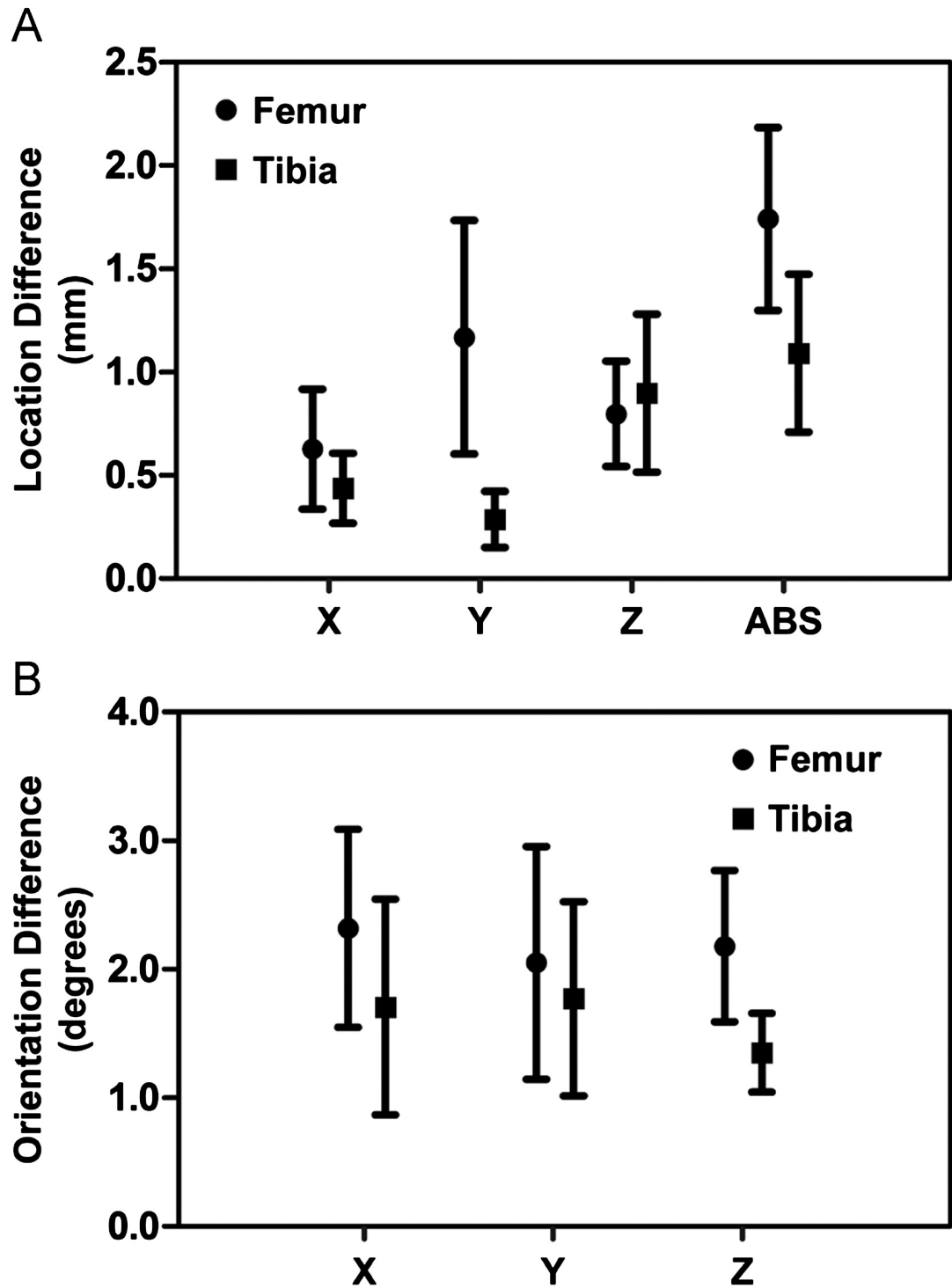


Figure 3.4: (A) Average location variability of the ACS origins in mm. (B) Average orientation variability of the ACS axes in degrees. Circles correspond to the femur ACSs and squares correspond to the tibia ACSs. Error bars are 95% confidence intervals. Variability between the ACSs is thought to be influenced primarily by differences in bone morphology between specimens.

this motion using two right-handed orthogonal coordinate systems allows angulations between the femur and tibia to be captured in all three anatomical planes.

Traditional methods for defining femoral and tibial ACSs employ data from the hip and ankle joints that may not be available for *ex vivo* biomechanics studies of isolated knee preparations [5,6]. Aspects of previous axis definitions were translated to the presented algorithm. Specifically, the axis of the cylinder fit to the femoral condyles was used to build the knee’s FE axis (Figure 3.2B). Additionally, the axis through the femoral diaphysis (anatomical axis) was used in conjunction with the “cylindrical axis” to create the third axis of the femoral ACS. As a result of the cross-product, the third axis of the femoral ACS is tilted medially, pointing from the center of the condyles toward the femoral head (Figure 3.1A.1). However, these observations require further quantitative investigation.

While the ACSs were designed with traditional definitions in mind, the absence of hip and ankle data combined with the lack of any gold standard complicate the evaluation of the proposed method. McPherson et al. described a method for creating a JCS for the distal femur and proximal tibia; however, the method is based on an extended knee position and assumes no variation of relative position between the femur and tibia between individuals. This assumption may not hold when investigating abnormal knee mechanics [10]. In these cases, a subject-specific method for creating position-independent femoral and tibial ACSs is necessary.

The algorithm’s repeatability with regard to subject-specific ACSs was evaluated by comparing inter-specimen ACSs. The majority of variability was likely a consequence of between-subject morphology differences. These differences contributed to increased variability because exact bone morphology was used to build the ACSs. Furthermore, surface discrepancies between specimens visibly affected the surface registration technique, creating imperfections in the alignment that contributed to increased variabil-

ity. While only one template knee position was used, this evaluation emphasizes the proposed method’s sensitivity to the anatomical and morphological differences between specimens, highlighting its capability of generating consistent, subject-specific ACSs.

An essential component of the femoral ACS is the smallest inertial axis of the femoral diaphysis (Figure 3.2A.2). This axis is consistently oriented through the diaphysis; however, its orientation is affected when the diaphyseal length approaches its width. We determined that a length larger than approximately 55 mm consistently orients the inertial axes with deviations less than 1° . In circumstances where the diaphyseal length approaches its diameter, the smallest inertial axis may be oriented in many directions. In these circumstances, it is appropriate to use the center axis of a cylinder fit to the diaphysis. The average diaphysis length used in this study was 146 ± 21 mm.

Accurately capturing all three anatomical planes of the knee is the first step in the assessment of knee kinematics. Differences between the presented ACS definitions and previous methods will likely result in slightly different kinematic interpretations; however, an attempt to minimize these differences was made by translating traditional axis definitions. While differences are unavoidable, providing a clear definition is required to interpret joint motion [11]. Herein, we defined the first, second, and third axes for the femur and tibia ACSs in detail.

3.5 Conflicts of interest

None.

3.6 Acknowledgements

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3.7 References

- [1] Andriacchi T. P., Mündermann A., Smith R. L., Alexander E. J., Dyrby C. O., and Koo S., 2004, “A framework for the in vivo pathomechanics of osteoarthritis at the knee,” *Ann Biomed Eng*, 32(3), pp. 447–457.
- [2] Andriacchi T. P., Koo S., and Scanlan S. F., 2009, “Gait mechanics influence healthy cartilage morphology and osteoarthritis of the knee,” *J Bone Joint Surg Am*, 91 Suppl 1, pp. 95–101.
- [3] Andriacchi T. P., and Mündermann A., 2006, “The role of ambulatory mechanics in the initiation and progression of knee osteoarthritis,” *Curr Opin Rheumatol*, 18(5), pp. 514–518.
- [4] Jordan S. S., DeFrate L. E., Nha K. W., Papannagari R., Gill T. J., and Li G., 2007, “The in vivo kinematics of the anteromedial and posterolateral bundles of the anterior cruciate ligament during weightbearing knee flexion,” *Am J Sports Med*, 35(4), pp. 547–554.
- [5] Eckhoff D. G., Bach J. M., Spitzer V. M., Reinig K. D., Bagur M. M., Baldini T. H., and Flannery N. M. P., 2005, “Three-dimensional mechanics, kinematics, and morphology of the knee viewed in virtual reality,” *J Bone Joint Surg Am*, 87 Suppl 2, pp. 71–80.
- [6] Fernandez J. W., Akbarshahi M., Kim H. J., and Pandy M. G., 2008, “Integrating modelling, motion capture and x-ray fluoroscopy to investigate patellofemoral function during dynamic activity,” *Comput Methods Biomech Biomed Engin*, 11(1), pp. 41–53.

- [7] Morrison J. B., 1970, “The mechanics of the knee joint in relation to normal walking,” *J Biomech*, 3(1), pp. 51–61.
- [8] Pennock G. R., and Clark K. J., 1990, “An anatomy-based coordinate system for the description of the kinematic displacements in the human knee,” *J Biomech*, 23(12), pp. 1209–1218.
- [9] Gonzalez-Ochoa C., McCammon S., and Peters J., 1998, “Computing moments of objects enclosed by piecewise polynomial surfaces,” *ACM Trans. Graph.*, 17, pp. 143–157.
- [10] McPherson A., Kärrholm J., Pinskerova V., Sosna A., and Martelli S., 2005, “Imaging knee position using MRI, RSA/CT and 3D digitisation,” *J Biomech*, 38(2), pp. 263–268.
- [11] Most E., Axe J., Rubash H., and Li G., 2004, “Sensitivity of the knee joint kinematics calculation to selection of flexion axes,” *J Biomech*, 37(11), pp. 1743–1748.

Chapter 4

Kinematic differences between optical motion capture and biplanar videoradiography during a jump-cut maneuver

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2012 July

Abstract

Jumping and cutting activities are investigated in many laboratories attempting to better understand the biomechanics associated with non-contact deceleration ACL injury. Optical motion capture (OMC) is widely used; however, it is subject to soft tissue artifact (STA). Biplanar videoradiography offers a unique approach to collecting skeletal motion without STA. The goal of this study was to compare how STA affects the six-degree-of-freedom motion of the femur and tibia during a jump-cut maneuver associated with non-contact ACL injury. Ten volunteers performed a jump-cut maneuver while their landing leg was imaged using optical motion capture (OMC) and biplanar videoradiography. The within-bone motion differences were compared using anatomical coordinate systems for the femur and tibia, respectively. The knee joint kinematic measurements were compared during two periods: before and after ground contact. Over the entire activity, the within-bone motion differences between the two motion capture techniques were significantly lower for the tibia than the femur for two of the rotational axes (flexion/extension, internal/external) and the origin. The OMC and biplanar videoradiography knee joint kinematics were in best agreement before landing. Kinematic deviations between the two techniques increased significantly after contact. This study provides information on kinematic discrepancies between OMC and biplanar videoradiography that can be used to optimize methods employing both technologies for studying dynamic *in vivo* knee kinematics and kinetics during a jump-cut maneuver.

Keywords: *knee, biomechanics, soft tissue artifact, x-ray, in vivo*

4.1 Introduction

Activities involving jumping, landing, and cutting are commonly associated with non-contact deceleration anterior cruciate ligament (ACL) injuries [1,2]. Non-contact deceleration ACL injuries are those sustained during activity without contact with another athlete or object. They account for approximately 70% of the estimated 400,000 ACL injuries sustained in the United States each year [3,4]. The mechanisms associated with non-contact deceleration ACL injury are not well understood and most likely occur from a combination of risk factors including environmental, anatomical, hormonal, and biomechanical factors [2]. Understanding the biomechanics of activities associated with non-contact deceleration ACL injury would provide insight into injury mechanisms and may provide evidence in support of specialized training and rehabilitation techniques to minimize injury and optimize treatment [3,5].

Optical motion capture (OMC) technologies have been used to non-invasively quantify 3-D joint kinematics, including jumping and cutting [6]. The field-of-view (FOV) of OMC systems is large enough to capture the motion of multiple joints over an entire jumping and cutting movement. When combined with ground reaction forces and a biomechanical model, OMC is a powerful tool for providing insight into the biomechanics of the knee during these high speed maneuvers. However, OMC is sensitive to soft tissue artifact (STA), where motion of skin-mounted markers move relative to the underlying bones, particularly during the landing and support phases of a movement [7,8]. STA limits the accuracy of joint metrics at the level of the ligaments and articulating surfaces. Cluster based motion capture techniques have been developed to mitigate the effect of STA [9]. While these techniques improve segment tracking, they do not eliminate STA. Attempts to eliminate STA using percutaneous bone fixtures have been successful; however, these methods are highly invasive and likely alter joint motion, especially during jumping and cutting maneuvers [7,10,11].

Elucidating how joint arthrokinematics (e.g. joint translation, joint space, contact regions) may contribute to a sports injury remains challenging without an accurate, non-invasive method for understanding the bone to bone kinematics during landing and cutting.

High-speed biplanar videoradiography motion capture techniques allow investigators to accurately quantify 3-D bone motion during dynamic activities without STA [12–15]. Currently, biplanar videoradiography systems are capable of capturing 3-D kinematics of a single joint within a basketball sized FOV. It would be advantageous to capture an entire jump-cut maneuver with OMC while using biplanar videoradiography to focus on knee joint kinematics at the level of the articulating surfaces during impact with the ground. This would provide highly accurate bone kinematics during an important phase of the maneuver while quantifying the kinematics of the surrounding joints throughout the entire movement.

Recently, Myers et al. [16] and Taylor et al. [17] made use of both biplanar videoradiography and OMC to study tibiofemoral kinematics during drop and jump landings. While these studies provide a foundation for combining the two motion capture techniques, comparing the respective kinematic outcomes was beyond their scope. Understanding how STA affects each body segment (e.g. thigh and shank) during different periods of an activity will strengthen the interpretation of kinematic outcomes associated with high demand activities. Moreover, by better understanding the limitations of a combined technique, kinetic outcomes may be estimated by effectively expanding the FOV using OMC. In order to optimize a combined OMC and biplanar videoradiography technique, their measurement differences during landing activities (i.e. those which are more susceptible to STA) must be quantified.

The goal of this study was to compare how STA affects the six-degree-of-freedom (DOF) motion of the femur and tibia during a jump-cut maneuver. For this assess-

ment, we: (1) quantified the within-bone motion differences as measured by OMC and biplanar videoradiography; and (2) evaluated how these within-bone motion differences affected the clinically relevant knee joint kinematics (tibia with respect to the femur) before and after ground contact. It has previously been shown that soft tissue surrounding the femur affects kinematic measurements more significantly than the soft tissue surrounding the tibia during knee flexion, axial hip rotation, and stair-ascent [18-20]. Additionally, Garling et al. has reported an increase in STA after ground contact during stair-ascent [18]. Based on these data, we hypothesized that the soft tissue surrounding the femur would have a greater effect on the kinematic measurements than the soft tissue surrounding the tibia during the jump-cut maneuver. Moreover, we hypothesized that STA would significantly influence the OMC joint kinematic measurements after ground contact during the jump-cut maneuver when compared to the biplanar videoradiography measurements.

4.2 Methods

4.2.1 Subjects

All experimental procedures were approved by the Institutional Review Board. Ten recreational athletes (5 males, 5 females; age 25 ± 3.3 years; height 1.73 ± 0.10 m; weight 73.17 ± 10.15 kg) were enrolled in this study. The inclusion criteria were: (1) no neurological disease(s); (2) no pregnancy; (3) no history of lower extremity injury; and (4) a Tegner activity score of five or greater [21]. After granting their informed consent, each subject was outfitted with two marker clusters containing five retro-reflective markers. These marker clusters were placed on the thigh and shank to track the six degree-of-freedom motions of the femur and tibia using OMC. This marker set was chosen because it is a subset of a larger marker set commonly used to

track segment motion using OMC [22]. The outfitted leg was chosen randomly (6L and 4R).

4.2.2 Jump-cut maneuver

Volunteers performed a jump-cut maneuver, originally described by Ford et al., that was designed to mimic maneuvers associated with non-contact deceleration ACL injury [6]. Three targets were placed on the floor within the testing environment: a jump-landing target on the center of the force plate and two targets placed two meters beyond the jump-landing target at an angle of approximately 45° . These two targets provided a reference for the subject to cut toward and jog past after landing. Before beginning the maneuver, the subject was asked to stand approximately one meter from the force plate with their knees bent at approximately 45° . A verbal prompt was used to cue the subject to jump upward and forward toward the landing target. A visual directional prompt (L or R) cued the subject to perform a sidestep cut toward and jog past one of the angled targets. For example, when a subject began their jump and was signaled to cut to the left, they landed and pushed off with their right foot and led with their left (Figure 4.1A). Ten trials (five in each direction) were performed. The subject was unaware of the directional prompt prior to each trial.

4.2.3 Data collection and processing

All biomechanical data were collected in the W.M. Keck Foundation X-Ray Reconstruction of Moving Morphology (XROMM) Facility at Brown University, (Providence, RI, USA). OMC data was collected at 250 Hz using a four camera Qualisys Oqus 5-series system (Gothenburg, Sweden). GRF data were time-synchronized with

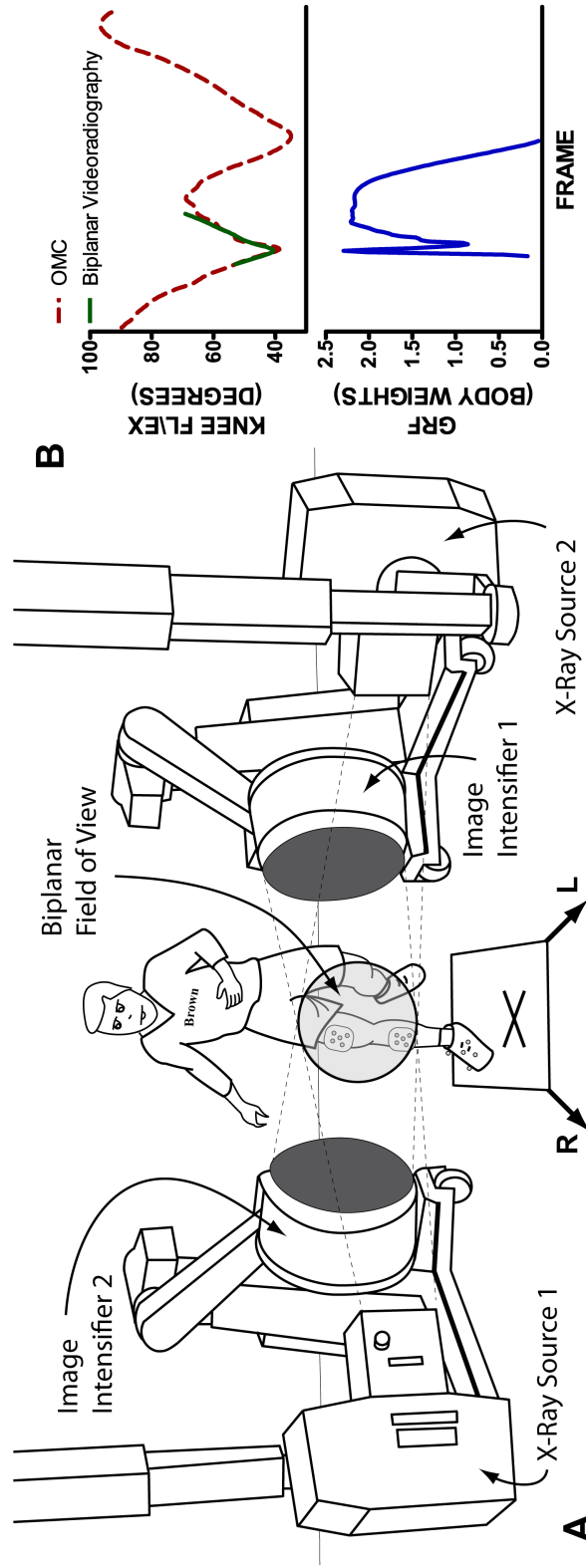


Figure 4.1: (A) Experimental set-up including image intensifiers and x-ray sources. The optical motion capture cameras are not shown. The subject is performing the jump-cut maneuver. In this example they were cued to cut to their left upon landing on the force plate. The 'X' marks the landing location and the arrows represent the left (L) and right (R) cut directions. (B) The OMC (dotted red) and biplanar videoradiography (solid green) knee flexion/extension and GRF (solid blue) for the entire jump-cut activity including the flight phase, landing, rotation, cut, and toe-off. The field of view for the biplanar videoradiography limits its ability to collect kinematic data for the entire jump-cut activity. However, it can be tailored to measure motion for specific periods of an activity where OMC is more sensitive to soft tissue artifact.

the OMC system and collected at 5,000 Hz using a Kistler model 9281B force plate (Amherst, NY, USA).

Biplanar videoradiography data were time-synchronized with the OMC system using an electrical trigger and collected at 250 Hz during three of the jump-cut trials where the subject landed and cut with their outfitted leg. The biplanar videoradiography system has been previously described in detail [14]. X-rays were generated using an exposure of 70 kVp and 100 mA. Digital radiographs were captured using shutter speeds of 400 - 800 μ s. The x-ray sources were positioned at 120° with a source to image distance of 165 cm. Image de-distortion and 3-D space calibration (XrayProject; Brown University, Providence, RI, USA) was performed using previously described methods [13].

Clinical CT scans (Lightspeed; GE, Piscataway, NJ, USA) for each subject's outfitted knee were acquired in the axial plane at 80 kVp, using GE's SMART mA and Bone Plus reconstruction algorithm. The voxel resolution was less than 0.381x0.381x0.625 mm³. For each scan, the femur and tibia were isolated from the entire volume using threshold and segmentation tools available in Mimics v14 (Materialise, Ann Arbor, MI, USA).

The biplanar videoradiography data were processed using custom markerless tracking software (Autoscooper; Brown University, Providence, RI, USA) detailed previously [14]. Briefly, the CT volumes for the femur and tibia were registered with the biplanar videoradiography data and the bones were tracked across all captured frames using the Autoscooper software (Figure 4.2A-D). These techniques track *in vivo* bone motion within 0.25 mm and 0.25° [14]. The knee joint kinematics of the tibia with respect to the femur were described using a pair of anatomical coordinate systems (ACSSs) created from the 3-D CT bone models of the femur and tibia using previously described methods (Figure 4.2E) [23]. This pair of ACSs was used to interpret knee

joint kinematics for both OMC and biplanar videoradiography.

In order to use the same ACSs for both imaging modalities, the global coordinate spaces of the OMC and biplanar videoradiography systems were registered using a rigid lattice that consisted of 11 radio-opaque, spherical markers outfitted with retroreflective tape (3M, St. Paul, MN, USA). The OMC and biplanar videoradiography systems simultaneously captured a static position of the lattice and the transformation matrix for co-registration of each global coordinate space was computed [24,25] and applied to the respective kinematic data sets.

4.2.4 Data analysis

In order to compare the within-bone motion differences, the CT-based ACSs were independently driven by the OMC and biplanar videoradiography systems. This resulted in the motion of the CT-based femoral and tibial ACSs as determined by the biplanar videoradiography system and as determined by the OMC system (Figure 4.2F). The biplanar videoradiography ACSs were compared to the OMC ACSs by determining the axes (Rx, Ry, and Rz) and origin (T) differences at each frame. The range (max - min) of axes and origin differences were calculated for each trial across all subjects and these data were described using sample means and standard deviations. The range differences of the femoral ACS were compared to the range differences of the tibial ACS using a paired t-test. Additionally, the overall median and maximum range differences of the femoral and tibial ACSs were computed for all subject trials.

Knee joint kinematics were compared for each trial across all subjects using both sets of ACSs. Joint rotations in flexion/extension (FL/EX), adduction/abduction (AD/AB), and internal/external (IN/EX) rotations of the tibia relative to the femur were interpreted using Grood and Suntay's method [26]. Joint translations in

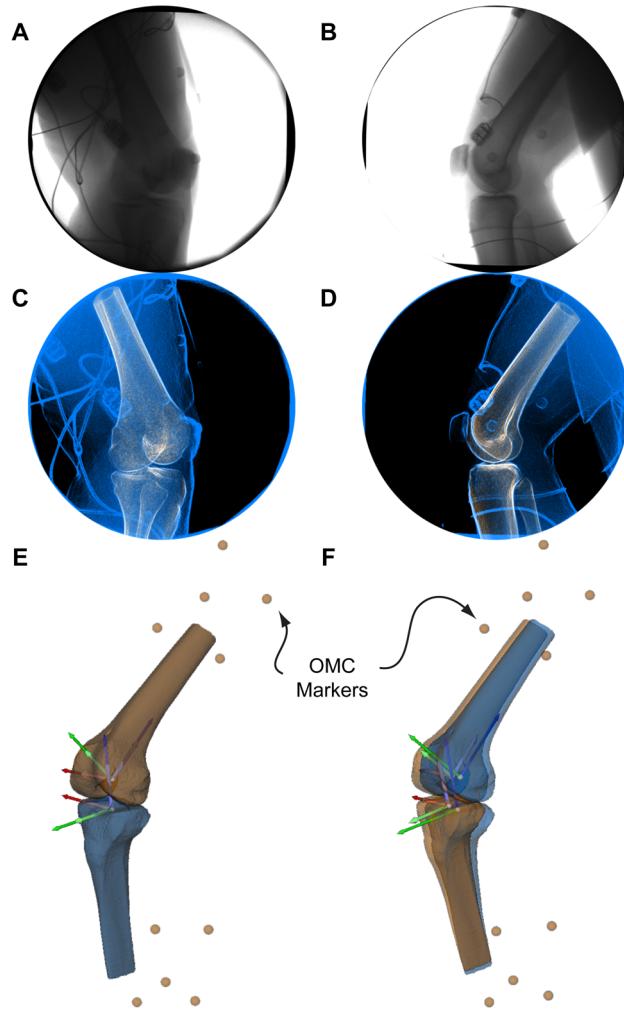


Figure 4.2: Panels A and B represent a single frame of the biplanar videoradiography data for x-ray source 1 and x-ray source 2, respectively. Panels C and D represent the same frame of the biplanar videoradiography data (blue) after image processing. Contrast and edge detection is used to enhance the images. Additionally, the digitally reconstructed radiographs generated from the CT volume are displayed in tan and are superimposed on the blue and black biplanar videoradiography data. The images represent the outcome of the Autoscooper software after bone tracking is completed for the current frame. The 3-D models of the tibia and femur driven by optical motion capture (tan) and biplanar motion capture (blue) are shown in panels E and F. All four independently tracked anatomical coordinate systems are also shown. The short and lighter coordinate systems are being driven by OMC and the long and darker coordinate systems are being driven by biplanar videoradiography. The external markers for the thigh and shank are also shown in tan. Panel E represents the initial frame, where OMC and biplanar videoradiography are perfectly aligned. Panel F represents a frame where soft tissue artifact is affecting the OMC driven bones and coordinate systems.

medial/lateral (ME/LA), anterior/posterior (AN/PO), and compression/distraction (CO/DI) displacements of the tibia relative to the femur were determined by a vector originating at the origin of the femoral ACS and terminating at the origin of the tibial ACS [27]. All kinematic measurements were processed using Visual3D (C-Motion, Germantown, MD, USA).

The joint rotations and translations were separated into two periods: (A) flight phase to contact; and (B) following contact. Period A began and period B ended when the femur and tibia entered and exited the FOV of the biplanar videoradiography system. For each period, the differences in kinematic excursions (max-min) between OMC and biplanar videoradiography were calculated for each trial. The absolute differences in kinematic excursions were described using sample means and standard deviations. A paired t-test was used to determine if the differences in values among periods A and B were greater than would be expected by chance. Additionally, the overall median and maximum differences in knee joint rotational and translational excursions were computed for both periods across all subject trials.

4.3 Results

The ranges of femoral ACS axes (Rx, Ry, Rz) and origin (T) differences between OMC and biplanar videoradiography were higher than the range of tibial ACS axes and origin differences (Figure 4.3 Top). The ranges of femoral differences were only significantly higher for Rx, Rz, and T; however, the p-value for Ry was 0.19. The maximum range of axes differences for the femoral and tibial ACSs were as high as 18° and 13°, respectively (Table 4.1). The maximum range of origin differences for the femoral and tibial ACSs were as high as 34 mm and 29 mm, respectively (Table 4.1).

The joint excursion differences between OMC and biplanar videoradiography were larger after contact (period B) for all knee joint parameters (Figure 4.3 Middle: FL/EX, AD/AB, IN/EX; Figure 4.3 Bottom: ME/LA, AN/PO, and CO/DI). The FL/EX excursion differences were not found to be statistically significant between period A and B ($p = 0.13$). For AD/AB and IN/EX, the increase in excursion difference between OMC and biplanar videoradiography was more striking. These increases were statistically significant ($p < 0.0001$). The maximum excursion differences for joint rotations were as high as 7° for periods A and 15° for period B (Table 4.1).

For ME/LA, AN/PO, and CO/DI, excursion differences between OMC and biplanar videoradiography increased from period A to period B. These increases were all statistically significant ($p < 0.001$). The maximum excursion differences for joint translations were as high as 11 mm and 28 mm for periods A and B, respectively (Table 4.1).

ANATOMICAL COORDINATE SYSTEM DIFFERENCES								
FEMUR					TIBIA			
	R_x ($^\circ$)	R_y ($^\circ$)	R_z ($^\circ$)	T (mm)	R_x ($^\circ$)	R_y ($^\circ$)	R_z ($^\circ$)	T (mm)
MAX	10.23	8.88	18.18	33.70	9.25	7.47	13.43	29.11
MEDIAN	5.07	3.67	11.97	24.91	4.24	3.60	7.48	14.18

KNEE JOINT KINEMATIC DIFFERENCES						
JOINT ROTATIONS				JOINT TRANSLATIONS		
	$FL\backslash EX$ ($^\circ$)	$AD\backslash AB$ ($^\circ$)	$IN\backslash EX$ ($^\circ$)	$ME\backslash LA$ (mm)	$AN\backslash PO$ (mm)	$CO\backslash DI$ (mm)
PERIOD A						
MAX	7.08	5.12	6.34	9.31	10.71	10.66
MEDIAN	0.51	0.02	0.06	1.70	1.21	0.68
PERIOD B						
MAX	9.28	11.80	14.76	16.84	21.62	27.60
MEDIAN	2.21	5.52	5.04	8.61	6.74	18.72

Table 4.1: (Top) Overall median and maximum range differences of the femoral and tibial ACS axes (R_x , R_y , and R_z) and origin (T). (Bottom) overall median and maximum difference in knee joint rotational and translational excursion for periods A and B.

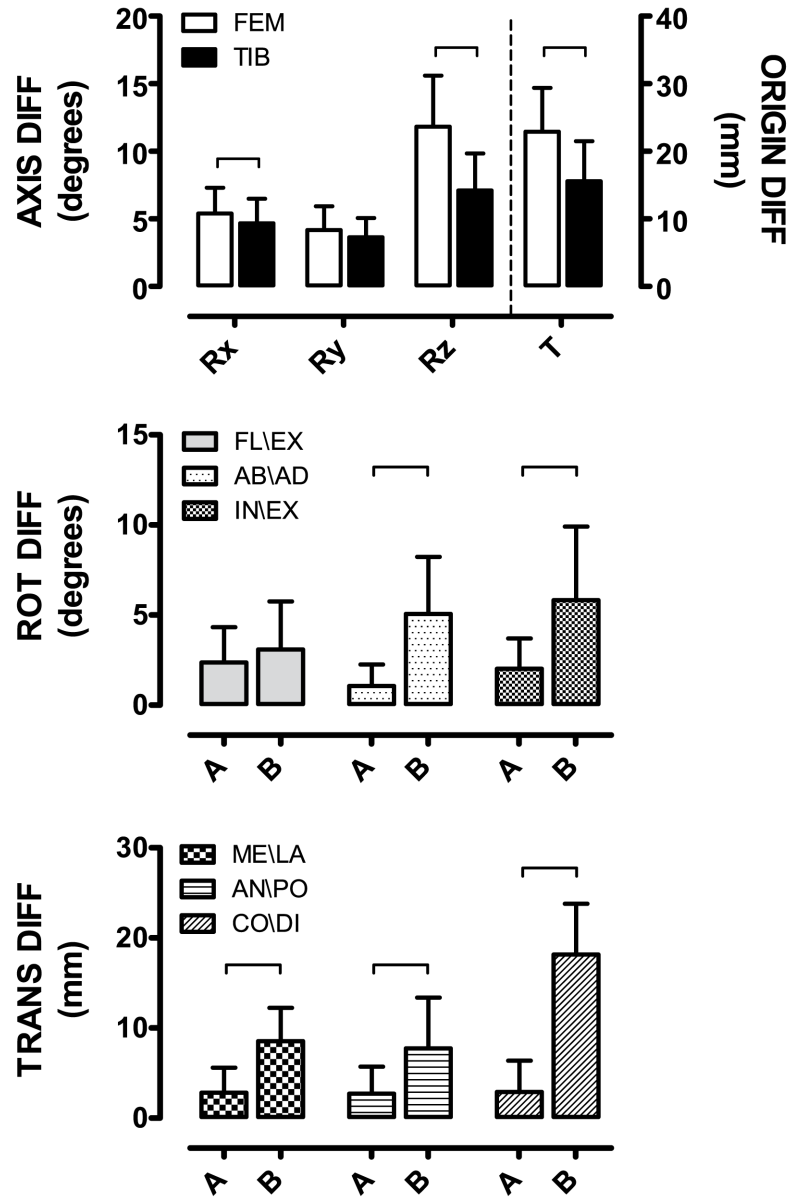


Figure 4.3: (Top) The difference between OMC and biplanar videoradiography for the three rotational axes (Rx, Ry, and Rz) and origin (T) of the independently driven femoral and tibial ACSs. (Middle) The rotation difference (FL\EX, AB\AD, IN\EX) between OMC and biplanar videoradiography. (Bottom) The translational difference (ME\LA, AN\PO, CO\DI) between OMC and biplanar videoradiography. For all knee joint rotational and translational differences, the data is displayed for the period before contact (A) and the period after contact (B). For each graph, the data is summarized using means plus standard deviations and the brackets represent significant differences ($p \leq 0.05$) between groups.

4.4 Discussion

We compared the kinematic measurements obtained from OMC and biplanar videoradiography during a jump-cut maneuver. The results support our first hypothesis that the soft tissue surrounding the femur would have a greater effect on the kinematic measurements than the soft tissue surrounding the tibia during a jump-cut maneuver. The results also support our second hypothesis that that STA would significantly influence the OMC joint kinematic measurements after ground contact (period B) during a jump-cut maneuver.

Peak OMC joint rotations and translations were observed to differ as much as 15° and 28 mm (Table 4.1), which is consistent with previous studies that report deviations of 2 - 20° and 10 - 30 mm [7,8,11,18,20]. Herein, the majority of deviations occur during the period after contact (period B), where the skin and muscle is reacting to the landing in an oscillatory manner (Figure 4.4). The OMC kinematic measurements followed the biplanar videoradiography measurements more closely during the period before contact (period A).

The overall distribution of OMC differences, as compared to biplanar videoradiography, was of similar magnitude within each period for all joint rotations and translations. For example, the FL/EX differences during period B were similar to the AB/AD and IN/EX differences during period B. These findings are especially important when evaluating secondary rotational motions such as AB/AD or IN/EX rotation, because the total range of motion is significantly less than FL/EX. This leads to OMC deviations of 100% and above for the secondary rotational motions as well as the three translational components. This amount of kinematic deviation will introduce errors in joint kinetics, determined through inverse dynamics. A study performed by Tsai et al. [20] comparing single-plane videoradiography motion capture to OMC during slow stair ascent reported significant deviations in joint moments at

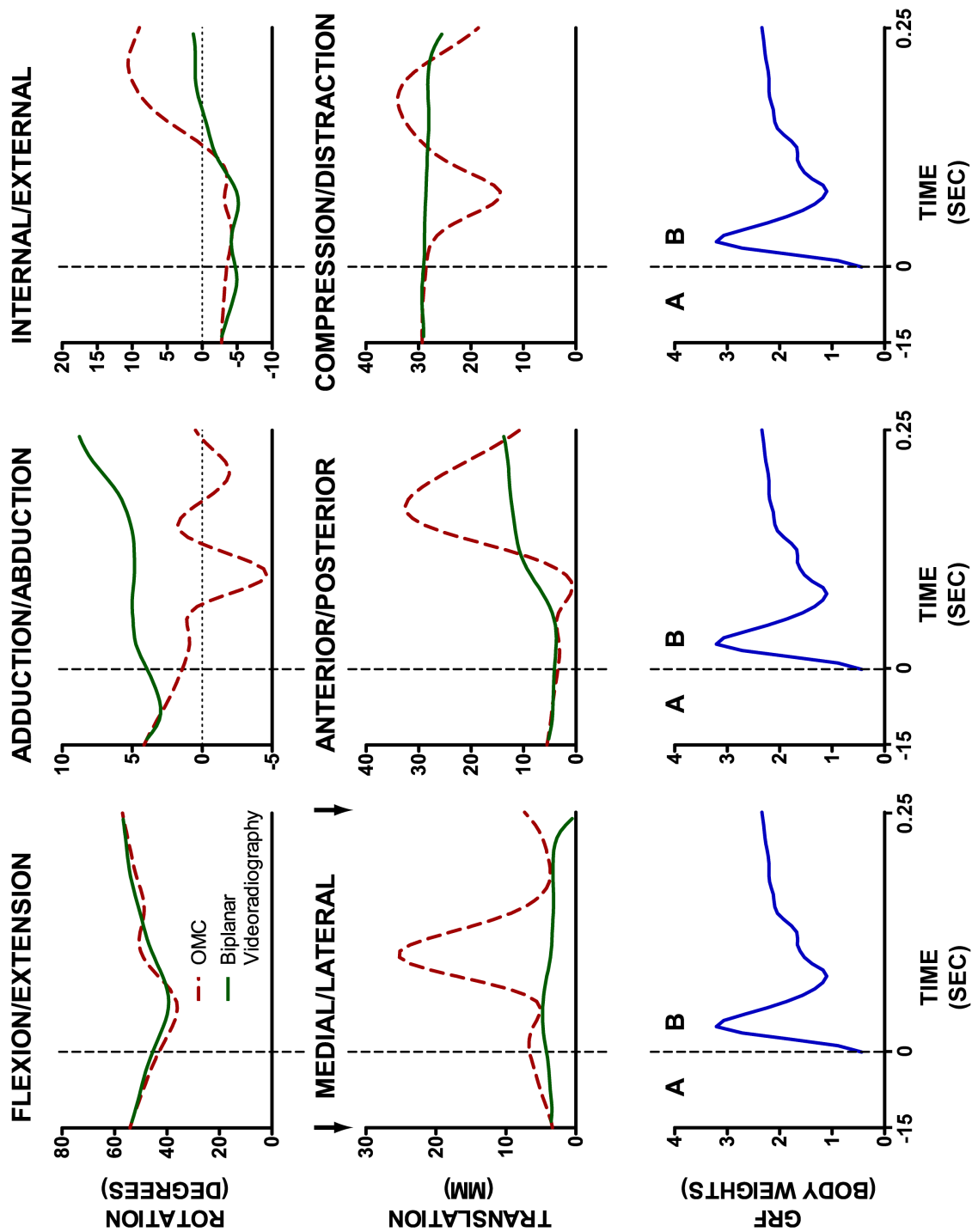


Figure 4.4: Example OMC (dotted red) and biplanar videoradiography (solid green) knee flexion angle and ground reaction force (solid blue) data versus time. The dotted vertical lines represent the contact event. The period before contact is period A and the period after contact is period B.

the knee ($> 12\%$), particularly the knee extensor moment. The effect OMC had on the abduction and internal rotational moment at the knee may have been mitigated by the restrictions of both single-plane videoradiography and the shank marker set used for OMC, which make out of plane knee kinematics difficult to elucidate. Additionally, the stair ascent activity studied may not produce impacts large enough to produce influential STA. We expect these kinetic parameters to be significantly effected, especially in high demand landing conditions such as jumping and cutting.

Currently, biplanar videoradiography is limited in its FOV and typically allows only a single joint to be imaged. Thus, a union between OMC and biplanar videoradiography is required for extracting 3-D kinetics at the knee during jumping and cutting. The data presented herein show that the soft tissue surrounding the femur affects joint kinematics more significantly than the soft tissue surrounding the tibia, which aligns with the results presented by Garling et al. [18] and Reinschmidt et al. [8]. This makes intuitive sense because the soft tissue mass surrounding the femur is more than twice that of the tibia [28]. Moreover, work from Okita et al. [29] and Reinschmidt et al. [30] support the expectation that motion artifact associated with surface markers rigidly affixed to the bony prominences of the ankle malioli and to the foot segment would be small. By using a calibration lattice for co-registration of OMC and biplanar videoradiography, we avoid using heavier radio-opaque markers on the surface of the skin and facilitate full body OMC modeling. The combined technique presented herein allows investigators to collect femur and tibia motion without STA while collecting OMC data for the foot and ankle to preserve the kinematic chain required for calculating inverse dynamics. These results highlight the importance of reporting knee joint kinetics in the tibial coordinate system to mitigate the effect of STA when using OMC alone.

Furthermore, an entire maneuver involving jumping and cutting will include kinematic

outcomes before and after landing and while beginning a cut (Figure 4.1B). These parts of the maneuver are more easily captured using OMC because the FOV is significantly larger. The combination of both OMC and biplanar videoradiography becomes advantageous in assessing the entire maneuver from start to finish. Based on the reported results, OMC can sufficiently measure the sagittal plane kinematics and kinetics before and after landing while biplanar videoradiography allows investigators to focus in on the knee during landing, where the bones, cartilage, and ligamentous structures are under the most stress.

The limitations of biplanar videoradiography have been thoroughly documented [12]. Specifically, biplanar videoradiography is not as readily available as traditional OMC, and the x-ray exposure increases the risk to subjects, albeit the total dose applied in this study was low (< 15 mrem). Furthermore, the relatively small imaging volume limits the range of activities that can be studied. Other potential study limitations include the placement of the OMC markers to optimize for the least squares technique and not the point cluster technique. Only a single jump-cut activity was tested, and results for walking, running, or other activities may be vastly different. Additionally, our subject population was relatively small and consisted of healthy recreational athletes with similar height and weight demographics. This excluded the possibility of directly correlating STA with body mass index (BMI). However, it would be expected that higher BMI would correlate with higher STA based on the results presented herein. Finally, the lack of a common and independent static calibration frame required that we co-register the two imaging modalities using a frame during the flight phase. This may produce a shift in the data along the y-axis of the Cartesian plane; however, any effects of this shift are eliminated using the excursion (max - min) analysis executed herein.

In this study, we captured an entire jump-cut maneuver with OMC while using bi-

planar videoradiography to focus on knee joint kinematics during impact with the ground. To our knowledge, this is the first use of biplanar videoradiography to assess six-DOF STA in the knee during a jump-cut maneuver. This study is a significant step toward understanding the ways OMC and biplanar videoradiography can be used together for answering questions about the biomechanics of the joint, especially during rapid movements and direction changes that are associated with injury. Additionally, this study provides a foundation for creating methods for modeling soft tissue motion in order to mitigate its affect on kinematic outcome measures. The results presented in this study should be considered when interpreting knee mechanics from OMC at and directly after impact during a jump-cut maneuver, specifically those associated with joint translations and secondary rotations (AB/AD and IN/EX). Finally, we recommend that all knee kinetics be interpreted in the tibial coordinate system based on the lower STA observed for the shank. It is our hope that a combination of OMC and biplanar videoradiography motion capture can be used to investigate entire jumping and cutting activities in order to better understand the kinematic and kinetic factors associated with non-contact deceleration ACL injury.

4.5 Conflicts of interest

None.

4.6 Acknowledgements

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4.7 References

- [1] Gianotti S. M., Marshall S. W., Hume P. A., and Bunt L., 2009, “Incidence of anterior cruciate ligament injury and other knee ligament injuries: a national population-based study,” *J Sci Med Sport*, 12(6), pp. 622–627.
- [2] Griffin L. Y., Albohm M. J., Arendt E. A., Bahr R., Beynnon B. D., Demaio M., Dick R. W., Engebretsen L., Garrett W. E., Hannafin J. A., Hewett T. E., Huston L. J., Ireland M. L., Johnson R. J., Lephart S., Mandelbaum B. R., Mann B. J., Marks P. H., Marshall S. W., Myklebust G., Noyes F. R., Powers C., Shields C., Shultz S. J., Silvers H., Slauterbeck J., Taylor D. C., Teitz C. C., Wojtys E. M., and Yu B., 2006, “Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting, January 2005,” *Am J Sports Med*, 34(9), pp. 1512–1532.
- [3] McNair P. J., Marshall R. N., and Matheson J. A., 1990, “Important features associated with acute anterior cruciate ligament injury,” *N. Z. Med. J.*, 103(901), pp. 537–539.
- [4] Junkin D. M., Johnson D. J., Fu F. H., Mark M. D., Willenborn M., Fanelli G. C., and Wascher D. W., 2009, “Knee Ligament Injuries,” *Orthopaedic Knowledge Update: Sports Medicine 4*, Amer Academy of Orthopaedic, pp. 135–153.

- [5] Boden B. P., Dean G. S., Feagin J. A., and Garrett W. E., 2000, "Mechanisms of anterior cruciate ligament injury," *Orthopedics*, 23(6), pp. 573–578.
- [6] Ford K. R., Myer G. D., Toms H. E., and Hewett T. E., 2005, "Gender differences in the kinematics of unanticipated cutting in young athletes," *Med Sci Sports Exerc*, 37(1), pp. 124–129.
- [7] Cappozzo A., Catani F., Leardini A., Benedetti M., and Della Croce U., 1996, "Position and orientation in space of bones during movement: experimental artefacts," *Clinical Biomechanics*, 11(2), pp. 90–100.
- [8] Reinschmidt C., van den Bogert A. J., Nigg B. M., Lundberg A., and Murphy N., 1997, "Effect of skin movement on the analysis of skeletal knee joint motion during running," *J Biomech*, 30(7), pp. 729–732.
- [9] Andriacchi T. P., Alexander E. J., Toney M. K., Dyrby C., and Sum J., 1998, "A point cluster method for in vivo motion analysis: applied to a study of knee kinematics," *J Biomech Eng*, 120(6), pp. 743–749.
- [10] Lafortune M. A., Cavanagh P. R., Sommer H. J., and Kalenak A., 1992, "Three-dimensional kinematics of the human knee during walking," *J Biomech*, 25(4), pp. 347–357.
- [11] Holden J. P., Orsini J. A., Siegel K. L., Kepple T. M., Gerber L. H., and Stanhope S. J., 1997, "Surface movement errors in shank kinematics and knee kinetics during gait," *Gait & Posture*, 5(3), pp. 217–227.
- [12] Tashman S., and Anderst W., 2003, "In-vivo measurement of dynamic joint motion using high speed biplane radiography and CT: application to canine ACL deficiency," *J Biomech Eng*, 125(2), pp. 238–245.
- [13] Brainerd E. L., Baier D. B., Gatesy S. M., Hedrick T. L., Metzger K. A., Gilbert S. L., and Crisco J. J., 2010, "X-ray reconstruction of moving morphology (XROMM): precision, accuracy and applications in comparative biomechanics research," *J Exp Zool A Ecol Genet Physiol*, 313(5), pp. 262–279.
- [14] Miranda D. L., Schwartz J. B., Loomis A. C., Brainerd E. L., Fleming B. C., and Crisco J. J., 2011, "Static and Dynamic Error of a Biplanar Videoradiography System Using Marker-Based and Markerless Tracking Techniques," *J Biomech Eng*, 133(12), p. 121002.
- [15] Torry M. R., Shelburne K. B., Peterson D. S., Giphart J. E., Krong J. P., Myers C., Steadman J. R., and Woo S. L.-Y., 2011, "Knee kinematic profiles during drop landings: a biplane fluoroscopy study," *Med Sci Sports Exerc*, 43(3), pp. 533–541.
- [16] Myers C. A., Torry M. R., Peterson D. S., Shelburne K. B., Giphart J. E., Krong J. P., Woo S. L.-Y., and Steadman J. R., 2011, "Measurements of Tibiofemoral

Kinematics During Soft and Stiff Drop Landings Using Biplane Fluoroscopy,” *Am J Sports Med.*

[17] Taylor K. A., Terry M. E., Utturkar G. M., Spritzer C. E., Queen R. M., Iribarra L. A., Garrett W. E., and DeFrate L. E., 2011, “Measurement of in vivo anterior cruciate ligament strain during dynamic jump landing,” *J Biomech*, 44(3), pp. 365–371.

[18] Garling E. H., Kaptein B. L., Mertens B., Barendregt W., Veeger H. E. J., Nelissen R. G. H. H., and Valstar E. R., 2007, “Soft-tissue artefact assessment during step-up using fluoroscopy and skin-mounted markers,” *Journal of Biomechanics*, 40(Supplement 1), pp. S18–S24.

[19] Akbarshahi M., Schache A. G., Fernandez J. W., Baker R., Banks S., and Pandy M. G., 2010, “Non-invasive assessment of soft-tissue artifact and its effect on knee joint kinematics during functional activity,” *J Biomech*, 43(7), pp. 1292–1301.

[20] Tsai T.-Y., Lu T.-W., Kuo M.-Y., and Lin C.-C., 2011, “Effects of soft tissue artifacts on the calculated kinematics and kinetics of the knee during stair-ascent,” *J Biomech*, 44(6), pp. 1182–1188.

[21] Tegner Y., and Lysholm J., 1985, “Rating systems in the evaluation of knee ligament injuries,” *Clin. Orthop. Relat. Res.*, (198), pp. 43–49.

[22] Buczek F. L., Rainbow M. J., Cooney K. M., Walker M. R., and Sanders J. O., 2010, “Implications of using hierarchical and six degree-of-freedom models for normal gait analyses,” *Gait & Posture*, 31(1), pp. 57–63.

[23] Miranda D. L., Rainbow M. J., Leventhal E. L., Crisco J. J., and Fleming B. C., 2010, “Automatic determination of anatomical coordinate systems for three-dimensional bone models of the isolated human knee,” *J Biomech*, 43(8), pp. 1623–1626.

[24] Söderkvist I., and Wedin P. A., 1993, “Determining the movements of the skeleton using well-configured markers,” *J Biomech*, 26(12), pp. 1473–1477.

[25] Challis J. H., 1995, “A procedure for determining rigid body transformation parameters,” *J Biomech*, 28(6), pp. 733–737.

[26] Grood E. S., and Suntay W. J., 1983, “A joint coordinate system for the clinical description of three-dimensional motions: application to the knee,” *J Biomech Eng*, 105(2), pp. 136–144.

[27] Deneweth J. M., Bey M. J., McLean S. G., Lock T. R., Kolowich P. A., and Tashman S., 2010, “Tibiofemoral joint kinematics of the anterior cruciate ligament-reconstructed knee during a single-legged hop landing,” *Am J Sports Med*, 38(9), pp. 1820–1828.

- [28] Cheng C.-K., Chen H.-H., Chen C.-S., Lee C.-L., and Chen C.-Y., 2000, “Segment inertial properties of Chinese adults determined from magnetic resonance imaging,” *Clinical Biomechanics*, 15(8), pp. 559–566.
- [29] Okita N., Meyers S. A., Challis J. H., and Sharkey N. A., 2009, “An objective evaluation of a segmented foot model,” *Gait Posture*, 30(1), pp. 27–34.
- [30] Reinschmidt C., van Den Bogert A. J., Murphy N., Lundberg A., and Nigg B. M., 1997, “Tibiocalcaneal motion during running, measured with external and bone markers,” *Clin Biomech (Bristol, Avon)*, 12(1), pp. 8–16.

Chapter 5

Knee biomechanics during a jump-cut maneuver: effects of gender and ACL surgery

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Abstract

The purpose of this study was to compare kinetic and kinematic measurements of the knee from male and female ACL-intact (ACL_{INT}) and ACL-reconstructed (ACL_{REC}) subjects during a jump-cut maneuver using biplanar videoradiography. Twenty subjects were recruited, 10 ACL_{INT} (5 males, 5 females) and 10 ACL_{REC} (4 males, 6 females; five years post surgery). Each subject performed a jump-cut maneuver by landing on a single leg and performing a 45° side-step cut. Ground reaction force was measured by a force plate and expressed relative to body weight. Six-degree-of-freedom knee kinematics were determined from a biplanar videoradiography system and an optical motion capture system. ACL_{INT} female subjects landed with a larger peak vertical GRF ($p < 0.001$) compared to ACL_{INT} male subjects. ACL_{INT} subjects landed with a larger peak vertical GRF ($p \leq 0.036$) compared to ACL_{REC} subjects. Regardless of ACL reconstruction status, female subjects underwent less knee flexion angle excursion ($p = 0.002$) and had an increased average rate of anterior tibial translation ($p = 0.037$) after contact compared to male subjects. Furthermore, ACL_{REC} subjects had a lower rate of anterior tibial translation compared to ACL_{INT} subjects ($p = 0.035$). Finally, no striking differences were observed in other knee motion parameters. Our results show that women permit a smaller amount of knee flexion angle excursion during a jump-cut maneuver, resulting in a larger peak vertical GRF and an increased rate of anterior tibial translation. Notably, ACL_{REC} subjects five years post surgery also perform the jump cut maneuver with lower GRF than ACL_{INT} subjects. This study proposes a causal sequence whereby increased landing stiffness (larger peak vertical GRF combined with less knee flexion angle excursion) leads to an increased rate of anterior tibial translation while performing a jump-cut maneuver.

Keywords: *kinematics, kinetics, landing stiffness, ground reaction force, anterior tibial translation, biplanar videoradiography*

5.1 Introduction

Injuries to the anterior cruciate ligament (ACL) are commonly associated with sport maneuvers involving jumping, landing, and cutting [1]. These maneuvers result in a sudden loading of the ACL due to the deceleration of the tibia that occurs after landing but just prior to a rapid direction change [2]. Approximately 70% of ACL injuries occur during deceleration maneuvers without contact from another athlete [3]. Although males suffer non-contact deceleration injury, females are reported to be up to ten times more prone when participating in the same high-risk activities [4]. Although many theories exist, the ACL failure mechanism and the associated gender bias remain unclear.

During normal function, the ACL restrains excessive anterior tibial translation and stabilizes secondary knee rotations (i.e., internal/external and abduction/adduction) [5]. ACL reconstruction has become the gold standard of treatment for athletes with an ACL tear in an attempt to restore joint stability and to return patients to a high functional level [6]. Unfortunately, of the 400,000 patients that undergo ACL reconstruction in the United States each year, up to 5% are at risk for re-injury [7], 45% fail to return to their pre-injury sport level [8], and 80% to 90% will develop radiographic evidence of osteoarthritis as early as seven years post surgery [9].

Given the unexplained greater risk of non-contact deceleration ACL injury in female subjects, any differences between gender and ACL reconstruction status in the kinematic and kinetic factors during associated sport activities may point to root causes for injury and re-injury, and avenues for prevention and rehabilitation. Unfortunately, the biomechanics of male and female ACL-intact (ACL_{INT}) and ACL-reconstructed (ACL_{REC}) knees during high risk non-contact deceleration activities, such as a jump-cut maneuver, are not well understood. These data have previously been difficult to obtain, in part, because non-invasive measurement of kinematics has been limited to

optical motion capture (OMC), which depend on surface markers that are prone to artifact from soft tissue oscillation immediately following landing [10].

Biplanar videoradiography, however, allows for direct measurement of *in vivo* bone motion, circumventing the effect of soft tissue artifact [11–17]. Biplanar videoradiography has recently been used to study dynamic ACL_{INT} and ACL_{REC} knee motion during running [11,12], two-legged drop landings [13–16], and single-leg hopping [17]. While these studies have made significant contributions to our understanding of both ACL_{INT} and ACL_{REC} knee function during running, drop landing, and hopping, the combined jump-cut maneuver, which is more commonly associated with non-contact deceleration ACL injury, has not been investigated [2,18]. Additionally, the biomechanics of ACL_{REC} subjects during these other dynamic tasks were investigated between four and twelve months after surgery [11,12,17]. While these time points are crucial for quantifying the immediate effects of ACL reconstruction, understanding the biomechanics of the knee more than five years after surgery may provide further insight into the long-term recovery process.

The purpose of this study was to compare force plate kinetic data and knee kinematic measurements from male and female ACL_{INT} and ACL_{REC} recreational athletes during a jump-cut maneuver in hopes that differences would point to plausible risk factors for injury. Knee kinematic measurements were primarily obtained from biplanar videoradiography; however, knee flexion/extension outside the field of view of the biplanar videoradiography system was obtained from traditional optical motion capture. The specific aims were to determine differences due to both gender and ACL reconstruction status between ACL_{REC} patients who were at least five years post-surgery and ACL_{INT} control subjects. More specifically, it was anticipated that ACL_{INT} women would tend to perform the jump-cut maneuver more upright and with more landing stiffness than ACL_{INT} men. This would be evident as decreased knee flexion angle

excursion and increased peak vertical ground reaction force (GRF), relative to their body weight, resulting in greater tibial translation (particularly anterior). In contrast, it was not known whether or not ACL_{REC} females and males five years post reconstruction would follow a similar pattern or if their injury and subsequent repair and rehabilitation would have resulted in altered kinetic and kinematic parameters (tested as an interaction between gender and ACL reconstruction status).

5.2 Methods

5.2.1 Subjects

All experimental procedures were approved by the Institutional Review Board. Twenty recreational athletes were enrolled in this study. Of these subjects, 10 were ACL_{INT} (5 males, 5 females) and 10 were ACL_{REC} (4 males, 6 females; 7 bone-patellar tendon-bone autografts, 3 hamstring tendon autografts). Age, weight, and height for all subjects are displayed in Table 5.1. The inclusion criteria for the ACL_{INT} subjects were: (1) no history of lower extremity injury; (2) no neurological disease(s); (3) no pregnancy; and (4) a Tegner activity score of five or greater [19]. It should be noted that the ACL_{INT} subjects were part of a separate study investigating the effects of soft tissue artifact on kinematic outcomes during a combined jump-cut maneuver [10]. The inclusion criteria for the ACL_{REC} patients were: (1) unilateral ACL reconstruction using bone-patellar tendon-bone or four-stranded hamstring tendon autograft (looped semitendinosus and gracilis); (2) at least five years post ACL reconstruction; (3) no systemic infection; (4) no neurological disease(s); (5) no pregnancy; and (6) a Tegner activity score of five or greater. The ACL reconstruction surgery type was confirmed from patient records. After granting their informed consent, each subject

was outfitted with 23 retro-reflective surface markers on a single leg to permit measurement of foot, shank, and thigh motion using OMC [20]. The outfitted leg was chosen at random for the ACL_{INT} subjects (6L and 4R). For the ACL_{REC} subjects, the ACL reconstructed leg was outfitted (7L and 3R).

5.2.2 Jump-Cut Maneuver

Each subject performed a jump-cut maneuver that was adapted from Ford et al. [18], and previously described in detail [10]. Briefly, three targets were placed on the floor within the testing environment (Figure 5.1A). The first target was located in the center of a force plate (model 9281B; Kistler, Amherst, NY, USA). The other two targets were placed toward the left and right of the landing target at an angle of 45°. Before beginning the maneuver, the subject was asked to stand approximately one meter from the force plate with their knees bent approximately 45°. Upon hearing a verbal “GO” prompt, the subject jumped upward and forward toward the first landing target. At the same time as the verbal “GO” prompt, a visual directional prompt, left (L) or right (R), cued the subject as to which direction to cut after landing on the target with one leg. Upon landing, the subject performed a sidestep cut and then jogged past the respective angled targets. For example, if a subject was prompted to cut to the left they would land, cut, and push-off with their right leg. A trial was excluded if the subject incorrectly performed the jump-cut maneuver (landing outside the target area, incorrect cut direction, crossover cut, etc...). A total of ten correctly executed trials were performed, and the subject was unaware of the directional prompt prior to a given trial.

PARAMETER	ACL STATUS	MEAN	SEM	GENDER	MEAN	SEM	INTERACTION	MEAN	SEM
AGE (YEARS)	ACL _{INT}	25.20	1.64	M	27.53	1.74	INT x M	25.80	2.32
	ACL _{REC}	26.96	1.67		24.63	1.57	INT x F	24.60	2.32
				F			REC x M	29.25	2.59
							REC x F	24.67	2.12
MASS (KG)	ACL _{INT}	73.16	2.93	M	84.88	3.11	INT x M	81.53	4.15
	ACL _{REC}	78.26	2.99		66.55	2.81	INT x F	65.07	4.15
				F			REC x M	88.49	4.64
							REC x F	68.03	3.79
HEIGHT (CM)	ACL _{INT}	172.85	2.09	M	178.70	2.21	INT x M	177.40	2.95
	ACL _{REC}	173.88	2.13		168.03	2.00	INT x F	168.30	2.95
				F			REC x M	180.00	3.30
							REC x F	167.75	2.70

Table 5.1: ACL_{INT} and ACL_{REC} male and female age, mass, and height demographics. Ages are in years, mass is in kilograms, and height is in centimeters. A two-way analysis of variance was performed on each set of demographics data, and no statistically significant differences ($p \leq 0.05$) were found between gender and condition. Statistically significant differences ($p \leq 0.05$) are highlighted using a dark gray background fill.

5.2.3 Data Collection and Processing

The jump-cut maneuvers were carried out and kinetic and kinematic data were gathered in the W.M. Keck Foundation XROMM Facility at Brown University (Providence, RI, USA). An OMC system (Oqus 5; Qualisys, Gothenburg, Sweden) was used to track the retro-reflective surface markers on each subject's outfitted leg during the entire jump-cut maneuver at a capture rate of 250 Hz. A force plate (model 9281B; Kistler, Amherst, NY, USA) was used to measure the GRF at 5,000 Hz. The biplanar videoradiography system was engaged for a maximum of six trials and measured motion at 250 Hz within a restricted field of view (FOV) above the force plate [21]. This was done to reduce radiation exposure and maximize the likelihood that the jump-cut maneuver occurred within the FOV of the biplanar videoradiography system. All devices were time synchronized. Image de-distortion and 3-D space calibration followed established protocols using custom MATLAB software (XrayProject; Brown University, Providence, RI, USA) [22].

Additionally, a single static clinical computed tomography (CT) scan was collected for each subject's outfitted knee. Image volumes were captured in the axial plane at 80 kVp while using GE's SMART mA and Bone Plus reconstruction algorithms. The voxel resolution (slice thickness and in-plane resolution) for each scan was less than $0.625 \times 0.465 \times 0.465$ mm³. The voxels corresponding to the femur and tibia were isolated from each CT volume using previously described methods [23] implemented in commercially available image segmentation software (Mimics v14; Materialise, Ann Arbor, MI, USA).

Custom markerless tracking software (Autoscooper; Brown University, Providence, RI) was used to process the biplanar videoradiography data [21]. Briefly, isolated CT volumes for the femur and tibia were input into a virtual 3-D environment containing the biplanar videoradiography sequences and their calibration information. Digitally

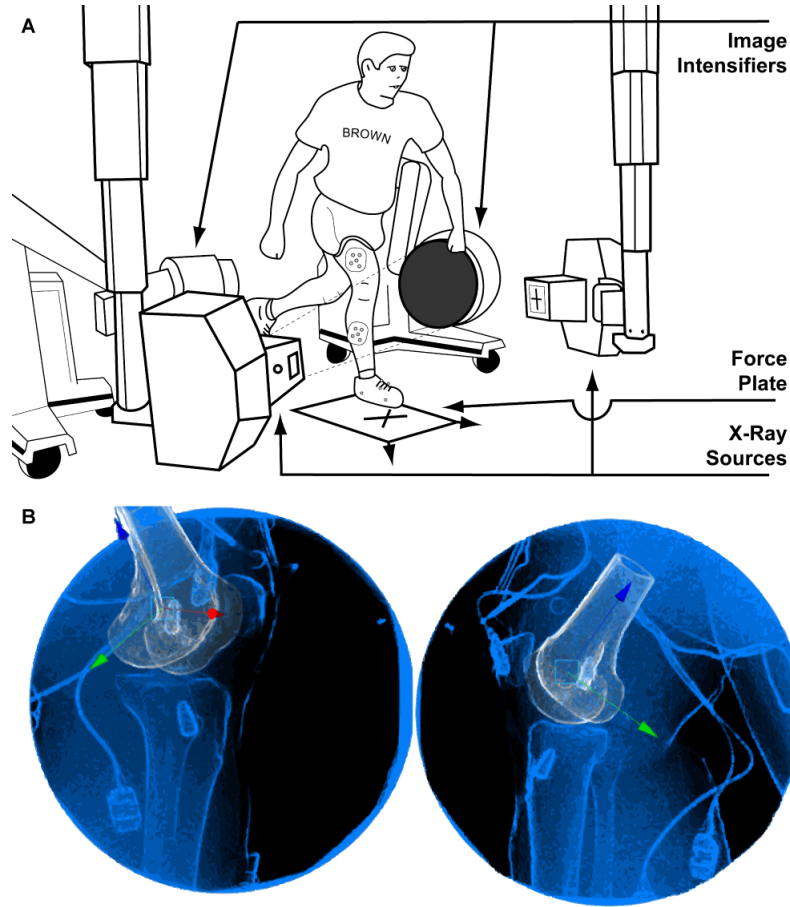


Figure 5.1: (A) Illustration depicting the experimental set-up used to capture both biplanar videoradiography and OMC data during a jump-cut maneuver. A screen directly in front of the subject prompted them with the directional arrow. The subject would land and cut in the direction they were prompted using the opposite leg. For example, if prompted with the left arrow, the subject would land and cut to the left using their right leg. The four OMC cameras are not displayed in this figure; however, they were positioned to capture the retro-reflective markers shown on the subject's right leg. (B) An example frame from the Autoscopers markerless tracking software. Each view represents one frame from each of the two videoradiographs generated from the two image intensifiers (Figure 5.1A). The blue and black portions of the images represent the actual radiographs. The orange femur represents the DRR. Both the DRR and videoradiographs have been enhanced with a sobel edge detection filter and a contact filter. This was done to create a strong visual match between the DRR and the actual radiograph. The translational manipulator is shown. This manipulator allowed the user to translate the DRR within the 3-D environment. A rotational manipulator was also available to the user. The DRR is shown here after performing markerless registration. The knee shown in this image is from one of the ACL_{REC} subjects. Both interference screws are visible in the femur and tibia.

reconstructed radiographs (DRRs) were generated from the CT volumes, and the kinematic transforms from CT space to each radiograph frame were determined after optimally matching the DRRs with the two views from the biplanar videoradiography system (Figure 5.1B). It has previously been shown that *in vivo* bone motion can be determined within 0.25° and 0.25 mm using these methods [21,24]. Furthermore, the rotational and translational tracking precision for this study was estimated at 0.08° and 0.45 mm, respectively.

The retroreflective marker data from the OMC system were filtered using a digital low-pass Butterworth filter with a cutoff frequency of 25 Hz. The kinematic transforms of the femur and tibia obtained from the biplanar videoradiography system were converted into quaternions. A quaternion is represented by four parameters that can be filtered [25]. A digital Butterworth filter with a 25 Hz cutoff frequency was applied to the three kinematic translation parameters and the four quaternion parameters. The filtered quaternion parameters were converted back to rotation matrices and recombined with the filtered kinematic translations. The GRF data was filtered using a digital Butterworth filter with a 100 Hz cutoff frequency.

5.2.4 Data Analysis

For comparison between subjects, the vertical GRF was normalized by body weight. A characteristic peak (Figure 5.2A) was observed in the vertical GRF within the first 25 milliseconds. This peak vertical GRF was quantified by its time after contact (peak vertical GRF time) and its magnitude (peak vertical GRF magnitude).

The kinematics of the tibia with respect to the femur were described for both OMC and biplanar videoradiography data sets using two independent anatomical coordinate systems (ACSs). These ACSs were determined from the 3-D CT models of the femur

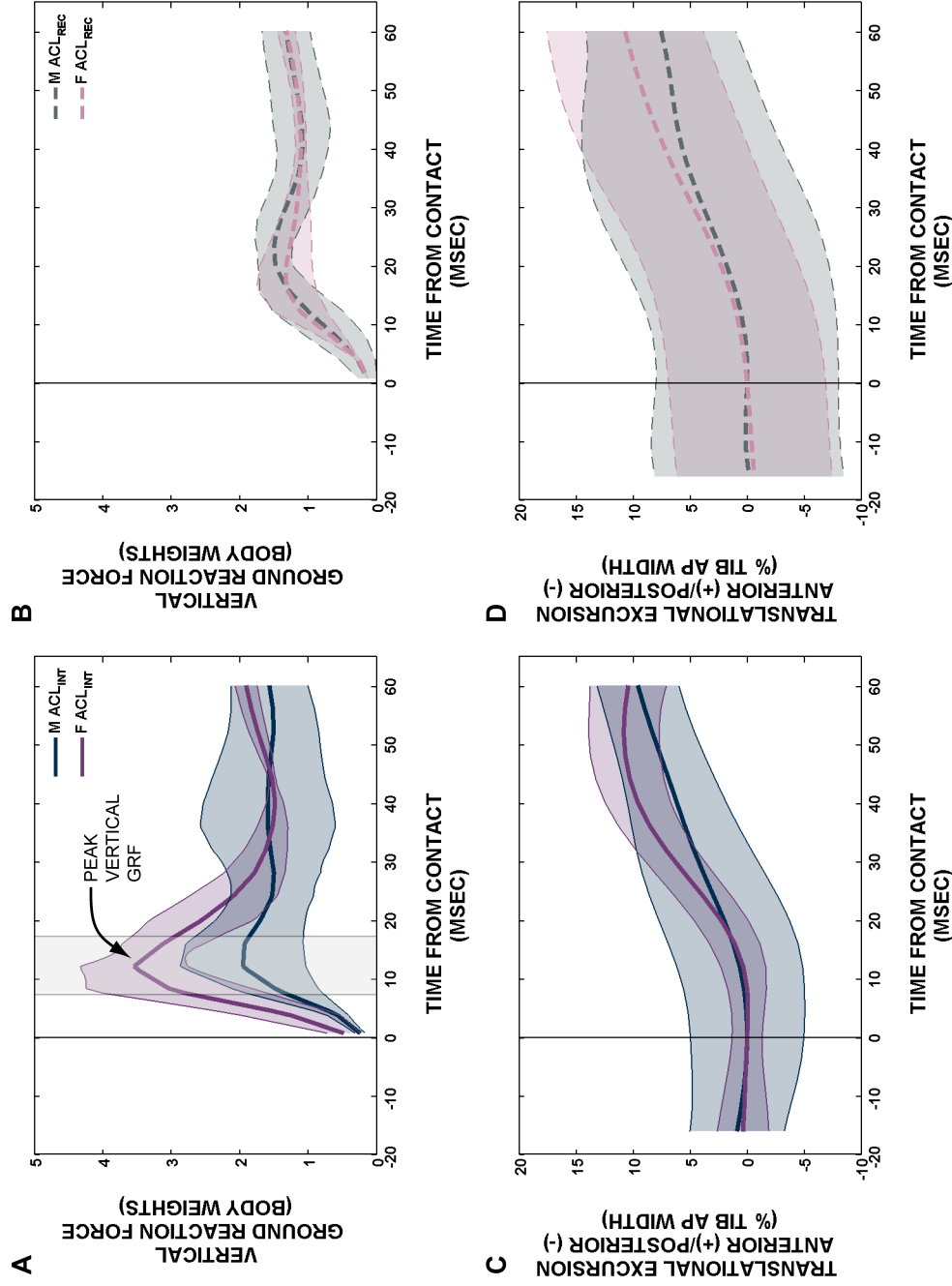


Figure 5.2: (A) ACL_{INT} vertical GRF. (B) ACL_{REC} vertical GRF. Each subject's GRF was normalized by their respective weight. Thus, vertical GRF units are in body weights. Notice the highlighted peak in the ACL_{INT} vertical GRF graph. All curves are displayed as mean ± 1 SD. The vertical line on each graph represents the time at contact. (C) ACL_{INT} AN/PO translational excursion. (D) ACL_{REC} AN/PO translational excursion. Anterior is positive and posterior is negative. All AN/PO translations were normalized for each subject by their respective tibial plateau width. Thus, translational units are defined as a percent of the total tibial width in the anterior-posterior direction. It should be noted that the AN/PO translational excursion data were obtained from the biplanar videoradiography system.

and tibia using previously described methods [23]. In order to use the same ACSs for both OMC and biplanar videoradiography, their global coordinate spaces were co-registered using a rigid lattice containing spherical markers that were radio-opaque and retro-reflective [10]. The mean and standard deviation for the root mean square fit error of the co-registration transforms was 0.31 ± 0.09 mm.

Knee joint rotations in flexion/extension (FL/EX), adduction/abduction (AD/AB), and internal/external (IN/EX) rotations of the tibia relative to the femur were interpreted using the method described by Grood and Suntay [26]. Joint translations in medial/lateral (ME/LA) and anterior/posterior (AN/PO) displacements of the tibia relative to the femur were determined by a vector originating at the origin of femoral ACS and terminating at the origin of the tibial ACS [17]. The ME/LA and AN/PO translations were normalized for each subject according to the ME/LA or AN/PO width of their tibial plateau, similar to the method reported by Tanifuji et al. [27]. These translations are interpreted as percent ME/LA or AN/PO tibial plateau width. Normalization was performed in order to make kinematic evaluations on individuals of different sizes.

Due to the limited field of view (FOV) of the biplanar videoradiography system, the joint rotations and translations were time normalized from 16 milliseconds prior to contact to 60 milliseconds after contact. This window was selected since it was common to all subjects for at least one trial. For comparison, all joint rotations and translations were zeroed at contact and interpreted as excursion. The average rate and maximum rate of AN/PO excursion was determined for each subject. Average rate was calculated as the total range divided by the change in time, and the maximum rate was calculated as the maximum time derivative. Additionally, the area under the curve (AUC), which simplifies time-series curve comparisons, was calculated for each time-series kinematic excursion trace by integrating the signal with respect to

time.

The FOV of the OMC system is significantly larger than that of the biplanar videoradiography system, permitting the measurement of knee FL/EX angle outside the time period containing the biplanar videoradiography data. Despite the soft tissue artifact observed in secondary rotations (AB/AD, IN/EX rotation) and translations (ME/LA, AN/PO) obtained from OMC, FL/EX remains relatively unaffected [10]. Using the OMC data, the minimum flexion angle after contact was determined. Additionally, the change from minimum flexion angle to maximum flexion angle after contact was calculated and interpreted as excursion. The OMC data were presented for only knee joint FL/EX. The biplanar videoradiography data are presented for all other kinematic parameters (AD/AB and IN/EX rotations, ME/LA and AN/PO translations).

The described kinematic and GRF outcomes were determined for each applicable subject trial, and then all trials were ensemble averaged for each subject. Comparisons between gender (M and F) and ACL reconstruction status (ACL_{INT} and ACL_{REC}) were made for all kinematic and kinetic variables using two way analyses of variance. These tests were performed with a significance level (alpha) of 0.05. Pairwise multiple comparisons were made using the Holm-Sidak method when a significant gender and ACL reconstruction status interaction was determined. The Holm-Sidak method maintains alpha at 0.05 across a set of hypothesis tests and adjusts p-values differently depending on their values ranked against each other. This is effective at maintaining alpha and avoiding beta inflation.

5.3 Results

A statistically significant interaction ($p = 0.003$) between gender and ACL reconstruction status was observed for the peak vertical GRF (Figure 5.2). Within the ACL_{INT} subjects, the females had a peak vertical GRF that was 1.45 body weights larger than the male subjects ($p < 0.001$). In contrast, the ACL_{REC} male and female subjects had nearly equal peak vertical GRFs. The ACL_{REC} male subjects' peak vertical GRF was 0.22 body weights larger than the female ACL_{REC} subjects but was not statistically significant ($p = 0.522$). When comparing within ACL reconstruction status, both the male and female ACL_{INT} subjects had a larger peak vertical GRF than the male and female ACL_{REC} subjects, respectively. The male ACL_{INT} subjects' peak vertical GRF were 0.80 body weights larger than the male ACL_{REC} subjects ($p = 0.036$), and the female ACL_{INT} subjects' peak vertical GRF were 2.46 body weights larger than the female ACL_{REC} subjects ($p < 0.001$).

The peak vertical GRF for the female subjects occurred 6.24 milliseconds earlier than the male subjects ($p = 0.021$). The interaction between gender and ACL reconstruction status approached, but was not statistically significant ($p = 0.117$); the peak vertical GRF for the female ACL_{REC} subjects occurred only 2.21 milliseconds before the male ACL_{REC} subjects. Conversely, the peak vertical GRF for the female ACL_{INT} subjects occurred 10.27 milliseconds before the male ACL_{INT} subjects. Moreover, the peak vertical GRF appears to occur earlier in ACL_{INT} subjects than ACL_{REC} subjects (4.80 milliseconds; $p = 0.066$).

The average rate of AN/PO translational excursion (Figure 5.2) was 0.05 %/millisecond larger for ACL_{INT} subjects compared to ACL_{REC} subjects ($p = 0.035$), and 0.05 %/millisecond larger for female subjects compared to male subjects ($p = 0.037$). The maximum rate of AN/PO translational excursion was 0.13 %/millisecond larger in female ACL_{INT} subjects as compared to male ACL_{INT} subjects; however, the difference

OUTCOME	ACL STATUS	MEAN	SEM	GENDER	MEAN	SEM	INTERACTION	MEAN	SEM
AD\AB EXCURSION AUC	ACL _{INT}	-5.18	13.74	M	26.03	14.58	INT x M	25.88	19.43
	ACL _{REC}	11.57	14.03	F	-19.65	13.16	INT x F	-36.24	19.43
							REC x M	26.19	21.73
IN\EX EXCURSION AUC	ACL _{INT}	76.59	32.91	M	38.06	34.21	REC x F	-3.05	17.74
	ACL _{REC}	10.59	32.25	F	49.04	30.88	INT x M	86.81	45.61
							INT x F	66.37	45.61
ME\LA EXCURSION AUC	ACL _{INT}	6.56	12.65	M	-17.35	13.41	REC x M	-10.63	50.99
	ACL _{REC}	-20.57	12.91	F	3.34	12.11	REC x F	31.71	41.63
		(-15.92)	(9.50)		(2.319)	(8.91)	INT x M	-6.85 (-5.72)	17.89 (13.16)
AN\PO EXCURSION AUC	ACL _{INT}	303.85	27.417	M	244.38	29.08	INT x F	19.97 (13.83)	17.89 (13.16)
	ACL _{REC}	(162.80)	(15.08)	F	(139.89)	(15.99)	REC x M	-27.84 (22.79)	20.00 (14.72)
		248.96	27.98		308.42	26.25	REC x F	-13.292 (-9.20)	16.33 (12.02)
AN\PO AVERAGE RATE	ACL _{INT}	0.22	0.02	M	0.17	0.02	INT x M	270.85 (150.74)	38.77 (21.33)
	ACL _{REC}	0.17	0.02	F	0.22	0.01	INT x F	336.84 (174.85)	38.77 (21.33)
		(0.10)	(0.01)		(0.12)	(0.01)	REC x M	217.91 (129.04)	43.35 (23.84)
AN\PO MAXIMUM RATE	ACL _{INT}	0.39	0.03	M	0.31	0.03	REC x F	280.01 (148.08)	35.40 (19.47)
	ACL _{REC}	0.30	0.03	F	0.37	0.03	INT x M	0.18 (0.10)	0.02 (0.01)
		(0.17)	(0.02)		(0.19)	(0.01)	INT x F	0.26 (0.13)	0.02 (0.01)
PEAK MAG BW**	ACL _{INT}	1.53	0.17	M	2.04	0.17	REC x M	0.16 (0.10)	0.02 (0.01)
	ACL _{REC}	3.16	0.16	F	2.65	0.16	REC x F	0.18 (0.10)	0.02 (0.01)
							INT x M	0.32 (0.18)	0.04 (0.02)
PEAK TIME	ACL _{INT}	16.73	1.70	M	22.25	1.80	INT x F	0.45 (0.23)	0.04 (0.02)
	ACL _{REC}	21.53	1.74	F	16.01	1.63	REC x M	0.30 (0.18)	0.04 (0.02)
							REC x F	0.29 (0.16)	0.03 (0.02)
* MvF in I: p=0.027; MvF in R: p=0.863; IvR in M: p=0.751; IvR in F: p=0.007									
** MvF in I: p<0.001; MvF in R: p=0.522; IvR in M: p=0.036; IvR in F: p<0.001									

Table 5.2: ACL_{INT} and ACL_{REC} male and female kinematic AUC, rate, and peak vertical GRF results. A two-way analysis of variance was performed on each set of data. Statistically significant differences ($p \leq 0.05$) are highlighted using a dark gray background fill. Near statistical significance ($p \leq 0.10$) is highlighted using a light gray background fill. Pair-wise comparisons were made using the Holm-Sidak method when a significant (or near significant) gender and ACL reconstruction status interaction was determined. These results, denoted by the * or **, are shown in the bottom row of the table. The data presented in the parentheses (ME/LA and AN/PO AUC, AN/PO rates) are the raw translational values in millimeters. Finally, it should be noted that the kinematic data presented in Table 2 were obtained from the biplanar videoradiography system.

between genders in ACL_{REC} subjects was only 0.01 %/millisecond. Pairwise multiple comparisons revealed that maximum AN/PO translational excursion rate differences were significant for males versus females within ACL_{INT} subjects ($p = 0.027$) and for ACL_{INT} versus ACL_{REC} within female subjects ($p = 0.007$). Additionally, the AUC of the AN/PO translational excursion was observed to be 55 % · millisecond larger for ACL_{INT} subjects than ACL_{REC} subjects ($p = 0.180$). The AUC for the female subjects was also larger than the AUC for the male subjects (64 % · millisecond; $p = 0.122$). No significant interaction between gender and ACL reconstruction status was observed ($p = 0.961$). However, the AUC of the AN/PO translational excursion was observed to be 66 % · millisecond larger for female ACL_{INT} subjects than for male ACL_{INT} subjects, and the AUC for the female ACL_{REC} subjects was also larger than the AUC for the male ACL_{REC} subjects (62.1 % · millisecond). These AN/PO translational kinematic data were determined from the biplanar videoradiography system.

The AD/AB rotational excursion (Figure 5.3) was relatively constant after contact, changing less than two degrees for both male and female ACL_{REC} and ACL_{INT} subjects. Despite the minimal rotational change after contact, the female subjects were slightly abducted (valgus) after contact while the male subjects were slightly adducted (varus) after contact (average female abduction excursion equal to 1.02 degrees, average male adduction excursion equal to 1.07 degrees; $p = 0.033$). The IN/EX rotational excursion (Figure 5.3) for all subjects followed a consistent pattern. Specifically, the male and female ACL_{INT} and ACL_{REC} subjects all began internally rotating after contact. While no statistically significant differences were observed in any group, the ACL_{INT} male and female subjects had a 76 degree · millisecond larger AUC than the ACL_{REC} male and female subjects ($p = 0.171$). These AD/AB and IN/EX rotational kinematic data were determined from the biplanar videoradiography system.

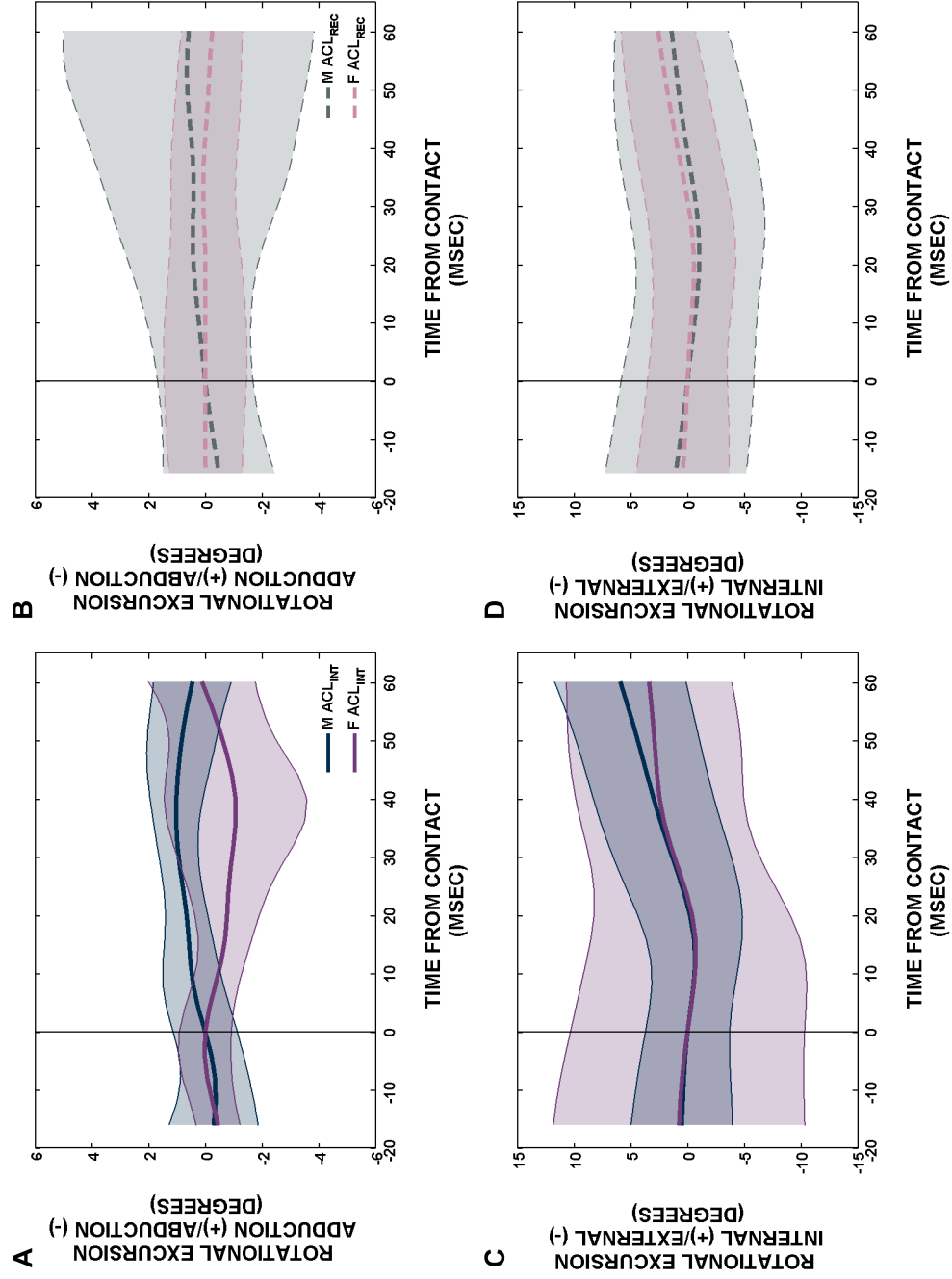


Figure 5.3: (A) ACL_{INT} AD/AB rotational excursion. (B) ACL_{REC} AD/AB rotational excursion. Adduction is positive and abduction is negative. (C) ACL_{INT} IN/EX rotational excursion. (D) ACL_{REC} IN/EX rotational excursion. All rotational excursion units are in degrees. All curves are displayed as mean $\pm$ 1 SD. The vertical line on each graph represents the time at contact. It should be noted that these data were obtained from the biplanar videoradiography system.

The minimum flexion angle occurred at or immediately following ground contact. Following this, all of the subjects absorbed the landing and continued the cut by flexing through stance phase to a maximum flexion angle. Using the OMC data to quantify the minimum flexion angle, maximum flexion angle, and the flexion angle excursion (change from minimum to maximum flexion angle), we observed that females tended to be more flexed at contact ($p = 0.054$), but their total excursion was significantly less ($p = 0.002$) (Figure 5.4). These FL/EX kinematic data were determined from the OMC system.

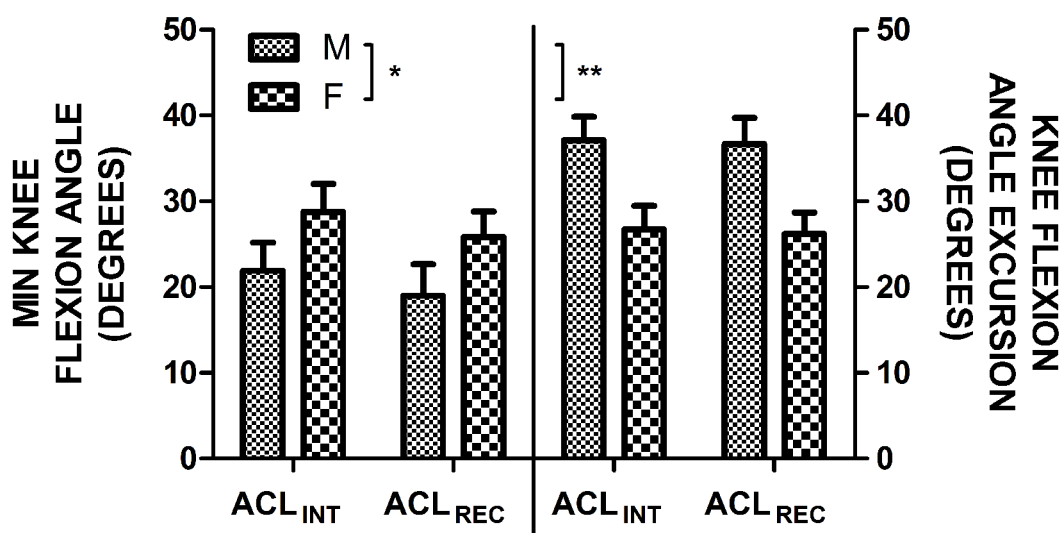


Figure 5.4: (Left y-axis) Minimum knee flexion angle for ACL_{INT} and ACL_{REC} male and female subjects. Minimum flexion occurred at or immediately following ground contact. No statistically significant differences between gender and condition were observed for minimum knee flexion angle values; however, the * represents a p-value of 0.054 denoting an apparent gender difference. (Right y-axis) Knee flexion angle excursion for ACL_{INT} and ACL_{REC} male and female subjects. Knee flexion angle excursion was defined as the change in knee flexion angle from minimum flexion to maximum flexion. A statistically significant difference was observed between male and female subjects. This is highlighted by the **, which represents a p-value of 0.002. The minimum knee flexion angle and knee flexion angle excursion units are in degrees. It should be noted that these data were obtained from the OMC system.

5.4 Discussion

We have compared knee kinematic and kinetic measurements from male and female ACL_{INT} and ACL_{REC} recreational athletes during a jump-cut maneuver associated with non-contact deceleration ACL injury. Two major findings were observed in our study. First, female subjects who have never had an ACL reconstruction appeared to perform the jump-cut maneuver with greater landing stiffness (smaller amount of knee flexion angle excursion combined with larger peak vertical GRF [16]) than males with or without a history of ACL reconstruction and other females with a history of ACL reconstruction. This was evidenced by the differences observed in the knee flexion angle excursion, which translated to qualitatively comparable differences in peak vertical GRF. Second, the male and female ACL_{REC} subjects appear to perform the jump-cut maneuver with less energy than the ACL_{INT} subjects, resulting in a lower peak vertical GRF even five years or more after their reconstruction. This may be a result of differences in strength, confidence, habit, and/or training following their injury.

These kinetic differences likely influence the differences observed in the rate of anterior tibial translation after contact, which is a common instigator of ACL injury. Specifically, we observed that anterior tibial translation increased at a faster rate in female ACL_{INT} subjects compared to their male ACL_{INT} counterparts (Figure 5.2). Notably, peak anterior tibial translation for the ACL_{INT} female subjects occurred within 60 milliseconds. A similar peak is not observed in the male ACL_{INT} subjects or the male and female ACL_{REC} subjects. Interestingly, the time to peak vertical GRF was significantly less in female subjects as compared to male subjects regardless of ACL reconstruction status. Also, the time to peak vertical GRF appears to be smaller in ACL_{INT} subjects as compared to ACL_{REC} subjects. The increased rate of anterior tibial translation observed in female ACL_{INT} subjects is likely a result

of the larger and more rapid peak vertical GRF observed immediately after ground contact. This rapid and large peak vertical GRF appears to produce a ‘snapping’ motion that differs from the more gradual increase in peak vertical GRF and anterior tibial translation observed in male ACL_{INT} subjects and male and female ACL_{REC} subjects. It may be that there is a reliable tendency for females to absorb less energy upon landing, which, through greater peak vertical GRF, resultant forces, and/or abnormal kinematics, may increase the risk for ACL injury.

Previous research has suggested that increased landing stiffness, as characterized by a smaller amount of knee flexion angle excursion combined with a high vertical GRF during landing and cutting activities, place individuals at increased risk of ACL injury [28–30]. Attempts have been made to correlate increased landing stiffness with increased anterior tibial translation with the goal of developing knee injury prevention training and rehabilitation programs [16]. During the jump-cut maneuver in our study, the female subjects landed and cut with less knee flexion angle excursion after contact. This result, when interpreted in the context of the faster and larger peak vertical GRF, confirms that the females are performing the jump-cut maneuver with more landing stiffness. This finding is in contrast to the observations made for the male subjects, who appear to be absorbing the energy they are applying at ground contact by flexing through the landing and subsequent cut. Moreover, the AN/PO translation never reached a maximum (within 60 milliseconds after contact) and increased at a lower rate after contact for the male ACL_{INT} and male and female ACL_{REC} subjects. This combination of increased knee flexion angle excursion and/or reduced peak vertical GRF (decreased landing stiffness) may contribute to the slower time to peak anterior tibial translation after contact for these subjects.

In a similar study investigating the knee kinematics of ACL_{INT} females during four functional tasks, Myers et al. observed that anterior tibial translation was increased

with activities of increasing external loading [15]. Specifically, they observed a 2.4 mm increase in anterior tibial translation during landing maneuvers as compared to walking. Moreover, their results show the same characteristic peak vertical GRF immediately following ground contact during the landing tasks. This peak is absent in the vertical GRF walking trace. Conversely, another study by Myers et al. showed no differences in anterior tibial translation between soft and stiff drop landings [16]. The authors attributed these findings to the ability of the ligaments and musculature about the knee to keep joint translations within a safe envelope of motion during controlled activities where external loading conditions are anticipated. While no excessive rotational or translation motion was observed in our study, the increased rate of anterior translation in female ACL_{INT} subjects suggests that stiffer landings under more unanticipated cutting activities may affect kinematic translations more than controlled, anticipated activities. Furthermore, the lower rate of anterior tibial translation seen in the ACL_{REC} subjects immediately following contact may be influenced by the lower peak vertical GRF.

This low peak vertical GRF observed in both male and female ACL_{REC} subjects matches results from Paterno et al. [31,32] and Vairo et al. [33]. In two separate studies, Paterno et al. showed ACL_{REC} male and female subjects decreased peak vertical GRF when performing landing activities two years after reconstruction. Vairo et al. reported decreased vertical peak GRF upon landing from a vertical drop among ACL_{REC} subjects approximately two years post surgery. These results are consistent with those reported in our study for ACL_{REC} men and women who are at least five years post-reconstruction. Additionally, no gender differences were observed in the peak GRF within ACL_{REC} subjects. Even after five years of strengthening activities, including formal rehabilitation, functional exercise, and return to sports, the ACL_{REC} subjects performed the jump-cut maneuver with less energy when compared to the ACL_{INT} subjects. While the exact mechanisms for this are unknown, it is

possible both behavioral and neuromechanical deficiencies are present in the ACL_{REC} subjects when performing jump-cut maneuvers on their previously injured limb. Alternatively, it is possible that the altered mechanics are a result of protective habits obtained during the ACL_{REC} subjects' rehabilitation. Additional research investigating neuromuscular activity and contralateral biomechanics may provide additional insight into the reduced vertical GRF observed in both male and female ACL_{REC} subjects.

In addition to the larger peak vertical GRF and rate of anterior tibial translation, the female subjects were generally more abducted after contact (Figure 5.3A-B). This is an interesting finding because videographic studies have suggested that a valgus (abduction) collapse is involved in the non-contact deceleration ACL injury mechanism [34]. Furthermore, the results presented herein are consistent with previous reports suggesting that females land with more knee abduction compared to males [18]. While the female subjects in our study were abducted after contact compared to the male subjects, the total amount of abduction (< 2 degrees) does not seem to correspond to a valgus collapse position. This may be a result of the subjects' ability to safely perform the jump-cut maneuver, which was implemented in our study to challenge the ACL. A valgus collapse position was not observed, neither were any adverse events (injury).

No significant differences were observed between gender and ACL reconstruction status for both IN/EX rotation or ME/LA translation after contact (Table 5.2). In general, all subjects began internally rotating after contact and remained stable in the ME/LA direction. With exception to the gender difference observed in AD/AB angle, the similar kinematic outcomes observed between gender and ACL reconstruction status after contact does not support our hypothesis. Deneweth et al. showed that, as compared to the ACL_{INT} contralateral knees, ACL_{REC} knees were more ex-

ternally rotated and less laterally translated during a single-leg hopped landing [17]. Unfortunately, obtaining contralateral limb kinematics was not feasible for our study. This makes direct comparisons difficult; however, Deneweth et al. does report total IN/EX excursion to be approximately five degrees and total ME/LA translation to be less than one millimeter for both reconstructed and contralateral knees. These excursion values align with the results presented in our study.

Despite the kinematic similarities, additional research investigating surface interactions between the medial and lateral compartments of the knee may provide more specific insight into subtle kinematic differences between both gender and ACL reconstruction status. Specifically, methods have been developed to identify distance weighted proximity centroids, regions of closest proximity [35], and point based surface velocities [36,37]. These techniques take advantage of the accuracy associated with biplanar videoradiography to make inferences about biomechanical changes at the articulating surfaces of the femur and tibia with the hope of better understanding the initiation and progression of osteoarthritis in ACL_{REC} individuals [38].

As investigators, we are limited to studying potential injury mechanisms in a laboratory testing environment without many of the situations presented in a sporting environment. Our study investigated male and female ACL_{INT} and ACL_{REC} subjects while they performed an activity in a controlled laboratory setting that has been associated with non-contact deceleration ACL injury. The incorporation of the ‘unanticipated’ element to the jump-cut maneuver mimicked the deceleration and cutting action associated with many sporting events. Despite the design, studies that employ biplanar videoradiography will be hindered by the inability to capture subjects performing sports activities in their native environments.

We acknowledge the small sample size for investigating kinematic and kinetic interactions between gender and ACL reconstruction status. While we did ensure that all

subjects were recreational athletes, we were not able to control for surgeon or rehabilitation protocol for the ACL_{REC} subjects. Additionally, we recognize the limitation of using two different graft types in our study. Based on biomechanical studies [11,39] and randomized clinical trials [40], we assumed that both bone-patellar tendon-bone and four-stranded hamstring tendon grafts would respond similarly during the jump-cut maneuver studied herein but recognize this as a study limitation.

Limitations associated with biplanar videoradiography should be noted. Specifically, the FOV restricted our ability to capture kinematic data for all subjects from 16 milliseconds before contact to 60 milliseconds after contact. Previous research has shown that peak anterior tibial translation occurs between 40 milliseconds to 50 milliseconds after contact [14], within the temporal range studied. Finally, each subject received up to 22 millirem of radiation exposure as a result of the biplanar videoradiography system and CT scan. While this falls well below the guidelines instituted by the NIH Radiation Safety Committee for acceptable radiation exposure to research subjects within a year (5 rem), it does limit the number of data collection trials. All subjects were aware of and gave informed consent to radiation exposure.

In conclusion, the results presented in our study support our hypothesis that kinematic and kinetic differences would be observed between both gender and ACL reconstruction status during a jump-cut maneuver. Specifically, we found that female ACL_{INT} subjects landed and cut with a smaller amount of knee flexion angle excursion and larger peak vertical GRF than the male ACL_{INT} subjects. Furthermore, we observed that the ACL_{REC} subjects had a significantly lower peak vertical GRF just after impact as compared to the ACL_{INT} subjects. We also noted that female ACL_{INT} subjects appear to have an increased rate of anterior tibial translation just after contact. Our study associates the increased rate of anterior tibial translation to increased landing stiffness (larger peak vertical GRF combined with smaller knee

flexion angle excursion) while performing the jump-cut maneuver. With respect to AD/AB, IN/EX rotation, and ME/LA translation, differences were only observed AD/AB angle. The female subjects in our study were abducted after contact compared to the male subjects; albeit, the amount of abduction does not appear to correspond to a valgus collapse position. Finally, no significant interactions were found between gender and ACL reconstruction status for IN/EX rotation or ME/LA translation after contact.

5.5 Conflicts of Interest

None.

5.6 Acknowledgements

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5.7 References

- [1] Gianotti S. M., Marshall S. W., Hume P. A., and Bunt L., 2009, “Incidence of anterior cruciate ligament injury and other knee ligament injuries: a national population-based study,” *J Sci Med Sport*, 12(6), pp. 622–627.
- [2] Griffin L. Y., Albohm M. J., Arendt E. A., Bahr R., Beynnon B. D., Demaio M., Dick R. W., Engebretsen L., Garrett W. E., Hannafin J. A., Hewett T. E., Huston L. J., Ireland M. L., Johnson R. J., Lephart S., Mandelbaum B. R., Mann B. J., Marks P. H., Marshall S. W., Myklebust G., Noyes F. R., Powers C., Shields C., Shultz S. J., Silvers H., Slauterbeck J., Taylor D. C., Teitz C. C., Wojtys E. M., and Yu B., 2006, “Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting, January 2005,” *Am J Sports Med*, 34(9), pp. 1512–1532.
- [3] McNair P. J., Marshall R. N., and Matheson J. A., 1990, “Important features associated with acute anterior cruciate ligament injury,” *N. Z. Med. J.*, 103(901), pp. 537–539.

- [4] Gwinn D. E., Wilckens J. H., McDevitt E. R., Ross G., and Kao T. C., 2000, "The relative incidence of anterior cruciate ligament injury in men and women at the United States Naval Academy," *Am J Sports Med*, 28(1), pp. 98–102.
- [5] Markolf K. L., Kochan A., and Amstutz H. C., 1984, "Measurement of knee stiffness and laxity in patients with documented absence of the anterior cruciate ligament," *J Bone Joint Surg Am*, 66(2), pp. 242–252.
- [6] Dargel J., Gotter M., Mader K., Pennig D., Koebke J., and Schmidt-Wiethoff R., 2007, "Biomechanics of the anterior cruciate ligament and implications for surgical reconstruction," *Strategies Trauma Limb Reconstr*, 2(1), pp. 1–12.
- [7] Wright R. W., Dunn W. R., Amendola A., Andrish J. T., Bergfeld J., Kaeding C. C., Marx R. G., McCarty E. C., Parker R. D., Wolcott M., Wolf B. R., and Spindler K. P., 2007, "Risk of tearing the intact anterior cruciate ligament in the contralateral knee and rupturing the anterior cruciate ligament graft during the first 2 years after anterior cruciate ligament reconstruction: a prospective MOON cohort study," *Am J Sports Med*, 35(7), pp. 1131–1134.
- [8] Ardern C. L., Taylor N. F., Feller J. A., and Webster K. E., 2012, "Return-to-sport outcomes at 2 to 7 years after anterior cruciate ligament reconstruction surgery," *Am J Sports Med*, 40(1), pp. 41–48.
- [9] Junkin D. M., Johnson D. J., Fu F. H., Mark M. D., Willenborn M., Fanelli G. C., and Wascher D. W., 2009, "Knee Ligament Injuries," *Orthopaedic Knowledge Update: Sports Medicine 4*, Amer Academy of Orthopaedic, pp. 135–153.
- [10] Miranda D. L., Rainbow M. L., Crisco J. J., and Fleming B. C., 2012, "Kinematic Errors due to Soft Tissue Artifact during a Jump-Cut Maneuver: A Comparison between Optical Motion Capture and Bi-planar X-ray Technologies," In: *Proceedings of the 58th Annual Meeting of the Orthopaedic Research Society*, San Francisco, CA, p. 887.
- [11] Tashman S., Kolowich P., Collon D., Anderson K., and Anderst W., 2007, "Dynamic function of the ACL-reconstructed knee during running," *Clin. Orthop. Relat. Res*, 454, pp. 66–73.
- [12] Tashman S., Collon D., Anderson K., Kolowich P., and Anderst W., 2004, "Abnormal rotational knee motion during running after anterior cruciate ligament reconstruction," *Am J Sports Med*, 32(4), pp. 975–983.
- [13] Torry M. R., Myers C., Shelburne K. B., Peterson D., Giphart J. E., Pennington W. W., Krong J. P., Woo S. L.-Y., and Steadman J. R., 2011, "Relationship of knee shear force and extensor moment on knee translations in females performing drop landings: a biplane fluoroscopy study," *Clin Biomech (Bristol, Avon)*, 26(10), pp. 1019–1024.

- [14] Torry M. R., Shelburne K. B., Peterson D. S., Giphart J. E., Krong J. P., Myers C., Steadman J. R., and Woo S. L.-Y., 2011, "Knee kinematic profiles during drop landings: a biplane fluoroscopy study," *Med Sci Sports Exerc*, 43(3), pp. 533–541.
- [15] Myers C. A., Torry M. R., Shelburne K. B., Giphart J. E., LaPrade R. F., Woo S. L.-Y., and Steadman J. R., 2012, "In vivo tibiofemoral kinematics during 4 functional tasks of increasing demand using biplane fluoroscopy," *Am J Sports Med*, 40(1), pp. 170–178.
- [16] Myers C. A., Torry M. R., Peterson D. S., Shelburne K. B., Giphart J. E., Krong J. P., Woo S. L.-Y., and Steadman J. R., 2011, "Measurements of tibiofemoral kinematics during soft and stiff drop landings using biplane fluoroscopy," *Am J Sports Med*, 39(8), pp. 1714–1722.
- [17] Deneweth J. M., Bey M. J., McLean S. G., Lock T. R., Kolowich P. A., and Tashman S., 2010, "Tibiofemoral joint kinematics of the anterior cruciate ligament-reconstructed knee during a single-legged hop landing," *Am J Sports Med*, 38(9), pp. 1820–1828.
- [18] Ford K. R., Myer G. D., Toms H. E., and Hewett T. E., 2005, "Gender differences in the kinematics of unanticipated cutting in young athletes," *Med Sci Sports Exerc*, 37(1), pp. 124–129.
- [19] Tegner Y., and Lysholm J., 1985, "Rating systems in the evaluation of knee ligament injuries," *Clin. Orthop. Relat. Res.*, (198), pp. 43–49.
- [20] Buczek F. L., Rainbow M. J., Cooney K. M., Walker M. R., and Sanders J. O., 2010, "Implications of using hierarchical and six degree-of-freedom models for normal gait analyses," *Gait & Posture*, 31(1), pp. 57–63.
- [21] Miranda D. L., Schwartz J. B., Loomis A. C., Brainerd E. L., Fleming B. C., and Crisco J. J., 2011, "Static and Dynamic Error of a Biplanar Videoradiography System Using Marker-Based and Markerless Tracking Techniques," *J Biomech Eng*, 133(12), p. 121002.
- [22] Brainerd E. L., Baier D. B., Gatesy S. M., Hedrick T. L., Metzger K. A., Gilbert S. L., and Crisco J. J., 2010, "X-ray reconstruction of moving morphology (XROMM): precision, accuracy and applications in comparative biomechanics research," *J Exp Zool A Ecol Genet Physiol*, 313(5), pp. 262–279.
- [23] Miranda D. L., Rainbow M. J., Leventhal E. L., Crisco J. J., and Fleming B. C., 2010, "Automatic determination of anatomical coordinate systems for three-dimensional bone models of the isolated human knee," *J Biomech*, 43(8), pp. 1623–1626.
- [24] Bey M. J., Zauel R., Brock S. K., and Tashman S., 2006, "Validation of a new model-based tracking technique for measuring three-dimensional, in vivo glenohumeral joint kinematics," *J Biomech Eng*, 128(4), pp. 604–609.

- [25] Coburn J., and Crisco J. J., 2005, "Interpolating three-dimensional kinematic data using quaternion splines and hermite curves," *J Biomech Eng*, 127(2), pp. 311–317.
- [26] Grood E. S., and Suntay W. J., 1983, "A joint coordinate system for the clinical description of three-dimensional motions: application to the knee," *J Biomech Eng*, 105(2), pp. 136–144.
- [27] Tanifuji O., Sato T., Kobayashi K., Mochizuki T., Koga Y., Yamagiwa H., Omori G., and Endo N., 2011, "Three-dimensional in vivo motion analysis of normal knees using single-plane fluoroscopy," *J Orthop Sci*, 16(6), pp. 710–718.
- [28] Beutler A., de la Motte S., Marshall S., Padua D., and Boden B., 2009, "Muscle strength and qualitative jump-landing differences in male and female military cadets: the jump-acl study," *J Sports Sci Med*, 8, pp. 663–671.
- [29] Boden B. P., Dean G. S., Feagin J. A., and Garrett W. E., 2000, "Mechanisms of anterior cruciate ligament injury," *Orthopedics*, 23(6), pp. 573–578.
- [30] Chappell J. D., Creighton R. A., Giuliani C., Yu B., and Garrett W. E., 2007, "Kinematics and electromyography of landing preparation in vertical stop-jump: risks for noncontact anterior cruciate ligament injury," *Am J Sports Med*, 35(2), pp. 235–241.
- [31] Paterno M. V., Schmitt L. C., Ford K. R., Rauh M. J., Myer G. D., and Hewett T. E., 2011, "Effects of sex on compensatory landing strategies upon return to sport after anterior cruciate ligament reconstruction," *J Orthop Sports Phys Ther*, 41(8), pp. 553–559.
- [32] Paterno M. V., Ford K. R., Myer G. D., Heyl R., and Hewett T. E., 2007, "Limb asymmetries in landing and jumping 2 years following anterior cruciate ligament reconstruction," *Clin J Sport Med*, 17(4), pp. 258–262.
- [33] Vairo G. L., Myers J. B., Sell T. C., Fu F. H., Harner C. D., and Lephart S. M., 2008, "Neuromuscular and biomechanical landing performance subsequent to ipsilateral semitendinosus and gracilis autograft anterior cruciate ligament reconstruction," *Knee Surg Sports Traumatol Arthrosc*, 16(1), pp. 2–14.
- [34] Krosshaug T., Nakamae A., Boden B. P., Engebretsen L., Smith G., Slauterbeck J. R., Hewett T. E., and Bahr R., 2007, "Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases," *Am J Sports Med*, 35(3), pp. 359–367.
- [35] Anderst W. J., and Tashman S., 2003, "A method to estimate in vivo dynamic articular surface interaction," *J Biomech*, 36(9), pp. 1291–1299.
- [36] Anderst W. J., and Tashman S., 2009, "The association between velocity of the center of closest proximity on subchondral bones and osteoarthritis progression," *J. Orthop. Res.*, 27(1), pp. 71–77.

- [37] Anderst W. J., and Tashman S., 2010, “Using relative velocity vectors to reveal axial rotation about the medial and lateral compartment of the knee,” *J Biomech*, 43(5), pp. 994–997.
- [38] Andriacchi T. P., Mündermann A., Smith R. L., Alexander E. J., Dyrby C. O., and Koo S., 2004, “A framework for the in vivo pathomechanics of osteoarthritis at the knee,” *Ann Biomed Eng*, 32(3), pp. 447–457.
- [39] Wilson T. W., Zafuta M. P., and Zobitz M., 1999, “A biomechanical analysis of matched bone-patellar tendon-bone and double-looped semitendinosus and gracilis tendon grafts,” *Am J Sports Med*, 27(2), pp. 202–207.
- [40] Mohtadi N. G., Chan D. S., Dainty K. N., and Whelan D. B., 2011, “Patellar tendon versus hamstring tendon autograft for anterior cruciate ligament rupture in adults,” *Cochrane Database Syst Rev*, (9), p. CD005960.

Chapter 6

Conclusions, related studies, and future directions

6.1 Conclusions

The studies and data presented in this dissertation describe and apply methods for using biplanar videoradiography to study dynamic *in vivo* knee motion. The techniques were applied with the goal of better understanding the kinematics and kinetics associated with the non-contact deceleration ACL injury mechanism during jumping and cutting in anterior cruciate ligament (ACL) -intact (ACL_{INT}) and ACL-reconstructed (ACL_{REC}) male and female recreational athletes. The long term objective of this work was to inform novel injury prevention and rehabilitation programs that mitigate the injury risk, improve injury recovery, and/or reduce risks associated with re-injury and post-traumatic joint degradation. The specific aims of the chapters presented in this dissertation were to: (1) develop and validate methods for accurately quantifying *in vivo* knee bone motion using biplanar videoradiography; (2) collect and quantify the kinematics of ACL_{INT} individuals performing an activity associated with non-contact deceleration ACL injury; and (3) investigate gender and ACL reconstruction status differences in kinematics and kinetics of ACL_{INT} and ACL_{REC} subjects performing an activity associated with non-contact deceleration ACL injury. The topics covered in Chapter 1 outline the background and significance of the studies presented in this dissertation and explain the purpose and hypotheses of the specific aims in detail.

6.1.1 Specific Aim 1

Develop and validate methods for accurately quantifying in vivo knee bone motion using biplanar videoradiography

The first aim was accomplished with the studies presented in Chapter 2 and Chapter 3. The goals of the study presented in Chapter 2 were to design and validate software techniques for tracking and quantifying *in vivo* three-dimensional (3-D) bone motion

using biplanar videoradiography. In this study, we validated new markerless tracking software (Autoscooper) that implemented previously described markerless tracking algorithms on a graphics processor. Additionally, we successfully developed protocols for evaluating the systematic tracking error of a biplanar videoradiography system using gold standard instrumentation. These protocols allowed static and dynamic tracking errors to be evaluated for a multitude of bones and system configurations. We were able to show that the dynamic markerless systematic tracking error of the biplanar videoradiography system fell below 0.25° for the distal femur, distal ulna, and distal radius. While unpublished, data have also been collected and processed for the proximal tibia, patella, distal humerus, proximal humerus, proximal ulna, proximal radius, first metacarpal, and third metacarpal. The dynamic markerless systematic errors associated with tracking these bones were similar to the tracking errors associated with the distal femur, distal ulna, and distal radius presented in Chapter 2. The methods and results presented in Chapter 2 indicate that we could accurately quantify 3-D bone motion non-invasively without the implantation of radio-opaque markers during dynamic jumping and cutting activities.

Our ability to dynamically track *in vivo* bone motion accurately, as established with the results presented in Chapter 2, requires robust anatomical coordinate systems to interpret the bone motion into clinically meaningful six degree-of-freedom (DOF) joint kinematics. The goal of the study presented in Chapter 3 was to develop and implement an algorithm to consistently and repeatedly define anatomical coordinate systems based on the 3-D geometry of the distal femur and proximal tibia in order to extract clinically relevant joint kinematics. The algorithm was successful in determining independent coordinate systems of the femur and tibia using bone models extracted from a 3-D imaging modality, which is required for any biplanar videoradiography study. These coordinate systems were based off the articulating surfaces of the femoral condyles and tibial plateau and were designed to define a knee joint

coordinate system based on the ‘roll-glide’ motion of a cylinder (femoral condyles) moving on top of a plane (tibial plateau).

Overall, we present, implement, and validate new software available to the biomechanics community for studying and interpreting dynamic *in vivo* bone motion. We hope that these methods help researchers focus on understanding and solving today’s biomechanical and orthopaedic challenges through strong collaborative efforts.

6.1.2 Specific Aim 2

Collect and quantify the kinematics of ACL_{INT} individuals performing an activity associated with non-contact deceleration ACL injury

The second aim was accomplished with the study presented in Chapter 4. The goal of this study was to determine the feasibility of collecting and quantifying the leg motion of ACL_{INT} recreational athletes performing an activity associated with non-contact deceleration ACL injury using the tools and techniques presented in Chapter 2 and Chapter 3. Additionally, we wanted to optimize methods for combining motion capture modalities to expand the field of view (FOV) of the biplanar videoradiography system as well as to understand how soft tissue affects kinematic measurements obtained from traditional optical motion capture (OMC) techniques during the proposed activity.

We were successful in designing an experimental set-up capable of synchronously measuring leg motion using biplanar videoradiography and OMC as well as electromyography (EMG) and ground reaction force (GRF) data. For this study, we recruited ten ACL_{INT} subjects and collected kinematic, kinetic, and neuromuscular data while they performed a jump-cut maneuver which was designed to mimic sport activities that are

associated with non-contact deceleration ACL injury. When comparing motion capture modalities, we determined that the soft tissue surrounding the femur influenced kinematic measurements more than the soft tissue surrounding the tibia. Overall, we observed that the soft tissue surrounding the knee produced kinematic deviations as high as 15° and 28 mm after landing. From these data, we concluded that the secondary knee rotations (adduction/abduction and internal/external rotation) and all knee translations (medial/lateral, anterior/posterior, and compression/distraction) determined from OMC should be interpreted with caution, especially following contact. However, it should be noted that OMC can sufficiently measure the sagittal plane kinematics before and after landing based on the results reported in this study. This permits the measurement of knee flexion/extension outside the FOV of the biplanar videoradiography system. This provides valuable information regarding the flexion angle during the entire maneuver while biplanar videoradiography can be used to focus on the knee during landing, where it is under the most stress. The methods and results presented and discussed in Chapter 4, which address the second aim, laid the foundation for the study presented in Chapter 5.

6.1.3 Specific Aim 3

Investigate gender and ACL reconstruction status differences in kinematics and kinetics of ACL_{INT} and ACL_{REC} subjects performing an activity associated with non-contact deceleration ACL injury

The final aim of this dissertation was accomplished with the study presented in Chapter 5. The goal of this study was to investigate the kinematic and kinetic gender and ACL reconstruction status differences using the methods developed and implemented in Specific Aim 1 and Specific Aim 2. To complete this aim, we recruited and collected biplanar videoradiography, OMC, GRF, and EMG data on an additional ten

ACL_{REC} subjects who were at least five years post-surgery. The experimental protocol and data collection were performed in the same manner as described in Chapter 4. For this aim we compared the knee kinematics and vertical GRF data between male and female ACL_{INT} and ACL_{REC} subjects with the hope of elucidating differences between gender and ACL reconstruction status. The results presented in Chapter 5 showed that female ACL_{INT} subjects landed and pivoted with a smaller range of knee flexion angle and a larger peak vertical GRF than the male ACL_{INT} subjects. The results also showed that the ACL_{REC} subjects had a significantly lower peak vertical GRF just after impact as compared to the ACL_{INT} subjects. We also noted that female ACL_{INT} subjects appear to have an increased rate of anterior tibial translation just after contact, and the increased rate of anterior tibial translation was associated with an increase in landing stiffness. The increase in landing stiffness was defined as a larger peak vertical GRF combined with a smaller range of knee flexion angle, while performing the jump-cut maneuver. These results support future studies aimed at reducing landing stiffness in women with the hope of decreasing the rate of anterior tibial translation, and consequently, the risk of ACL injury. Moreover, the reduced energy with which the ACL_{REC} subjects were performing the jump-cut maneuver may be countered with focused rehabilitation, where a restoration of psychological and physical mechanisms may help ACL_{REC} patients return to pre-injury activity levels (Section 6.2.4).

Finally, it should be mentioned that as part of this study we also modeled the vertical GRF traces from each subject trial using a non-linear parameterized piecewise model. This was done to provide insight into the different characteristics of the vertical GRF immediately after contact. These methods and results are presented in Appendix A. Based on the findings from this study, we elected to simplify the methods and analysis as presented in Chapter 5. However, the exercise provided a platform for our final analysis and may prove to be an effective tool for future analyses.

6.2 Related studies and future directions

There are a number of related studies and future directions that seek to further address the aims presented in this dissertation. The kinematic, kinetic, and neuromuscular data obtained in Chapter 4 and Chapter 5 provides a wealth of information regarding the function of the ACL_{INT} and ACL_{REC} knee during a jump-cut maneuver. A number of related studies have been performed to further address the aims of this dissertation as described below. Overall, the methods developed and results obtained herein provide a strong foundation for furthering the long term objective of this work. Specifically, the design and implementation of prospective studies aimed at developing subject specific injury prevention and rehabilitation training programs is an exciting avenue of future research. The following subsections provide a brief overview of the additional research avenues afforded by the primary and related studies presented in this dissertation.

6.2.1 Articular surface interactions (Appendix B)

The first related study was conducted to compare the interactions between the femoral and tibial articulating surfaces of ACL_{INT} and ACL_{REC} recreational athletes during a jump-cut maneuver (Appendix B). In addition to the traditional kinematic outcome measures (knee joint rotations and translations), dynamic articular surface interactions can be calculated using the set of tibiofemoral bone kinematics and the morphology information from the 3-D CT bone models [1,2]. Articular surface interaction outcomes such as tibiofemoral joint proximity, proximity center location, proximity center path, proximity area, and surface velocity may provide valuable information on how changes in bone kinematics affect the tibiofemoral articular surface [3]. A better understanding of subtle changes in articular surface deformation may provide insight

into the pathomechanics associated with the onset and progression of osteoarthritis. This is especially relevant to ACL_{REC} subjects, who are at an increased risk of joint degradation. Preliminary work has been done to investigate the effect of ACL reconstruction status on tibiofemoral articular surface interactions during the jump-cut maneuver as highlighted in Appendix B.

The overall goal of this analysis is to correlate dynamic articular surface interactions to the initiation and progression of post-traumatic osteoarthritis (PTOA) with the hope of developing new detection and treatment methodology. The results presented in Appendix B provide pilot data for designing a study aimed at following a cohort of ACL_{REC} , ACL-deficient (ACL_{DEF}), and/or ACL_{INT} matched control subjects to correlate changes in articular surface interactions to serial changes in cartilage and meniscus degradation. The kinematic and kinetic techniques used in this dissertation can be combined with techniques developed by Bowers et al. [4–7] which take advantage of 3-D soft tissue information obtained from magnetic resonance imaging (MRI) to quantify cartilage and meniscus damage at different time points following surgery. Exciting work quantifying the association of cartilage thickness distributions to the 3-D kinematics of the femur and tibia [8] stress the importance of understanding how ACL deficiency or reconstruction shift contact locations [1,9]. Recently, Beveridge [10] has correlated dynamic surface interactions with cartilage damage in sheep. Using these techniques to correlate the exact location and amount of cartilaginous damage to alterations in dynamic surface interactions in humans may provide invaluable information for modified surgical and rehabilitation techniques.

6.2.2 Ligament elongations (Appendix C)

The second related study was conducted to determine how the native ACL elongates in relation to secondary knee rotations during the jump-cut maneuver (Appendix C).

Quantifying the elongation of the ACL during the jump-cut maneuver may provide insight into the dynamic *in vivo* loading of the ligament during an activity associated with non-contact deceleration ACL injury. These data are needed for understanding how specific motions directly impact the ACL. The preliminary work investigating ACL elongations in ACL_{INT} subjects during the jump-cut maneuver is highlighted in Appendix C.

Additional work needs to be done to improve the anatomical accuracy of ACL insertion sites used in Appendix C. This may be accomplished by merging soft tissue measurements from ultrasound or MRI with the CT bone models [9,11–13]. Moreover, using these soft tissue measurements to improve the simple ligament model used in Appendix C may help incorporate information about the ACL bundles (anteromedial (AM) and posterolateral (PL)), individual ligament fibers, and ligament wrapping into the model. It is apparent that more realistic models of the ACL, specifically ones that incorporate both the AM and the PL bundles as well as the contact and friction caused by the wrapping around the bone, more closely predict actual ligament properties [14–16]. Notably, it has been shown that subject-specific ligament modeling improves the accuracy of these predictive models [17]; however, while these complex ligament models exist, the kinematic boundary conditions are determined from cadaveric studies that cannot match data obtained from *in vivo* jumping and cutting activities. Combining the reliable *in vivo* kinematics from biplanar videoradiography with advanced ligament modeling techniques may enable mechanical properties to be determined for activities that are known to put the ACL at risk. These data have the potential to identify what kinematic parameters place the intact ACL at the greatest risk of injury under a variety motion conditions [11–13]. A better understanding of the dynamic reconstructed ligament properties in ACL_{REC} subjects may improve current surgical procedures following ACL injury.

6.2.3 Muscle activity (Appendix D)

The third related study was conducted to elucidate muscle timing and activity of the quadriceps (Q), hamstring (H), and gastrocnemius (G) muscles between male and female ACL_{INT} and ACL_{REC} subjects performing the jump-cut maneuver (Appendix D). Surface EMG signals were recorded for each jump-cut trial collected on all ACL_{INT} and ACL_{REC} subjects using a wireless EMG system (Myomonitor IV; Delsys, Boston, MA). Quadriceps muscle activity was measured for the rectus femorus (RF) and vastus medialis (VM). Hamstring muscle activity was measured for the biceps femorus (BF) and semitendinosis (ST). Additionally, the muscle activity was measured for both the medial and lateral head of the gastrocnemius. These muscles were chosen based on previous studies, their direct influence on the knee, and the ease with which they can be palpated [18–23]. Bipolar re-usable surface electrodes were applied to the surface of the skin to collect muscle activity data. These electrodes required no gel and could be easily affixed on the surface of the muscle of interest after simple skin preparation.

Each muscle’s electrode placement was marked with a washable marker. The subject was then asked to shave, lightly abrade, and clean (Electrode Prep Pad; PDI, Orangeburg, NY, USA) the marked areas. The EMG electrodes were then affixed to the surface of the prepared skin using removable electrode specific adhesive tape. The surface electrode wires ran from the electrode to a control box that was fastened to the subject’s waist. The EMG system was capable of wirelessly syncing with the OMC and biplanar videoradiography systems in real time. As with the analog force plate data, each electrode channel of the EMG system was sampled at 5,000 Hz.

An example set of processed EMG data is shown in Figure D.1. The traces are from one ACL_{INT} subject. The traces include the rectified and filtered muscle activity from the rectus femorus and semitendinosis during one trial. The flexion angle during the

jump-cut maneuver is also shown. Note that the minimum amount of activity for the rectus femoris is occurring directly after ground contact and the maximum amount of activity is occurring at the maximum flexion angle during the stance phase of the activity. Conversely, the semitendinosis activity is at a maximum and minimum, respectively. These observations in muscle activity are expected. During the first half of contact, the subject is decelerating their tibia and flexing to absorb ground contact. Functionally, this requires hamstring contraction. Once the subject has reached the maximum flexion angle, they transition from flexing their knee to extending it. As a knee extensor, the quadriceps is responsible for extending the leg through the rest of stance and toe-off. A more detailed analysis of muscle timing and activity is highlighted in Appendix D.

The work presented in Appendix D summarizes the neuromuscular data collected during the studies presented in Chapter 4 and Chapter 5. The results suggest that there is both a gender bias in antagonistic ACL mechanisms and a sensory deficit in ACL_{REC} subjects. These data are particularly interesting when interpreted in the context of the increased rate of anterior tibial translation observed in female ACL_{INT} subjects (Chapter 5). Furthermore, the delays observed in muscle timing support the theory that a neuromuscular deficiency may contribute to the decreased energy with which the ACL_{REC} subjects perform the jump-cut maneuver (Chapter 5).

6.2.4 Injury prevention and rehabilitation

It is apparent from the work presented herein that kinematic (Chapter 5), kinetic (Chapter 5), and neuromuscular differences (Appendix D) are present between ACL_{INT} and ACL_{REC} men and women. We successfully identified potential ACL injury risk factors in ACL_{INT} women. Specifically, the increased leg stiffness, the higher rate of anterior tibial translation, and the apparent muscle activity timing disparity following

ground contact are variables that may be modified given a specific injury prevention training program. Moreover, the kinematic, kinetic, and neuromuscular disparities between ACL_{INT} and ACL_{REC} subjects five years after surgery suggest that focused long-term rehabilitation may benefit the ACL_{REC} subject population and promote a return to their pre-injury neuromechanics [24–29].

The work presented in this dissertation lays the technical groundwork and provides pilot data supporting a prospective injury prevention and rehabilitation study aimed at addressing the identified risk factors in female and ACL_{REC} subjects. These types of studies would involve pre-training/rehabilitation testing and the implementation of a subject specific injury prevention/rehabilitation training program, followed by post-training/rehabilitation testing to quantify the effects of training/rehabilitation. It has been shown that neuromuscular training programs may reduce ACL injury risk by as much as 50% [30–32]. Furthermore, training interventions aimed at reducing leg stiffness and loading in runners has shown to mitigate injury and positively alter biomechanics [33–35]. Modifying and applying similar techniques to ACL_{INT} female athletes may be a viable option for reducing their injury risk.

6.2.5 ACL deficient subject population

Finally, in addition to the data collected from the ACL_{INT} and ACL_{REC} subjects, we have a similar ongoing study applying the jump-cut protocol from Chapter 4 and Chapter 5 to collect kinematic, kinetic, and neuromuscular data from ACL-deficient (ACL_{DEF}) subjects. The ACL_{DEF} subject population is unique in that the inclusion criteria for these subjects are: (1) unilateral ACL injury; (2) continued participation in sport activities; (3) no previous knee injury; (4) no concomitant knee injury; (5) willingness to perform a one-legged hop test on their injured leg (an apprehension test); and (6) no pregnancy. By recruiting subjects who are able

to continue with recreational sports and perform a one-legged hop test despite the injury, we are able to screen out ACL_{DEF} subjects from the study that are physically or mentally incapable of performing the jump-cut maneuver. These inclusion criteria bias our subject population by eliminating subjects with very unstable knees; however, it allows us to study a specific subject population that has a unique ability to maintain functional stability despite being ACL_{DEF} . To our knowledge, the data provided from this subject population when performing the jump-cut maneuver does not exist in the literature. The kinematic, kinetic, and/or neuromuscular differences between subject populations (ACL_{INT} , ACL_{REC} , ACL_{DEF}) may provide insight into the mechanisms that allow these ACL_{DEF} subjects to functionally ‘cope’ with their injury. To date, we have successfully collected data on three ACL_{DEF} subjects who fit the inclusion criteria (Figure 6.1). Based on these data, there appears to be no gross kinematic instability. These subjects appear to be coping with the ligament disruption. It should be noted that no adverse outcomes (i.e. knee slippage) occurred during any collected jump-cut trials. Additional work needs to be done to obtain, process, and analyze the data from additional ACL_{DEF} subjects using the methods presented in this dissertation. Moreover, processing and analyzing the neuromuscular data from this subject population may provide insight into the apparent functional stability of the knee despite ACL deficiency. Clinically, these data may help us understand how ACL_{DEF} copers are successful with non-surgical treatment, which may provide clinicians with additional tools for identifying non-copers who may be candidates for non-surgical rehabilitation [28,36,37].

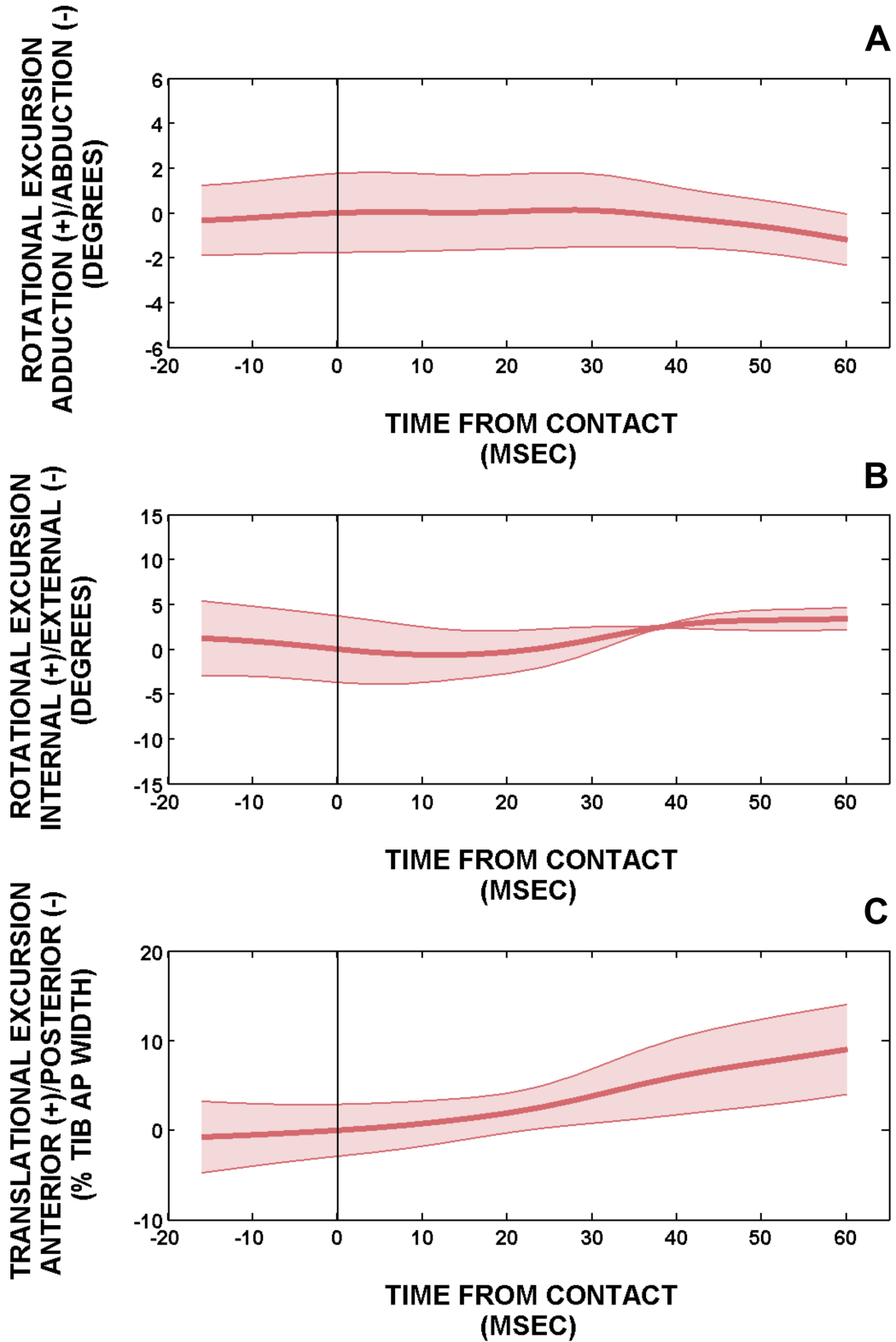


Figure 6.1: (A) ACL_{DEF} adduction/abduction rotational excursion. (B) ACL_{DEF} internal/external rotational excursion. All rotational excursion units are in degrees. (C) ACL_{DEF} anterior/posterior translational excursion. All curves are displayed as mean $\pm$ 1 SD. The vertical line on each graph represents the time at contact.

6.3 Summary

The collection of work presented in this dissertation served to develop and validate methods for tracking dynamic *in vivo* bone motion using biplanar videoradiography. The application of these methods was an essential component of this dissertation. Specifically, we strove to apply these methods to better understand how differences in knee kinematics and kinetics between gender and ACL reconstruction status may influence the gender injury bias, re-injury rate, and development of early onset post-traumatic osteoarthritis. We chose to study a jump-cut maneuver that attempts to mimic activities associated with non-contact deceleration ACL injury. This maneuver was perfectly suited for study with biplanar videoradiography because it combined a jump, landing, and 45° directional cut into one trial that can be safely and reliably performed within a confined environment. The culmination of the results presented in this dissertation relates increased anterior tibial translation to increased landing stiffness, which may be a contributing factor to the non-contact deceleration ACL injury. Ultimately, the long term goals of this work are to inform novel injury prevention and rehabilitation strategies. It is our hope that the ongoing and future directions related to the studies presented in this dissertation will further this goal. These data provide the foundation for a prospective injury prevention and rehabilitation training study that attempts to reduce injury risk in females as well as increase activity tolerance in ACL_{REC} patients to their pre-injury level. Overall, the tools and techniques developed in this work will provide a platform for future studies investigating *in vivo* bone motion in the knee, ankle, shoulder, elbow, wrist, and hand.

6.4 References

- [1] Anderst W. J., and Tashman S., 2003, “A method to estimate in vivo dynamic articular surface interaction,” *J Biomech*, 36(9), pp. 1291–1299.
- [2] Anderst W. J., and Tashman S., 2010, “Using relative velocity vectors to reveal axial rotation about the medial and lateral compartment of the knee,” *J Biomech*, 43(5), pp. 994–997.
- [3] Andriacchi T. P., Mündermann A., Smith R. L., Alexander E. J., Dyrby C. O., and Koo S., 2004, “A framework for the in vivo pathomechanics of osteoarthritis at the knee,” *Ann Biomed Eng*, 32(3), pp. 447–457.
- [4] Bowers M. E., Tung G. A., Trinh N., Leventhal E., Crisco J. J., Kimia B., and Fleming B. C., 2008, “Effects of ACL interference screws on articular cartilage volume and thickness measurements with 1.5 T and 3 T MRI,” *Osteoarthr. Cartil*, 16(5), pp. 572–578.
- [5] Bowers M. E., Trinh N., Tung G. A., Crisco J. J., Kimia B. B., and Fleming B. C., 2008, “Quantitative MR imaging using ‘LiveWire’ to measure tibiofemoral articular cartilage thickness,” *Osteoarthr. Cartil*, 16(10), pp. 1167–1173.
- [6] Bowers M. E., Tung G. A., Fleming B. C., Crisco J. J., and Rey J., 2007, “Quantification of meniscal volume by segmentation of 3T magnetic resonance images,” *J Biomech*, 40(12), pp. 2811–2815.
- [7] Bowers M. E., Tung G. A., Oksendahl H. L., Hulstyn M. J., Fadale P. D., Machan J. T., and Fleming B. C., 2010, “Quantitative magnetic resonance imaging detects changes in meniscal volume in vivo after partial meniscectomy,” *Am J Sports Med*, 38(8), pp. 1631–1637.
- [8] Koo S., Rylander J. H., and Andriacchi T. P., 2011, “Knee joint kinematics during walking influences the spatial cartilage thickness distribution in the knee,” *J Biomech*, 44(7), pp. 1405–1409.
- [9] Li G., Moses J. M., Papannagari R., Pathare N. P., DeFrate L. E., and Gill T. J., 2006, “Anterior cruciate ligament deficiency alters the in vivo motion of the tibiofemoral cartilage contact points in both the anteroposterior and mediolateral directions,” *J Bone Joint Surg Am*, 88(8), pp. 1826–1834.
- [10] Beveridge J. E., 2012, “Surface Interactions and Cartilage Damage in Two Ovine Models of Stifle Injury,” Ph.D., University of Calgary.
- [11] Taylor K. A., Terry M. E., Utturkar G. M., Spritzer C. E., Queen R. M., Iribarra L. A., Garrett W. E., and DeFrate L. E., 2011, “Measurement of in vivo

anterior cruciate ligament strain during dynamic jump landing,” *J Biomech*, 44(3), pp. 365–371.

[12] Utturkar G. M., Irribarra L. A., Taylor K. A., Spritzer C. E., Taylor D. C., Garrett W. E., and Defrate L. E., 2012, “The Effects of a Valgus Collapse Knee Position on In Vivo ACL Elongation,” *Ann Biomed Eng*.

[13] Hosseini A., Gill T. J., Van de Velde S. K., and Li G., 2011, “Estimation of In Vivo ACL Force Changes in Response to Increased Weightbearing,” *J Biomech Eng*, 133(5), p. 051004.

[14] Zhang X., Wu C., Jiang G., and Woo S. L.-Y., 2008, “The effects of geometry and fiber bundle orientation on the finite element modeling of the anterior cruciate ligament,” *Conf Proc IEEE Eng Med Biol Soc*, 2008, pp. 899–902.

[15] Song Y., Debski R. E., Musahl V., Thomas M., and Woo S. L.-Y., 2004, “A three-dimensional finite element model of the human anterior cruciate ligament: a computational analysis with experimental validation,” *J Biomech*, 37(3), pp. 383–390.

[16] Marai G. E., Laidlaw D. H., Demiralp C., Andrews S., Grimm C. M., and Crisco J. J., 2004, “Estimating joint contact areas and ligament lengths from bone kinematics and surfaces,” *IEEE Trans Biomed Eng*, 51(5), pp. 790–799.

[17] Zhang X., Jiang G., Wu C., and Woo S. L.-Y., 2008, “A subject-specific finite element model of the anterior cruciate ligament,” *Conf Proc IEEE Eng Med Biol Soc*, 2008, pp. 891–894.

[18] Beaulieu M. L., Lamontagne M., and Xu L., 2008, “Gender differences in time-frequency EMG analysis of unanticipated cutting maneuvers,” *Med Sci Sports Exerc*, 40(10), pp. 1795–1804.

[19] Colby S., Francisco A., Yu B., Kirkendall D., Finch M., and Garrett W., 2000, “Electromyographic and kinematic analysis of cutting maneuvers. Implications for anterior cruciate ligament injury,” *Am J Sports Med*, 28(2), pp. 234–240.

[20] DeMorat G., Weinhold P., Blackburn T., Chudik S., and Garrett W., 2004, “Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury,” *Am J Sports Med*, 32(2), pp. 477–483.

[21] Malinzak R. A., Colby S. M., Kirkendall D. T., Yu B., and Garrett W. E., 2001, “A comparison of knee joint motion patterns between men and women in selected athletic tasks,” *Clin Biomech (Bristol, Avon)*, 16(5), pp. 438–445.

[22] White K. K., Lee S. S., Cutuk A., Hargens A. R., and Pedowitz R. A., 2003, “EMG power spectra of intercollegiate athletes and anterior cruciate ligament injury risk in females,” *Med Sci Sports Exerc*, 35(3), pp. 371–376.

- [23] Benjaminse A., Gokeler A., Fleisig G., Sell T., and Otten B., 2011, “What is the true evidence for gender-related differences during plant and cut maneuvers? A systematic review,” *Knee Surgery, Sports Traumatology, Arthroscopy*, 19(1), pp. 42–54.
- [24] Myer G. D., Chu D. A., Brent J. L., and Hewett T. E., 2008, “Trunk and hip control neuromuscular training for the prevention of knee joint injury,” *Clin Sports Med*, 27(3), pp. 425–448, ix.
- [25] Cochrane J. L., Lloyd D. G., Besier T. F., Elliott B. C., Doyle T. L. A., and Ackland T. R., 2010, “Training affects knee kinematics and kinetics in cutting maneuvers in sport,” *Med Sci Sports Exerc*, 42(8), pp. 1535–1544.
- [26] Risberg M. A., Mørk M., Jenssen H. K., and Holm I., 2001, “Design and implementation of a neuromuscular training program following anterior cruciate ligament reconstruction,” *J Orthop Sports Phys Ther*, 31(11), pp. 620–631.
- [27] Bizzini M., Hancock D., and Impellizzeri F., 2012, “Suggestions from the field for return to sports participation following anterior cruciate ligament reconstruction: soccer,” *J Orthop Sports Phys Ther*, 42(4), pp. 304–312.
- [28] Gardinier E. S., Manal K., Buchanan T. S., and Snyder-Mackler L., 2012, “Gait and neuromuscular asymmetries after acute anterior cruciate ligament rupture,” *Med Sci Sports Exerc*, 44(8), pp. 1490–1496.
- [29] Roewer B. D., Di Stasi S. L., and Snyder-Mackler L., 2011, “Quadriceps strength and weight acceptance strategies continue to improve two years after anterior cruciate ligament reconstruction,” *J Biomech*, 44(10), pp. 1948–1953.
- [30] Myer G. D., Ford K. R., Brent J. L., and Hewett T. E., 2007, “Differential neuromuscular training effects on ACL injury risk factors in ‘high-risk’ versus ‘low-risk’ athletes,” *BMC Musculoskelet Disord*, 8, p. 39.
- [31] Hewett T. E., Lindenfeld T. N., Riccobene J. V., and Noyes F. R., 1999, “The effect of neuromuscular training on the incidence of knee injury in female athletes. A prospective study,” *Am J Sports Med*, 27(6), pp. 699–706.
- [32] Hewett T. E., Ford K. R., and Myer G. D., 2006, “Anterior cruciate ligament injuries in female athletes: Part 2, a meta-analysis of neuromuscular interventions aimed at injury prevention,” *Am J Sports Med*, 34(3), pp. 490–498.
- [33] Cheung R. T. H., and Davis I. S., 2011, “Landing pattern modification to improve patellofemoral pain in runners: a case series,” *J Orthop Sports Phys Ther*, 41(12), pp. 914–919.
- [34] Crowell H. P., and Davis I. S., 2011, “Gait retraining to reduce lower extremity loading in runners,” *Clin Biomech (Bristol, Avon)*, 26(1), pp. 78–83.

- [35] Barrios J. A., Crossley K. M., and Davis I. S., 2010, “Gait retraining to reduce the knee adduction moment through real-time visual feedback of dynamic knee alignment,” *J Biomech*, 43(11), pp. 2208–2213.
- [36] Di Stasi S. L., and Snyder-Mackler L., 2012, “The effects of neuromuscular training on the gait patterns of ACL-deficient men and women,” *Clin Biomech (Bristol, Avon)*, 27(4), pp. 360–365.
- [37] Moksnes H., Snyder-Mackler L., and Risberg M. A., 2008, “Individuals with an anterior cruciate ligament-deficient knee classified as noncopers may be candidates for nonsurgical rehabilitation,” *J Orthop Sports Phys Ther*, 38(10), pp. 586–595.

Appendix A

Non-linear piecewise modeling of
the vertical ground reaction force
observed during a jump-cut
maneuver

A.1 Introduction

Sport activities associated with non-contact deceleration anterior cruciate ligament (ACL) injury, like a jump-cut maneuver, produce a vertical ground reaction force (GRF) that contains information relating to the energy with which an individual performs the maneuver. These data are typically consistent and contain two overlaid loading patterns during the period immediately following ground contact. First is a large and relatively fast loading impulse that mimics a gamma distribution, and second is a more gradual linear loading increase. The consistency of the vertical GRF data combined with its apparent relationship to both the gamma distribution function and the linear function suggest that these data can be modeled to obtain descriptive scalar parameters. Obtaining these parameters from GRF data may allow more robust comparisons to be made between subject populations while performing dynamic jumping and cutting activities. The purpose of this study was to design and implement a parameterized model to capture the key components of a vertical GRF time series trace from a jump-cut maneuver in order to better understand and compare these data among different subjects and subject populations.

A.2 Methods

After granting their informed consent, 10 ACL-intact (ACL_{INT}) subjects (5 M, 25.8 ± 4.2 years; 5 F, 24.6 ± 2.3 years) and 10 ACL-reconstructed (ACL_{REC}) subjects (4 M, 29.3 ± 6.1 years; 6 F 24.7 ± 6.7 years) were instructed to perform a jump-cut maneuver associated with a non-contact deceleration ACL injury [1]. Three targets were placed in the testing environment for the subject to land on and jog toward: one on the center of a force plate and the other two 45° and two meters from the landing target. For each trial, the subject stood one meter from the force plate with

their knees bent at 45°. When the subject jumped toward the landing target, a visual prompt instructed the subject to perform a sidestep cut (L or R) and jog toward the respective angled target. Ten trials were performed and the subject was unaware of the directional prompt prior to each trial.

The force plate data was used to collect vertical GRF data at 5,000 Hz. These data were filtered using a digital low-pass Butterworth filter with a cut-off frequency of 100 Hz. For the purposes of this study, the GRF data were cropped and time normalized from 16 milliseconds prior to contact to 60 milliseconds after contact. This temporal window was chosen because it matched the window where biplanar videoradiography kinematic data was collected.

The vertical GRF data was modeled as the sum of two separate functions (Figure A.1). The gradual loading increase after ground contact was modeled using a linear function:

$$f_1(t \mid P_1, P_2) = \left(\frac{P_2 - P_1}{100} \right) \cdot (x + 1)$$

where

t is the time period; and

P_1 & P_2 are the two vertical GRF values shown in Figure A.1.

The characteristic peak observed immediately after ground contact was modeled using a scaled gamma distribution:

$$f_2(t \mid P_3, P_4, P_5) = P_5 \cdot \frac{1}{P_4^{P_3} \Gamma(P_3)} \cdot t^{P_3-1} \cdot e^{\frac{-t}{P_4}}$$

where

t is the time period;

P_3 is the shape parameter;

P_4 is the scale parameter;

P_5 is the magnitude parameter; and

Γ is the Gamma function:

$$\Gamma(P_5) = \int_0^{\infty} e^{-t} t^{P_5-1} dt$$

The final function was defined as the sum of f_1 and f_2 and contains five parameters:

$$f(t \mid P_1, P_2, P_3, P_4, P_5) = f_1(t \mid P_1, P_2) + f_2(t \mid P_3, P_4, P_5)$$

The five parameters of f were estimated using the nonlinear regression algorithm in MATLAB (`nlinfit`). This algorithm estimates the parameters of the function by iteratively adjusting them to obtain a minimum fit residual. The algorithm had an initial guess for each parameter and the maximum number of iterations was set at 1,000. An example of the raw vertical GRF, f using the initial guesses for the five parameters and f using the fitted values for the five parameters, is shown in Figure A.2.

In addition to this model, the total impulse and total stance time were calculated. The total impulse was calculated as the integral of the vertical GRF data with respect to time. The total stance time was calculated as the time change from contact to toe-off.

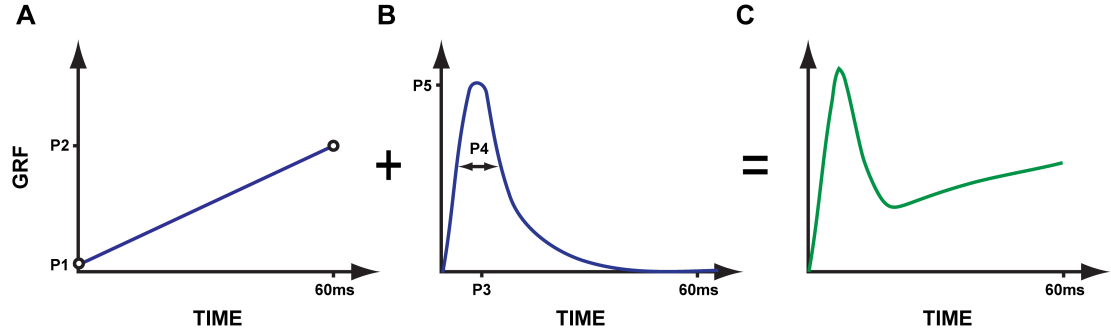


Figure A.1: Representation of functions f_1 and f_2 and the resulting function, f . The GRF intercept (P_1) and GRF at 60 ms (P_2) form the primary components of f_1 . Note that the shape (P_3) and scale (P_4) parameters of relating to the Gamma distribution (f_2) are shown here as the time point and width of the peak. This is an approximation of their influence, and both influence the location and height of the peak as well as the spread of the distribution. However, P_5 does parameterize the maximum height of the peak, as depicted in the figure.

Each estimated parameter from the model as well as the impulse and stance time were determined for each applicable subject trial, and then all trials were ensemble averaged for each subject. Comparisons between gender (M and F) and ACL reconstruction status (ACL_{INT} and ACL_{REC}) were made for all kinematic and kinetic variables using a two way analysis of variance. These tests were performed with a significance level of $\alpha = 0.05$. Pairwise multiple comparisons were made using the Holm-Sidak method when a significant gender and ACL reconstruction status interaction was determined.

A.3 Results

The average mean squared error for the model was 0.02 ± 0.08 body weights. The parameter, P_2 , denoting the final height (within the temporal window studied herein) of the gradually increasing vertical GRF was 0.5 body weights larger in ACL_{INT} subjects compared to ACL_{REC} subjects ($p = 0.03$). A statistically significant interaction between gender and ACL reconstruction status was observed for the parameter, P_5 ,

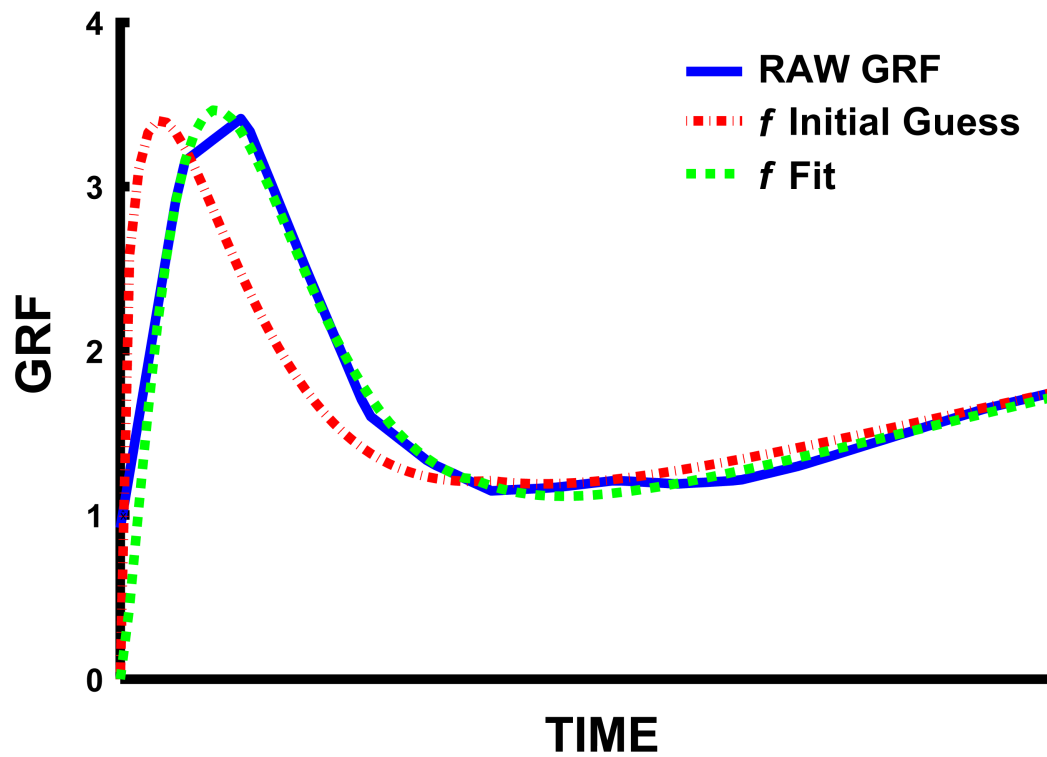


Figure A.2: Representative graph showing an example trial from one subject's raw GRF data (solid blue line). The function, f , is shown using the initial guesses for the five parameters (dot-hashed red line) and the final fitted values for the five parameters (hashed green line).

denoting the peak vertical GRF observed immediately after ground contact ($p = 0.006$). The ACL_{INT} subjects had a P_5 parameter that was 1.43 body weights larger than the ACL_{REC} subjects. The male subjects had a P_5 parameter that was 0.71 body weights lower than the female subjects' P_5 parameter. Within ACL_{REC} subjects, the male and female P_5 parameters were only separated by 0.24 body weights ($p = 0.599$). Conversely, the male and female P_5 parameters were separated by 1.66 body weights within ACL_{INT} subjects ($p = 0.001$). All other model parameters were not observed to be different among gender and ACL reconstruction status (Table A.1).

In addition to the differences observed in the model parameters P_2 and P_5 , an apparent gender difference was observed for the total vertical GRF impulse. The female subjects had a 50.66 body weight $\cdot$ second smaller impulse than the male subjects ($p = 0.059$), with the strongest influence likely arising from the ACL_{REC} subjects. The female ACL_{REC} subjects had an 82.41 body weight $\cdot$ second smaller impulse than the male ACL_{REC} subjects, while the female ACL_{INT} subjects only had a 18.92 body weight $\cdot$ second smaller impulse than the male ACL_{INT} subjects. With regard to stance time, the ACL_{REC} subjects performed the jump-cut maneuver 0.77 seconds faster than the ACL_{INT} subjects ($p < 0.001$).

A.4 Discussion

We successfully applied a five parameter non-linear piecewise model to the vertical GRF from male and female ACL_{INT} and ACL_{REC} subjects performing a jump-cut maneuver. To our knowledge, this is the first attempt at modeling the vertical GRF using a gamma distribution to capture the peak vertical GRF observed immediately following ground contact and a linear function to capture the gradual increase in vertical GRF during the initial stance period. The ACL_{REC} subjects appear to have a

OUTCOME	CONDITION	MEAN	SEM	GENDER	MEAN	SEM	INTERACTION	MEAN	SEM
P_1	ACL _{INT}	0.27	0.07	M	0.13	0.08	I x M	0.15	0.10
	ACL _{REC}	0.13	0.07	F	0.27	0.07	I x F R x M R x F	0.39 0.11 0.16	0.10 0.12 0.09
P_2	ACL _{INT}	1.80	0.15	M	1.50	0.16	I x M	1.72	0.21
	ACL _{REC}	1.30	0.15	F	1.60	0.14	I x F R x M R x F	1.88 1.28 1.33	0.21 0.23 0.19
P_3	ACL _{INT}	48.49	5.79	M	57.53	6.10	I x M	50.62	8.63
	ACL _{REC}	58.20	5.79	F	49.15	5.46	I x F R x M R x F	46.35 64.45 51.96	7.72 8.63 7.72
P_4	ACL _{INT}	0.74	0.10	M	0.80	0.11	I x M	0.93	0.15
	ACL _{REC}	0.76	0.10	F	0.69	0.09	I x F R x M R x F	0.59 0.68 0.08	0.13 0.15 0.13
P_5^*	ACL _{INT}	2.50	0.21	M	1.43	0.22	I x M	1.67	0.30
	ACL _{REC}	1.07	0.21	F	2.14	0.20	I x F R x M R x F	3.33 1.19 0.96	0.30 0.33 0.27
IMPULSE (BW-SEC)	ACL _{INT}	513.60	17.48	M	542.66	18.54	I x M	523.06	24.72
	ACL _{REC}	521.06	17.48	F	492.00	16.73	I x F R x M R x F	504.14 562.27 479.86	24.72 27.64 22.56
STANCE TIME (SEC)	ACL _{INT}	1.86	0.04	M	1.43	0.04	I x M	1.81	0.06
	ACL _{REC}	1.09	0.04	F	1.53	0.04	I x F R x M R x F	1.92 1.04 1.14	0.06 0.06 0.05
* MVF in I: p=0.001; MVF in R: p=0.599; IvR in M: p=0.294; IvR in F: p<0.001									

Table A.1: ACL_{INT} and ACL_{REC} male and female model parameter, vertical GRF impulse, and stance time results. A two-way analysis of variance was performed on each set of data. Statistically significant differences ($p \leq 0.05$) are highlighted using a dark gray background fill. Near statistical significance ($p \leq 0.10$) is highlighted using a light gray background fill. Pair-wise comparisons were made using the Holm-Sidak method when a significant gender and ACL reconstruction status interaction was determined. These results, denoted by the *, are shown in the bottom row of the table.

more gradual increase their vertical GRF compared to the ACL_{INT} subjects. This is evidenced by the similar starting vertical GRF (P_1) regardless of ACL reconstruction status combined with the lower ending vertical GRF (P_2) in ACL_{REC} subjects as compared to the ACL_{INT} subjects. In addition, the ACL_{REC} subjects have a smaller peak vertical GRF parameter (P_5) compared to the ACL_{INT} subjects. Within the ACL_{INT} subjects, the females' peak vertical GRF parameter is strikingly larger than the peak vertical GRF parameter observed for male subjects. The decreased energy with which the ACL_{REC} subjects are performing the jump-cut maneuver is further emphasized by the significantly lower stance time compared to the ACL_{INT} subjects. The combination of these findings provides further evidence that, even after five years, ACL_{REC} subjects have physiological deficiencies, neuromechanical deficiencies, and/or protective habits obtained during rehabilitation that contribute to the decreased energy with which they perform the jump-cut maneuver.

The model used in this study was designed to parameterize the data within the first 60 milliseconds after contact, and it provides insight into the initial loading period of a jump-cut maneuver. In the future, these modeling techniques can be applied to the entire vertical GRF trace to quantify different aspects of the curve. The simplest additions would likely be in the form of a piecewise linear model that parameterizes the transition from the gradual loading phase through the stance phase and then toe-off; however, a quadratic or cubic function may be more effective. It is important to note that the insights gained during this analysis provided the foundation for the analysis presented in Chapter 5. Based on the results for the scale and shape parameters of the gamma distribution, we were able to simplify the analysis without sacrificing outcomes and interpretation.

A.5 Conflicts of interest

None.

A.6 Acknowledgements

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A.7 References

[1] Ford K. R., Myer G. D., Toms H. E., and Hewett T. E., 2005, “Gender differences in the kinematics of unanticipated cutting in young athletes,” *Med Sci Sports Exerc*, 37(1), pp. 124–129.

Appendix B

Bone surface interactions during a jump-cut maneuver: effect of ACL reconstruction surgery

Daniel L. Miranda, Michael J. Rainbow, Braden C. Fleming

The following appendix was submitted to the *59th Annual Meeting of the Orthopaedic Research Society* in San Antonio, Texas on January 26-29, 2013.

B.1 Introduction

Non-contact deceleration anterior cruciate ligament (ACL) injuries that occur during jump-cut maneuvers often result in multi-dimensional knee instability. This instability is typically treated by arthroscopic ligament reconstruction. Despite the restoration of functional knee stability, 80% to 90% of patients with ACL reconstruction have early signs and faster progression of osteoarthritis after surgery. The mechanics at the articular surfaces of the knee are not fully understood. A better understanding of the interactions between the articulating surfaces of the femur and tibia may help to explain why ACL injured patients are at higher risk for premature joint degradation.

The purpose of this study was to compare the interactions between the femoral and tibial articulating surfaces of ACL-intact (ACL_{INT}) and ACL-reconstructed (ACL_{REC}) recreational athletes during a jump-cut maneuver using biplanar videoradiography. The specific aims were to determine differences in bone surface interactions due to ACL status between ACL_{REC} patients who were at least five years post-surgery and ACL_{INT} subjects.

B.2 Methods

All procedures were approved by the IRB. Twenty recreational athletes were enrolled in this study: 10 ACL_{INT} (5 males, 5 females); 10 ACL_{REC} (4 males, 6 females; 7 bone-patellar tendon-bone autografts, 3 hamstring tendon autografts). The inclusion criteria for the ACL_{INT} subjects were: (1) no lower extremity injury; (2) no neurological disease(s); (3) no pregnancy; and (4) a Tegner activity score of five or greater [2]. The additional inclusion criteria for the ACL_{REC} patients were: (1) unilateral ACL reconstruction; and (2) at least five years post ACL reconstruction. One leg was

imaged for each subject. The leg was chosen at random for the ACL_{INT} subjects (6L and 4R). The ACL reconstructed leg was chosen for the ACL_{REC} subjects (7L and 3R).

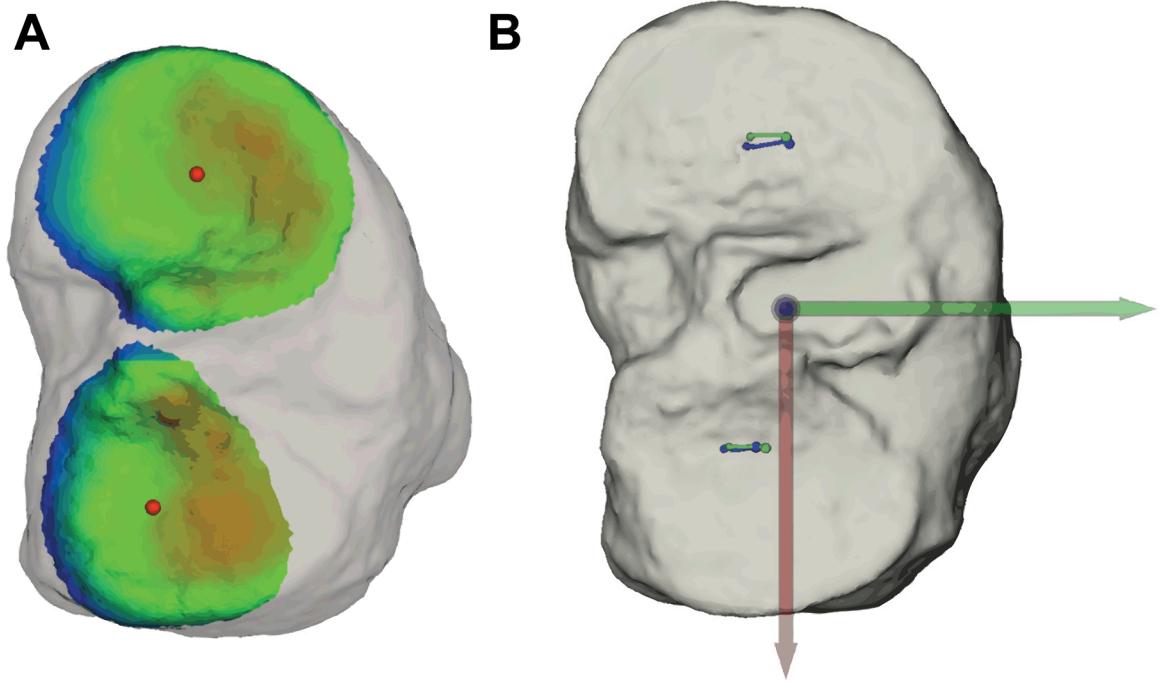


Figure B.1: (A) Example proximity area and proximity center (red sphere) for the medial and lateral compartments of the tibia. Blue and red are farthest and closest to the opposing surface, respectively. (B) Mean proximity center path for the ACL_{INT} (green) and ACL_{REC} (blue) subjects. The large and small spheres represent the beginning and end of the paths, respectively. Most of the proximity center translation is occurring in the AN/PO direction. The vectors represent the ME/LA (red) and AN/PO (green) axes of the tibial ACS.

The jump-cut maneuver was designed to mimic maneuvers associated with non-contact deceleration ACL injury [3]. Three targets were placed in the testing environment: one on the center of a force plate and the other two toward the left and right of the landing target at an angle of 45° and a distance of two meters. A verbal prompt was used to cue the subject to jump upward and forward toward the landing target. A visual directional prompt (L or R) instructed the subject to perform a sidestep cut and jog toward one of the angled targets. For example, when subjects cut to the left, they pushed off with their right foot and led with their left. Ten trials

were performed and the subject was unaware of the directional prompt. Biplanar videoradiography data were collected during a subset of these trials.

Biplanar videoradiography measurements were collected at 250 Hz and time synchronized with the force plate, which collected ground reaction force (GRF) data at 5,000 Hz. These data were collected in the W.M. Keck Facility at Brown University. Additionally, a CT scan was obtained for each subject's instrumented leg. The biplanar videoradiography data were processed using custom markerless tracking software [4]. Anatomical coordinate systems (ACSs) were created from 3-D CT bone models [5]. All kinematic data were filtered using a digital low-pass Butterworth filter with a cutoff frequency of 25 Hz.

The knee surface interactions were based on the articular surfaces from the 3-D CT bone models. These interactions were determined at every kinematic frame and summarized by the proximity area, proximity center location, and proximity center distance. The proximity area was defined as the total polygonal surface area within a threshold distance of nine millimeters from the opposing surface. The proximity center location was defined as the central point on the proximity area weighted by the average of the distances of each polygon to the closest point on the opposing surface (Figure B.1). The proximity center distance was defined as the minimum distance of the proximity center to the opposing surface. The proximity area, proximity center, and proximity center distance were calculated between the medial and lateral compartments of the tibia using the opposing surfaces of the femur, respectively. The two compartments were isolated using the sagittal plane formed by the x- and y-axes of the tibial ACS.

The proximity area, proximity center position, and proximity center distance were cropped from 16 ms prior to contact to 60 milliseconds after contact. The total post-contact medial/lateral (ME/LA) and anterior/posterior (AN/PO) proximity center

excursions were calculated. The proximity center distance was zeroed at contact, and the post-contact excursions were calculated. The proximity area was binned in one millimeter distance increments, and the post-contact time integral (AUC) for each bin was determined. These outcomes were calculated for each trial and ensemble averaged for each subject. Comparisons between ACL_{INT} and ACL_{REC} were made using a one way analysis of variance. Comparisons between the medial and lateral compartments were made using paired t-tests.

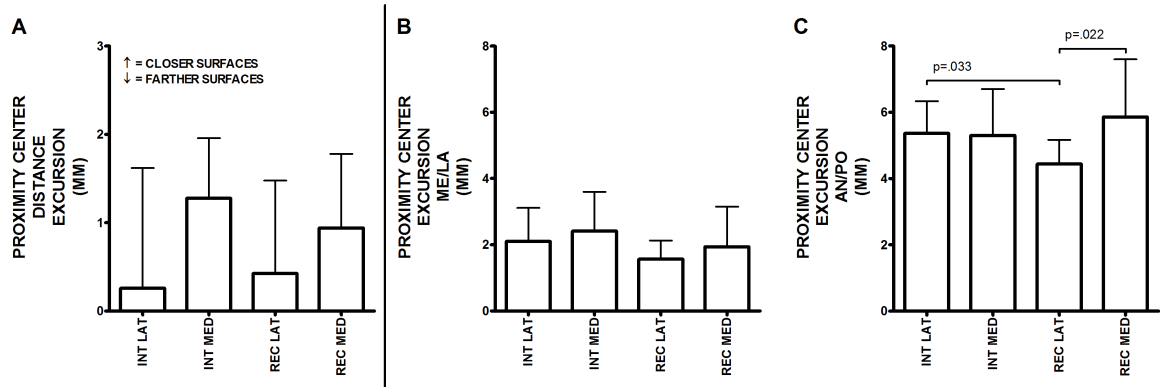


Figure B.2: (A) The mean proximity center distance excursion after contact. All proximity center distance excursions were decreasing, meaning the surfaces were moving closer together. (B) The mean ME/LA proximity center excursion. (C) The AN/PO proximity center excursion. The units are in millimeters for the proximity center distance excursion plot and both proximity center excursion plots. All error bars represent one standard deviation from the mean. Any statistically significant differences are labeled with a bracket.

B.3 Results

The proximity center distance decreased after contact for all subjects (Figure B.2A). While not significant, the distance decreased more for the medial compartment than the lateral compartment for all subjects. For ME/LA proximity center excursion, no differences were observed between ACL_{INT} and ACL_{REC} subjects (Figure B.2B). The mean AN/PO proximity center excursions were higher than the ME/LA excursions for

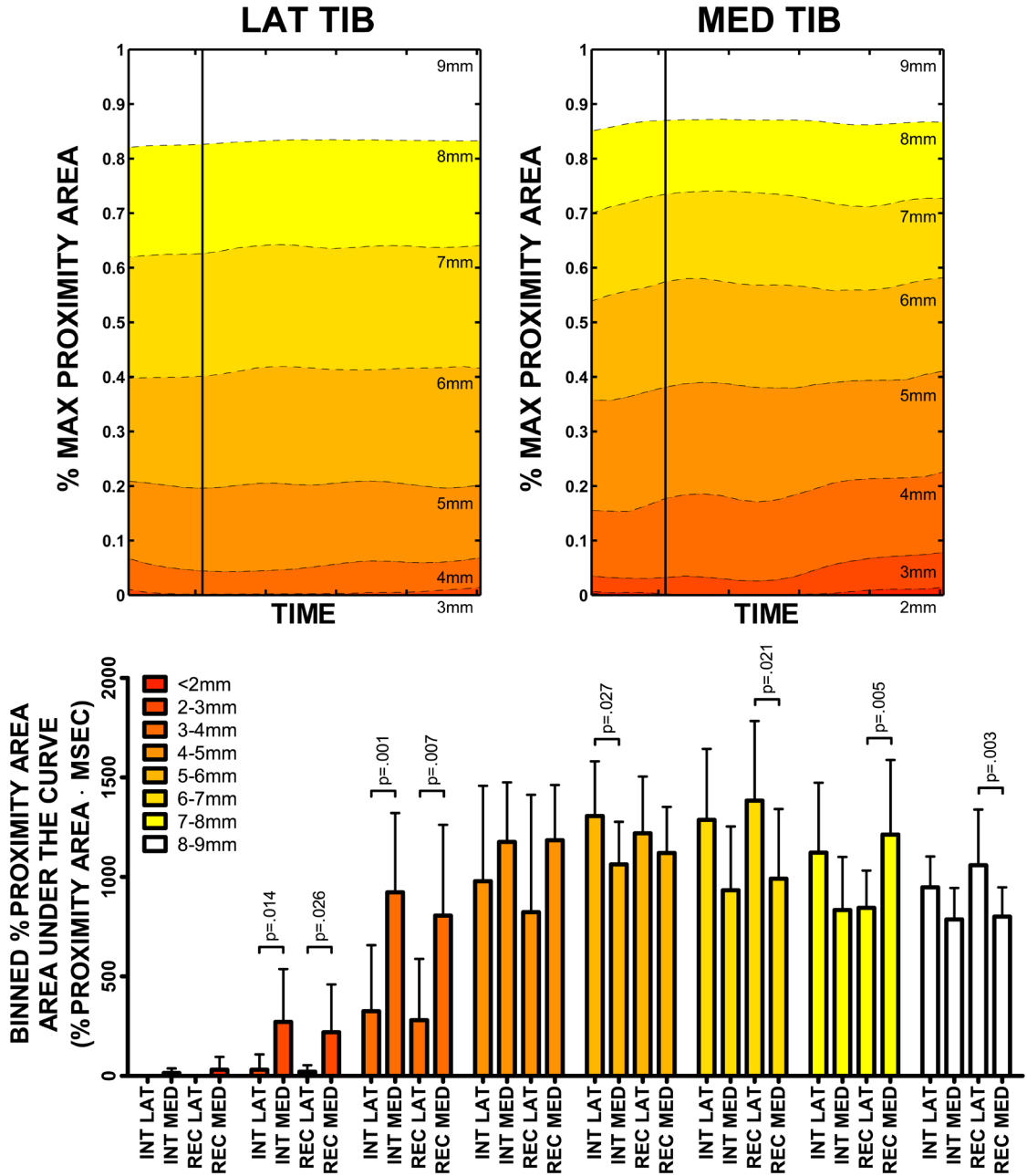


Figure B.3: (Top) Mean binned percent max proximity area for all subjects. The hashed lines are labeled with the bin distance. The vertical line is the time of ground contact. (Bottom) Bar plot representing the binned percent proximity area's AUC. The units are percent proximity area · milliseconds. All bars represent the mean, and the error bars denote one standard deviation from the mean. Any statistically significant differences are labeled with a bracket.

each compartment (Figure B.2C). For the ACL_{REC} subjects, the AN/PO proximity center excursion was 1.42 mm larger for the medial compartment compared to the lateral compartment. Furthermore, the AN/PO proximity center excursion was 0.93 mm larger for the lateral compartment of the ACL_{INT} subjects compared to the lateral compartment of the ACL_{REC} subjects.

No significant differences between ACL_{INT} and ACL_{REC} subjects were observed for the binned proximity area measurements. In general, the medial compartment for both ACL_{INT} and ACL_{REC} subjects had a larger percentage of area within four millimeters of the opposing femoral surface compared to the lateral compartment (Figure B.3).

B.4 Discussion

We have compared three articular surface interaction parameters from ACL_{INT} and ACL_{REC} recreational athletes during a jump-cut maneuver associated with non-contact deceleration ACL injury. We found that both subject populations experienced a decrease in proximity center distance after contact and had larger AN/PO proximity center excursion compared to ME/LA excursion. Furthermore, we observed a larger percentage of the proximity area to be within four millimeters of the opposing surface in the medial compartment compared to the lateral compartment for all subjects. These findings are consistent with results from Anderst et al. during a one-legged hop [6]. While a significant difference was only observed between ACL_{INT} and ACL_{REC} subjects for the AN/PO proximity center excursion in the lateral compartment, there was a trend for the proximity center distance to decrease in the medial compartment compared to lateral compartment for both ACL_{INT} and ACL_{REC} subjects. In the future, we hope to apply and build on these techniques to elucidate subtle articular surface interaction differences between ACL intact, deficient, and reconstructed recre-

ational athletes. These non-traditional kinematic outcomes may help us understand how kinematic changes affect the articulating surfaces of the knee, and may be useful for studying the onset and progression of post-traumatic osteoarthritis.

B.5 Conflicts of interest

None.

B.6 Acknowledgements

This study was funded by grants from the W.M. Keck Foundation, NIH R01-AR047910, and NIH COBRE P20-RR02484. The authors wish to thank Dr. Michael J. Bey and Mr. Roger Zauel at the Henry Ford Hospital for providing the proximity center algorithm.

B.7 References

- [1] Ben W., and Kibler M. D., 2009, Orthopaedic Knowledge Update: Sports Medicine 4, Amer Academy of Orthopaedic.
- [2] Tegner Y., and Lysholm J., 1985, "Rating systems in the evaluation of knee ligament injuries," Clin. Orthop. Relat. Res, (198), pp. 43–49.
- [3] Ford K. R., Myer G. D., Toms H. E., and Hewett T. E., 2005, "Gender differences in the kinematics of unanticipated cutting in young athletes," Med Sci Sports Exerc, 37(1), pp. 124–129.
- [4] Miranda D. L., Schwartz J. B., Loomis A. C., Brainerd E. L., Fleming B. C., and Crisco J. J., 2011, "Static and Dynamic Error of a Biplanar Videoradiography System Using Marker-Based and Markerless Tracking Techniques," J Biomech Eng, 133(12), p. 121002.
- [5] Miranda D. L., Rainbow M. J., Leventhal E. L., Crisco J. J., and Fleming B. C., 2010, "Automatic determination of anatomical coordinate systems for three-dimensional bone models of the isolated human knee," J Biomech, 43(8), pp. 1623–1626.
- [6] Anderst W. J., and Tashman S., 2003, "A method to estimate in vivo dynamic articular surface interaction," J Biomech, 36(9), pp. 1291–1299.

Appendix C

Biplanar videoradiography derived ACL excursions during a jump-cut maneuver associated with ACL injury

Daniel L. Miranda, Michael J. Rainbow, Joseph J. Crisco, Braden C. Fleming

The following appendix was presented as a podium presentation at the *12th Annual International Symposium on Ligaments and Tendons* in San Francisco, California on February 3, 2012.

C.1 Introduction

Activities involving jumping and cutting are commonly associated with non-contact deceleration ACL injury. High-speed biplanar videoradiography provides a means to accurately quantify joint kinematics and ligament length changes during dynamic activities. The purpose of this study was to compare how ACL length and knee kinematics change across three phases of a jump-cut maneuver.

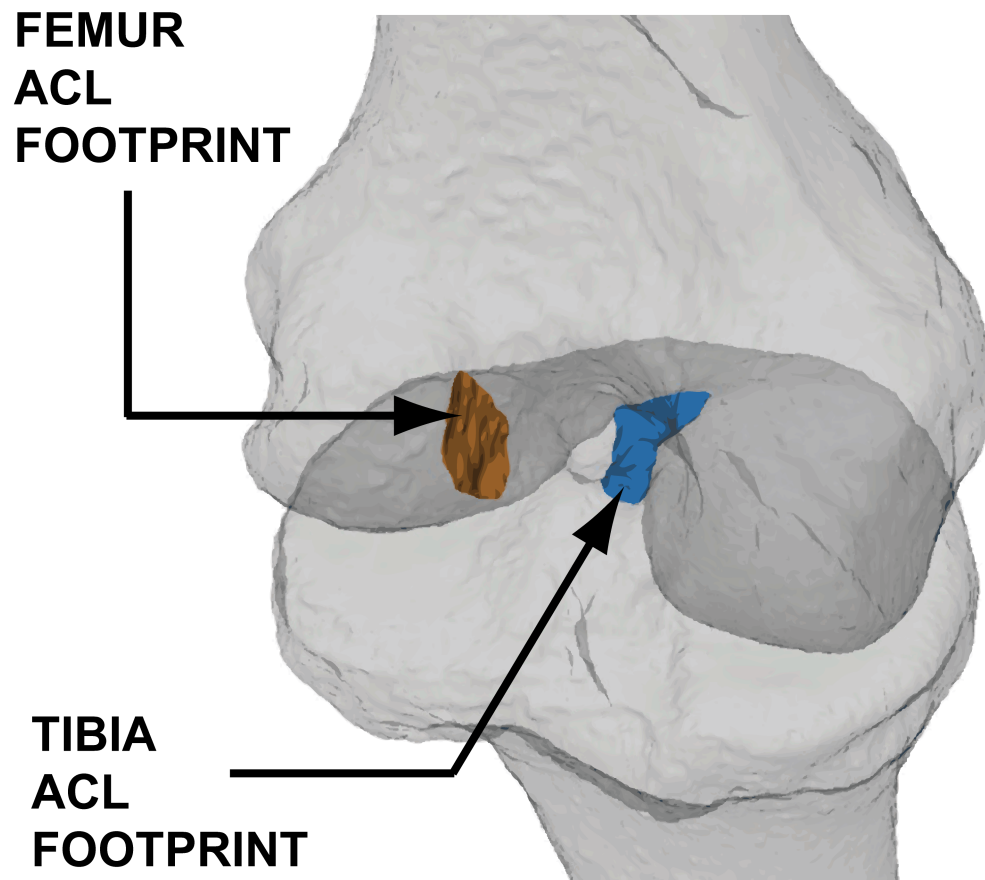


Figure C.1: Illustration depicting an example set of ACL insertion sites for the femur in orange and tibia in blue.

C.2 Methods

After granting their informed consent, 10 healthy volunteers (5 M, 25.8 ± 4.2 years; 5 F, 24.6 ± 2.3 years) were instructed to perform a jump-cut maneuver associated with a non-contact deceleration ACL injury [1]. Three targets were placed in the testing environment for the subject to land on and jog toward: one on the center of a force plate and the other two 45° and two meters from the landing target. For each trial, the subject stood one meter from the force plate with their knees bent at 45° . When the subject jumped toward the landing target, a visual prompt instructed the subject to perform a sidestep cut (L or R) and jog toward the respective angled target. Ten trials were performed and the subject was unaware of the directional prompt prior to each trial. Three-dimensional femur and tibia models were created using Materialise Mimics 13 (Ann Arbor, MI, USA) and their respective ACL insertion sites (Figure C.1) were outlined according to anatomic references from Girgis et al. [2] and Tsukada et al. [3] using Geomagic Studio 11 (Morrisville, NC, USA).

Biplanar videoradiography data was collected at 250 Hz during three trials. Ground reaction forces (GRF) were time synchronized and collected at 5,000 Hz. CT scans were obtained for each subject's imaged leg. The biplanar videoradiography data were processed using custom markerless tracking software (Autoscooper; Brown University, Providence, RI, USA). The center of each insertion site was then tracked using the biplanar videoradiography motion data, and the ACL length was determined as the distance between the insertion site centroids. Percent ACL elongation was calculated as:

$$ACL_e = 100 \times \left(\frac{L_{ACL_i} - L_{ACL_{ref}}}{L_{ACL_{ref}}} \right)$$

where

ACL_e is the percent ACL elongation;

L_{ACL_i} is the length of the ACL at a given time-point; and

$L_{ACL_{ref}}$ is the length of the ACL as determined from the CT scan.

Knee joint adduction/abduction (AD\AB) and internal/external (IN\EX) rotations were determined in Visual3D using the method described by Grood and Suntay [4]. All data were separated into three periods: flight phase to contact (A), contact to peak GRF (B), and following peak GRF (C) (Figure C.2). All trials were time normalized within each period and ensemble averaged across each subject's trials. The total excursions of the ACL (max-min), AD\AB, and IN\EX rotations were determined for each subject across each period. Kruskal-Wallis and Dunn's multiple comparison tests were used to test for significant differences between periods.

C.3 Results

The median ACL excursion values varied significantly across the three periods ($p = 0.026$). The ACL lengthened to the greatest extent (Figure C.3A) after peak ground reaction force (period C). Median AD\AB excursion (Figure C.3B) was not found to be statistically significant across periods ($p = 0.117$). However, the median IN\EX rotational excursion varied significantly across the three periods ($p < 0.0001$). External rotation of the tibia occurred prior to contact. Internal rotation of the tibia began after contact and through peak GRF (Figure C.3C).

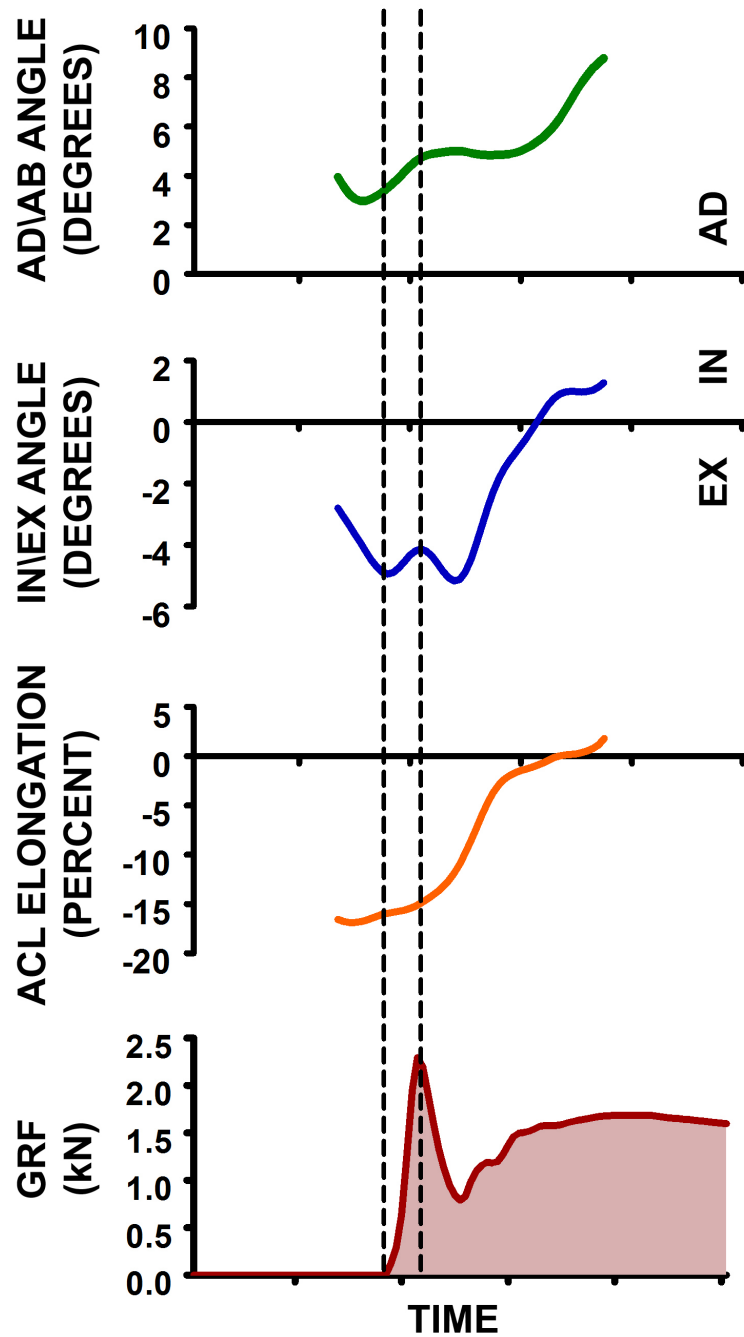


Figure C.2: (A) Knee AD/AB angle in degrees. (B) Knee IN/EX angle in degrees. (C) Percent ACL elongation. (D) Ground reaction force in kilonewtons. All kinetic and kinematic traces are from a single subject trail. The vertical hashed lines represent ground contact and peak vertical GRF. The time period before the first vertical hashed line is period A. The time period between the two vertical hashed lines is period B. The time period following the second vertical hashed line is period C.

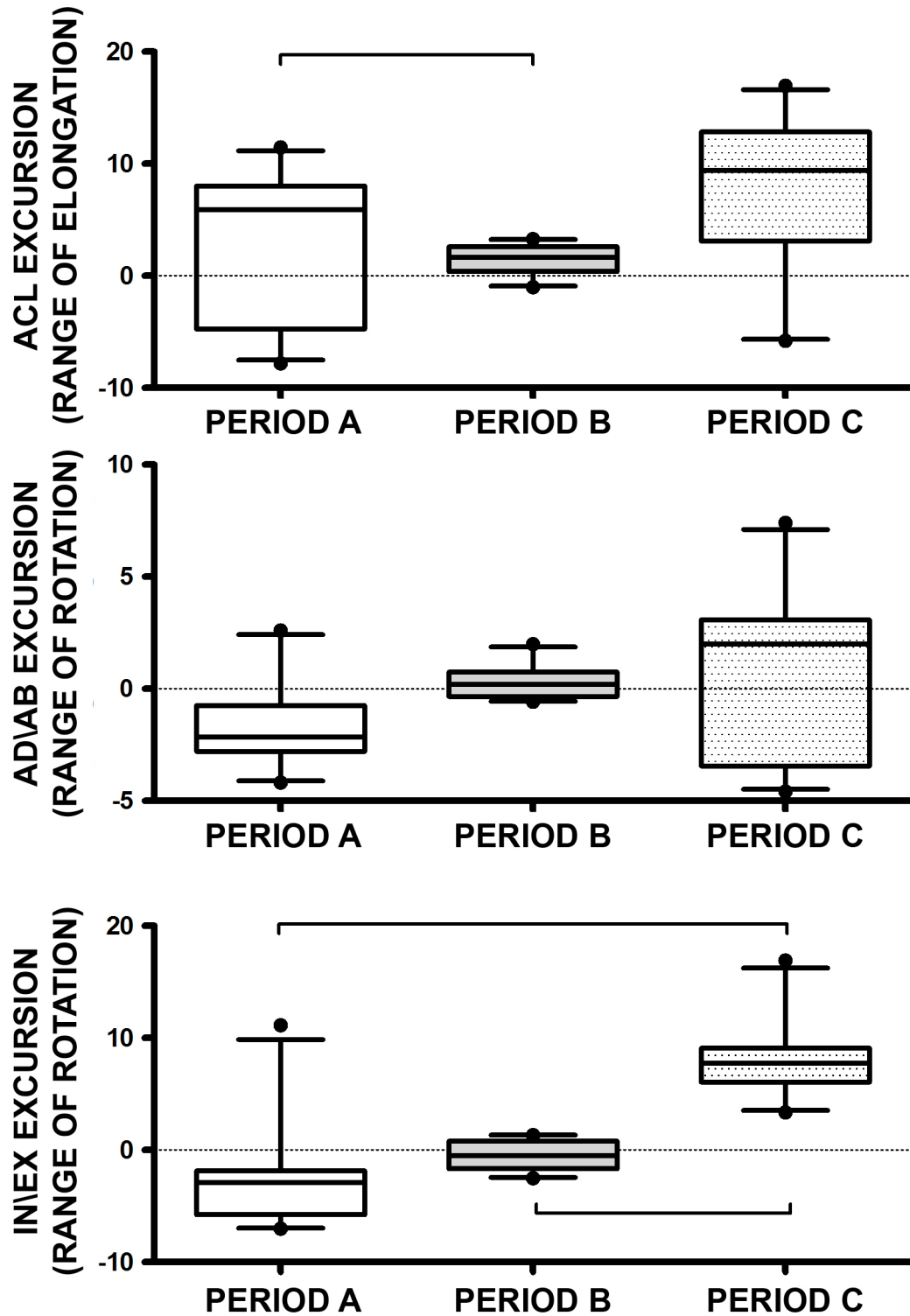


Figure C.3: (A) Signed ACL elongation excursion. (B) Signed knee adduction/abduction angle excursion. (C) Signed internal/external angle excursion for periods A, B, and C. The bracketed lines connecting box plots denotes that a pairwise statistical difference was found.

C.4 Discussion

We have successfully tracked the *in vivo* kinematics of the femur and tibia during a jump-cut maneuver using biplanar videoradiography technology. Using this technology, we successfully generated reliable secondary knee rotations and estimated ACL elongation during three periods of the jump-cut maneuver. An increase in ACL elongation and internal rotation of the tibia occurred after ground contact, most notably after the peak GRF. This is consistent with results of Tashman et al., where the ACL lengthens and the tibia internally rotates after foot strike during downhill running [5]. Additionally, these results support the proposition that excessive internal tibial rotation could produce strains contributing to ACL failure.

While this study is limited by subject number, it provides insight into how knee rotations affect ACL biomechanics during an activity associated with non-contact deceleration ACL injury. Additionally, it is our intent to incorporate improved modeling techniques that take full advantage of the accuracy of biplanar videoradiography technology. Specifically, the model presented herein is a simple linear approximation of the ACL. Modeling the anteromedial bundle (AMB) and posterolateral bundle (PLB) using multiple piecewise linear fibers may provide additional insight into how the ligament is functioning across its cross-section during the different periods of the jump-cut maneuver. Moreover, computational methods exist that use the bony morphology to estimate ligament lengths [6]. Using these methods to account for ligament and fiber wrapping, specifically around the medial aspect of the lateral femoral condyle, will provide more accurate elongation estimates over the entire jump-cut kinematic profile. Ultimately, it is our hope that these techniques will help elucidate differences between gender and injury condition (ACL deficiency and reconstruction) to inform novel injury prevention and rehabilitation strategies.

C.5 Conflicts of interest

None.

C.6 Acknowledgements

This study was funded by grants from the W.M. Keck Foundation and the NIH (R01-AR047910-07; P20-RR02484).

C.7 References

- [1] Ford K. R., Myer G. D., Toms H. E., and Hewett T. E., 2005, “Gender differences in the kinematics of unanticipated cutting in young athletes,” *Med Sci Sports Exerc*, 37(1), pp. 124–129.
- [2] Girgis F. G., Marshall J. L., and Monajem A., 1975, “The cruciate ligaments of the knee joint. Anatomical, functional and experimental analysis,” *Clin. Orthop. Relat. Res*, (106), pp. 216–231.
- [3] Tsukada H., Ishibashi Y., Tsuda E., Fukuda A., and Toh S., 2008, “Anatomical analysis of the anterior cruciate ligament femoral and tibial footprints,” *J Orthop Sci*, 13(2), pp. 122–129.
- [4] Grood E. S., and Suntay W. J., 1983, “A joint coordinate system for the clinical description of three-dimensional motions: application to the knee,” *J Biomech Eng*, 105(2), pp. 136–144.
- [5] Tashman S., Collon D., Anderson K., Kolowich P., and Anderst W., 2004, “Abnormal rotational knee motion during running after anterior cruciate ligament reconstruction,” *Am J Sports Med*, 32(4), pp. 975–983.
- [6] Marai G. E., Laidlaw D. H., Demiralp C., Andrews S., Grimm C. M., and Crisco J. J., 2004, “Estimating joint contact areas and ligament lengths from bone kinematics and surfaces,” *IEEE Trans Biomed Eng*, 51(5), pp. 790–799.

Appendix D

Effects of gender and ACL reconstruction status on muscle activity of the quadriceps, hamstring, and gastrocnemius during a jump-cut maneuver

Margaret S. Coats-Thomas, Daniel L. Miranda, Braden C. Fleming

The following appendix was submitted to the *59th Annual Meeting of the Orthopaedic Research Society* in San Antonio, Texas on January 26-29, 2013.

D.1 Introduction

The ACL is the most common fully disrupted ligament in the knee, and the risk of injury is greatest in sports that involve cutting maneuvers without contact from another player. ACL injuries occur 2-10 times more frequently in female athletes compared to male athletes, and patients following ACL reconstruction surgery are at high risk for tearing their graft [1,2]. It is believed that the alterations in muscle activity of the quadriceps (Q), hamstring (H), and gastrocnemius (G) muscles are factors contributing to the risk of injury. To further understand mechanisms of ACL injuries, it is necessary to explore the individual and combined functions of these muscles during activities associated with non-contact ACL injury in both ACL-intact (ACL_{INT}) and ACL-reconstructed (ACL_{REC}) subjects.

The purpose of this study was to compare timing and activity levels of the Q, H, and G muscles when male and female ACL_{INT} and ACL_{REC} subjects perform a jump-cut maneuver. Outcome measures include the muscle onset time, the time to peak activity, and the Q:H and G:H muscle activity ratios. We hypothesize that differences in all of these outcomes will be observed between gender and ACL reconstruction status.

D.2 Methods

All subjects granted their informed consent following Institutional Review Board approval. Ten ACL_{INT} subjects (5 females ages 24.6 ± 2.3 years, 5 male ages 25.8 ± 4.2 years) and ten ACL_{REC} subjects (6 females ages 24.7 ± 6.7 years, 4 males ages 29.3 ± 6.1 years) participated in the study. For ACL_{INT} subjects, one leg was chosen at random (6 left and 4 right), and for ACL_{REC} subjects the reconstructed leg was evaluated (7 left, 3 right). The chosen leg was outfitted with 23 retro-reflective surface

markers and six EMG electrodes. The surface electrodes were applied to two quadriceps muscles: rectus femoris (RF) and vastus medialis (VM), two hamstring muscles: semitendinosus (ST) and biceps femoris (BF), and both heads of the gastrocnemius muscle: medial (MG) and lateral (LG) following a standardized EMG protocol.

Subjects were asked to perform a jump-cut maneuver intended to recreate a motion associated with non-contact ACL injury (Figure 4.1A & Figure 5.1A). Subjects stood with knees bent at a 45° flexion angle approximately two meters from the force plate. The subject was cued by a verbal prompt to jump toward the landing target at the center of the force plate. Then, a visual directional cue instructed the subject to cut toward an additional target placed one meter away from the landing target and at a 45° angle to either the right or left. Each subject then jogged past the target after cutting. The subject performed ten trials.

EMG data were recorded for the RF, VM, ST, BF, MG and LG using a Myomonitor IV Wireless EMG System and Single Differential Detection surface EMG sensors (Delsys, Boston, MA, USA) at a sampling frequency of 5,000 Hz. This study was part of a larger study where kinematic data were collected using optical motion capture and biplanar videoradiography at 250 Hz. All EMG signals were full-wave rectified then high-pass filtered at 20 Hz and low-pass filtered at 500 Hz. Filtered signals were smoothed by calculating the root mean square value within a 500 millisecond moving window (Figure D.1). All signals were processed using Visual3D (C-Motion, Germantown, MD, USA).

Muscle onset times were determined by finding the point at which the signal reached a value more than three standard deviations (SD) above the mean for all muscles except the BF, for which it was two SD. The mean was calculated from a manually defined rest period (time at the start of the trial when the subject was standing still). Muscle peak times were determined by finding the global maximum of the

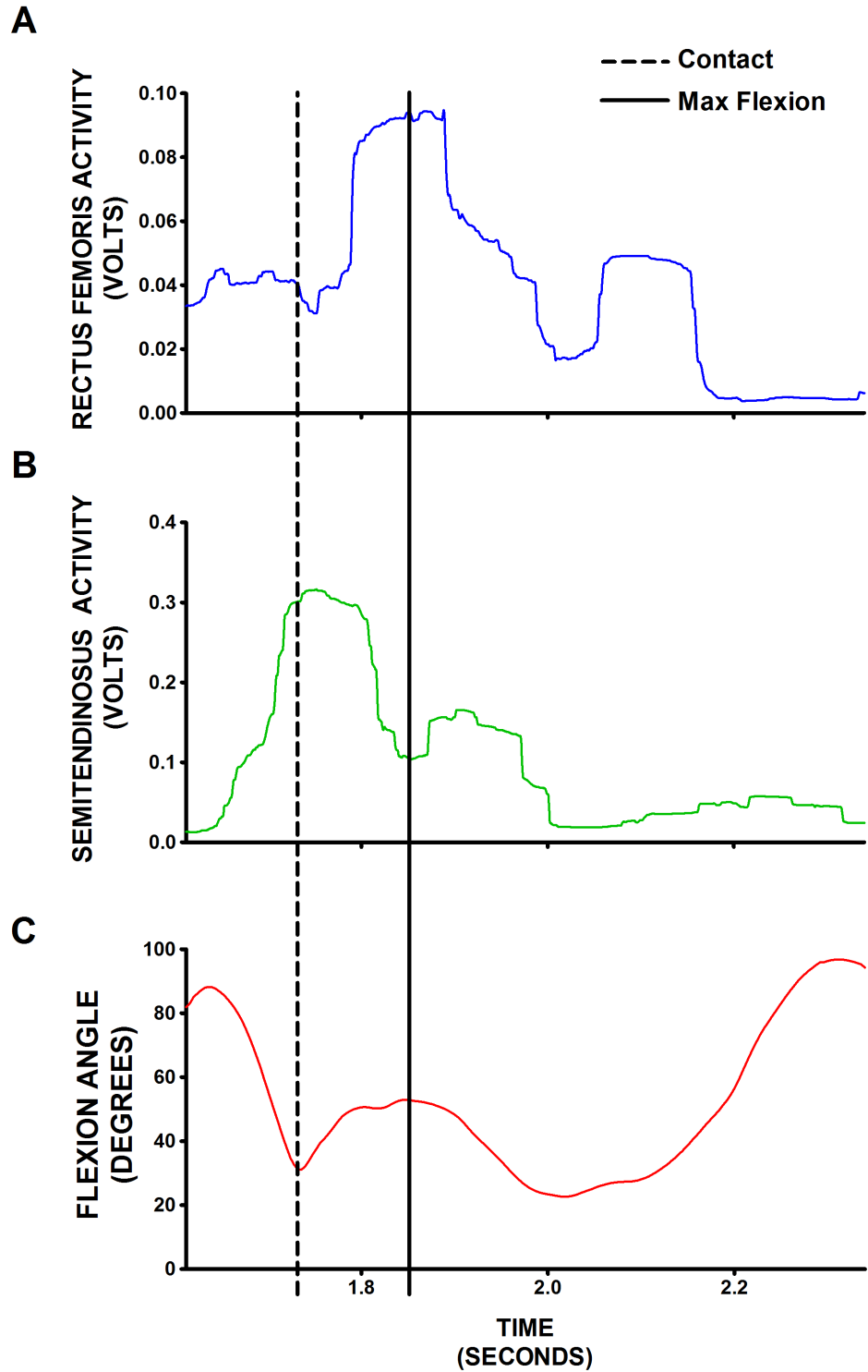


Figure D.1: (A) Sample processed, smoothed EMG trace for the RF. (B) Sample processed, smoothed EMG trace for the ST. (C) Trace of the flexion angle of the knee during the jump-cut maneuver. For A, B, and C: the dotted line shows the time at which the subject made contact with the force plate, and the solid line shows the point at which the knee reached maximum flexion.

processed, smoothed signals. The signals for each muscle were integrated over the preparatory phase (PREP; 50 milliseconds before contact to contact) and the loading phase (LOAD; contact to max flexion). Integration values were recorded to determine the ratios of Q:H and G:H muscle activity for each trial during PREP and LOAD; then the average of the trials for each subject during each phase was calculated. A two-way analysis of variance ($p\text{-value} \leq 0.05$) was used to test for significance of each outcome measure.

D.3 Results

We observed RF and VM peak timing to be 71 milliseconds and 78 milliseconds earlier in ACL_{INT} subjects than in ACL_{REC} subjects, respectively (RF: $p = 0.036$; VM: $p = 0.026$) (Figure D.2). We also found BF peak timing to be 75 milliseconds earlier in females than in males ($p = 0.035$) and 90 milliseconds earlier in ACL_{INT} subjects than in ACL_{REC} subjects ($p = 0.014$). LG and MG peak times were observed to be 106 milliseconds and 87 milliseconds earlier in females than in males, respectively (LG: $p = 0.012$; MG: $p = 0.010$). Finally, we found that MG peak timing was 77 milliseconds earlier in ACL_{INT} subjects than in ACL_{REC} subjects ($p = 0.020$).

D.4 Discussion

We successfully measured the Q, H, and G muscle activity during a jump-cut maneuver. The significant differences we observed suggest the BF peaks earlier in females than in males, a finding supported by a previous EMG study of a similar unanticipated cutting maneuver [3]. Our results show that the G muscle (MG and LG) peaked earlier in females than in males as well. Considering the gender-bias of ACL injury

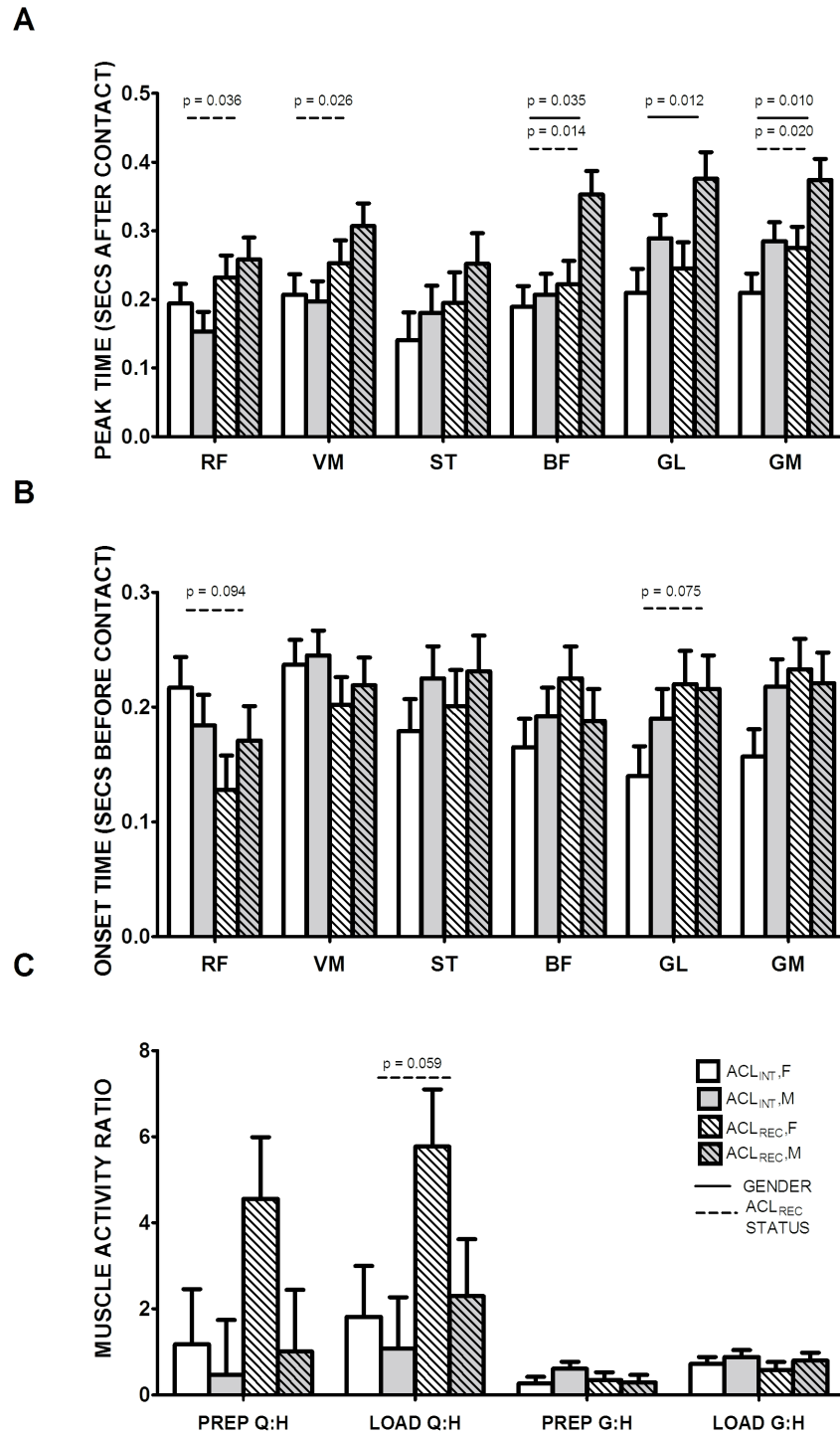


Figure D.2: Bar graphs showing the mean peak times (A) and the mean onset times (B) for all 6 muscles studied. (C) Bar graph showing the PREP and LOAD Q:H and G:H muscle activity ratios. For A, B, and C: dotted and solid lines above the bars display p-values for differences among gender (male versus female) or ACL reconstruction status (ACL_{INT} versus ACL_{REC}) that were either significant or close to being significant.

and the fact that the G is an antagonist of the ACL, this is an interesting finding [4]. Past studies have shown there is a somatosensory deficit in an ACL_{REC} knee, which leads to a prolonged reaction time due to impaired proprioception [5]. Our results suggest that the RF, VM, BF and MG in ACL_{REC} subjects peak later after contact than those in ACL_{INT} subjects. A somatosensory deficit in the ACL_{REC} knees could explain these delayed peak times. While many studies have compared Q:H muscle activity ratios across gender, a comparison of these ratios between ACL_{REC} and ACL_{INT} subjects has not been performed. While not significant ($p = 0.059$), our findings suggest the LOAD Q:H muscle activity ratio is greater in ACL_{REC} subjects than in ACL_{INT} subjects. This result suggests ACL_{REC} subjects have a greater amount of Q activity in relation to H activity than ACL_{INT} subjects during the LOAD phase. While not significant ($p = 0.094$), it was found that RF activity began earlier before contact in ACL_{INT} subjects than in ACL_{REC} subjects. Our findings also suggest the LG activated earlier before contact in ACL_{REC} subjects than in ACL_{INT} subjects, though the difference was not significant ($p = 0.075$). One limitation of this study is that no maximum voluntary contraction data were recorded, so we were unable to look at the relative magnitudes of each muscle. Another limitation is that we used surface electrodes to obtain the EMG signals instead of more precise needle electrodes. Finally, the sample size was small and may have affected our statistical analysis.

The results of our study show that differences exist between both gender and ACL reconstruction status for peak and onset timing and muscle activity ratios for the Q, H, and G muscle groups. From this study we learned that certain knee musculature peak earlier in ACL_{INT} and female subjects than in ACL_{REC} and male subjects. Future studies will focus on exploring this trend and the mechanisms that make it true.

D.5 Conflicts of interest

None.

D.6 Acknowledgements

This study was funded by grants from the W.M. Keck Foundation, NIH R01-AR047910, NIH COBRE P20-RR02484, and the Brown University UTRA program. The authors wish to thank Dr. Susan D’Andrea at the Center for Restorative and Regenerative Medicine for providing the Delsys EMG system.

D.7 References

- [1] Tsai L.-C., McLean S., Colletti P. M., and Powers C. M., 2012, “Greater muscle co-contraction results in increased tibiofemoral compressive forces in females who have undergone anterior cruciate ligament reconstruction,” *Journal of orthopaedic research: official publication of the Orthopaedic Research Society*.
- [2] Wright R. W., Dunn W. R., Amendola A., Andrish J. T., Bergfeld J., Kaeding C. C., Marx R. G., McCarty E. C., Parker R. D., Wolcott M., Wolf B. R., and Spindler K. P., 2007, “Risk of tearing the intact anterior cruciate ligament in the contralateral knee and rupturing the anterior cruciate ligament graft during the first 2 years after anterior cruciate ligament reconstruction: a prospective MOON cohort study,” *Am J Sports Med*, 35(7), pp. 1131–1134.
- [3] Beaulieu M. L., Lamontagne M., and Xu L., 2008, “Gender differences in time-frequency EMG analysis of unanticipated cutting maneuvers,” *Med Sci Sports Exerc*, 40(10), pp. 1795–1804.
- [4] Fleming B. C., Renstrom P. A., Ohlen G., Johnson R. J., Peura G. D., Beynnon B. D., and Badger G. J., 2001, “The gastrocnemius muscle is an antagonist of the anterior cruciate ligament,” *J. Orthop. Res.*, 19(6), pp. 1178–1184.
- [5] Kaneko F., Onari K., Kawaguchi K., Tsukisaka K., and Roy S. H., 2002, “Electromechanical delay after ACL reconstruction: an innovative method for investigating central and peripheral contributions,” *J Orthop Sports Phys Ther*, 32(4), pp. 158–165.

