Welcome to the 2018 I-FAB meeting in New York City. We have an exciting scientific agenda in a truly unique venue planned. The meeting is problem based and multidisciplinary. The foot and ankle pathology or concern will be the focus of each session with presentations from the epidemiological, experimental, computational, and clinical perspectives. All related disciplines are encouraged to participate to enhance the breadth and depth of our conference.

This meeting is comprised of a pre-conference day that includes HSS laboratory tours and 4 tutorials. The conference includes 3 keynote addresses, the Alex Stacoff memorial award lecture, 4 special sessions, with 110 contributed abstracts that will be disseminated as podium and poster presentations. At the conclusion there will be a Best Podium, Best Poster, Best Student Podium, and Best Student Poster Award. We are fortunate to have 12 sponsors that have contributed substantially to the success of this meeting.

The social program includes a welcome reception to be held at the Wyndham New Yorker Hotel on the Mezzanine (Sunday 4/8/18 at 6:30PM – 8:30PM) and a banquet to be held on the Hornblower Hybrid Yacht (Tuesday 4/10/18 6:30PM – 10:30PM). Do not hesitate to ask our conference staff at the registration desk or your conference and program chairs for assistance.

Best wishes,

Howard J Hillstrom, PhD, Conference Chair  HillstromH@HSS.edu
Jinsup Song, DPM, PhD, Program Chair  JSong@Temple.edu
i-FAB 2018 Sponsor Summary

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The 2018 I-FAB meeting will be problem based.

- The concept is to encourage multi-disciplinary participation and interaction.
- Participants will include foot & ankle Orthopedic surgeons, Podiatric physicians, Physical Therapists, Biomedical engineers, scientists, trainees, and related disciplines.
- Each session will be organized by the anatomical site of injury or pathology.
- In any given podium session there may be epidemiological, experimental, computational, or clinical outcomes based presentations.
- Posters will be organized in a similar manner.
- The program committee will be pleased to discuss any suggestions that you have to enhance the scientific program
I-FAB 2018 Program Committee

Howard J Hillstrom, PhD  Hospital for Special Surgery, New York, NY, USA (Conf. Chair)
Jinsup Song, DPM, PhD  Temple University, Philadelphia, PA, USA (Program Chair)
Rajshree Hillstrom, PhD, MBA  Anglia Ruskin University, Chelmsford, Essex, UK
Ellis, Scott MD  Hospital for Special Surgery, New York, NY, USA
Robert Turner, PT, DPT  Hospital for Special Surgery, New York, NY, USA
Julie Steele, PhD  University of Wollongong, Wollongong, Australia
Julie Stebbins, PhD  Oxford University, Oxford, UK
Marian T Hannan, DSc, MPH  Hebrew Senior Life, Harvard Medical School, Boston, MA, USA
Robin M Queen, PhD  Virginia Tech, Blacksburg, VA, USA
Irene S Davis, PT, PhD  Spaulding Rehab, Harvard Medical School, Cambridge, MA, USA
William R Ledoux, PhD  VA Puget Sound, Seattle, WA, USA (I-FAB Board)
Chris J Nester, BSc, PhD  University of Salford, Manchester, Salford, UK (I-FAB Board)
Taeyong Lee, PhD  Women's College, Busan, South Korea (I-FAB Board)
Dieter Rosenbaum, PhD  University of Muenster, Muenster, Germany (I-FAB Board)
Joshua Burns, SCHN  Children's Hospital at Westmead, Westmead, Australia, (I-FAB Board)
Alberto Leardini, PhD  Istituto Ortopedico Rizzoli, Bologna, Italy, (I-FAB Board)

I-FAB 2018 Local Committee

Seth L Hillstrom, BA  BMA Global  Conference Financial Officer
Kika Hillstrom, PhD  BMA Global  Assistant Conference Officer
Mehnaz Shahid, MS  HSS  i-FAB 2018 'Can do person'
Howard J Hillstrom, PhD  HSS  i-FAB 2018 Program Liaison
Rajshree Hillstrom, PhD, MBA  ARU  European Guest Liaison
Linda Reid, BA  Consultant  Manhattan Aficionado
Jinsup Song, DPM, PhD  TUSPM  Asian Guest Liaison
Jung-Eon Song, BA  Consultant  Social Event Planner
Ki-eun Song  TU  Dessert Connoisseur
HSS Lab Tours - 9AM to 11AM
Sunday 4/8/18

Meet in Wyndham New Yorker Lobby at 8:15AM or
Leon Root, MD Motion Analysis Laboratory at 510 East 73rd St - 1st floor at 9AM

1. Cone Beam Weight-Bearing CT - Cesar De Cesar Netto, MD
2. Leon Root, MD Motion Analysis Laboratory - Howard Hillstrom, PhD & Mehnaz Shahid, MS
3. Bike Fitting - Happy Freedman
4. Shearwave Elastography - Ogonna Kenechi Nwaka, MD
5. High Fidelity Cadaveric Simulator - Daniel Sturnick, MS
6. HUNOVA by Movendo - Robert Turner, PT, DPT
7. MRI Center - Matthew Koff, PhD
Tutorial A: “Finite element modelling to predict joint stress”
Chair: Rajshree Hillstrom, PhD, MBA (Anglia Ruskin University)
Rajshree.Hillstrom@anglia.ac.uk
Moderator: Howard J. Hillstrom, PhD (Hospital for Special Surgery)
1. Rajshree Hillstrom, PhD, MBA          Anglia Ruskin University              Rajshree.Hillstrom@anglia.ac.uk
2. Oliver Morgan, BSc                            Anglia Ruskin University              Oliver.Morgan@anglia.ac.uk
3. Howard Hillstrom, PhD                      Hospital for Special Surgery              HillstromH@HSS.edu

Tutorial B: “Foot/Ankle Assessment Strategies Across Disciplines: The Search for Common Ground”
Chair: Robert Turner (Hospital for Special Surgery) Turnerr@hss.edu
Moderator: Craig Payne, Podiatrist (La Trobe University)
1. Robert Turner PT DPT OCS                 Hospital for Special Surgery              Turnerr@hss.edu
2. Scott Ellis MD                                     Hospital for Special Surgery          Elliss@hss.edu
3. Jinsup Song                             Temple University                          JSONG@tuspm.temple.edu

Tutorial C: “Multi-Segment Foot Modeling for In-vivo Kinematic Measurements"
Chair. Alberto Leardini, PhD (Movement Analysis Laboratory, Rizzoli Orthopedic Institute) leardini@ior.it
Moderator: Julie Stebbins, PhD (University of Oxford)
1. Alberto Leardini, D.Phil.               Rizzoli Orthopedic Institute                   leardini@ior.it
2. Julie Stebbins, D.Phil.                   University of Oxford                       Julie.Stebbins@ouh.nhs.uk

Tutorial D: "Understanding Coordination and Movement Variability: Application of Vector Coding in Dynamical Systems"
Chair: Nachiappan Chockalingam, PhD (Staffordshire University)  n.chockalingam@staffs.ac.uk
Moderator: Joseph Hammil, PhD (University of Massachusetts)
1. Robert A. Needham,                           Staffordshire University                   r.needham@staffs.ac.uk
2. Nachiappan Chockalingam,                    Staffordshire University                   n.chockalingam@staffs.ac.uk
3. Roozbeh Naemi,                             Staffordshire University                   r.naemi@staffs.ac.uk
4. Joseph Hamill                               University of Massachusetts                   jhamill@kin.umass.edu

Schedule:
Tutorial A: Sutton Place 3rd Floor          Tutorial B: Gramercy Park 3rd Floor 1PM – 3PM
Tutorial C: Sutton Place 3rd Floor          Tutorial D: Gramercy Park 3rd Floor 3:30PM – 5:30PM
3rd Floor Ballrooms

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<th>Rooms</th>
<th>Ceiling</th>
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<th>Classroom</th>
<th>Banquet</th>
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Grand Ballroom Balcony
Tutorial A: Finite Element Modelling to Predict Joint Stress

Chair: Rajshree Hillstrom (Anglia Ruskin University)
Rajshree.Hillstrom@anglia.ac.uk

1. Hillstrom, Rajshree, PhD, MBA          Anglia Ruskin University                   Rajshree.Hillstrom@anglia.ac.uk
2. Morgan, Oliver, BS                             Anglia Ruskin University                   Oliver.Morgan@anglia.ac.uk
3. Hillstrom, Howard, PhD                     Hospital for Special Surgery             HillstromH@HSS.edu

Osteoarthritis (OA), the number one cause of disability, exceeds 1% of the gross national product in both the UK and US and is increasing in prevalence as the population ages and becomes more obese. Excessive stress damages tissues within the joint, irrespective of the specific biomechanical etiology, and is considered a risk factor for OA. Hence, it is important to monitor joint stress. However, stress cannot be measured in vivo. In vitro experiments are expensive and may not accurately represent in vivo tissue properties. Finite element modelling is a useful computational method to predict joint stress before and after surgical interventions. This is a powerful tool to investigate the impact of surgical procedures on joint stress. The effect of factors, such as tissue geometry, material properties, type of activity and loading conditions, which could affect joint stress, may be investigated in the model.

When modelling human tissues, assumptions are made, for example on tissue material properties, which are subject specific and challenging to measure non-destructively. Sensitivity analyses are conducted to identify and simulate, with higher accuracy, those material properties that drive changes in stress magnitude. In this tutorial, participants will use FEBio finite element software, learn how to create simple finite element models and analyse joint stress. Participants will also have the opportunity to work with a first metatarsophalangeal joint finite element model to investigate the effects of different tissue material properties, foot types, and surgical interventions on joint stress.

- Dr. Howard Hillstrom (Hospital for Special Surgery) will introduce the application of finite element modelling and the clinical rationale for its use, based upon his 90 studies in lower extremity biomechanics, including the assessment of foot type and musculoskeletal pathology, employing experimental and computational approaches. This introduction will include contemporary examples of biomechanical problems that could be addressed, using finite element models.
- Dr. Rajshree Hillstrom (Anglia Ruskin University) will present a step-by-step method for developing a simple finite element model of a joint, using FEBio software. This will include creating and meshing simple geometry representing joint tissues, assigning tissue material properties, boundary conditions to simulate tissue interactions, loading conditions and to predict joint stress. During this hands-on session, participants will develop their own finite element models and run analyses. Dr R Hillstrom will then explain how to develop anatomically complex models. Participants will split in groups and solve for stress in the 1st metatarsophalangeal joint for different tissue properties and surgical corrections.
- Oliver Morgan (Anglia Ruskin University) will assist participants with troubleshooting any problems during the hands-on aspect of this tutorial. He will make sure that FEBio is running smoothly on either windows or mac laptops and that participants have all the supporting files to participate in the tutorial.
Foot and Ankle assessment strategies across disciplines are highly variable by practitioner as well as specialty field. Orthopedic assessments will be compared to podiatric and physical therapy assessments to illustrate commonalities as well as differences in each profession. Three common diagnoses (moderate 1st metatarsophalangeal joint osteoarthritis (MTPJ OA), adult acquired flatfoot deformity and hallux valgus) will be used as case examples of information gleaned from each discipline. We will compare and contrast specific assessment styles, treatment interventions and goals in an effort to elicit discussion from the participants. This workshop will highlight the necessity of an interdisciplinary approach to the totality of foot care to maximize treatment outcomes. Using prepared video assessments each speaker will offer their unique assessment strategies and discuss their treatment planning process for each diagnosis:

**Scott Ellis MD** will demonstrate and discuss his thought process and evaluation criteria from a **Orthopedic Surgeon**’s perspective. For hallux rigidus, the time and timing of surgical treatment is dictated by the presence or absence of pain and crepitus with mid-range of motion as well as the appearance of the MTP joint on plain radiographs and other imaging. For the adult acquired flatfoot deformity, the ability to perform a single heel raise, passive hindfoot range of motion, and the amount of progression of deformity dictate the timing and time of surgical intervention. MRI and weight-bearing CT scan have become essential tools as well.

**Jinsup Song DPM, PhD.** While Doctor of Podiatric Medicine are trained to manage foot and ankle pathologies using both surgical and conservative interventions, Dr. Song will demonstrate and discuss his thought process and evaluation criteria from a podiatric physician’s conservative perspective. In particular, decision process related to prescription of foot orthoses will be discussed.

**Robert Turner PT DPT OCS** will then discuss his thought process and evaluation criteria from a Doctor of Physical Therapy perspective. Utilizing functional testing, manual motion testing and palpation of the tissues of the foot and ankle, deficiencies will be noted and addressed with specific exercises prescribed for each case. Footwear patterns will be assessed and modified as needed for each case.
Tutorial C: Multi-Segment Foot Modeling for In-vivo Kinematic Measurements

Chair: Alberto Leardini, (Movement Analysis Laboratory, Rizzoli Orthopedic Institute) leardini@ior.it
Names: Institution: email:
1. Alberto Leardini, D.Phil. Rizzoli Orthopedic Institute leardini@ior.it
2. Julie Stebbins, D.Phil. University of Oxford Julie.Stebbins@ouh.nhs.uk

Tutorial D: Understanding Coordination and Movement Variability: Application of Vector Coding in Dynamical Systems

Chair: Nachiappan Chockalingam, PhD (Staffordshire University) n.chockalingam@staffs.ac.uk
Name: Affiliation/Institution: Email:
1. Robert A. Needham, PhD Staffordshire University r.needham@staffs.ac.uk
2. Nachiappan Chockalingam, PhD Staffordshire University n.chockalingam@staffs.ac.uk
3. Roozbeh Naemi, PhD Staffordshire University r.naemi@staffs.ac.uk
4. Joseph Hamill, PhD University of Massachusetts jhamill@kin.umass.edu

The application of dynamical systems theory to describe the complex behavior of human movement is becoming increasingly popular in biomechanics and motor control research. A dynamical systems framework offers non-linear data analysis techniques that can quantify movement coordination and coordination variability over time. An angle-angle diagram illustrates the continuous interaction between two segments and provides a qualitative view on coordination. Vector coding is a data analysis technique that calculates the vector orientation between data points on an angle-angle diagram. The vector orientation, which can range between 0-360°, is a quantifiable measure that is referred to as the coupling angle. Each coupling angle across a normalized movement cycle can be assigned to a coordination pattern classification. This tutorial will provide a comprehensive review on methodological considerations, and on theory and practice of vector coding. This includes recent developments on a new coordination classification that expands on current ideologies, the introduction of a novel approach to quantify segmental dominancy during a movement cycle, comment on the appropriate use of circular statistics, and provide new data reporting techniques that have the capability to visually display multiple trials and segmental couplings. In addition, this tutorial will provide an insight on how vector coding offers a new perspective on the functional workings of the foot that traditional linear data analysis and reporting techniques cannot detail. Finally, current theories on coordination variability and overuse injuries will be presented.

Nachiappan Chockalingam (Staffordshire University/ University of Malta): Whilst outlining the need for changes to clinical data reporting, Nachi will introduce dynamical systems and provide an overview of the theoretical background to coordination and movement variability. He will also highlight the differences between linear and non-linear analysis. Whilst emphasizing the usefulness of single subject research designs
which forms the basis of dynamical systems approach, he will introduce novel concepts for clinical data reporting such as coordination profiling.

Robert A. Needham (Staffordshire University): Rob will introduce and define vector coding. He will provide an overview of our previous work and the development of a new coordination pattern classification. His lecture will also introduce concepts relating to quantification of segmental dominancy using lower limb data. In addition, Rob will touch on statistical approaches with a focus on circular statistics. He will conclude by presenting novel approaches on reporting several outcome measures using vector coding.

Joseph Hamill (University of Massachusetts/ Staffordshire University): Joe will outline his pioneering work in this area of dynamical systems and provide an overview of the developments and its applications. He will outline the use of such approaches in various clinical scenarios with a focus on foot and ankle. Whilst discussing coordination patterns during various foot fall patterns during running, Joe will also touch on coordination variability and its link to incidence of injury and its continuum.

Roozbeh Naemi (Staffordshire University): Roozbeh will introduce advanced mathematical calculations behind dynamical systems and vector coding. He will provide a critical analysis of previous papers and provide an overview of how our current approaches provide clarity to the background calculations in vector coding. Where appropriate, Roozbeh will use examples from foot and ankle data to provide an understanding of the concepts.
2018 I-FAB Program

Sunday April 8, 2018
9-11AM  HSS Lab Tours
11:30AM  Lunch in NYC
1-3PM    Tutorials A & B
3:00PM   Coffee Break
3:30PM   Tutorials C & D
5:30PM   Pre-conference Didactics Concluded
6-9PM    Welcome Reception

Monday April 9, 2018
8AM      Conference Welcome & Introduction
8:05AM   Keynote 1: Michael J. Coughlin, MD
9AM      Session 1 - Total Ankle Replacement (SS)
10:30AM  Coffee Break with Vendors
11AM     Session 2 - Neuropathy & Motor Control
Noon     Lunch in NYC
1:30PM   Session 3 - Pediatric Foot (SS)
3:10PM   Session 4 - Vendor Technology Update
3:20PM   Coffee Break with Vendors
3:50PM   Session 5 - Military Biomechanics
5:00PM   Session 6 - Tendon & Ligament Injuries
6:00PM   Adjourn

Tuesday April 10, 2018
8AM      Daily Game Plan
8:05AM   Keynote 2: Marian T. Hannan, DSc, MPH
9AM      Session 7 - Foot Type I
10:10AM  Coffee Break with Vendors
10:40AM  Session 8 - National Biomechanics Day & STEM
11:00AM  Session 9 - Foot Type II
12:10PM  Lunch in NYC
1:30PM   Alex Staff Lecture: Don Anderson, PhD
2:00PM   Session 10 - Ankle
3:00PM   Session 11 - Hallux Rigidus and Hallux Valgus
4:00PM   ISB Footwear Biomechanics 2019
4:05PM   Session 12a - Poster Teasers
4:15PM   Session 12b - Posters and Coffee
5:30PM   Adjourn
7-10PM   Congress Banquet

Wednesday April 11, 2018
8AM      Daily Game Plan
8:05AM   Keynote 3: Marcus G. Pandy, PhD
9AM      Session 13 - Healthy Locomotion
10:00AM  Coffee Break with Vendors
10:30AM  Session 14 - Good Vibrations (SS)
12:00PM  Lunch in NYC
1:30PM   Session 15 - Sports Injuries
3:20PM   Coffee Break
3:40PM   Session 16 - Minimalist Shoes (SS)
5:10PM   Awards and Congress Wrap-up
5:30PM   Adjourn

Note: SS = Special Session
2nd Floor Ballrooms

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Grand Ballroom

MEzzanine

Open to Lobby

Foyer

Crystal Ballroom

Line of Balcony Above

Elevators
Keynotes

**Keynote 1:** Monday 8:05AM - 9AM  **Michael J Coughlin, MD**
“Total Ankle Design: What is Being Done in America and Around the World”
Moderator: Scott Ellis, MD (HSS)

**Keynote 2:** Tuesday 8:05AM - 9AM  **Marian T. Hannan, DSc, MPH**
"Out of the Lab and Into the Streets: Population-Based Epidemiology Of Foot Pathologies"
Moderator: Howard Hillstrom, PhD (HSS)

**Stacoff Memorial Lecture:** Tuesday 2PM - 2:30PM  **Donald D. Anderson, PhD**
"Enabling Post-Traumatic Osteoarthritis Risk Prediction from Pathomechanics"
Moderator: William Ledoux, PhD (University of Washington, Seattle VA)

**Keynote 3:** Wednesday 8:05 - 9AM  **Marcus G. Pandy, PhD**
"Muscle and Joint Function in Human Gait"
Moderator: Alberto Leardini, PhD (Instituto Ortopedico Rizzoli)
Keynotes

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"Muscle and Joint Function in Human Gait"
Moderator: Alberto Leardini, PhD (Instituto Ortopedico Rizzoli)
Michael J. Coughlin, MD

Michael J. Coughlin graduated from the University of Oregon Medical School in 1974 and did his orthopaedic training at the University of California-San Francisco and a foot and ankle fellowship in 1978. Following this, he returned to orthopaedic foot and ankle practice in Boise, Idaho. He was the youngest President of the American Orthopaedic Foot and Ankle Society, serving from 1990 – 1991. He was also president of the International Federation of Foot and Ankle Society’s from 2002-2005. He is the co-editor of Surgery of the Foot and Ankle 9th Edition and edited three prior editions. It is considered “the bible” of foot and ankle surgery in the US. He has published over 200 scientific articles and over 500 scientific presentations on issues relating to the foot and ankle. He is a Professor of Orthopedic Surgery at University of California San Francisco; he is the Director of the Idaho Orthopedic Foot and Ankle Fellowship and practices orthopedic foot and ankle surgery in Boise, Idaho at the St Alphonsus Coughlin Foot and Ankle Clinic. He has been a consultant for the NFL for 9 years, and Co-chair of the Foot and Ankle subcommittee for Health and Safety for 7 years. He consults for the Green Bay Packers, and examines for them at the NFL combine yearly. He also consults for the Oregon Ducks and Boise State Broncos.

“Total ankle design: what is being done in America and around the world”

Total ankle replacement has evolved substantially since first performed in 1970 (Lord and Marotte). Early attempts in the US and Europe centered around cemented prostheses, which by design, were non-anatomic, and had little regard for soft tissue balancing. After a hiatus of two decades, improved un-cemented designs in Asia, Europe and the U.S. dramatically changed the orthopaedic landscape, and provided a real alternative to arthrodesis for end stage ankle arthritis.

The introduction of three part metal/polyethylene prostheses by Buechel (U.S.), Koføed (Denmark), and ceramic designs by Takakura (Japan) lead to a plethora of new and innovative designs around the world.

The popularity of a mobile bearing total ankles in Europe is contrasted with the difficulty of introducing a Class-3 (new) device in the U.S. At this time, the Scandinavian (STAR) (Mann, Coughlin) total ankle is the only three part ankle permitted by the FDA. Several three part total ankles from Europe have been modified to two part prostheses in order to circumvent the
rigorous investigative process required by the FDA. Several two-part ankles have been authorized for implantation using the 510-k process.

While early results with first generation ankle replacements demonstrated high failure rates, current U.S. reports using uncemented with titanium spray surfaces for bony ingrowth have noted to have >80% retention rates at ten years or more follow-up. Total ankle replacement is considered now to be a realistic alternative to ankle arthrodesis.


Marian T. Hannan, ScD, MPH

Dr. Hannan is a Professor of Medicine at Harvard Medical School and a Senior Scientist at the Institute for Aging Research at Hebrew SeniorLife in Boston, MA, USA. She is the co-Director of the large Musculoskeletal Research Center at the Institute. Dr. Marian T. Hannan received her undergraduate degree at the University of California, Berkeley, the Master of Public Health degree at Yale University School of Medicine and her doctorate in Epidemiology at Boston University School of Medicine. Dr. Hannan is the Editor-in-Chief for the highly regarded journal, Arthritis Care & Research.

Dr. Hannan is currently conducting research on risk factors for arthritis, foot disorders and biomechanics, hip fracture and osteoporosis. She is particularly interested in the effect of biomechanics upon physical function and the influence of body composition. She is widely published with her work represented in over 50 scientific journals in the medical field. Dr. Hannan is the principal investigator on a number of National Institutes of Health grants, and has had continuous NIH grant funding since 1996.

At Harvard Medical School, Dr. Hannan teaches Clinical Epidemiology and Population Health to first-year medical students as well as nutrition seminars. She also lectures in the HMS geriatrics fellowship program. Dr. Hannan directs the Frailty Course at Harvard School of Public Health. Since 2004, Dr. Hannan has served on NIH study sections reviewing national grant applications, as well as providing reviews for many international science organizations. Her mentoring of young investigators includes many scientists and medical fellows in the Boston area as well as 8-10 per year across the U.S. and Canada through the U.S. Bone & Joint Initiative’s Young Investigator Initiative. She has been a keynote speaker at many venues and is always happy to speak about how the foot connects humans to the world.

“Out of the Lab and Into the Streets’: Population-Based Epidemiology of Foot Pathologies”

This presentation will provide an epidemiological overview of the current population levels of foot & ankle pathologies, including current challenges and viewpoints. This session will also address the need for a data-driven approach as we consider common pathways, for example, the links between obesity and pedal pathologies as well as special considerations regarding clinical trial designs for foot and ankle research.
The past decade has brought many new insights to the epidemiology of foot and ankle disorders as well as insights into early pathology, and even possible prevention of impaired foot structure and function. Clinical cases and laboratory studies have provided information on treatment and insights into mechanisms. Yet, we still know relatively little of the population impact and informed prevention that may help people NOT become patients.

The objective of this presentation is to provide a population-based understanding of foot type, pathologies and function in the population. We will consider how populations inform science & medicine, how to obtain complex measurements from large groups outside the laboratory, and highlight major findings of population-based foot studies. A better understanding of these issues can help to inform the public as well as disseminate clinical and scientific information.

How does all of this inform our understanding of foot biomechanics and translation of research? Epidemiology may serve as a bridge between our current knowledge base and how to grow this foundation to the next level of insights and interventions. Such a focus will encourage the integration of our knowledge of biomechanics and movement with "Big Data" collections, taking our field to the next level.
Donald D. Anderson, PhD

Don Anderson is Professor and Vice-Chair of Research in the Department of Orthopedics & Rehabilitation at the University of Iowa, where he directs the Orthopaedic Biomechanics Laboratory. Dr. Anderson holds a BSE in Biomedical Engineering, as well as an MS and PhD in Mechanical Engineering, all from the University of Iowa. He has nearly 30 years of post-doctoral experience with image analysis, computer modeling, and computational stress analysis in musculoskeletal applications. Dr. Anderson's primary research focus is articular joint biomechanics, and his most recent work involves the scaling up of methods for patient-specific articular joint modeling in the ankle.

“Enabling Post-Traumatic Osteoarthritis Risk Prediction from Pathomechanics”

The long-term goal of our research is to forestall post-traumatic osteoarthritis (PTOA), the disabling condition that often develops after joint injuries like an intra-articular fracture (IAF) of the tibial plafond. PTOA leads to permanent disability in nearly 30% of individuals having sustained an IAF, with those of the foot and ankle being the most disabling. The impairment associated with ankle OA is comparable to that caused by end-stage kidney disease or congestive heart failure. The vast majority of ankle OA is post-traumatic, with tibial plafond IAFs often leading to disabling PTOA within two to five years. As a result, patients with ankle injuries provide an ideal population in which to study this degenerative pathway so that we can optimize treatment. We have developed patient-specific precision medicine approaches to predict PTOA risk in the ankle using CT-based measures of pathomechanical factors associated with IAFs (fracture severity and elevated contact stress post-treatment) of the tibial plafond. A primary objective of the group’s present work is to enable the use of these innovative methods for assessing IAFs to better inform patient care and to guide future clinical trials of new therapies directed at mitigating or arresting the environment that triggers progressive joint degeneration.
Marcus G. Pandy, PhD

Marcus Pandy is appointed as Chair of Mechanical and Biomedical Engineering at the University of Melbourne and formerly served on the faculty of the Department of Biomedical Engineering at the University of Texas at Austin. A focus of Dr Pandy’s research career has been the development, validation and implementation of experimental and computational tools for accurate assessment of muscle, ligament and joint function in vivo. Dr Pandy is a Fellow of the American Institute of Medical and Biological Engineering, the American Society of Mechanical Engineers and Engineers Australia.

“Muscle and Joint Function in Human Gait”

Gait-analysis techniques have been used for more than a century to provide information on the kinematics and kinetics of human gait, yet the ability of this approach to evaluate functional performance is limited because it cannot be used to discern the actions of individual muscles. Rapid increases in computing power combined with recent advances in medical imaging, more accurate methods for measuring dynamic joint motion, and more efficient algorithms for modelling the human neuromusculoskeletal system have enabled detailed analyses of musculoskeletal function. The aim of this presentation is to illustrate how computational modelling may be combined with novel imaging methods such as mobile X-ray fluoroscopy to provide a more comprehensive understanding of muscle and joint function in healthy and pathological gait.
Session 1: Total Ankle Replacement (SS)

Chair: Sorin Siegler, PhD (Drexel University)
Co-Moderators: Robin Queen, PhD (Virginia Tech)

9AM Keith Wapner, MD (Invited)


9:30AM  I-2: Belvedere, Claudio, et al. Experimental evaluation of custom-made morphological approximations of the ankle articular surfaces

9:40AM  I-3: Leardini, Alberto, et al. Flexibility in the ankle joint after implantation of custom-made artificial articular surfaces

9:50AM  I-4: Sturnick, Daniel R., et al. Influence of tibial component position on altered kinematics following total ankle arthroplasty during simulated gait

10:00AM I-5: Saito, Guilherme H., et al. Differences in gait mechanics after total ankle arthroplasty and ankle arthrodesis

10:10AM I-6: Kraszewski, Andrew, et al. Differences in Mechanics During Stair Ascent After Ankle Arthrodesis and Total Ankle Arthroplasty

10:20AM I-7: Seo, Hansol, et al. Characterization of Ankle Joint Motions Patterns for Main Twelve Activities of Daily Living (ADLs) of Elderly
Experimental evaluation of custom-made morphological approximations of the ankle articular surfaces

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INTRODUCTION: Severe arthritis of the ankle joint is often treated by replacing the affected articular surfaces with prosthetic components aimed at replicating the joint natural mechanics. The resulting kinematics and flexibility in replaced ankles are influenced by the geometry of these artificial articulating surfaces. Current total ankle prosthesis designs nowadays offer only cylindrical or truncated-cone geometries with medially-oriented apex (TCM). A recent image-based study [1], proposed an original saddle-shaped, skewed, truncated cone with laterally-oriented apex (SSCL), differently to what proposed by Inman in 1976. The goal of this study was to experimentally compare the two traditional and the original designs in terms of their ability to replicate natural mechanical characteristics of the ankle joint.

METHODS: Ten cadaver specimens underwent a validated full process of custom design of these three geometries, including medical imaging, 3D modeling and printing of the three implantable sets of customized articular surfaces for tibiotalar joint replacement. Tests were performed on each specimen under cyclic loading, following an established technique [2,3] which implied measures of continuous torque across the ankle complex while tracking motion of the tibia, talus, and calcaneus by means of a stereophotogrammetric system for surgical navigation (Stryker Knee Navigation System, Stryker®, Kalamazoo, MI-USA). Torques in out-of-sagittal plane directions were applied in neutral and in the two extremes of flexion. 3D kinematics and flexibility at the overall ankle complex, as well as at the separate ankle and subtalar joints were thus determined. The performance of the three custom made artificial surfaces was compared with the natural joint surface.

RESULTS: Results showed that SSCL surfaces replicated the natural mechanical characteristics better that the cylindrical and the TCM, both at the ankle and subtalar joints, in all anatomical planes. In the large majority of the specimens and joints observed and torques applied, SSCL was superior on a statistical basis compared to the other two surface approximations, in particular for the responses in maximum dorsiflexion (see Figure below).

DISCUSSION: Originally, all these three implant sets were designed to match the specific morphology of each tested specimens, also according to the corresponding designing approach. Therefore this study may also represent a demonstration case also for possible future processes of ankle prosthesis designing and manufacturing on a patient-specific basis. The best replication of natural kinematics and flexibility was observed for the SSCL surface approximation, particularly in maximum dorsiflexion, which is in fact the joint position where the ankle articular surfaces are most congruent, and therefore where the contribution of the articular surfaces is expected to be the largest.

SIGNIFICANCE/CLINICAL RELEVANCE: These results demonstrate that replacing the natural ankle joint by custom-made surfaces in the shape of a saddle skewed truncated cone with apex laterally [1], produce a more natural mobility and stability behavior than those obtained by surfaces mimicking traditional total ankle designs, thus establishing the potential of this original geometry for novel total ankle replacement designs. In future publications, the design parameters, the full load transfer capacities, i.e. joint stiffness, the joint linear displacements, and the contact surfaces will be addressed.

REFERENCES:
Anteroposterior Translational Malalignment of Ankle Arthrodesis in Cadaveric Gait Simulation

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INTRODUCTION: Tibiotalar arthrodesis is a common surgical treatment for end stage ankle arthritis. Proper alignment is an important consideration, as malalignment can lead to complications that may require a revision surgery [1]. The biomechanical effects of malalignment are not well understood. The purpose of this study was to determine how anteroposterior (AP) translational malalignment of ankle arthrodesis affects the kinematics of the foot.

METHODS: Ankle arthrodesis was performed on nine cadaveric foot specimens with a custom device that could fuse the ankle neutrally and induce discrete malalignments of 3, 6, and 9 mm anteriorly (3A, 6A, 9A) and posteriorly (3P, 6P, 9P) (Figure 1). Gait was simulated for each specimen under each alignment using the robotic gait simulator (RGS) [2]. The RGS consists of a force plate mounted to a six degree-of-freedom robot that recreates the relative tibia to ground motion seen during the stance phase of gait. The target tibia kinematics and vertical ground reaction force were taken from in vivo data averaged across ten ankle arthrodesis subjects who were one year postoperative. Actuators were connected to nine extrinsic foot tendons to simulate muscle forces. An eight-camera motion capture system tracked reflective marker sets that were rigidly attached to the bones of interest. Bone kinematics were computed with a custom ten-segment foot model. The range of motion (ROM) and joint angle at each percent stance phase were determined for nine joints and one bone-to-bone relationship. Statistical analysis was performed with a linear mixed effects regression model that tested for differences in kinematics by condition. When omnibus testing was significant, pair-wise comparisons were carried out.

RESULTS: ROM: AP malalignment caused significant differences in sagittal plane ROM for two joints. Talonavicular ROM was significantly decreased for the 9P alignment (1.6°, p=0.014), and the 9A and 9P alignments significantly decreased the ROM of the first metatarsophalangeal joint (2.5°, p=0.0222 and 0.0313, respectively). Joint Position: There were several significant differences in joint position at various intervals within stance. The 6P and 9P alignments significantly inverted the talocalcaneal joint throughout stance phase and adducted it during late stance. Every posterior malalignment significantly plantar flexed the talocalcaneal joint in late stance, while the 6A and 9A alignments dorsiflexed it. During late stance, the 6P and 9P conditions also caused significant plantar flexion of the talonavicular joint. The 9A condition significantly dorsiflexed it, but this occurred at the very end of stance. Within the forefoot, differences in joint position were seen on the medial side of the foot. In early and late stance, the 6P and 9P alignments significantly inverted the first metatarsal relative to the talus. Both conditions also significantly adducted and plantar flexed the first metatarsal during the second half of stance phase. First metatarsal dorsiflexion relative to the talus was observed at the beginning and end of stance under the 6A and 9A alignments. Finally, the 6P and 9P alignments significantly everted the first metatarsophalangeal joint throughout stance phase.

DISCUSSION: AP malalignment did not substantially affect joint ROM but did cause significant differences in joint position throughout stance phase. Differences were seen in the talocalcaneal, talonavicular, and first metatarsophalangeal joints, and the first metatarsal relative to the talus. In general, the 6P and 9P alignments had the greatest effect on foot kinematics. Aberrant motion may lead to altered joint loading and facilitate cartilage degeneration. These findings may thus have important implications for the postoperative health of these joints. This study had several limitations. Simulations were performed at 25% body weight and 1/6th the speed of physiological gait to maintain fixation of the fusion device. The 9A and 9P conditions were only achieved with eight and seven specimens, respectively, which decreased statistical power. Also, because the two extreme malalignments were difficult, they were always done last so testing was only partially randomized.

SIGNIFICANCE: AP malalignment of ankle arthrodesis altered the kinematics of three joints and one bone-to-bone relationship within the foot. The most widespread effects were seen when the talus was displaced 6 mm or more in the posterior direction. In vivo, this may lead to changes in joint loading, which could negatively impact patient outcomes.

REFERENCES:

ACKNOWLEDGEMENTS: Thanks to Jane Shofer for her help with the statistics.

FIGURES AND TABLES:

Figure 1: Fusion device in neutral and 9P alignments.
Flexibility in the ankle joint after implantation of custom-made artificial articular surfaces

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INTRODUCTION: In total ankle replacement the resulting joint kinematics and flexibility are influenced by the geometry of the artificial articulating surfaces. Current total ankle prosthesis designs offer only cylindrical (CYL), or truncated-cone geometries with medially-oriented apex (TCM), according to what proposed by Inman in 1976. Differently from the latter, an original saddle-shaped, skewed, truncated cone with laterally-oriented apex (SSCL) has been recently proposed [1]. An extensive experimental study compared the effects of these different surfaces, defined with a custom-made approach, in terms of their ability to replicate natural mechanical characteristics of the ankle joint.

METHODS: Ten cadaver specimens underwent a validated full process of custom design and experimental tests of these three artificial surfaces [2]. This included CT scans of the specimen, 3D bone modeling, surface designing and printing in ABS. Tests were performed on each specimen and for each of the three sets of customized articular surfaces implanted sequentially in the tibiotalar joint. In these conditions, torque across the ankle complex under cyclic loading was measured with a torque sensor, synchronized with a stereophotogrammetric system for surgical navigation (Stryker Knee Navigation System, Stryker®, Kalamazoo, MI-USA) for motion at the tibio-talar and sub-talar joints. Torques in out-of-sagittal plane directions were applied in neutral joint position and in the two extremes of flexion. In the joint rotation vs torque plots for the tibio-talar joint, regression lines were calculated over relevant data from the three repetitions in the loading run of the cycle; therefore these represent the correlation between torque and corresponding rotation at the joint.

RESULTS: Flexibility plots (see Figure below) showed that all three artificial surfaces replicated well the corresponding patterns in the natural joint; this was observed for both for the Int/Ext and Inv/Eve rotations. Interestingly, tibio-talar joints after replacements were stiffer than the natural in max Dorsiflexion and neutral joint positions, more lax in max Plantarflexion, which is in fact the joint position where the ankle is less congruent. In a large number of specimens the ratio between rotation and torque at the regression lines was in favor of the SSCL, thus replicating the corresponding patterns in the natural joint better than CYL and TCM, both at the ankle and subtalar joints, in both out-of-sagittal anatomical planes.

DISCUSSION: Originally, three implant sets were designed according to corresponding different concepts and particularly also to match the specific morphology of each specimen. The best replication of natural flexibility was observed for the SSCL surface approximation, particularly at the extremes of the range of flexion.

SIGNIFICANCE/CLINICAL RELEVANCE: These results demonstrate that replacing the natural ankle joint by custom-made surfaces in the shape of a saddle skewed truncated cone with apex located laterally [1], produce a more natural joint flexibility than those obtained by surfaces mimicking traditional total ankle designs, i.e. CYL or TCM, thus establishing the potential of the SSCL geometry for novel total ankle replacement designs. In future work, the effect of friction at the articulating surfaces will be addressed more carefully. The present study may also represent a demonstration case also for possible future processes of ankle prosthesis designing and manufacturing on a patient-specific basis.

REFERENCES:

Figure 1: Flexibility curves of the tibio-talar joint in Int/Ext rotation from the neutral joint position in a typical specimen: in the natural (left), and after implant of the TCM (centre) and SSCL (right) articular surface approximations; the regression lines are also shown.
Differences in gait mechanics after total ankle arthroplasty and ankle arthrodesis

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INTRODUCTION: In the past, ankle arthrodesis was considered the gold-standard treatment for end-stage ankle arthritis. However, with the development of modern total ankle arthroplasty (TAA) designs and a better understanding of surgical techniques, TAA has become a reliable treatment for end-stage ankle arthritis. Both treatment options are effective in providing pain relief and function improvement. However, patients following TAA demonstrate more normalized gait and a more normal ground reaction force curve than patients after ankle arthrodesis. In addition, patients after ankle arthrodesis compensate for loss of sagittal range of motion through the ankle by increasing motion at the subtalar joint, conceivably leading to early degeneration of the adjacent joints. It is unclear if this similarly occurs in TAA patients, since the gains in motion after TAA compared to ankle arthrodesis are rather modest in previous gait studies. This study aims to quantitate the three-dimensional foot and ankle kinematics and calculate the moments at the joints to determine power that is generated during the task of walking on level ground.

METHODS: Ten patients who previously underwent TAA with a modern fixed-bearing ankle replacement (Salto Talaris - Tornier or INBONE 2 - Wright Medical Technology) and ten patients who previously underwent ankle arthrodesis were recruited for participation in the study. Patients were matched for age, sex, BMI, time from surgery and pre-operative diagnosis. A minimum of 1-year follow up was required for inclusion. The modified Oxford multi-segment foot model was applied to determine hindfoot, midfoot, and forefoot kinematics (Figure 1). A standard 6-degree of freedom marker set was simultaneously applied with the modified Oxford multi-segment foot kinematics measurements to determine hindfoot kinetics (Figure 2). Each participant performed the gait analysis at their self-selected walking speed.

RESULTS: In the sagittal plane, motion of the tibia in relation to the axis of the foot was significantly higher in the TAA group (20.9 ± 4.9 vs 14.6 ± 2.8 degrees, p = .003). Forefoot-tibia motion was also significantly higher in the TAA group (26.7 ± 6.5 vs 20.0 ± 5.4 degrees, p = .024). Differences in forefoot-hindfoot motion and hindfoot-tibia motion between groups were not significant. In the frontal plane, the arthrodesis group presented a significantly greater forefoot-hindfoot motion (7.4 ± 1.3 vs 7.2 ± 1.4 degrees, p = .015). No other statistically significant difference was observed in the frontal plane between groups. No significant differences were observed in ankle moment when comparing the TAA and the arthrodesis group (-1.3 ± 0.3 vs -1.2 ± 0.2, p = .505). Ankle power was greater in the TAA group (2.0 ± 0.6) in comparison to the arthrodesis group (1.3 ± 0.4), but the difference was not significant (p = .104).

DISCUSSION: Hindfoot-tibia motion in the sagittal plane was similar between groups, suggesting that following ankle arthrodesis patients have an increased motion at the subtalar joint. However, no significant differences were observed in forefoot-hindfoot motion between groups. Furthermore, the TAA group did not show significant improvement in ankle moment and ankle power. TAA have shown to provide improved clinical outcomes on uneven surfaces and stairs, whereas the present study was performed on level ground. Gait analysis of patients performing more demanding tasks such as ascending and descending stairs may demonstrate different results.

SIGNIFICANCE/CLINICAL RELEVANCE: Compensatory increased motion at the subtalar joint may explain the high rates of subtalar joint arthritis observed following ankle arthrodesis. Gait comparison of ankle moment and power on uneven surfaces and stairs should be the focus of future research.
Characterization of Ankle Joint Motions Patterns for Main Twelve Activities of Daily Living (ADLs) of Elderly

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INTRODUCTION: As we enter the Ageing Society, the decrease of activities of daily living (ADLs) ability performance of the elderly is inevitable as the proportion of the elderly population increases significantly. This results in the loss of quality of life by forming the difficulties of independent living. As a result, the development of aged-friendly products considering the convenience of everyday life is increasing rapidly. Prior to this development, analysis of joint motion by various motions enables systematic description and quantitative evaluation of human motion. The purpose of this study was to analyze the characterization of ankle joint motions patterns for main twelve ADLs of elderly.

METHODS: Following Institutional Review Board approval, 25 healthy males elderly (height: 170.1±4.5cm, weight: 68.8±9.5kg, age: 72.2±4.3) and 25 healthy females elderly participated. At the time of the experiment, twelve ADLs were selected based on Katz’s ADLs index, and the ADLs that generated the ankle joint motions on the anterior-posterior and medial-lateral. A 3-axis IMU sensor (Seedtech) was used to kinematic measure the ankle joint motions for main ADLs. The IMU sensor had an angle and an angular acceleration value and is attached to the tibia, instep, medial malleolus, and heel such that the local x axis is parallel to the sagittal plane vertical vector and the local y axis is parallel to the coronal plane vertical vector. The range of motion (ROMs) data were analyzed on the basis of major ankle motions. Depending on the results of the ROMs, the representative patterns between ankle joint motion patterns for the ADLs was quantified using the k-means clustering methods and similarity of patterns was analyzed using the cross-correlation methods (MATLAB R2016b Software). The closer the result of the similarity is to '1', the higher the similarity.

RESULTS: During the selected ADLs, the ankle joint motion of the dorsi/plantar flexion was relatively higher than that of the inversion/eversion and the adduction/abduction. In particular, it showed the maximum ROMs in the 'Sit down Cross-Legged and Stand up' motion (dorsi/plantar flexion: 96.9±9.2°, inversion/eversion: 57.1±8.9°, adduction/abduction: 58.9±11.0°), and 'Sit down and Stand up on Chair' motion with little motion of the ankle joint showed the minimum range of motion. (dorsi/plantar flexion: 16.7±2.9°, inversion/eversion: 11.6±2.5°, and adduction/abduction: 11.9±2.8°) (p <0.05). In the main motions of the ankle joint, equally, the similarity between 'Lying on the bed' motion and Sit down squatting and Stand up motions pattern was the highest, and the value was 0.94. The overall similarity value of each ADLs was 0.5 or less.

DISCUSSION: According to the results, there is a statistically significant difference of the ankle joint motions for main twelve ADLs of elderly. Compared with previous studies on the young adults, the ROMs of the elderly ankle joints of the 'normal walking' and 'stair up/down' showed a lower tendency. In the 'Sit down Cross-Legged and Stand up' and 'Sit down squatting and Stand up' motions, which had relatively large motion, the elderly ROM tended to be higher than the young adults. The low similarity results of the overall ankle motion patterns in the twelve ADLs may serve as a basis for qualitatively describing the ADL behavior according to each pattern. And the characteristic range of ankle joint motions according to these ADLs performance is expected to utilized by reflecting on elderly-friendly products development and care.

SIGNIFICANCE/CLINICAL RELEVANCE: The characteristic range of ankle joint motions according to these ADLs performance is expected to utilized by reflecting on elderly-friendly products development and care.

REFERENCES:

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Influence of tibial component position on altered kinematics following total ankle arthroplasty during simulated gait

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INTRODUCTION: Malposition of total ankle arthroplasty (TAA) components has been shown to affect periarticular ligament balance and joint contact mechanics in previous biomechanical studies. However, it is unclear how component position influences ankle joint motion. The objective of this study is to assess the effect of component position on ankle joint kinematics following TAA during simulated gait.

METHODS: Eight mid-tibia cadaveric specimens were utilized in this IRB approved study. The stance phase of gait was simulated both pre- and post-TAA in each specimen using a six-degree of freedom robotic platform. Ground reaction forces and tibial kinematic from in vivo data were replicated while physiologic tendon force profiles were applied to each extrinsic ankle tendons by linear actuators instrumented. Ankle kinematics was measured from reflective markers attached to bones via surgical pins. TAAs were completed using a common fixed-bearing total ankle system following the manufacturer recommended protocol (Salto Talaris, Integra LifeSciences). Using reconstructed CT data, 3D tibial component position relative to a standard ankle joint reference was characterized (Figure 1A). The effect of tibial component position on absolute differences in ankle kinematics (pre – post TAA) was assessed using linear regression with a level of significance set to p = 0.05.

RESULTS: Differences in ankle joint kinematics were only identified in the transverse plane, where internal talar rotation was significantly increased following TAA compared to the native condition (Figure 1B). The medial position of TAA tibial components was found to be positively associated with increased internal talar rotation (Figure 1C; β = 1.861 degrees/mm, R² = 0.72, p = 0.008). No other measurements of tibial component position (anterior-posterior/inferior-superior position, sagittal/frontal/transverse plane angle) were found to be significantly associated with altered ankle kinematics following TAA (All β < 0.1 and p > 0.05).

DISCUSSION: A large proportion (72%) of altered transverse plane kinematics following TAA was explained by medial-lateral component position. While previous studies evaluated the effects of tibial component varus/valgus and anterior/posterior slope malposition on clinical outcomes, this is the first report to identify the influence of medial-lateral position on ankle kinematics. Medial-lateral position of the tibial component is often a neglected parameter during operative procedures, where implants are usually positioned in order to preserve bone stock of the medial malleolus. However, little attention is given to the position of the center of the tibial component in relation to the center of the tibial axis. Further understanding the relationship between implant position and resultant joint function is a specific goal of future research.

SIGNIFICANCE/CLINICAL RELEVANCE: This study suggests that the medial-lateral position of the tibial implant may have an influence on final ankle kinematics. This finding could have clinical implications for techniques implemented during surgical procedures and for the development of new instrumentation systems.
Differences in mechanics during stair ascent after ankle arthrodesis and total ankle arthroplasty

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INTRODUCTION: Ankle arthrodesis, once the standard treatment for end-stage ankle arthritis, today competes with modern total ankle arthroplasty (TAA) which is considered a reliable treatment. Both are effective in providing pain relief and functional improvement. However, patients following TAA demonstrate more normalized gait than patients after ankle arthrodesis {1}; in addition, patients after ankle arthrodesis compensate for loss of ankle sagittal range of motion through increased movement in neighboring joints. Previous studies examined level walking, but not stair navigation. Our objective was to investigate ankle and foot mechanics during stair ascent between ankle replacement and arthrodesis patients.

METHODS: Institutional IRB approval was obtained. A total of 20 patients were recruited and gave informed consent: Ten previously underwent TAA with a modern fixed-bearing ankle replacement (Salto Talaris - Integra Lifesciences or INBONE 2-Wright Medical Technology) and ten patients who previously underwent Fusion. Patients were matched for age, sex, BMI, time from surgery and pre-operative diagnosis. A minimum of 1-year follow-up was required for inclusion. They underwent instrumented 3D motion analysis during a short three-stair ascent at a comfortable speed five times. For the surgically affected limb, sagittal ankle range-of-motion (ROM) (deg), peak moment (Nm/kg), and peak power (W/kg) were calculated during stance along with modified Oxford foot kinematics {2} in the sagittal and frontal planes: hindfoot-to-tibia, forefoot-to-tibia, forefoot-to-hindfoot, and hindfoot-to-lab. Cycle time (s) of the affected limb was included as a proxy for climbing speed. Comparisons were conducted with Mann-Whitney tests for independent samples; significance was set at p<0.05.

RESULTS: We report the results from 17 patients: 9 TAA vs 8 Fusion; three patients could not complete the stair trials with a reciprocal gait pattern and were excluded from analysis. Ankle ROM (25±7 vs 17±5 °, p=.021) and peak ankle plantarflexion power (2.3±0.5 vs 1.4±0.2 W/kg, p=.002) were significantly higher in TAA than Fusion patients (see Figure 1), but peak moment was not significantly different between groups. Oxford foot ROM outcomes were not significantly different between groups for either plane of motion. Climbing speed (1.5±0.3 vs 1.6±0.2 sec, p=.236) was not different.

DISCUSSION: The merits of ankle replacement and fusion are debated among surgeons. TAA patients showed 64% greater peak plantarflexion ankle power despite that Fusion patients ascended 6% faster. Power is a product of the joint’s moment and angular velocity (rad/s) – thus one or both could contribute to the observed difference. Further analysis found both ankle plantarflexion angular velocity and moment were greater on average in TAA patients at the time of peak ankle power, but only angular velocity (164±34 vs 117±23 °/s, p=.011) was significantly different. Still, it is likely both account for the power difference. Higher TAA angular velocity is attributable to their greater ankle ROM.

SIGNIFICANCE/CLINICAL RELEVANCE: Even though previous studies demonstrated similar functional outcomes during level walking between TAA and arthrodesis, TAA patients potentially have improved performance walking on stairs. Future research should compare TAA and arthrodesis during specific daily activities, and not only during level walking.

REFERENCES:

Figure 1. Average ankle flexion power across stance ± standard deviation bands: Fusion (red) and TAA (blue). Box plots show group distributions of late-stance peak values, p-value given.
Session 2: Neuropathy and Motor Control

Co-Moderators: Sicco Bus, PhD (University of Amsterdam) and Paolo Caravaggio, PhD (Instituto Ortopedico Rizzoli)

11:00AM  2-1:  Bus, Sicco A., et al. Plantar pressures, footwear adherence and ulcer recurrence in patients with diabetes and a Charcot midfoot deformity

11:10AM  2-2:  Caravaggi, Paolo, et al. Multisegment foot kinematics and EMG analysis in the type 1 and type 2 diabetic patients


11:40AM  2-5:  Needham, Robert A. et al. Coupling angle mapping to assess multi-segment foot coordination and coordination variability during gait

11:50AM  2-6:  Reeves, Joanna, et al. What are the immediate effects of foot orthosis geometry on tibialis posterior EMG activity and foot biomechanics?
**Plantar pressures, footwear adherence and ulcer recurrence in patients with diabetes and a Charcot midfoot deformity**

Sicco A. Bus¹, Renske Keukenkamp¹, Ruth Barn², Tessa E. Busch-Westbroek¹, Jim Woodburn²

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Disclosures: None

INTRODUCTION: Charcot midfoot deformity is a severe complication of diabetes and a significant risk factor for plantar foot ulceration. However, minimal data is available on footwear management and clinical outcomes in patients with Charcot neuroarthropathy and midfoot rocker deformity, while footwear is an important component in the prevention of foot ulcer recurrence. The aim was to analyze plantar foot pressures, footwear adherence and plantar foot ulcer recurrence in diabetic patients with a Charcot midfoot deformity.

METHODS: Data from a previous footwear trial¹ was used to compare 20 patients with diabetes, peripheral neuropathy, plantar foot ulcer history, and Charcot midfoot deformity, to 118 diabetic patients with the same risk factors but without Charcot diagnosis and midfoot deformity. The institutional review board approved the study and informed consent was obtained from all patients. All patients wore fully custom-made footwear. Barefoot (Emed-X) and in-shoe plantar pressures (Pedar-X) were measured at trial entry. Daily step count (StepWatch) and custom-made footwear use (@monitor) was measured over 7 days, with footwear adherence defined as the percentage of steps that the custom-made footwear was worn. Plantar foot ulcer recurrence was assessed at 18 months. Depending upon the distribution of the data, independent samples t-tests or Mann-Whitney U tests were used to compare differences between study groups. Proportions were compared using Fisher’s exact test. For all tests, a significance level of $P<0.05$ was use.

RESULTS: Median [IQR] barefoot and in-shoe midfoot peak pressures were significantly higher in the Charcot than in the non-Charcot group (barefoot: 756[260,1267] vs. 146[100,208] kPa, $P<0.001$; in-shoe: 152[104,201] vs. 119[94,160] kPa, $P=0.03$). Other foot regions showed significantly lower plantar pressures in the Charcot group. Both groups exhibited similar activity levels (approximately 6600 step counts per day), with no significant group differences present ($P=0.82$). The Charcot group was significantly more adherent (95 [82, 98] % vs. 78 [55,92] %), especially when being at home (94 [86, 95] % vs. 68 [27,89] %) compared to the non-Charcot group ($P=0.001$). Forty percent of the Charcot group patients had a recurrent plantar foot ulcer in 18 months, versus 47% in the non-Charcot group ($P=0.63$); midfoot ulcers occurred more in the Charcot group (4 out of 8 vs. 1 out of 55, $P=0.002$).

DISCUSSION: The data suggest that while in-shoe midfoot peak pressures are considered to be low and substantially improved from barefoot peak pressure, and while footwear adherence is almost optimal, this does not necessarily protect patients with midfoot Charcot deformity against plantar foot ulcer recurrence. Significantly higher midfoot plantar pressures may explain the relatively more midfoot recurrent ulcers in the Charcot group. Further improvement of the custom-made footwear for the midfoot region and the use of region-specific target pressures may be required to improve clinical outcome for these patients.

SIGNIFICANCE/CLINICAL RELEVANCE: Relatively low in-shoe peak pressures and high adherence does not necessarily protect against plantar foot ulcer recurrence in patients with diabetes, midfoot Charcot deformity and plantar foot ulcer history. Further improvement of the custom-made footwear seems indicated, in particular in the midfoot region

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REFERENCES:

Multisegment foot kinematics and EMG analysis in the type 1 and type 2 diabetic patients

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INTRODUCTION: Diabetic foot refers to a complex set of physiological and mechanical alterations characterizing feet of type 1 and type 2 diabetic patients. Providing a comprehensive scenario of the effects of diabetes mellitus on foot biomechanics is not trivial, due to the number of factors that characterize this pathology, such as illness duration and glycosylated hemoglobin level which is associated to the presence and severity of neuropathy. Kinematic, kinetics and EMG analyses have often been performed in isolation, or on a number of limited sample-size diabetic subgroups. This situation has created a fragmented scenario where the available information can hardly be arranged in a coherent picture of diabetic foot biomechanics. In the present study, we aimed at collecting joints motion, force and plantar pressure data from a relatively large population of patients affected by different forms of diabetes mellitus. In this paper the effect of diabetes type 1 and type 2 on foot joint kinematics and EMG activation of leg muscles is reported.

METHODS: From January 2016 a wide sample of patients with diabetes mellitus were visited by an experienced diabetologist and were clinically classified as type 1 or type 2, with or without peripheral neuropathy. 74 patients (25 type 1, 49 type 2; 40 M, 34 F; age 57 ± 12 years; BMI 28.7 ± 6.4 kg/m²) underwent functional evaluation via gait analysis using a validated foot and ankle kinematic protocol with integrated pressure measurements [1, 2]. This allowed measurement of ankle, midtarsal, tarso-metatarsal, and the first MTP joint kinematics during normal walking (Vicon, 100hz). Maximum voluntary contraction (MVC) and gait-cycle activation of the tibialis anterior and gastrocnemius medial head muscles were recorded via wireless sEMG (Cometa, 2000hz). EMG envelopes during walking were normalized in amplitude to the corresponding peak of EMG in MVC. 27 healthy subjects (11 M, 16 F; age 53 ± 9 years, BMI 24.2 ± 3.5 kg/m²) were analysed according to the same protocol and were used as control. Principal component analysis was performed on the foot joints range of motion (ROM), separately, for the control, type 1, and type 2 diabetic groups. Non-parametric Mann-Whitney and Kruskal-Wallis tests were used to assess differences in ROM and EMG maximum activation between diabetic groups and control. Approval was granted by the Ethical committee of the hosting Institute for the gait analysis, and informed consent was signed by all participants in the study.

RESULTS: The peak of tibialis anterior and of gastrocnemius activation during MVC tests were lower in the diabetic type 2 group with respect to the control (p < 0.05). In stance, however, the tibialis anterior showed larger MVC-normalized EMG activation than control. The pooled type 1 and type 2 diabetic patients presented lower mobility in several foot joints with respect to the control (e.g. midtarsal flexion ROM: diabetes=13.5 ± 3.7 deg; control = 15.5 ± 3.3 deg; p < 0.05). Transverse-plane motion of the ankle and first MTP joint dorsiflexion in type 2 diabetic group showed reduced motion with respect type 1. The first four principal components described 83%, 84% and 80% of the variance in foot joint kinematics for the control, type 1 and type 2 subgroups, respectively. The principal components of the control and of the diabetes type 1 joint kinematics were comprised of almost identical sets of ROM variables. Three principal components describing diabetes type 2 kinematics showed significant loadings (> 0.25) for ankle joint sagittal- and frontal-plane ROM and for midtarsal joint frontal plane ROM, which were not present in the principal components describing the variance of diabetes type 1 and control kinematics.

DISCUSSION: This paper is part of a larger investigation on the effects of diabetes mellitus on foot and ankle biomechanics. In accordance with previous kinematic investigations, diabetic feet showed restricted ROM in stance compared to age-matched control feet, but differences were also found between type 1 and type 2 diabetes. It could be speculated that a larger activation of the main plantar/dorsiflexors at the ankle joint during stance would be necessary to compensate for the reduced maximum force exerted by these muscles as recorded in MVC tests. Sagittal- and frontal-plane ROM of the ankle and frontal plane ROM of the midtarsal joint resulted significant factors to describe foot mobility in type 2 diabetes.

SIGNIFICANCE/CLINICAL RELEVANCE: Deeper understanding of multisegment foot kinematics and muscle performance in type 1 and type 2 diabetes may help driving early rehabilitation treatment, as well as a better understanding of the multiple alterations in foot loading which likely occur at a later stage of the disease.

REFERENCES:

Disclosures: Paolo Caravaggi (N), Claudia Giacomozzi (N), Giada Lullini (N), Giulio Marchesini Reggiani (N), Luca Baccolini (N), Maurizio Ortolani (N), Alberto Leardini (N), Lisa Berti (N).
Combined finite element modeling and musculoskeletal modeling techniques can improve diabetic foot preventive management

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INTRODUCTION: Complications of diabetes that affect the lower extremities are common, among them foot ulceration is the most frequently recognized one, and unfortunately also ulcers recurrence is common [1]. Therefore prevention of ulcer and its recurrence is one of the most important topics in the current approach to diabetic foot disease [1]. The best approach is considered a early detection of subjects at risk through a good understanding of the factors that predict ulcers and its recurrence. Knowledge of the predictors can be used to optimize the preventive management [1]. Repetitive stresses both on the plantar surface of the foot and at the level of internal tissues can be detected through finite element modeling (FEM) techniques. Such measurements can also be used to plan therapeutic footwear for preventive management. Recently the authors demonstrated that the development of foot FEM can be enhanced on diabetic subjects by adopting both subject specific geometries (MRI based) and boundary conditions acquired during gait [2]. Furthermore by including lower limb muscle forces as further boundary conditions, the foot FEM simulation results were improved even more on a healthy subject [3]. The aim of this study was twofold: first to evaluate the possibility of improving the performances of a foot FEM applied to a cohort of neuropathic subjects by including lower limb muscle forces computed in OpenSim, and second to verify the impact of the novel approach on the internal stresses estimation.

METHODS: Eight neuropathic foot FEMs and 10 healthy subjects were developed as in [2] by applying subjects specific boundary conditions acquired during gait to two, previously developed, foot FEMs respectively of a healthy and of a neuropathic subjects [2]. Three gait trials per subjects were acquired through 2 force plates (Bertec FP4060), a motion capture system (BTS S.r.l), 2 plantar pressure plates (Imago Ortesi), an 8 channels electromyographic system (BTS) [2, 4]. Hence subjects specific muscle forces were determined as in [4] in OpenSim for both healthy and diabetic subjects and compared through T-Test (p<0.05). FEM simulations were run by considering only the kinematics and the ground reaction forces as boundary conditions [2] or by including the muscles forces generated with the Gait 2392 model in OpenSim. Plantar pressure data obtained through the FEMs were compared with the experimentally measured ones in both conditions for validation purposes.

RESULTS: Results showed that a better approximation of the experimentally measured plantar pressure was obtained when adopting the FEM driven with the muscle forces as boundary conditions together with the ground reaction forces and the kinematics (Figure 1). Furthermore these models leaded to lower Von Mises stresses, thus confirming that the important role of muscle forces in foot biomechanics shouldn’t be neglected when developing foot FEMs.

DISCUSSION: The adoption of foot FEMs driven with lower limb muscles forces showed the possibility to improve prediction of both internal stresses and strain not only on healthy subjects [3] but also on diabetics.

SIGNIFICANCE/CLINICAL RELEVANCE: This methodology can be adopted to optimize the preventive management by providing a early detection of subjects at risk and in order to develop specific insoles that can improve gait biomechanics and foot function aiming to reduce foot internal and external stresses.

REFERENCES:

Figure 1: Top: OpenSim Static Optimization results: comparison between mean muscles forces (dashed lines) plus and minus 1 standard deviation of both CS (in blue) and of DPNS (in red). Results of T-Test have been reported in term of the instants of the gait cycle where significant differences were found (blue asterisk means p < 0.05). Bottom left: Von Mises stresses on plantar soft tissues for one Neuropathic subject: simulations with the GRF FEM model and the one with GRF and muscle forces. Bottom right: comparison between simulated and experimental plantar pressures on the 8 neuropathic subjects.
**Plantar System and Dysproprioception. Clinical evaluation of participation.**

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**INTRODUCTION:** Regulation of the balance function is organized in hierarchical patterns and chronologically requires the central sensory integration of information on vestibular, visual, cutaneous and proprioceptive inputs, as well as the motor command and component (1; 2). The plantar system (PS) provides cues and feedback coming from tactile and proprioceptive information (3). Their sensory integration contributes to the balance function control. Sensory integration can be affected by syndromes such as: the Proprioceptive Dysfunction Syndrome (PSD; 4), the Oral Dysperception Syndrome (ODS, 4), the Postural Deficiency Syndrome (5-7) or the Sensory Processing Disorders (8; 9). In the aforesaid syndromes, PS participation is usually assessed by clinical evaluation such as delays in crawling, standing, walking or running (8; 9) or more recently by Vertical Heterophoria (VH; 4; 10). To evaluate the sensory processing disorder, called “dysesthesiology” in patients presenting PDS and ODS (4), clinicians and podiatrists usually use VH, separated into 7 conditions: 5 in sitting positions without plantar contact and 2 in natural standing position with and without foam (4). The last 2 conditions require some PS participation but do not provide podiatrist with enough information on disruptor and/or regulator factors.

In order to improve clinical practice, we propose a new framework, adding 14 conditions to allow the evaluation PS participation with the MP, on foam thickness: 1) 3 mm, reduction of exteroceptive and nociceptive cues, 2) 5 mm, reduction of exteroceptive cues and recruitment of plantar surface, 3) 8 mm, recruitment of proprioceptive cues, 4) 10 mm, induction of proprioceptive instability; 5) with stimulation (11-13) to maximize plantar cutaneous sensory awareness and increase feedback from cutaneous receptors; 6) with the patient’s current shoes; 7) with therapeutic foot orthoses/insoles; in standing and sitting position to compare results with the control conditions (standing and sitting in spontaneous position, 4).

**DISCUSSION:** The hereby proposed clinical score aims to follow the scoring defined by Quercia et al.(10), e.g. the sum of the number of conditions giving the Vertical Orthophoria (VO) and changing MP Index of Lability (IL; 10): a number between 0 and 14 for VO and IL for sitting and standing. Higher numbers would expose a significant relevance of implication of the PS in regulation for VO, while it could suggest perturbation or neutral for IL. The efficient sensory integration would be obtained with a higher VO and lower LI score. Low numbers would expose poor sensory integration. Consequently the VO and LI score variations could first indicate where the sensory processing disorder is, and then the level of participation of the PS in terms of perturbators and/or regulators.

This new clinical score could be used to improve the podiatry therapeutic proposition, help communication between clinicians of different domains, and help follow patients’ foot health in clinical routine. However, this framework of the PS participation still needs to be validated in clinical trials.

**SIGNIFICANCE/CLINICAL RELEVANCE:** Improvement of the clinical practice to evaluate the PS participation in PDS, ODS and Sensory Processing Disorders. High score of vertical orthophoria show good sensory integration. Creation of a clinical score to evaluate the efficiency/inefficiency of the sensory integration of the PS cues in sensory processing disorders.

**REFERENCES:**


Key-words: learning disorder, foot insoles, sensory processing disorders, Proprioceptive Dysfunction Syndrome, podiatry.
Coupling angle mapping to assess multi-segment foot coordination and coordination variability during gait

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INTRODUCTION: Vector coding (VC) provides a quantitative measure of movement coordination and coordination variability (CAV). Traditional time-series reporting of VC data can be difficult to interpret when multiple trials are overlapped. Coupling angle mapping (CAM) is a reporting approach that uses color to depict movement coordination and CAV across multiple participants and trials (Needham et al., 2017). This study is the first of its kind to apply this technique on the multi-segmented foot to report on novel insights into the coordination pattern (CP).

METHODS: Data were obtained from ten male participants ((mean ± standard deviation) age: 22.4 ± 2.46 years, height: 180.3 ± 7.18 cm, mass: 74.97 ± 11.02 kg). Ethical approval was sought and granted by the University Research Ethics Committee. Rearfoot and medial-forefoot kinematic data was collected (100Hz) using an 8-camera motion capture system (VICON, Oxford, UK). Procedures and data processing for VC in addition to CAM are reported elsewhere (Needham et al. 2017). At each instant in time, the coupling angle (CA) is assigned to a CP classification (Figure 1c).

RESULTS: In figure 1, subtle differences in the CP were noted between trials (a), while the CP for several participants did not coincide with the group (b). CAV was higher and extended for the group (b) in comparison with participant 10 data (a).

DISCUSSION: Although CAM highlights the CP, this novel approach showcases the reporting of segmental dominancy that details the changes in the distribution of the CA within the CP classification. CAM noted differences in the CP between participants during early stance phase that questions the interpretation and clinical relevance of reporting group CAV data.

SIGNIFICANCE/CLINICAL RELEVANCE: This work presents CP, CAV and segmental dominancy data via illustrations that provides unique insights into segmental movements that may support the design of individualized clinical interventions.

FIGURES AND TABLES:

Figure 1. Coupling angle mapping during stance representing coupling angle data on forefoot-rearfoot coordination in the sagittal plane across five trials (T1-T5) for participant 10 (P10) (a); mean coupling angle data across 10 participants (P1-P10) (b). Coordination variability (CAV – legend ‘e/f’) and segmental dominancy (Seg. Dom. – legend ‘d’) are also presented.

What are the immediate effects of foot orthosis geometry on tibialis posterior EMG activity and foot biomechanics?

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INTRODUCTION: Foot orthoses used to treat conditions like tibial posterior tendon dysfunction can effectively reduce the external forces applied to the foot. However little is known about whether specific aspects of foot orthosis geometry can affect activation of the tibialis posterior. If an orthosis reduced the external eversion moment during stance, we might expect the force required from the tibialis posterior to resist eversion to be less and so a reduced EMG activation would be seen. Reduced activation could mean less force through the tendon which could facilitate healing in the case of tibialis posterior tendon dysfunction. The aim of this study was to establish if medial heel wedging and increased medial arch height have effects on EMG of tibialis posterior and other muscles of the lower limb, and foot and ankle moment/motion.

METHODS: Ethical approval was obtained from the University of Salford (HSR1617-36). Healthy participants (n=10) performed walking trials in standardised shoes with five inserts in a random order: i) a flat inlay and ii) a standard Salfordinsole®, and a Salfordinsole® with iii) a 6 mm increase in arch height, iv) an 8 mm increase in medial heel wedging and v) both a 6 mm increase in arch height and an 8 mm increase in medial wedging. Recording of the tibialis posterior was performed with bipolar fine-wire electrodes (44 gauge × 100 mm paired-hook wires, Teflon-coated stainless-steel wire) using the posterior approach. Kinematic and kinetics data was collected concurrently. The root mean squared (RMS) EMG signal was normalised to the peak of the average of the six gait cycles in the flat inlay condition. A repeated measures ANOVA, or non-parametric equivalent, will be performed to compare peak EMG between conditions for early and mid-stance.

RESULTS: Preliminary results (n=10) show that tibialis posterior activity reduced in early stance in the wedge and combined arch and wedge conditions and also reduced slightly in the arch and arch and wedge conditions in late stance relative to the flat inlay (FIGURE). However there was typically a reduction in moment across stance for all orthotic conditions.

DISCUSSION: The findings of this study so far are similar to that of Murley et al. (1) who found a reduction in tibialis posterior activity in early stance with customized and prefabricated orthoses relative to a shoe only. Large between trial variability exists in tibialis posterior activity, however there appears to be a trend towards phase specific reduction in tibialis posterior activity with orthoses. With a greater sample size the specific effects of heel wedging and arch height on tibialis posterior activity may become more apparent.

SIGNIFICANCE/CLINICAL RELEVANCE: By relating change in tibialis posterior activation with foot orthoses to change in the forces and motion at the foot and ankle we can better understand the mechanism by which foot orthoses can benefit patients. If we can better understand how a foot orthosis can offload the tibialis posterior tendon we may be able to improve the prescription of treatments for tibialis posterior tendon dysfunction.


FIGURE: Average RMS amplitude of the tibialis posterior over the gait cycle expressed as a percentage of the grand peak of the flat inlay condition. Red line: flat inlay; yellow line: standard Salfordinsole®; black line: 6 mm increase in arch height; green line: 8 mm increase in medial wedging and blue line: both a 6 mm increase in arch height and an 8 mm increase in medial wedging. Shaded pink area represents average standard deviation of flat inlay condition.
Session 3: Pediatric Foot (SS)

Chair: Carina Price, PhD (University of Salford)
Co-Moderators: Dieter Rosenbaum, PhD (Clinical Research & Services, Otto Bock) and Russel Volpe, DPM (NYCPM)

1:30PM Karen Adolph, PhD (invited)

2:00PM 3-1: Caravaggi, Paolo, et al. Postural and kinematic alterations in the paediatric asymptomatic plano-valgus foot joints

2:10PM 3-2: Chard, Angus, et al. Effect of thong style flip-flops and supportive shoes on children’s midfoot joint power during the propulsive phase of walking

2:20PM 3-3: Evans, Angela M. Paediatric Flatfeet – Identifying the Boomerangs

2:30PM 3-4: Holmes, Sarah J., et al, Impact of multilevel joint contractures of the hips, knees and ankles on the Gait Profile Score in children with cerebral palsy

2:40PM 3-5: McClymont, Juliet, et al. Average peak plantar pressure and variability in foot contact morphology (area) during two dynamic locomotor tasks in infants

2:50PM 3-6: Tulchin-Francis, Kirsten, et al. Quantifying Pediatric Foot Deformities using Multi-Segment Foot Kinematics & Pedobarograph

3:00PM 3-7: Amene, Juliet, et al. Kinematic Foot Types of Planovalgus Feet in Children with Cerebral Palsy
Postural and kinematic alterations in the paediatric asymptomatic plano-valgus foot joints

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INTRODUCTION: Plano-valgus (PV) is a common alteration of foot posture present in the paediatric population, characterized by valgus rearfoot, foot pronation and drop of the medial longitudinal arch (MLA). If misdiagnosed, this condition has the potential to cause pain and discomfort, and may hinder the lower limb kinematic chain. While a number of studies have investigated the kinematics of the paediatric PV foot, e.g. [1], no information is thus far available on postural and kinematic alterations of the major joints spanning the MLA - i.e. midtarsal and tarso-metatarsal.

METHODS: 20 children (13 M, 7 F; 13 ± 1 years) with bilateral asymptomatic PV foot were recruited in the study. Radiological indicators of PV condition, such as the calcaneal pitch, lateral talo-first metatarsal angle, and talo-navicular coverage, were measured from weight bearing X-rays. Gait analysis was conducted on the children’s feet with the Rizzoli Foot Model [2, 3]. This was applied to measuring double-leg support upright static posture and gait kinematics of the main foot joints, including midtarsal and tarso-metatarsal joints, along with MLA deformation. Range of motion and temporal profiles of joint rotations were compared to those from a control group of age-matched children with normally-developed (ND) feet (4 M, 6 F; age 13 ± 1 years). Mann-Whitney U test was used to assess differences in static posture and kinematic global parameters between PV and control. One-dimensional statistical parametric mapping was used to determine differences in stance-normalized foot joint rotations between PV and control. Acknowledgement of the Hospital’s IRB was granted (protocol n° 7/17) and parents’ informed consent was obtained for all children recruited in the study.

RESULTS: The PV midtarsal joint was more dorsiflexed, everted and abducted than that in the control group, but showed reduced median sagittal-plane ROM (PV= 15.9 [12.1 19.2] deg; ND = 22.2 [19.5 24.9] deg; p < 0.01). The tarso-metatarsal joint was more plantarflexed and adducted, and showed larger frontal-plane ROM. The MLA showed larger ROM (PV= 58 ± 18 deg; ND = 37 ± 9 deg; p < 0.05) and was more dropped throughout gait duration. A diagrammatic representation of sagittal-plane orientation of foot segments in static posture and at push-off in stance is shown in figure 1.

DISCUSSION: Similar to what reported in previous studies, the PV hindfoot resulted significantly everted and plantarflexed with respect to the tibia, and the MLA was more collapsed throughout stance duration. In addition, the Rizzoli Foot Model allowed investigation of the postural and kinematic alterations at the midtarsal and tarso-metatarsal joints. It should be highlighted that children with PV foot walked more slowly and with a reduced stride length than control, and this might have – albeit marginally - affected the differences observed in the joint rotation profiles.

SIGNIFICANCE/CLINICAL RELEVANCE: In the paediatric plano-valgus foot, a hindered windlass mechanism and/or insufficient activation of the intrinsic plantar muscles [4] may be responsible for larger dorsiflexion of midfoot joints and greater collapse of the MLA during gait. A better understanding of PV midfoot joints postural and kinematic alterations is necessary to improve diagnosis and treatment of PV condition.

REFERENCES:

Figure 1 Diagrammatic representation of the sagittal-plane intersegmental orientation in double-leg support static posture (left), and at push-off (right) for the normally-developed (top) and plano-valgus (bottom) feet according to the average intersegmental rotations from the two groups.
Effect of thong style flip-flops and supportive shoes on children’s midfoot joint power during the propulsive phase of walking

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INTRODUCTION: Thongs may be beneficial for children’s developing feet despite increased ankle dorsiflexion and midfoot plantarflexion seen during walking[1]. Enclosed footwear limits midfoot joint motion and power[2,3]. Ankle and midfoot joint plantarflexion is necessary for the generation of foot and ankle power during the propulsive phase of walking. The splinting effect of enclosed footwear significantly reduces the midfoot contribution. This effect may change intrinsic and extrinsic foot muscle contraction, thought necessary for developing feet and a child’s arch development. The purpose of this study was to evaluate the effect of wearing thongs and supportive shoes on children’s barefoot midfoot joint power during the propulsive phase of walking. It is hypothesized that wearing thongs will not affect midfoot power generation while wearing supportive shoes will reduce midfoot power generation during the propulsive phase of gait.

METHODS: Walking data were recorded from twelve healthy children aged 8-12 years at 200 Hz. A multi-segment foot model was used in an inverse dynamics analysis to derive ankle and midfoot power. Five trials of each participant and each footwear condition were collected while walking at self-selected speed with the order of barefoot, thong and supportive shoe randomized. A repeated measures ANOVA with a threshold of p<0.05 was used to determine the significance of peak midfoot power generation differences.

RESULTS: In the sagittal plane peak power generation at the midfoot was similar when thongs were worn (1.68 W/Kg; p>0.696, 95% CI [1.26, 2.10]) to that seen when children were barefoot (1.525 W/Kg 95% CI [1.067, 2.01]), while supportive shoes reduced midfoot peak power generation by 0.525W/kg (42%) to 1.00 W/kg (p<0.012, 95%CI [0.50, 1.50]) due to a reduced plantarflexing angular velocity by 148% (68%) p<0.0001, and plantarflexion moment by 0.10 Nm/kg (11%) p>0.018.

DISCUSSION: During the propulsive phase of walking, the midfoot functioned in the sagittal plane similarly to barefoot while thongs were worn. The supportive shoes, however, limited the power production of the midfoot. The long-term effect of wearing such supportive shoes should be investigated to determine what deleterious effects they may have on lower limb health.

SIGNIFICANCE/CLINICAL RELEVANCE: Children’s barefoot midfoot dynamics were unaffected during the thong condition, while supportive shoes had a splinted effect on the midfoot, reducing barefoot midfoot power generation. Reducing the splinting effect of supportive shoes may be beneficial for children’s midfoot joint dynamics.

REFERENCES:

Figure 1. Sagittal plane midfoot joint power (W/Kg), generation to positive, during the propulsive phase of walking, while barefoot (solid line), wearing thongs (dashed line) and supportive shoes (dotted line).
Paediatric Flatfeet – Identifying the Boomerangs

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INTRODUCTION: The paediatric flatfoot is a frequent presentation to clinicians across medical and allied health disciplines, and a common concern for parents. It has been hotly debated with respect to the need for, and indications directing, intervention. Whilst available level 1 evidence, clarifies the lack of support for bespoke foot orthoses for children with asymptomatic flatfeet, there remains regular query as to which degree of paediatric flatfoot is ‘normal’, and which are most likely to be problematic in the future?

METHODS: Review of current research for the evidence for treatment of flatfeet reveals a substantial quantity of literature focused on diagnosis, clinical significance, and intervention.

RESULTS: This paper examines the paediatric flatfoot and varied influences, including military history, hierarchical evidence, and contemporary research findings. It identifies three validated assessment tools for clinical use, and normal range parameters. Normative data for paediatric foot posture, utilizing the Foot Posture Index (FPI-6), has been expanded to include international populations from the UK, Spain, and Australia for ages three to 14 years (n>3000). The paediatric flatfoot proforma (p-FFP) and triage tool (3qq) both provide clinical frameworks for intervention and prioritizing respectively. In addition, three morphological attributes of the foot are identified as ‘predictive’ of symptoms: heel angle, talo-navicular joint coverage angle, ankle range/midfoot stability.

DISCUSSION: From the wider perspective of public health, the concept of the ‘boomerang’ will be introduced to emphasise the morphological criteria, which merit clinical attention, and which are most probably associated with the painful flatfoot. The evolving issue of paediatric foot strength is also considered.

SIGNIFICANCE/CLINICAL RELEVANCE: Appreciated within the context and range of normal foot development, the ‘boomerang’ concept helps clinicians to methodically identify the paediatric flatfoot which is more likely to ‘return’ as a problem. This concept is easily understood by parents and school-aged children.

REFERENCES:
Impact of multilevel joint contractures of the hips, knees and ankles on the Gait Profile Score in children with cerebral palsy

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DISCLOSURES: This research did not receive any funding from agencies in the public, commercial, or not-for-profit sector.

INTRODUCTION:

Three dimensional gait analysis (3DGA) generates a large and complex dataset, often presenting a barrier to its effective use in clinical decision making when assessing the walking ability in children with cerebral palsy. The Gait Profile Score (GPS) is a single index measure of 3DGA kinematic data, helping to enhance clinical interpretation for patients, families and clinicians. With the increasing use of gait indices, it is necessary to determine whether these indices can represent the severity of such clinical impairments. Therefore, the primary aim of this study was to investigate the extent to which sagittal plane multilevel joint contractures influence the GPS in children with cerebral palsy.

METHODS:

Children who attended our gait analysis service from January 2011 to July 2016 and provided informed consent were eligible for inclusion in this study (Sydney Children’s Hospitals Network Human Research Ethics Committee LNR/12/SCHN/146). Children with a diagnosis of cerebral palsy, who were 18 years or younger and classified as a Gross Motor Function Classification System level I-III were included. Previous surgery warranted exclusion from this study.

Range of motion in the hip, knee and ankle was measured in the sagittal plane using a goniometer. A contracture was defined as any angle below the minimum range of values collected from 50 typically developing children. The children underwent 3DGA, walking along an 8m walkway, using a Vicon MX system with eight infrared cameras and three AMTI force plates. The GPS was calculated using MATLAB 2016b with a higher score indicating a greater deviation from normal.

Analyses were performed with SPSS software v22. The study sample was characterised using descriptive statistics. One sample t-tests compared the participant’s range of motion and GPS to normative values. Pearson’s product correlation (r) assessed the relationship between patient characteristics, physical examination measures and the GPS. One way analysis of variance (ANOVA) assessed the influence of the number and type of contracture on the participants’ GPS (ipsilateral limb). Stepwise multiple regression analysed which physical examination measures contributed to an elevated GPS (ipsilateral limb). Significance was set at p<.05.

RESULTS:

145 children (mean age: 11 years, 4 months; SD: 2 years, 10 months; range 5-18 years; 83 males) were included. The GPS of the participants was significantly higher than the normative values (p<.05). There was a significant correlation (p<.001) between an elevated GPS (ipsilateral limb) and reduced range of hip extension (r=-.348), knee extension (r=.353) and ankle dorsiflexion (knee extended) (r=.466).

As the number of joints affected by contracture increased, the GPS also increased. Children with combined ankle, knee and hip contractures had an increased GPS (mean 17.5°, SD 6.2°) compared to the participants with ankle contractures only (GPS mean 12.8°, SD 5.1°) or no contractures (GPS mean 11.0°, SD 2.3°) (p<.05). In multiple regression, reduced hip extension (β=-.244, p<.001) and ankle dorsiflexion (knee extended) (β=-.421, p<.001) range of motion, and knee flexion weakness (β=-.346, p<.001) predicted 47% of the variance in the GPS (r²=.465).

DISCUSSION:

The GPS provides a simple measure for clinicians and families to quantify the impact of contractures on kinematic gait. Reduced ankle dorsiflexion (knee extended), knee extension and hip extension range of motion were significantly correlated with an increased GPS. This is consistent with previous studies correlating joint range of motion to kinematic variables during gait. Children with more joint contractures had a significantly increased GPS compared to those with none or limited contractures. This is the first study to use the GPS as a measure of the severity of physical impairments in this population.

Restrictions in ankle dorsiflexion (knee extended) range of motion, hip extension range of motion and knee flexion weakness, predicted almost half of the variation in the GPS. Targeting and regularly monitoring these impairments may guide therapists in referring patients to appropriate and timely medical management.

There were some limitations to this study. First, as range of motion and spasticity were highly correlated both measures could not be included in the multiple regression analysis. Second, coronal and transverse plane abnormalities were not included however are likely to contribute to an elevated GPS. Third, there was a lack of consensus in the literature when defining contracture, presenting a barrier to standardising outcome measures for research and for clinicians managing possible contractures. Finally, future studies should assess the impact of multilevel contractures on measures of everyday function and quality of life.

CLINICAL RELEVANCE:

The GPS demonstrates the impact of multilevel sagittal plane joint contractures on kinematic gait parameters in children with cerebral palsy. This supports the use of the GPS as a simplified measure to quantify the contribution of contractures to gait limitations and may direct clinicians in deciding appropriate interventions for these children.
Average peak plantar pressure and variability in foot contact morphology (area) during two dynamic locomotor tasks in infants

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Introduction: Throughout infancy the foot continually develops as a weight bearing structure and plantar pressure profiles of independently walking infants are available (Bosch et al. 2007). However, we lack information about how the foot develops prior to independent walking, including the distribution and magnitude of average peak pressure and variation in foot contact morphologies (areas) during supported walking (a precursor to independent walking). We therefore aimed to investigate plantar pressure and contact morphologies during supported and independent walking in infants.

Methods: Ethical approval for data collection was obtained by the Universities of Brighton and Salford and informed consent collected from the parent of the infant participants. Contact area parameters were collected within two custom designed baby-friendly lab spaces (4m²). Pressure records were collected on a Novel Emed XL pressure platform embedded in a 4m² area. Data has been processed on a total of 6 babies (4 independent walkers and 2 supported walkers) and total N = 198 steps during supported and independent walking to date (aged 11-15 months). Supported walking was defined as the participant walking willingly whilst the parent supporting the child by hand above their head. Each step was extracted individually and masked to define rearfoot, mid- and forefoot. As an initial starting point, contact areas were defined when each region was loaded and in contact with the pressure mat for more than 50% of the overall stance phase. This method was thought to allow for the regular posterior shift (e.g., initial forefoot strike with a posterior shift to the midfoot area) in contact area during supported walking. The dataset is comprised of a tally of all steps collected during supported and unsupported walking.

Results: Initial results demonstrate high and comparable variation in contact area during both supported, and independent walking. During supported walking, approximately 70% of steps spent >50% of stance time in the midfoot, 55% of steps utilised the forefoot and 55% of steps spent >50% of stance phase loading the rearfoot. During independent walking, approximately 94% of steps spent >50% of stance time loading the heel and midfoot, whilst 82% of steps utilised the forefoot.

Discussion: Compared to independent walking, supported walking might be characterised by a more variable distribution of load exploiting as biomechanical variability develops step-to-step in accordance with dynamic systems. However, we used 50% of stance as a cut off for accepting the rear, mid- or forefoot as being in contact with the pressure mat, and future work will analyse how sensitive results are to different criteria. Also, regional analysis of plantar pressures (i.e. rear, mid-, forefoot) may be sensitive to the effects of assumptions in the distribution of pressure. We propose to employ statistical parametric mapping as a more objective approach to plantar pressure analysis.

Clinical relevance: Clinical decisions about normal or otherwise development of the paediatric foot require an understanding of how and why pressure distributions develop. Variations in contact morphologies presented here provide insight into the development of the functional units of the foot as it matures into a weight bearing structure.

References:

Acknowledgements: This work is funded by the William M Scholl Foundation (UK) as part of the Great Foundations initiative (www.greatfoundations.org.uk).

Figure 1: Pressure images (a,b) represent continuous supported walking in one infant, showing the variation in contact area step-to-step, and compared to independent walking (c).
Quantifying Pediatric Foot Deformities using Multi-Segment Foot Kinematics & Pedobarograph

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Texas Scottish Rite Hospital for Children, Dallas, TX
Kirsten.Tulchin-Francis@tsrh.org

Disclosures: Kirsten Tulchin-Francis (N), Ashley Erdman (N), Sophia Ulman (N), Jacob Zide (N), Anthony Riccio (N)

INTRODUCTION: Multi-segment foot (MSF) kinematics and plantar pressures have been shown to identify biomechanical variances in foot function in children and adults with foot pathology. Structural differences and abnormal kinematic motion often dictate changes in plantar pressures, however, direct correlation of measures is often difficult due to complex foot deformities and compensatory motion. The purpose of this study was to evaluate the relationship between plantar pressures and coronal plane MSF kinematics across a wide range of pediatric and adolescent foot pathology.

METHODS: Fifty patients with foot deformities were enrolled in prospective, IRB approved research aimed at evaluating foot structure and function. Foot deformities ranged from severe planovalgus to cavovarus in patients presenting with a variety of structural (tarsal coalition, clubfoot sequela), functional (deformity due to trauma) and neurological (cerebral palsy, myelomeningocele) diagnoses. Feet were divided into three categories based on the average coronal hindfoot position during the single limb stance (SLS) phase of gait using a custom MSF model [1]. Categories were defined using multi-segment foot kinematic data of N=20 typically developing individuals without history of foot pathology, musculoskeletal disease, lower extremity injury or neuromuscular disorder. Correlations were evaluated between average coronal hindfoot (HF) position and coronal forefoot (FF) position during SLS, and the hindfoot to forefoot angle (HFA) and the distribution of lateral to medial forefoot peak pressure [LM_FF, defined as (L-M)/(L+M)], based on plantar pressure analysis.

RESULTS: Using the control group (mean ± one stdev) as reference, the subject cohort consisted of 13/50 feet which had significant hindfoot valgus (Valgus) during gait, 17/50 feet exhibiting hindfoot varus (Varus), and the remaining feet within normal limits (HFNorm). Due to variation in sagittal and coronal compensations, the average forefoot position during stance was not significantly different between Valgus and HFNorm feet (Table 1). However, Varus feet had significantly less forefoot inversion compared to both Valgus and HFNorm. There was a moderate correlation between coronal hindfoot and forefoot position during SLS (r= -.60). HFA and LM_FF also demonstrated moderate correlation to hindfoot position (r= -.6 and 0.52, respectively), and were able to distinguish between Valgus feet and both HFNorm and Varus feet. However, these variables were not significantly different between Varus and HFNorm.

DISCUSSION: While anatomical abnormalities and/or changes in kinematic motion often lead to changes in plantar pressures, multiple variations in hindfoot and forefoot position, across all three planes, can lead to considerably different plantar pressure patterns. For example, two feet with similar amounts of hindfoot valgus can present with significantly different forefoot plantar pressure patterns based on the amount of forefoot compensation. This study demonstrates that, even in the presence of large differences in hindfoot deformity, it is important to look at both motion and plantar pressures to examine the complex foot structure.

SIGNIFICANCE/CLINICAL RELEVANCE: The application of a multi-segment foot model and plantar pressures can be used to detect variations in complex, 3D foot mechanics in children with pathology. However, it is important that clinicians and researchers examine the entire foot complex to identify compensations and deficiencies among all relevant variables. The additional information gained can be used to aid clinical decision making and to assess treatment outcomes.


ACKNOWLEDGEMENTS: The authors wish to acknowledge support from the TSRHC Research Program Fund.

Table 1: MSF and Plantar pressure measures based on coronal hindfoot position during gait

<table>
<thead>
<tr>
<th></th>
<th>Valgus Hindfoot (N=13)</th>
<th>Valgus Hindfoot (N=20)</th>
<th>Valgus Hindfoot (N=17)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ave Cor HF (SLS)</td>
<td>-13.6 ± 5.1</td>
<td>-2.9 ± 2.3</td>
<td>12.1 ± 8.9</td>
</tr>
<tr>
<td>Ave Cor FF (SLS)</td>
<td>12.0 ± 8.5</td>
<td>7.4 ± 7.4</td>
<td>-2.9 ± 8.2 * Δ</td>
</tr>
<tr>
<td>HFA</td>
<td>189.0 ± 7.4</td>
<td>177.8 ± 9.4 *</td>
<td>173.6 ± 5.5 *</td>
</tr>
<tr>
<td>LM_FF</td>
<td>-0.3 ± 0.5 *</td>
<td>0.3 ± 0.5 *</td>
<td>0.4 ± 0.4 *</td>
</tr>
</tbody>
</table>

* Significant difference from Valgus
Δ Significant difference HFNorm to Varus

Figure 1: HFA vs. coronal HF position during SLS. Individual Subjects shown as single data points. Group mean ± 2 stdev shown as colored rings.
Kinematic Foot Types of Planovalgus Feet in Children with Cerebral Palsy

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Disclosures: None

INTRODUCTION: Pes planovalgus (PV) is characterized by flattening of the medial longitudinal arch, along with hindfoot valgus. It is the most common foot deformity in ambulatory cerebral palsy and accounts for 25-30% of all surgical procedures in children with CP [1]. Previous investigations of the multi-segment foot and ankle kinematics of children with PV secondary to cerebral palsy identified variability in foot motion during gait [2]. Such variability in foot and ankle kinematics has been addressed by identifying foot types using multi-segmental foot and ankle analysis, principal component analysis (PCA) and cluster analysis [3]. Therefore, the purpose of the current project was to identify clinically relevant foot types among a sample of typically developing (TD) children and children with PV secondary to cerebral palsy.

METHODS: This retrospective study was approved by an institutional review board (IRB). The study sample (PV group) included 31 children (age = 11.5 ± 2.4 yrs) with pes planovalgus (12 unilateral and 19 bilateral, total of 50 feet) and a control group of 16 typically developing (TD group) children (32 feet total, age = 11.3 ± 2.0 yrs). Gait analysis was performed using the Milwaukee Foot Model to characterize kinematics of the foot and ankle [4]. Principle component analysis (PCA) was used as a data reduction technique on 32 variables. Variables described tibia, hindfoot and forefoot kinematics, as well as walking velocity. K-means clustering analysis was performed on the principal components (PCs) and a one-way analysis of variance (ANOVA) was done to evaluate the effect of foot type on PC scores. When a main effect of foot type was identified, post-hoc comparisons were made relative to the TD Group (rectus foot type) using Dunnett’s test.

RESULTS: PCA resulted in 7 PCs accounting for 91% of the variance of the original dataset. The PCs described the segmental/plane of involvement of the deformity and joint excursions across the gait cycle. K-means clustering identified seven foot types (Table 1, Figure 1). Type 1 was identified as the control group with a rectus foot type. A main effect of foot type was not identified for PCs 6 and 7; therefore, post-hoc analysis was not performed on those PC scores.

DISCUSSION: The current study identified 7 distinct, foot types among a sample of TD children and children with PV using multi-segment foot and ankle kinematics as inputs for PCA and K-means cluster analysis. PCA reduced 32 variables describing the location and plane of involvement in the foot and ankle to 7 PCs. Cluster analysis identified subgroups of participants with PV who presented with variable planes of involvement and severity. This analysis also provided insight into the intersegmental relationship between the hindfoot and forefoot in the presence of a midfoot break.

SIGNIFICANCE/CLINICAL RELEVANCE: The current study demonstrated that each subgroup presented unique characteristics of PV including segment(s) and plane(s) involved, as well as severity of deformity. These quantitative methods can ultimately be used to analyze severity of deformity and can be used to facilitate treatment planning when used in conjunction with other information.


ACKNOWLEDGEMENTS: The contents of this abstract were developed under NIDILRR grant 90AR5022-01-00.

Table 1. Comparisons of PC1-PC5

<table>
<thead>
<tr>
<th>Foot Type</th>
<th>PC1</th>
<th>PC2</th>
<th>PC3</th>
<th>PC4</th>
<th>PC5</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 (n=12, control; 2 CP) - Rectus Foot Type</td>
<td>↑</td>
<td>↓</td>
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<td>↓</td>
<td></td>
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<tr>
<td>2 (n=10; 2 control, 14 CP)</td>
<td>↓</td>
<td>↑</td>
<td>↑</td>
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<tr>
<td>3 (n=1; 2 CP) - Plane foot</td>
<td>↓</td>
<td>↑</td>
<td>↑</td>
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<tr>
<td>4 (n=12, 2 CP) - Classic PV</td>
<td>↓</td>
<td>↑</td>
<td>↑</td>
<td>↓</td>
<td></td>
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<tr>
<td>5 (n=3, 3 CP)</td>
<td>↓</td>
<td>↑</td>
<td>↓</td>
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<tr>
<td>6 (n=6, 6 CP)</td>
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<tr>
<td>7 (n=15, 15 control, 6 CP) - Planovalgus Foot Type</td>
<td>↓</td>
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<td></td>
</tr>
</tbody>
</table>

All comparisons made to Foot Type 1 (Rectus)
Session 4: Vendor Technology Update

3:10PM       Novel Electronics
Session 5: Military Biomechanics

Co-Moderators: Don Goss, PT, PhD, Kenneth Cameron, PhD, and Michael Neary, DPM (Keller Army Hospital and the United States Military Academy)

3:50PM  5-1: Song, Jinsup, et al. Assessment of foot structure, flexibility and function in USMA cadets

4:00PM  5-2: Cameron, Kenneth, et al. Association between Foot Structure & Subsequent Lower-extremity & Ankle Injury in a Young and Active Military Population

4:10PM  5-3: Giangrande, Alessia et al. Effect of weight type and carrying mode on in-shoe plantar pressure magnitude and distribution

4:20PM  5-4: Clarke, Tim, et al. The effect of a 3-week functional hip strengthening programme on centre of pressure progression during barefoot walking

4:30PM  5-5: Miller, Erin, et al. Accuracy of self-reported foot strike patterns and running cadence

4:40PM  5-6: Helton, Gary L., et al. Association Between Shoe Selection and Lower Extremity Injuries in United States Military Academy Cadets
Assessment of foot structure, flexibility and function in USMA cadets

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Disclosures: Howard Hillstrom and Jinsup Song are consultants for JakTool LLC. All other authors declare no potential conflict of interest.

INTRODUCTION: Lower extremity musculoskeletal (MSK) injuries are common, complex, and costly problems especially among military personnel. Between 15% to 35% of men and 38% to 67% of women sustain at least one injury during Basic Combat Training, with 77% located in the pelvis and lower extremity. (Knapik JJ 2006) Existing literature has demonstrated an association between static foot structure and dynamic foot function, as well as between overuse injury and demographic characteristics. Previous studies have failed to provide a comprehensive biomechanical foot characteristics.

METHODS: In this study, foot structure, function, and arch height flexibility (AHF) were objectively measured in 1,090 incoming cadets (16.3% female, mean age of 18.5 years and BMI of 24.5 kg/m\textsuperscript{2}) of the United States Military Academy (USMA) at the start of their training. USMA IRB approved the protocol and a consent was obtained prior to data collection. Each foot was categorized by foot type (planus, rectus, and cavus) based on standing arch height index (AHI). Based on quintile distribution of AHF, each foot was identified as flexible, referent, or stiff. Five trials of dynamic planter pressure distribution was captured (emem-X at 100 Hz, novel gmbh, Munich) for each foot using a second-step protocol during self-selected comfortable pace of barefoot walking. Peak pressure and the Center of Pressure Excursion Index (CPEI %) were calculated for each foot. A Generalized Linear Model with an identity link function was used to examine the effects of race, gender, foot types, and AHF while accounting for potential dependence in bilateral data.

RESULTS: Planus and flexible feet independently demonstrated over-pronation, as measured by reduced CPEI. When comparing across race, Black participants showed a significantly lower AHI, a larger malleolar valgus index (MVI), and a higher prevalence of pes planus (91.7% versus 73.3% overall). However, Asian participants with increased AHF displayed over-pronation in gait. Females showed no significant difference in standing AHI and MVI but demonstrated a significantly greater AHF and a reduced CPEI than male participants. Results demonstrated that (1) flexible arches are more prevalent in pes planus and (2) those with flexible arch, rather than lowered arch, may be associated with over-pronation in gait.

DISCUSSION: Those participants with flexible arches demonstrated over-pronation, even in the absence of lowered arch, suggest the potential importance of arch flexibility on foot function. Improved understanding of the factors that are associated with aberrant foot function may identify subjects at risk of lower limb MSK injuries and guide treatment strategies.

SIGNIFICANCE/CLINICAL RELEVANCE: This was the first large scale investigation that comprehensively characterized biomechanical foot in a cohort of young at-risk individuals for lower limb musculoskeletal injuries. Long-term goal is to examine the relationship between these biomechanical features and injuries, ultimately to develop effective preventive measures.

References
Association between Foot Structure and Subsequent Lower-extremity and Ankle Injury in a Young and Active Military Population

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¹Keller Army Hospital and the United States Military Academy, West Point, NY, ²Hospital for Special Surgery, New York, NY, USA, ³Temple University School of Podiatric Medicine, Philadelphia, PA, ⁴New York College of Podiatric Medicine, New York, NY, ⁵Harvard Medical School, Boston, MA

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INTRODUCTION: Injuries to the lower-extremity and ankle are common among athletes and military personnel; however, little is known about how measures of foot structure are associated with lower-extremity and ankle injury risk in young and physically active populations. The purpose of this study was to examine the association between foot structure upon entry to military service and subsequent ankle and lower-extremity injury during one year of follow-up.

METHODS: To accomplish this objective we conducted a prospective cohort study at the United States Military Academy (USMA) at West Point. The study was approved by the IRB at our site prior to initiation. All incoming cadets at the USMA in the summer of 2013 were recruited to participate in this study. Study volunteers completed foot structural assessments within 3 days of arrival. Arch height and foot length were measured in sitting and standing positions, using a custom-made jig to calculate Arch Height Index to assess foot structure (planus, neutral, cavus). The primary outcomes of interest were 1) time from baseline until incident lower-extremity injury and 2) time from baseline until incident ankle injury during one year follow-up period. Univariate and multivariable Cox Proportional Hazards regression models were used to analyze the data. Kaplan-Meier survival estimates, hazard ratios (HR) and 95% confidence intervals (CIs) were calculated by foot structure.

RESULTS: Complete data for foot structure were available on 1090 subjects or 2180 feet (18.5 ±1.1 years, 1.76±0.80 m, 76.1±12.6 kg, and 24.5±2.96 kg/m²), of which 174 (16%) were female. In univariate models, subjects with rectus foot structure were at the greatest risk for incident lower-extremity injury followed by planus foot structure during the follow-up period. Individuals with cavus foot structure were 18% less likely (HR=0.82; 95% CI=0.57, 1.18) to sustain a lower-extremity injury during follow-up when compared to those with rectus foot structure (Figure 1A). Results were similar in multivariable models controlling for sex and BMI for both sitting and standing measures of foot structure. In univariate models, subjects with rectus foot structure were also at the greatest risk for incident ankle injury followed by planus foot structure during the follow-up period. Individuals with cavus foot structure were 52% less likely (HR=0.48; 95% CI=0.21, 1.12) to sustain an ankle injury during follow-up when compared to those with rectus foot structure (Figure 1B). Again, results were similar in multivariable models controlling for sex and BMI for both sitting and standing measures of foot structure.

SIGNIFICANCE/CLINICAL RELEVANCE: These data suggest that cavus foot structure may be associated with reduced lower-extremity and ankle injury risk. These findings are preliminary but consistent across lower-extremity injury outcomes. As a result, further study is needed to fully understand the potential association between foot structure and subsequent lower-extremity injury risk.

FIGURES/TABLES:

Figure 1A. Kaplan-Meier survival estimates for lower-extremity injury during follow-up by foot type measured in a standing position.

Figure 1B. Kaplan-Meier survival estimates for ankle injury during follow-up by foot type measured in a standing position.
Effect of weight type and carrying mode on in-shoe plantar pressure magnitude and distribution

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Disclosures: Alessia Giangrande (N), Alberto Leardini (N), Maurizio Ortolani (N), Silvia Tamarri (N), Claudio Belvedere (N), Giada Lullini (N), Lisa Berti (N), Paolo Caravaggi (N)

INTRODUCTION: It has been demonstrated that excessive workloads cause stress, pain, and musculoskeletal disorders in the back and lower limbs. For example, Andersen et al. [1] have reported on the association between workload and increased incidence of low back pain. Scarce is however the current literature on the association between weight-carrying and force distribution in different plantar regions. Aim of this study was to assess how in-shoe plantar load distribution is affected by weight type and by different carrying modes.

METHODS: 13 young healthy subjects were thus far recruited and examined for this study (age 32.6 ± 6.2 years, BMI 22.7 ± 2.8 kg/m²). A capacitive insole system (Pedar, Novel) was used to measure peak pressure (kPa), pressure-time integral (kPa*s), force-time integral (%BW*s), and maximum force (%BW) at rearfoot, midfoot, metatarsal head, toes and total foot while subjects were performing the following motor tasks: normal walking; fast walking; stair ascending, and stair descending. Subjects wore their personal trainers and a latex flat insole. Three weight types (4 – 8 - 12 kg) were carried by each subject in three carrying modes: inside a box against the chest (BX); divided in two hand held light bags (HB), and inside a backpack (BP). The weight type and carrying modes were randomized for each subject. In-shoe plantar pressure distribution when subjects performed the same tasks with no weight was used as control. For each subject, average plantar load parameters across a minimum of four steps (2 left, 2 right) were calculated in each foot region. The percent differences in maximum force at each plantar region between the three carrying modes and the control were determined. N on-parametric paired Friedman test and Tukey-Kramer post-hoc with Bonferroni correction, were used to assess statistical differences in pedobarographic parameters between the three carrying modes and the three weights.

RESULTS: The Friedman test revealed no significant differences in contact time between the three carrying modes, and between the three weights, in any motor task. The maximum force recorded at the total foot during normal walking was linearly associated to the weight type. For example, in BX, differences in maximum force at total foot were between 74 and 91 % of the carried weight. Significant increases in maximum force associated to weight types were also detected in most foot regions. Carrying modes affected plantar loading in all foot regions. For the 12 kg weight, maximum force significantly increased at rearfoot, toes and total foot in all carrying modes, and at midfoot (37 ± 29%) only in BX (figure 1).

DISCUSSION: In this study we aimed at assessing the effects of weight type and carrying mode on in-shoe plantar loading magnitude and distribution. Pressure insoles appear suitable to detect weight carried in daily motor tasks or activities. As expected, in simple motor tasks, the maximum force recorded by pressure insoles is consistent with the carried weight. Moreover, load distribution across foot regions depends on the carrying mode. The same weight carried against the chest affects plantar load distribution differently than carried in a backpack, and this could be related to different joint torques and muscle forces acting at the main foot joints. A larger sample size will be sought to improve the statistical power of the current results.

SIGNIFICANCE/CLINICAL RELEVANCE: Carrying mode should be considered when choosing the most appropriate footwear/orthotics and in the assessment of the mechanical loading acting on the lower limb. Pressure insoles could be used to estimate the weight type and the number of occurrences it is carried in standard working environments.

REFERENCES:

Figure 1. Percent differences in maximum force between each carrying mode (BX, HB and BP) and control at different foot regions when carrying the 12 kg weight. On the left, statistical significant differences in maximum force between carrying mode and control are indicated with the task acronym within the relevant plantar region.
The effect of a 3-week functional hip strengthening programme on centre of pressure progression during barefoot walking
Tim Clarke1,2, Helen R Branthwaite1, Andrew Greenhalgh3, Nachiappan Chockalingam4
1Staffordshire University Stoke-on-Trent UK. 2 SO2 Med Op. MOD, London UK. 3 University of Hertfordshire, UK.
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Disclosures: This work was a part of the PG dissertation completed by Tim Clarke; Helen Branthwaite and Nachiappan Chockalingam were involved in the design and supervised this study

INTRODUCTION: There is growing evidence that weakness or fatigue in the proximal stability muscles, especially gluteus medius (Gmed) and gluteus maximus (Gmax), can influence distal lower limb kinematics (Homan et al, 2013) and foot posture (Barwick et al, 2012) causing increased abnormal distal joint loading and risk of lower limb injuries (Chuter et al, 2012). Strengthening interventions targeted at the proximal stability muscles of the hip and their effect on foot posture and lower limb injury rate have not been investigated in detail (Snyder et al, 2009) and the two segments of the lower limb are often treated independently of each other. The aim of this study, therefore, was to investigate if a 3 week hip strengthening programme altered the centre of pressure progression in a group of military personnel referred for rehabilitation due to musculoskeletal injury.

METHODS: Ethical approval was granted from Ministry of Defence Research Ethics Committee. 38 male British Army personnel, referred for a 3-week intensive rehabilitation course with non-traumatic lower limb pathology, were recruited and randomly allocated to either a functional hip strengthening intervention, (n= 21, age 29 +/- 4.80, weight 83.1 +/- 13.4 kg), or control group, (n = 17, age 28 +/- 5.36, weight 85.2 +/- 10.4 kg). Each group completed the standard rehabilitation regime, with the exercise group adding specific hip exercises (Gmed and Gmax) over a 3 week programme. Foot pressure measurement system (RS-scan® system, Belgium) was used to measure the COP coordinates for progression angle at admission and discharge. Four sub-phases of the stance phase of gait were calculated (Chui 2012), including (ICP - heel strike 0% of stance phase), Forefoot contact phase (FFCP - forefoot loading 35% of stance phase), Foot flat phase (FFP - mid-stance 50% of stance phase), Forefoot push off phase (FFPOP - toe off 100% of stance phase). Pre and post data was compared statistically with a repeated measures ANOVA (p≤0.05).

RESULTS: There was significant effects of centre of pressure progression observed (p 0.017) between the 2 groups at discharge for ICP. The hip strengthening group moved laterally (1.83° - 0.96° +/- 5.68) whilst the control group remained in a medial position (1.56°+2.8° +/- 5.94). At forefoot contact phase significant differences for both groups was also observed (p 0.035) after the 3 week period (-4.09° - 4.26° +/-2.46), however, the exercise group did not appear to be more lateral in the progression angle than the control.

DISCUSSION: Hip rehabilitation of Gmed and Gmax alters centre of pressure progression angle to be more lateral than those who do not specifically exercise this muscle group. This is in keeping with the action of the external rotation of hip muscle. Additionally, an intensive rehabilitation course, as used in the military rehabilitation care pathway for injured soldiers, significantly improves the lateral position of the centre of pressure at forefoot loading, indicating that foot function can be altered through rehabilitation. These findings support previous work (Snyder et al 2009) that a functional closed chain resistance hip rotation exercises may be effective in controlling the rearfoot in the early contact phases of walking gait.

SIGNIFICANCE/CLINICAL RELEVANCE: Implementation of a hip strengthening programme alters the centre of pressure progression at initial contact and forefoot contact phases and should be considered as part of a functional rehabilitation plan for foot function.

REFERENCES:
Chui et al 2013 Gait and Posture 37.p.408–12
Barwick et al 2012 The Foot 22.p.224-231
Accuracy of Self-Reported Foot Strike Pattern and Running Cadence

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INTRODUCTION: Running is the primary aerobic exercise method for our nation’s military. Runners in the military sustain injuries at approximately 30-37% annual incidence.1,2 Since the publication of Born to Run3 and the advent of minimalist footwear and barefoot running, many runners have attempted to change from a rearfoot strike pattern to a non-rearfoot strike. Cadence manipulation has been shown to effect negative joint work4 and plantar pressure.5 In an effort to further understand running related injury development and prevention, Goss et al. previously observed 64% accuracy of self-reported foot strike pattern dichotomized in a small laboratory sample.6

METHODS: Participants ran at a self-selected speed for 5 minutes. Data were collected in the final minute of a 5-minute run. Two-dimensional sagittal plane video data were collected using a Casio EXilim camera at 240 Hz on 529 subjects aged 18-60. These video data were collapsed and re-analyzed from 4 previous studies with local institutional review board approval.

RESULTS: The mean age of participants was 24.8 +/- 7.9. Mean height was 171.4 +/- 10 cm and mean weight was 76.3 +/- 11.9 kg. Self-reported average weekly running distance was 10.6 +/- 10.4 miles. Out of 529 participants (357 males, 172 females), only 57 reported they were familiar with step rate (10.8%). Only 49% of those 57 were able to accurately report their cadence within 10 steps per minute. A subsample of 129 runners were asked if they could describe their foot strike pattern prior to running data collection on the treadmill. Of the 129 runners, 99 (77%) could correctly describe their foot strike pattern. Out of the 30 participants that incorrectly identified their foot strike pattern, 80% of them (24) reported a non-rearfoot strike and then demonstrated a rearfoot strike pattern.

DISCUSSION: Most runners in these studies were not familiar with step rate or cadence. Of those who stated they were familiar with cadence, about ½ were able to estimate their cadence within 10 steps per minute. More runners here at West Point were able to accurately report their own foot strike pattern than previous studies have reported. For those incorrect foot strike classifications, most of them demonstrated a rearfoot strike pattern while reporting a non-rearfoot strike pattern.

SIGNIFICANCE/CLINICAL RELEVANCE: Runners may be more familiar with foot strike pattern than steps per minute or cadence.

REFERENCES:
Association Between Shoe Selection and Lower Extremity Injuries in United States Military Academy Cadets.

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DISCLOSURES: I affirm that I have no financial affiliation (including research funding) or involvement with any commercial organization that has a direct financial interest in any matter included in this manuscript.

INTRODUCTION: Running provides an inexpensive form of moderate to vigorous exercise in order to improve cardiac, metabolic, and mental health. However, running related overuse injuries are very common among recreational runners with the reported annual injury rate ranging from 39 – 85%. Soldiers within the Army are no exception where running accounts for 50% of exercise and sport related injuries. As the primary mode of physical endurance training within the civilian and military population, it is important to minimize running related injuries. Large prospective trials investigating injury risk associated with running in minimalist shoe wear are rather scarce and often contradictory in nature. There is currently no definition of the minimalist shoe vs. a conventional shoe addressing torsional shoe stiffness. Therefore, the purpose of this study was to investigate the relationship between lower extremity musculoskeletal injury and shoes with varying levels of torsional shoe stiffness and heel height.

METHODS: The study included 1025 of 1308 incoming United States Military Academy (USMA) cadets. Subjects were recruited after approval of the study and consent form from the Keller Army Community Hospital Institutional Review Board. All subjects’ shoe length and stiffness were recorded using a novel device, the shoe stiffness in torsion device (SySTM), which we manufactured for the purposes of this study. Shoe heel height was also recorded. Thereafter, lower extremity injuries sustained over nine weeks during cadet basic training were documented and Kaplan-Meier survival curves were estimated with time to incident of lower extremity injury as the primary outcome by level of the independent predictor variables, i.e. torsional shoe stiffness, heel height. Risk factors and known or potential covariates were carried forward into a multivariable Cox Proportional Hazards Regression Model.

RESULTS: Cadets wearing shoes with moderate lateral torsional stiffness were 49% less likely to incur any type of injury and 52% less likely to incur an overuse injury than cadets wearing shoes with minimal lateral torsional stiffness. Risk of injury was similar among cadets wearing shoes with minimal and extreme lateral torsional stiffness.

DISCUSSION: The results of this study indicate that a shoe with mild to moderate lateral torsional stiffness and mild to moderate heel height may be most appropriate in efforts to reduce the risk of lower extremity injuries during training in young physically active athletic and military populations. Furthermore, shoes with minimal lateral torsional stiffness and heel height should be discouraged within these populations due to the significantly increased risk of injury. One limitation of the study was the 19% dropout rate due to incomplete data. Also, we did not account for hysteresis when assessing shoe torsional stiffness. Lastly, the torque wrench used to rotate the shoe to the 30 degree offset in order to assess stiffness was very sensitive and constantly changing potentially introducing error into the stiffness measures taken.

SIGNIFICANCE/CLINICAL RELEVANCE: The current study provides guidance on key characteristics of shoe wear important in minimizing injury risk during physical fitness training. The SySTM provides for a reliable, portable, and inexpensive method of capturing footwear torsional stiffness and heel height objectively in the clinic setting.

References:

Session 6: Tendonopathies and Ligament Injuries

Co-Moderators: Irene Davies, PT, PhD (Spaulding Rehab, Harvard University) and Don Goss, PT, PhD (Keller Army Hospital and USMA)

5:00PM 6-1: Abdeen, Rawan, et al. Structural Characteristic of Selected Ankle Structures in Healthy and Chronic Ankle Instability

5:10PM 6-2: Eerdekens, Maarten, et al. Multi-segment Foot Kinetics in Chronic Ankle Instability & Healthy Subjects During Barefoot Running with Rearfoot striking Pattern

5:20PM 6-3: Mettler, Jeff H., et al. Effects of Speed, Grade, and Shoe Stiffness on Plantar Fascia Strain During Walking


5:40PM 6-5: Teoh, Jee Chin, et al. The influence of prolonged weight bearing physical activities on plantar tissue behavior

5:50PM 6-6: Ward, Erin D., et al. The Effects of Posterior Tibial Tendon Dysfunction on Foot Function
Structural Characteristic of Selected Ankle Structures in Healthy and Chronic Ankle Instability

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Introduction: Lateral ankle sprains are one of the most common musculo-skeletal injuries encountered both in clinical practice and in the sporting community. Even though it is understood that scar tissue forms in the ligament following a sprain, little is known about the morphologic alteration in the ligaments and tendons around the ankle after the injury. The objectives of the study were to (1) characterise and compare selected ankle structures in healthy and injured cohorts and (2) quantitatively measure the echogenicity of the anterior talofibular ligament (ATFL) in healthy and injured ankles.

Methods: 40 healthy participants and 30 participants with chronic ankle instability (CAI) were recruited from a university population. A musculoskeletal sonographic image of ATFL, calcenofibular ligament (CFL), peroneal tendons, tibialis posterior tendon and Achilles tendons was obtained using a Venue 40 ultrasound system with a 5-13 MHz transducer. Thickness, length, and cross sectional area (CSA) of each structure was measured for each ankle in a neutral and tension position. The analysis of ATFL echogenicity was implemented in Matlab and involved: image pre-processing, clustering segmentation of the image, and feature extraction. A series of independent T-test statistical was compared structural differences between healthy and injured ankles.

Results: The ATFL was significantly longer by 10% in CAI compared to healthy ankles with p-value of 0.04 and effect size Cohen’s d =1.53. The thickness of ATFL and CFL were significantly thicker in injured compared to healthy participants by 57% (p-value= 0, and Cohen’s d =2.688) and 10% (p-value= 0.003 and Cohen’s d = 0.8) respectively. The intensity and contrast of the injured ATFL was statistically lower than normal ligament with p-value <0.01 and Cohen’s d= 0.726. There were no statistically significant differences in the peroneal tendons, tibialis posterior tendon and Achilles tendon between healthy and injured groups.

Discussion: Lengthening of the ATFL will likely lead to reduced constraint on the talus relative to the fibula and tibia, allowing it to translate anteriorly or rotate medially relative the fibula (Croy et al., 2013). The thickness of ATFL was increased but by far more than other observed (15% in Liu et al., 2015). Differences in remodelled ligament matrix might explain a difference between healthy and CAI. These include changes in the types of collagen, decreased collagen crosslinks, increased vascularity, abnormal innervation, and presence of inflammatory cell pockets (Hauser et al., 2013). The scar tissue of the remodelled ATFL in the CAI group would explain the lower intensity data and be associated with weakness in the ligament strength, which is believed to persist for years after the initial injury (Hauser & Dolan, 2011). This may result in abnormal biomechanical, biochemical and ultrastructural properties and risk of further injury (Hauser & Dolan, 2011).

Clinical relevance: Understanding the structural differences between healthy and injured ankles and having a framework to use ultrasound to evaluate ligament tissue quality (as well as morphology) may improve objectivity in the clinical assessment of lateral ankle injuries. This may lead to clinical insights in terms of injury classification and treatments.

![Input images](image1.png) ![De-specked image](image2.png) ![Segmented image](image3.png)

Figure 1: Developed framework for ATFL segmentation

References:

Multi-segment foot kinetics in Chronic Ankle Instability and healthy subjects during barefoot running with a rearfoot-striking pattern

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Disclosures: None to declare

INTRODUCTION: Up to 32-74% of lateral ankle sprains (LAS) can result in chronic ankle instability (CAI) (1). Biomechanical abnormalities are a potential intrinsic risk factor in the occurrence of LAS. Recent evidence and personal clinical expertise highlighted that foot-striking patterns during running potentially impact lower limb kinetics and may result in running-related injuries, such as LAS. A review of literature exposed a lack of multi-segment kinetic foot measurements in a CAI population. Therefore, we aimed to investigate differences in joint kinetics of the rearfoot, Chopart, Lisfranc and Hallux segments between healthy and CAI subjects while running barefoot with a rearfoot-striking pattern.

METHODS: Seven healthy (4 men/ 3 women) and twelve CAI (5 men/ 7 women) participants signed the informed consent and the ethical committee of the University Hospital of Leuven approved the protocol. Reflective markers were placed on foot and lower limb anatomical landmarks according to the Rizzoli Foot Model (RFM) (2). All participants performed a running analysis at a speed of 3.3 m/s (+10%) in a clinical motion analysis laboratory that is instrumented with a 3D motion analysis system (Vicon®, 100 Hz) and a plantar pressure platform place (RSscan international, 200 Hz) on top of a force platform (AMTI®, 1000Hz). The kinetic computation combines the marker position, ground reaction force, and pedobarograph data. The center of pressure and resultant ground reaction force were distributed over each segment of the RFM (Ankle=Calcaneus-Shank, Chopart=Calcaneus-Midfoot, Lisfranc=Midfoot-Metatarsus and hallux) using the proportionality scheme validated by Saraswat et al. (3). Inertial parameter calculations of foot segments were based on the mass of the segments and on their geometric solids, whereas the mass of the foot was distributed at 30/30/30/10 percent. An inverse dynamic analysis program written in Matlab computed joint moments and powers starting from the hallux segment and progressing proximally using Newton-Euler equations, similar to a recent methodology published by Deschamps et al (4).

RESULTS: CAI subjects have maximum and minimum peak moments and powers that are similar to healthy subjects, except for a significant difference in their rearfoot minimum peak power in frontal plane, where CAI subjects absorb 0.33 Watt/kg in comparison to 0.56 Watt/kg in healthy subjects (table 1).

DISCUSSION: After visual inspection of the kinetic waveform data, the significant difference in rearfoot absorption in the frontal plane was found during the late stance phase. This phenomenon is in line with the study of Santilli et al. (2005), which highlighted a decreased activity of the Peroneus Longus (PL) during late stance phase in subjects with CAI. The impaired function of PL may result in lower rearfoot power absorptions in the frontal plane during the late stance phase in CAI subjects.

SIGNIFICANCE/CLINICAL RELEVANCE: A first step has been taken towards understanding and unraveling the biomechanical and kinetic characteristics of multiple foot segments during barefoot running in a CAI population, aiding in the optimization of clinical-decision making and interventional strategies.


ACKNOWLEDGEMENTS: None

Table 1: Kinetic comparisons of peak (maximum) and minimum (minimum) moments and powers

<table>
<thead>
<tr>
<th>Segment</th>
<th>Control</th>
<th>CAI</th>
<th>P*</th>
<th>Control</th>
<th>CAI</th>
<th>P*</th>
<th>Control</th>
<th>CAI</th>
<th>P*</th>
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</thead>
<tbody>
<tr>
<td>Sag</td>
<td>2.81±0.44</td>
<td>2.64±0.29</td>
<td>0.54</td>
<td>-0.41(0.19)</td>
<td>-0.39(0.13)</td>
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<td>1.39(1.27)</td>
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<td>Ankle</td>
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<td>0.21±0.14</td>
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<td>-0.16(0.11)</td>
<td>0.24</td>
<td>0.21(0.2)</td>
<td>0.22(0.11)</td>
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<tr>
<td>Trans</td>
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<td>0.21±0.19</td>
<td>0.37</td>
<td>-0.41(0.25)</td>
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<td>0.79</td>
<td>0.29(0.17)</td>
<td>0.34±0.25</td>
<td>0.96</td>
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<tr>
<td>Sag</td>
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<td>1.92±0.21</td>
<td>0.25</td>
<td>-0.03(0.01)</td>
<td>-0.02(0.19)</td>
<td>0.24</td>
<td>3.95±1.02</td>
<td>3.78±1.39</td>
<td>0.81</td>
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<td>0.18±0.12</td>
<td>0.72</td>
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<td>-0.21(0.13)</td>
<td>0.93</td>
<td>0.09(0.11)</td>
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<td>0.96±0.19</td>
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<td>0.68</td>
<td>1.19(0.46)</td>
<td>1.31±0.81</td>
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<tr>
<td>Lisfranc</td>
<td>0.16±0.09</td>
<td>0.19±0.19</td>
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<td>-0.13(0.17)</td>
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<tr>
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<td>-0.34(0.15)</td>
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<td>0.28(0.25)</td>
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<td>0.00(0.00)</td>
<td>0.29</td>
<td>0.05±0.04</td>
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<td>0.11</td>
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<tr>
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<td>0.051±0.048</td>
<td>0.72</td>
<td>-0.04(0.03)</td>
<td>-0.038(0.041)</td>
<td>0.79</td>
<td>0.05±0.04</td>
<td>0.03±0.023</td>
<td>0.35</td>
</tr>
</tbody>
</table>

* P-value outcomes of the Mann-Whitney U test (P<0.05 = significant difference between means). Italic: not significant; Bold: significant. CAI: Chronic Ankle Instability; Sag: sagittal plane; Front: frontal plane; Trans: transverse plane; SD: standard deviation
Effects of Speed, Grade, and Shoe Stiffness on Plantar Fascia Strain During Walking

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Disclosures: The authors have no conflicts of interest to disclose.

INTRODUCTION: Plantar fasciitis afflicts approximately 2 million people each year¹ and is thought to occur due to repetitive tensile overload of the plantar fascia. Both toe dorsiflexion due to the windlass mechanism²,³ and arch collapse⁴ contribute to plantar fascia strain (PFS). Arch collapse is caused by ground reaction forces as well as Achilles tendon forces¹,². Walking and running at various treadmill inclines results in increased Achilles tendon strain⁵, suggesting that gait speed and incline may affect PFS. In addition, the windlass effect may be reduced by shoe selection due to decreased toe dorsiflexion⁶. Therefore, the purpose of the study was to use a musculoskeletal model to determine the effects of speed, grade, and shoe stiffness on plantar fascia strain.

METHODS: 15 healthy subjects (age: 25.8±8.2yrs; height: 1.8±0.1m; mass: 72.2±16.2 kg) completed the study while walking on a treadmill. Subjects were fitted with marker triads glued over the right calcaneus, navicular, and hallux via holes cut in the shoe. Experimental conditions included all combinations of 2 speeds (preferred walking speed (PS) and 20% greater than preferred walking speed (PS+20%)), 3 inclines (0°, 5°, and 10°), and 2 shoe forefoot bending stiffnesses (stiff and flexible midsole). A four segment (rearfoot, forefoot, hallux, sesamoid) model of the foot (Figure 1) was used to deform the foot according to triad movement. Translation of the hallux and translation/rotation of the sesamoid were visually matched to the contour of the 1st metatarsal head and then calculated as a function of the first metatarsophalangeal (MTP) joint angle. PFS was then estimated as the change in length of the plantar fascia from normal standing and was due to both the windlass effect of the MTP joint and collapse of the arch (rearfoot-forefoot angle). A repeated measures ANOVA was used to determine statistical significance (α=0.05) with trend analysis used to compare the incline effect.

RESULTS: There were no significant interactions. The main effect of stiffness was not significant (stiff = 1.86 ± 2.85%, flexible = 1.92 ± 2.51%; p = 0.69) (Figure 2C), but speed was significant (PS = 1.81 ± 2.58%, PS+20% = 1.96 ± 2.74%; p=0.016) with greater strain in the PS+20% condition (Figure 2A). In addition, there was a linear trend to the incline effect (Figure 2B), with greater strains in the higher inclines (0° = 1.59 ± 2.68%, 5° = 1.86 ± 2.71%, 10° = 2.22 ± 2.66%; p=0.014).

DISCUSSION: As the MTP is extended, it tightens the plantar fascia and helps to re-establish the arch which has collapsed during mid-stance. The cross-correlation between the MTP angle and the rearfoot-forefoot angle was -0.58 over all conditions in this study. Changes in PFS appeared to be more greatly influenced by the MTP joint (r=0.57) than by the rearfoot-forefoot angle (r=-0.25).

SIGNIFICANCE/CLINICAL RELEVANCE: This study was conducted to better understand causative agents associated with increased tension in the plantar fascia, which may lead to the development of plantar fasciitis. Clinically, the results can be applied when recommending exercise or rehabilitating patients with plantar fasciitis.

REFERENCES:
Histological and biomechanical study of tendinopathy induced by serial injections of collagenase: a new experimental model in the Achilles tendon of rabbits.

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NOTHING TO DISCLOSE

ABSTRACT

PURPOSE: To compare biomechanical and histological findings of a new animal model of Achilles tendinopathy, induced by serial low-dose injections of bacterial collagenase, to the traditionally used single high-dose injection model and to controls.

METHODS: Forty-five rabbits were randomly assigned to three groups. Low-dose Group (LD) (n=18) underwent three serial low-dose (0.1mg) injections of collagenase in both Achilles tendons, High-dose Group (HD) (n=18) underwent single high-dose injection (0.3mg) and Control Group (C) (n=9) served control. Injections were separated by 2 weeks. Six rabbits were sacrificed from each experimental group (LD and HD) at 10, 12 and 16 weeks. Animals in Group C were sacrificed after 16 weeks. Histological and biomechanical properties were compared. P-values <0.05 were considered significant.

RESULTS: At 10 weeks, HD Group animals, when compared to LD Group, showed increased histological score and tendon cross sectional area, with impaired biomechanical properties, demonstrating an acute onset of the disease. At 12 weeks, histological score was higher in the LD Group, with similar biomechanical findings between the groups. After 16 weeks, histological score was significantly increased for both LD (11.8±2.28) and HD Groups (5.6±2.51), when compared to controls (2±0.76). Regarding biomechanical findings, groups differed in cross sectional area of the tendon, Young’s modulus, yield stress and ultimate tensile strength, with more pronounced findings in animals from Group A.

CONCLUSIONS: The animal model of Achilles tendinopathy induced by serial injections of low-dose collagenase showed more pronounced histological and biomechanical findings after 16 weeks, reproducing better the progressive characteristic of the chronic tendinopathy in humans.

CLINICAL RELEVANCE: The development of a proper animal model of Achilles tendinopathy is essential in the understanding of the pathology and search for better treatment options.
The influence of prolonged weight bearing physical activities on plantar tissue behavior

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INTRODUCTION: Activity level influences amount of accumulated mechanical trauma to the foot [1]. As high as 39% of marathon runner were reported to suffer the epidermal split [2]. The constant pounding on hard surfaces and shearing forces cause the localized injury frequently found at the foot [2]. Nonetheless, majority of investigations of plantar tissues focus on diseased population. The purpose of this study is to investigate the relationship between the localised cumulative stress due to prolonged weight bearing activities and plantar tissue stiffness in healthy individuals with the use of in-vivo indentation technique. It is hypothesized that excessive and repeated mechanical stress will alter the tissue properties over time.

METHODS: Subjects were recruited from engineering (control) and ballet schools (15 each). IRB approval and informed consent were obtained. The indenter probe had a 5mm diameter hemispherical tipped stylus which was driven by a stepper motor. The stylus was connected to a miniature compression load cell. 2nd sub metatarsal head (MTH) and heel pad were indented with maximum deformation depth of 2.65mm and a constant speed of 12.3mm/s. Tissue behaviour was characterized via stiffness constant (K). Barefoot plantar pressure distribution was recorded in standing mode.

RESULTS: As shown in Fig. 1a, plantar tissue of ballet students were found to be significantly higher as compared to the control at the forefoot region. The finding was the opposite the at the heel, with higher stiffness observed in control group. Similar results were obtained from plantar pressure measurement (Fig. 1b).

DISCUSSION: The occupational behavior of ballet dancers with anteriorly shifted COP during standing resulted in higher mechanical stress on the forefoot region. Apart from the process of tissue glycation due to ageing and metabolic disease, high frequency and large magnitude of localized stresses due to high impact physical activities may also stiffen the plantar tissue excessively [3].

SIGNIFICANCE/CLINICAL RELEVANCE: The relationship between localized mechanical stress and tissue stiffness was established. Excessive and repeated stresses stiffen the planter tissue not only in diseased population, but also healthy individuals who engages in high impact sports activities.

REFERENCES:

FIGURES AND TABLES:

Figure 1. (A) Plantar tissue stiffness and (B) plantar pressure distribution at the heel and forefoot regions of control and ballet subjects.
The Effects of Posterior Tibial Tendon Dysfunction on Foot Function
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Disclosures: The authors have no conflicts of interest to disclose.

INTRODUCTION
Numerous clinical studies and research articles have implicated the posterior tibial tendon¹-² as being an integral component in the development of flexible flatfoot deformity. Therefore, it was the aim of this study to determine if loss of function of the posterior tibial tendon does indeed significantly increase calcaneal eversion²-³ as well as impart increased strains upon the plantar fascia⁴and the spring ligament¹,⁴-⁵.

METHODS
An intra-medullary rod was driven into the tibia of 4 fresh cadaveric specimens connecting the specimen to a dynamic gait simulator. Eight extrinsic muscles crossing the ankle were connected to load cells in series with motors to simulate muscle activity. Microstrain⁶ micro strain gauges were placed in the medial and lateral bands of the plantar fascia as well as in the superomedial portion of the spring ligament. Two 0.062 inch k-wires were driven into the posterior calcaneus and two into the posterior tibia. A marker was attached to each k-wire for collecting kinematics. Each foot was walked 5 trials for 2 conditions (normal and posterior tibial tendon dysfunction (PTTD)). The PTTD condition was created by detaching the posterior tibial tendon from the simulator. Data were analysed using Cohen effect sizes which are 0.20, 0.50 and 0.80 representing small, medium and large differences respectively.

RESULTS
An increase in calcaneal eversion from forefoot loading to heel off and decreased inversion from heel off to toe off was observed during simulated PTTD with an effect size of 0.990 and 0.984 respectively (figure 1). The strain in the superomedial portion of the spring ligament was also, observed to increase in PTTD with a large effect size of 0.982 (figure 2). The strain in the lateral band of the plantar fascia increased in 3 feet with an effect size of 0.412. However, the strain in the medial band of the plantar fascia only increased in 1 foot with an effect size of 0.076.

DISCUSSION
This study confirmed that the function of the posterior tibial tendon is essential to the function of the foot. Kinematic changes associated with PTTD were confirmed with an increase in calcaneal eversion, as well as the inability of the calcaneus to invert in late midstance (figure 1). The study also demonstrates, for the first time, that strains increase in the superomedial spring ligament early during gait with PTTD and is consistent with injuries observed to the spring ligament on MRI of PTTD patients. No data previously existed for plantar fascial strain and PTTD. The authors hypothesized there would be an increase in plantar fascial strain due to abnormal kinematics. However, there was only a small increase in plantar fascial strain which appears to be consistent with MRI results for PTTD patients⁵. The study is limited by a small number of specimens.

SIGNIFICANCE/CLINICAL RELEVANCE
The study confirmed both kinematic data from gait studies on patient’s with PTTD as well as what is observed clinically with MRI results from PTTD patients.

REFERENCES

Figure 1 –Calcaneal eversion

Figure 2-Spring ligament change in strain
Session 7: Foot Type I

Co-Moderators: Jinsup Song, PDM, PhD (TUSPM) and Robert Turner, PT, DPT (HSS)

9:00AM 7-1: Fields, Cheryl, et al. Foot Structure, Function, and Flexibility in MS Patients


9:20AM 7-3: Krzak, Joseph J., et al. Foot posture is associated with plantar pressure during gait: a comparison of normal, planus and cavus feet

9:30AM 7-4: Olsen, Mark T., et al. Static Foot Structure May Predict Midfoot Mechanics


9:50PM 7-6: Roberts, Lauren, et al. WBCT Hindfoot Alignment of Adult Acquired Flatfoot Deformity: A Comparison of Clinical Assessment and Weightbearing ConeBeam CT Examinations

10:00AM 7-7: Michael Rainbow, et al. In Vivo Intrinsic Foot Bone Motion Measured with Bead Tracked Biplanar Videoradiography
INTRODUCTION: Multiple Sclerosis (MS) is a progressive inflammatory neurologic condition common in young adults with a high prevalence of unsteady gait and balance resulting in frequent falls. Since the foot is the first structure to interface with the ground and help support our mass during gait and posture and many of these patients have different degrees of drop foot, our goal of this cross-sectional pilot study was to characterize foot structure, function, and flexibility in those suffering from MS.

METHODS: The study population included 53 MS patients and 19 healthy controls each of whom underwent four sets of measurements. Foot structure was measured using Arch Height Index (AHI) device (JAK Tool, Cranbury, NJ) and participants were classified into planus, rectus, or cavus feet. Function was assessed by measuring the center of pressure excursion index (CPEI) during gait at one’s self-selected walking speed using a plantar pressure measuring device (emed-X, Novel electronics, St Paul, Mn.). Secondary function analyses were performed on masked regions of the maximum plantar loading (ie pressure, force) throughout stance phase plot. Flexibility was assessed by calculating arch height flexibility (AHF) from sitting and standing AHI measurements. All participants had a timed 25 foot walk test.

RESULTS: MS patients (x=6.8s, s=4.98s) required significantly (p=0.0036) longer times to walk 25 feet than controls (x=4.53s, s=4.45s). Foot structure was planus in arch height and not significantly different between MS and control subjects. MS subjects differed in foot function as evidenced by a greater pressure time integral (PTI), force time integral (FTI), contact area, and maximum force beneath the medial arch compared with controls during self-selected walking speed. MS subjects had a lower pronation supination index in midstance and early midstance indicative of over-pronation. In addition the MS subjects had lower maximum force and peak pressure under the 2nd metatarsal head compared with controls during gait. Although the MS subjects exhibited greater AHF this difference was not significant.

DISCUSSION: The findings support the postulate that MS patients have a planus foot structure that exhibit more over-pronation during gait than controls. MS subjects had higher 1st met head forces and pressures and lower 2nd met head forces and pressures than control subjects. Controls in this study had a lower 1st met head pressure and higher 2nd suggesting a hypermobile 1st ray as was seen in Hillstrom et al (2013). MS subjects had a different distribution with 1st met head forces and pressures similar to that of the 2nd met head. This unique pressure distribution in MS subjects may be a compensation for balance and proprioception deficits. By keeping the first metatarsal head firmly on the ground the MS patients may improve their balance and posture but this theory would need further testing.

SIGNIFICANCE/CLINICAL RELEVANCE: Understanding the differences in foot structure, function, and flexibility could enhance the therapeutic strategies for decreasing the risk of falls in MS patients. The more medially directed loading associated with the over-pronating foot may pre-dispose the MS patient to secondary pathologies (e.g. hallux rigidus, hallux valgus, posterior tibial dysfunction, and Achilles tendonitis).

REFERENCES:
Hindfoot Alignment in Stage II Adult-Acquired Flatfoot Deformity: Can Clinical Evaluation Predict Radiographic Measurements?

Cesar de Cesar Netto¹; Grace Kunas¹; Anca Marinescu¹; Dylan S Soukup¹, Lauren Roberts¹; Scott Ellis¹
¹Hospital for Special Surgery, NY, US

NOTHING TO DISCLOSE

Introduction: Previous work has demonstrated that the amount of radiographic hindfoot correction required at the time of adult acquired flatfoot deformity (AAFD) surgical treatment can be predicted by the amount of radiographic deformity present before surgery. Successful outcomes after reconstruction are closely correlated with hindfoot valgus correction. However, it is not clear if differences exist between clinical and radiographic assessment of hindfoot valgus. The purpose of this study was to evaluate the correlation between radiographic and clinical evaluation of hindfoot alignment in patients with stage II AAFD.

Methods: Twenty-nine patients (30 feet) with stage II AAFD, 17 men and 12 women, mean age of 51 (range, 20 to 71) years, were prospectively recruited. In a controlled and standardized fashion, bilateral weightbearing radiographic hindfoot alignment views were taken. Radiographic parameters were measured by two blinded and independent readers: hindfoot alignment angle (HAA) and hindfoot moment arm (HMA). Clinical photographs of hindfoot alignment were taken, in three different vertical camera angulations (0, 20 and 40 degrees). Pictures were assessed by the same readers for standing tibiocalcaneal angle (STCA) and resting calcaneal stance position (RCSP). Intra- and interobserver reliability were assessed by Pearson/Spearman's and intraclass correlation coefficient (ICC), respectively. Relationship between clinical and radiographic hindfoot alignment was evaluated by a linear regression model. Comparison between the different angles (RCSP, STCA and HAA) was performed using Wilcoxon rank sum test. P-values of less than 0.05 were considered significant.

Results: We found overall almost perfect intra- (range, 0.91-0.99) and interobserver reliability (range, 0.74-0.98) for all measures. Mean value and confidence interval (CI) for RCSP and STCA were 10.78 degrees (CI: 10.09-11.47) and 12.55 degrees (CI: 11.71-13.40), respectively. The position of the camera did not influence readings of clinical alignment (p>.05). The mean HMA was 18.74mm (CI: 16.34-21.14mm) and the mean HAA was 23.54 degrees (CI: 21.08-25.99). Clinical and radiographic hindfoot alignment were found to significantly correlate (p<0.05). However, the radiographic hindfoot alignment (HAA) demonstrated increased valgus when compared to both clinical alignment measurements, with a mean difference of 12.76 degrees from the RCSP (CI: 10.99-14.53, p<0.0001) and 10.98 degrees from the STCA (CI: 9.22-12.76, p<0.0001).

Conclusion: We found significant correlation between radiographic and clinical hindfoot alignment in patients with stage II AAFD. However, radiographic measurements of hindfoot alignment angle demonstrated significantly more pronounced valgus alignment than the clinical evaluation.

Clinical Relevance: The results of our study suggest that clinical evaluation of hindfoot alignment in patients with AAFD potentially underestimates the bony valgus deformity. One should consider these findings when using clinical evaluation in the surgical treatment algorithm for flatfoot patients.
Foot posture is associated with plantar pressure during gait: a comparison of normal, planus and cavus feet

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Disclosures: the authors have none to declare.

INTRODUCTION: The aim of this study was to compare plantar pressure between healthy individuals with normal, planus or cavus feet.

METHODS: Ninety-two healthy volunteers (aged 18 to 45) were classified as either normal (n=35), pes planus (n=31) or pes cavus (n=26) based on the Foot Posture Index, Arch Index and normalised navicular height truncated. Barefoot walking trials were conducted using an emed®-x400 plantar pressure system (Novel GmbH, Munich, Germany). A 10 region mask was used that included the heel, midfoot, 1st, 2nd, 3rd, 4th and 5th metatarsophalangeal joints, hallux, 2nd toe, and the 3rd, 4th and 5th toes. Peak pressure, pressure-time integral, maximum force, force-time integral and contact area were calculated for each region. To test for differences between groups, a one-way analysis of variance (ANOVA) was performed with significance level set at <0.05. Post-hoc comparisons of the mean differences (MD) between groups with Bonferroni adjustments were applied to all ANOVAs. Confidence intervals (CI) and effect sizes (ES) using Cohen’s d were calculated for all significant mean differences.

RESULTS: Overall, the largest differences were between the planus and cavus foot groups in forefoot pressure and force. In particular, peak pressures at the 4th and 5th metatarsophalangeal joints in the planus foot group were lower compared to the normal and cavus foot groups, and displayed the largest effect sizes (Figure 1). The cavus group demonstrated higher peak plantar pressure in the heel compared the planus group, and in the 1st MTPJ compared to both the normal and planus groups.

DISCUSSION: The findings from this study indicate that the largest difference between foot posture groups, both in the number of significant post-hoc comparisons and the magnitude of effect size, was between planus and cavus feet, mostly in the lateral and medial regions of the forefoot.

SIGNIFICANCE/CLINICAL RELEVANCE: This study confirms that foot posture does influence plantar pressures, and that each foot posture classification displays unique plantar pressure characteristics.

ACKNOWLEDGEMENTS: The data for this study was collected by AKB during his doctoral studies, which were funded by the Australian Government. In addition, HBM is currently a National Health and Medical Research Council Senior Research Fellow (ID: 1020925).

Figure 1. Peak pressure diagrams for all foot posture groups (top) and differences in peak pressure for all foot posture group comparisons (bottom). For the foot posture comparison diagrams, red indicates greater peak plantar pressure values and blue indicates lesser peak plantar pressure values.
Static Foot Structure May Predict Midfoot Mechanics

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INTRODUCTION: Clinical interventions for foot injury rehabilitation and prevention are often prescribed based on static measures of foot structure. However, this convention merits further investigation as the static-dynamic relationship has only been explored in walking and running and remains unclear in the literature. As such, a drop landing task may be a valuable mechanism to apply a greater stress to the foot to uncover a static-dynamic relationship. Therefore, the primary aim of this study was to explore the relationship between static foot structure and dynamic midfoot kinematics and kinetics during a barefoot single-leg landing.

METHODS: Forty-eight females (age = 20.4 ± 1.8 yr, height = 1.6 ± 0.06 m, weight = 57.3 ± 5.5 kg) signed an informed consent approved by the institutional review board before participating in this cross-sectional study. Static foot structure of the dominant leg was measured using the Arch Height Index Measurement System (AHIMS). Standing arch height index (AHI) was calculated as dorsum height divided by truncated foot length. Skin surface markers were attached to the dominant leg and foot according to a multi-segment foot model created by Bruening et al. A14-camera motion capture system (Vicon, Motion Capture Systems, Ltd.) was used to sample kinematic data at 250 Hz while 2 force platforms (Advanced Mechanical Technology, Inc.) were used to collect kinetic data at 1000 Hz. A static trial was captured with subjects in equal weight-bearing stance. Subjects then hung from wooden rings and performed barefoot single-leg drop landings onto 2 force platforms from a height of 0.4 m. A successful trial constituted a natural landing in which the navicular and cuboid markers aligned with the split between the 2 forces platforms, resulting in a rear foot and forefoot impact on separate plates. Data from the landing trials were imported into Visual 3D (C-motion, Inc.) for calculation of static midfoot angle (MA), midtarsal range of motion (ROM), and midtarsal work. Pearson correlation coefficients were calculated for static and dynamic variables using paired t-tests in SAS (SAS Institute, Inc.).

RESULTS: A significant inverse correlation was found between standing AHI using AHIMS and static MA using motion capture technology (r = -0.6087, p < 0.0001). Standing AHI was correlated negatively with sagittal plane midtarsal ROM (r = -0.32032, p = 0.0264) and positively with midtarsal work (r = 0.33180, p = 0.0212). Static MA was correlated positively with sagittal plane midtarsal ROM (r = 0.48336, p = 0.0005) and negatively with midtarsal work (r = -0.32321, p = 0.0250). Raw data can be referenced in Table 1.

DISCUSSION: The strong inverse correlation between the standing AHI and static MA suggests that either method is appropriate for characterizing static foot structure in clinical and research settings. However, static MA showed a stronger correlation to sagittal plane midtarsal ROM than standing AHI. When comparing static and dynamic foot measures, it may be beneficial to use the same technology to reveal any existing relationships that are present. We also found a relationship between both static foot structure measures and both midtarsal ROM and midtarsal work. More specifically, we observed that static foot structure was able to predict 31-48% of variation associated with dynamic midfoot function during a landing task. Our findings suggest that midfoot function may be predictable without having to administer dynamic testing accompanied by complex collection processes and analyses. However, a higher impact dynamic task, such as a drop landing, may required to properly observe a static-dynamic relationship. This type of landing could be performed clinically using a 2D high speed video camera and skin surface markers. We used a multi-segment foot model to measure midtarsal ROM, which is an improvement from traditional models but is still limited in identifying individual tarsal articulations.

SIGNIFICANCE/CLINICAL RELEVANCE: Static foot structure may be a valuable clinical tool in assessing midfoot function relating to injury risk in athletes, who participate in high impact, repeated loading activities, as well as in pathological populations.


FIGURES AND TABLES:

Table 1 Raw data for static and dynamic variables

<table>
<thead>
<tr>
<th>Standing AHI</th>
<th>Static MA (deg)</th>
<th>Midtarsal ROM (deg)</th>
<th>Midtarsal Work (J·kg⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean ± SD</td>
<td>0.324 ± 0.019</td>
<td>-22.475 ± 5.229</td>
<td>27.041 ± 6.916</td>
</tr>
</tbody>
</table>
Measurement of Foot Muscle Strength and Activation
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INTRODUCTION: Intrinsic and extrinsic foot muscles play an important role in proper foot function, yet functional methods of quantifying foot muscle strength are not readily available for clinicians and researchers. Understanding if weakness is present may influence injury prevention and/or rehabilitation efforts. While some functional movements that can isolate the foot muscles are habitual for most adults (toe flexion), others are unfamiliar (doming or short-foot) and may require training to get a reliable measurement. Therefore, the purpose of this study was multi-factorial: 1) to measure the reliability of two methods of measuring foot strength during doming, 2) to determine if there was a learning effect when testing unfamiliar movements, and 3) to measure foot muscle activity during each test.

METHODS: Forty-five subjects (21 M, 24 F; age 25.2±6.4 y; ht 171.5±10.1 cm; wt 71.0±13.1 kg) signed IRB approved consent forms, then completed 2 testing sessions, 2 weeks apart. During each testing session, subjects completed 3 trials of the following tests, in random order: Doming-pull (DL), Doming-push (DS), Great toe flexion (GT), and Lateral toes flexion (LT). DL and DS required the same movement, but the force transducer was secured below the foot during DL and above the foot during DS. Surface EMG data was collected from the gastrocnemius (GAS), fibularis longus (FL), fibularis brevis (FB), tibialis anterior (TA), and abductor hallucis (ABDH). EMG data were normalized to muscle activity during weight-bearing maximal plantarflexion (GAS, FL, FB, and ABDH) or maximal dorsiflexion (TA). During the 2 weeks between testing sessions, subjects were asked to practice the doming movement daily.

Force and EMG data were synchronized and averaged over 1000 points (force = 1000 Hz, EMG = 2000 Hz) surrounding peak force. Within day reliability of the doming strength measurements was assessed using interclass correlation coefficients (ICC2,k) across the three trials. Differences between doming strength during the two sessions were assessed using paired t-tests. A 2-way ANOVA determined differences in percentage of muscle activation between tests during Day 2 testing.

RESULTS: Within session reliability of DL and DS were good-excellent for the first and second days (Table 1). Strength increased between days for both devices (p≤.003). Muscle activity (Table 2) in the GAS was not significantly different between any of the exercises. The FL, FB, TA, and ABDH showed significantly more activity during DL and DS compared to GT and LT (p≤.01). The ABDH was activated significantly more than the other muscles during DL, DS, and GT (p≤.002).

DISCUSSION: Within session reliability of DL and DS were comparable and the strength measurements were strongly correlated (r=0.8), indicating that either method could be used. Comparing doming strength measurements between days, it appears that either a learning effect or strengthening during the 2 weeks of practice allowed subjects to produce more force during Day 2. The increased reliability of the DS measurement on Day 2 suggests less variation between trials per subject.

Muscle activity was normalized to a submaximal contraction, which represents a limitation of the study, but still allows for an indication of which muscles are being used to perform the doming and toe flexion actions. These tests are conducted to try to isolate intrinsic foot muscles, while minimizing the activity of the associated extrinsic muscles. Our data showed that the ABDH is activated during all of the movements we tested. Interestingly, it was more active during DL than DS, though not statistically significant. Force output was also greater in the DL than DS (p<.001).

SIGNIFICANCE/CLINICAL RELEVANCE: These tests are reliable for measuring foot muscle strength. Due to the apparent learning effect, subjects or patients should be taught the doming movement prior to testing.

FIGURES AND TABLES:
Table 1. Force (avg ± SD) and within-day ICCs for doming tests.

<table>
<thead>
<tr>
<th></th>
<th>DL (lbf)</th>
<th>ICC</th>
<th>DS (lbf)</th>
<th>ICC</th>
</tr>
</thead>
<tbody>
<tr>
<td>Day 1</td>
<td>24.7±13.2</td>
<td>0.937</td>
<td>20.4±6.2</td>
<td>0.866</td>
</tr>
<tr>
<td>Day 2</td>
<td>33.5±16.6</td>
<td>0.962</td>
<td>24.9±6.2</td>
<td>0.948</td>
</tr>
<tr>
<td>% change</td>
<td>+34.4</td>
<td></td>
<td>+22.1</td>
<td></td>
</tr>
</tbody>
</table>

Table 2. Percent activation for each muscle during each test on Day 2 (avg ± SD), represented as a percentage of muscle activation during weight-bearing plantarflexion (GAS, FL, FB, and ABDH) or dorsiflexion (TA).

<table>
<thead>
<tr>
<th></th>
<th>GAS</th>
<th>FL</th>
<th>FB</th>
<th>TA</th>
<th>ABDH</th>
</tr>
</thead>
<tbody>
<tr>
<td>DL</td>
<td>7.79±4.34</td>
<td>39.1±23.0</td>
<td>46.9±30.4</td>
<td>45.8±27.2</td>
<td>94.3±66.1</td>
</tr>
<tr>
<td>DS</td>
<td>7.38±4.64</td>
<td>38.0±25.4</td>
<td>42.3±28.9</td>
<td>40.9±22.9</td>
<td>78.9±69.4</td>
</tr>
<tr>
<td>GT</td>
<td>7.94±6.47</td>
<td>12.8±8.23</td>
<td>19.0±8.42</td>
<td>20.8±13.3</td>
<td>32.2±29.4</td>
</tr>
<tr>
<td>LT</td>
<td>10.1±8.64</td>
<td>13.0±9.21</td>
<td>25.4±13.8</td>
<td>27.2±18.9</td>
<td>25.8±22.7</td>
</tr>
</tbody>
</table>
WBCT Hindfoot Alignment of Adult Acquired Flatfoot Deformity: A Comparison of Clinical Assessment and Weightbearing ConeBeam CT Examinations

Lauren Roberts1, Cesar de Cesar Netto1, Scott Ellis1, Shadpour Demehri2, Lew Schon3

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Disclosures: No relevant disclosures.

INTRODUCTION:
Assessment of hindfoot alignment in adult acquired flatfoot deformity (AAFD) can be challenging. Clinical judgment and radiograph studies while important may not represent the accurate valgus alignment of the affected patients. Weightbearing (WB) ConeBeam CT (CBCT) is an emerging imaging modality that may potentially better demonstrate the three-dimensional (3D) deformity, facilitating visualization of important soft-tissue and bony landmarks and helping in surgical planning. Based on the relative position of bone and soft-tissue axes, different measurements of hindfoot alignment can be obtained with CT images. Therefore, we compared clinical assessment of hindfoot valgus alignment in AAFD patients with different measurements performed on WB CBCT images.

METHODS:
In this prospective, IRB-approved study, 20 patients (20 feet, 15 right and 5 left) with clinical diagnosis of flexible AAFD were included. There were 12 males and 8 females, with a mean age of 52.2 years (range, 20 – 88 years of age), and average BMI of 30.35 kg/m² (range, 19.00 – 46.09 kg/m²). Patients underwent “clinical” assessment of hindfoot alignment as well as WB CBCT. Two independent and blinded foot and ankle board-certified surgeons performed different hindfoot alignment measurements on the WB CBCT images that included: 3D “clinical” alignment using soft tissue axes of the tibia and calcaneus; Achilles tendon axis/calcaneal tuberosity angle; angles formed between the tibial axis and the calcaneal tuberosity, calcaneal axis and line connecting midpoint of subtalar joint and most inferior part of calcaneal tuberosity. Positive values were considered valgus alignment. Mean differences between the measurements modalities were compared by paired T-test. Intra- and Inter-observer reliability for the WB CBCT measurements were calculated using Pearson correlation.

RESULTS:
The mean clinical hindfoot valgus measured was 15.15° (SD 7.7°). It was found to be significantly different from the mean values of all WB CBCT angles modalities: 3D “clinical” alignment (10.42°, p<0.015); Achilles tendon/calcaneal tuberosity angle (2.96°, p<0.0001); tibial axis/calcaneal tuberosity angle (5.42°, p<0.0001); tibial axis/subtalar joint angle (7.52°, p<0.0001) and tibial axis/calcaneal axis angle (20.39°, p<0.017).

We found an excellent intra-observer agreement for all WB CBCT 3D measurements (range, 0.8863 – 0.9713, p<0.0001). There was also good to excellent inter-observer reliability, with the exception of the 3D “clinical” alignment (r=0.450, p<0.04), that showed moderate correlation.

CONCLUSIONS:
The use of 3D WB CBCT imaging can help characterize the valgus hindfoot alignment in patients with adult acquired flatfoot deformity. We found the different CBCT measurements modalities to be reliable and repeatable, and to significantly differ from the clinical evaluation of hindfoot valgus alignment.

SIGNIFICANCE/CLINICAL RELEVANCE:
Reliably characterizing hindfoot alignment is integral to surgical planning to ensure appropriate correction to obtain best possible patient outcomes.
In Vivo Intrinsic Foot Bone Motion Measured with Bead Tracked Biplanar Videoradiography

Michael J Rainbow¹, Lauren Welle¹, Andrew WL Dickinson¹, Toni Arndt²

Introduction: Three of the most ubiquitous models of foot function are the spring-arch [1], the windlass mechanism [2], and the transverse tarsal locking mechanism [3]. In recent years, some of these mechanisms have been revisited and refined. For example, Kelly et al. showed that the intrinsic foot muscles act on the arch more than previously thought [4], which calls into question the dominant role ascribed to the passive structures such as the plantar aponeurosis. Holowka et al. and others questioned the rigid lever concept by showing that the midtarsal joint is less stiff during push off than was previously assumed [5]. Likewise, Okita et al. found that the transverse tarsal joint moves throughout the gait cycle [6], which challenges the concept of the transverse tarsal locking mechanism. A mechanistic understanding of foot and ankle function is crucial because treatment strategies aim to restore or replace the foot’s versatile function; however, the current models seem to be insufficient and have not yet been replaced by a comprehensive unified model.

Two challenges in determining a unified theory of foot function are large inter-subject variability in the foot’s morphology, and difficulty in measuring the intricate motion of the foot bones during in vivo dynamic activities. X-ray Reconstruction of Moving Morphology (XROMM) is an emerging approach to measure in vivo foot and ankle bone motion during dynamic tasks; however, it remains challenging to track the foot bones with sufficient accuracy to analyze joint arthromechanics. The purpose of this study was to quantify intrinsic motion of the foot at high accuracy with a subject who had tantalum beads implanted into his foot bones. Here we report accuracy of scientific rotoscoping for markerless tracking, and we report preliminary results on the axes of rotation of the tarsus.

Methods: One male subject (49y) previously had a set of three tantalum beads implanted in the calcaneus, talus, navicular, cuboid, medial cuneiform, and first metatarsal (MT1). After ethics approval and informed consent, CT scans were acquired of the foot (0.441x0.441x0.625mm). The subject performed one-legged hopping at three frequencies (108/120/132 bpm). He also jogged slowly while barefoot. The beads were segmented in the CT scans (Mimics, Materialise) and tracked in XMA Lab (Brown University). The bones were segmented and registered to the beads. The bones were also tracked markerlessly using scientific rotoscoping in Autoscoper (Brown University). We computed RMS errors using the bead tracking as the gold standard. We also computed the instantaneous helical axis of motion and angular velocity of the medial longitudinal arch (MT1 wrt calcaneus), as well as the cuboid, navicular, and calcaneus relative to the talus.

Results and Discussion: RMS errors were small by motion capture standards but larger than previously reported tracking of isolated long bones (< 0.1 mm, 0.1 deg), and too large to generate stable instantaneous helical axes without taking larger steps (Table 1); therefore, manual rotoscoping techniques may not be adequate to establish a detailed understanding of joint velocities and accelerations, and these may be important for understanding how the foot bones loads. More robust global optimization may be required; for example, the simulated annealing 4D tracking algorithm implemented in DSX (C-Motion). Collision detection, multi-bone tracking, and model-based tracking may also be required.

The rotation axis of the medial longitudinal arch moved substantially in 3D from the navicular to the talar neck during weight acceptance and then moved back to the navicular during pushoff, indicating that the joints along the arch alter their relative contributions throughout the cycle. The cuboid, navicular, and calcaneus largely shared a common axis that swept through a range of 13 degrees throughout the cycle. Although the axes were similar, the amount of rotation of the talonavicular and cuboid navicular joint was up to 2.5 times larger than the sub talar joint. Given the unlikely scenario that only this specific subject possesses a single common axis at the tarsal complex, we hypothesize that some of the variation observed in foot bone morphology accommodates a single axis at the tarsal complex so that it can manage deceleration and acceleration of the talus during loading and propulsion.


Table 1: RMS errors between markerless and bead data across proximal tarsal joints.

<table>
<thead>
<tr>
<th>Joint</th>
<th>X (°)</th>
<th>Y (°)</th>
<th>Z (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Joints</td>
<td></td>
<td></td>
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<tr>
<td>X</td>
<td>2.65 ± 2.21°</td>
<td>3.12 ± 2.64°</td>
<td>2.40 ± 2.26°</td>
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Figure 1. Angular velocity and axis of rotation of the medial longitudinal arch (hopping).

Figure 2: Angular velocity during jogging of each joint in the talar coordinate system. A. Dorsiflexion B. Eversion C. Internal Rotation. D. Helical axes of the cuboid, navicular and calcaneus relative to the talar coordinate system during late stance.
Session 8: National Biomechanics Day and STEM

Co-Moderators: Howard Hillstrom, PhD (HSS) and Robin Queen, PhD (Virginia Tech)

10:30AM       Hillstrom, Howard, National Biomechanics Day

10:35AM       8-1: Drazan, John, et al. Low Cost Sports Biomechanics Lab for Accessible STEM Outreach
Low Cost Sports Biomechanics Lab for Accessible STEM Outreach

John F. Drazan¹, Amy Loya¹, Benjamin Horne¹, Joan Bechtold², Robin M Queen³

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INTRODUCTION: Youth are recruited into STEM careers through the “STEM Pipeline” which typically focuses on voluntary “STEM intensive” topics such as robotics and prosthetic design. As such, the STEM pipeline suffers from self-selection of participants; only those who are already interested in STEM voluntarily sign up for these programs. This may contribute to the persistent lack of gender and racial diversity within STEM fields. Biomechanics researchers are uniquely positioned within the scientific community to expand the reach and involvement in the STEM pipeline due to the direct relevance of movement science to youth sports. As such, the engagement of these youth in sports through biomechanics would represent a meaningful addition to the STEM pipeline. The purpose of this study was to evaluate the ability of youth sports to serve as a venue for STEM engagement through the use of sports science and biomechanics. We hypothesized that youth who were recruited through sports programs would self-report lower starting interest in STEM than youth recruited through a traditional out of school STEM program.

METHODS: Two cohorts of youth (Sports (n=145) and Traditional (n=69)) participated in the same sport science activity, however the sport cohort was recruited through youth sport coaches to participate in one of a series of “basketball training clinics”, while the STEM cohort participated in the program during a traditional “STEM enrichment day” at the university. Each youth's demographic information, self-reported interest in STEM, basketball, and sport science were each captured using a 15-question likert-type survey. The sports group completed the same survey following participation in the program to determine if their interests in STEM increased. Statistical analysis was completed using a Wilcoxon Signed-rank test for ordinal data with statistical significance set at p<0.05.

RESULTS: The Traditional cohort was composed of 51% females and 91% under represented minority students (URGS). The Sports cohort was composed of 28% females and 66% UGRS. The interest profile of each group was very different, with Traditional group reporting that 43% of the students were very interested in studying STEM in college, while this was reported in only 13% of the Sports cohort. In contrast, 54% of the Sports cohort reported that they were very interested in playing basketball in college, while this was reported in only 7% of Traditional students. The Sports cohort demonstrated an increase (13% to 20%) in self-reported interest in studying STEM in college following participation (p<0.001).

DISCUSSION: The difference in starting interests indicates that the Traditional group was self-selected for participation in STEM activities, while the Sports group was self selected for basketball interest and not STEM activities. While each group was very diverse, the higher starting interest in STEM among the Traditional cohort indicates that these youth may be already well served by the existing STEM pipeline initiatives. The increase in interest among the Sports cohort following participation provides evidence that biomechanists can reach a unique, underserved population with meaningful STEM outreach. This outreach can then change those students perspectives on pursuing STEM careers.

SIGNIFICANCE/CLINICAL RELEVANCE: This work demonstrates the issue of self-selection in STEM outreach and provides a model for biomechanics researchers to address this issue. Biomechanics researchers can apply their research directly to sports, and as such, can develop STEM outreach programs that can be deployed through youth sport initiatives, thus broadening the STEM pipeline.
Session 9: Foot Type II

Co-Moderators: Taeyong Lee, PhD (Busan Women’s College) and Howard Hillstrom, PhD (HSS)

11:00AM 9-1: Kruger, Karen M., et al. Preliminary Comparison of Hindfoot Kinematics Obtained from the Milwaukee Foot Model and Biplane Fluoroscopy

11:10AM 9-2: Lee, Yeokyeong, et al. The computational study of plantar soft tissue at heel region under weight-bearing condition

11:20AM 9-3: Martinez-Santos, A., et al. Identifying variables which may affect an individual’s response to insole design

11:30AM 9-4: McCahill, Jennifer, et al. Validation of the Foot Profile Score

11:40AM 9-5: Moisan, Gabriel, et al. Muscle activation with two types of foot orthoses in participants with cavus feet during walking


12:00PM 9-7: Brierty, Alexis, et al. Transfer load strategies in typically developed children
Preliminary Comparison of Hindfoot Kinematics Obtained from the Milwaukee Foot Model and Biplane Fluoroscopy

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Disclosures: None

INTRODUCTION: Recent advances in biplane fluoroscopy techniques allow for direct tracking of bony motion during walking and it is considered the gold standard in calculating hindfoot kinematics [1,2]. However, the expensive and time-consuming nature of it, along with radiation exposure, limit clinical use. Multiple marker-based models are currently available for assessing foot and ankle kinematics in clinical gait labs. The Milwaukee Foot Model (MFM) is unique in that it includes skeletal indexing of the foot and ankle using a series of weight bearing radiographs to improve tracking of segments that lack palpable landmarks. The purpose of this work was to compare hindfoot kinematics obtained from model-based tracking of biplane fluoroscopy and the marker-based Milwaukee Foot Model.

METHODS: Biplane fluoroscopy data was collected using a previously described method [3] in which a subject walked on a runway at a normal walking speed. An ankle CT scan was also obtained and the tibia, talus, and calcaneus segmented (Mimics) with a solid model constructed for each bone. Coordinate systems were defined using anatomical landmarks visible on the bone 3D surfaces (PostView). Model-based markerless tracking was used to obtain kinematics of each bone using DSX software (C-Motion, Figure 1). Marker data and radiographic offset measurements were also collected to obtain kinematics via the Milwaukee Foot Model [4]. Data was collected for a healthy subject with no ankle pathology.

RESULTS: Kinematics were calculated for the calcaneus relative to tibia using both the MFM and biplane fluoroscopy (Figure 2). Due to the limited field of view, only 85% of stance was able to be calculated using biplane fluoroscopy. Overall the shape of the kinematic curves showed excellent agreement. The most major discrepancy was observed in hindfoot rotation which showed roughly a 15° offset between the two curves.

DISCUSSION: The abstract shows a preliminary comparison of hindfoot kinematics obtained by a segmental foot model and dynamic fluoroscopy. The results were encouraging in showing similar curve shapes. Offsets between the curves may be due to inter-user variations in measuring radiographic offsets. Future work to automate measurement of these angles would help to improve the model’s accuracy. Moving forward with this method, further evaluation of the MFM and other segmental foot models could be completed to better determine which models are most appropriate for specific foot pathologies.

SIGNIFICANCE/CLINICAL RELEVANCE: Accurate knowledge of hindfoot kinematics during functional activities such as walking is critical for identifying the effects of injury and disease. Methods of obtaining reliable foot kinematics in a clinical setting are critical to the improved care of patients with a variety of orthopaedic impairments or foot deformities, such as cerebral palsy, spina bifida, club foot, and pes planovalgus.


ACKNOWLEDGEMENTS: The contents of this abstract were developed under HSS/NIDILRR grant 90AR5022-01-00 and Shriners grant 71005.
The computational study of plantar soft tissue at heel region under weight-bearing condition
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Disclosures: Yeokyeong Lee (N), Jee Chin Teoh (N), Hee Sun Kim (N), Taeyong Lee(N)

INTRODUCTION: Plantar soft tissue is a complicated composite material that protects the body from daily encountered impact caused by various weight bearing physical activities. The cushioning property of plantar tissue is often compensated by ageing mechanism and diseases that eventually expose the tissue to high risk of ulceration which can be fatal in diseased population such as diabetes mellitus patients. The purpose of the study is to create a heel pad FE model to facilitate the understanding of tissue behavior under various extreme circumstances which are not able to be conducted in experiments.

METHODS: In this study, the effective FE method was proposed to verify the indentation experiment. The experiment was conducted on 3 healthy subjects at the heel with a novel indentation system (Chen et al., 2011). A schematic FE model of the heel and the spherical indenter was generated. The heel was assumed to be a cylinder with a diameter of 50 mm, and 6 noded-wedge elements were used. Hyper-elastic material model obtained from the experiment was applied to the heel part. The indenter was modeled as a rigid sphere with a diameter of 5 mm. Before simulating indentation test, 122.50 N was uniformly loaded to heel model to generate body weight (50kg) bearing condition. Other simulation conditions were same as the experiment, and implicit dynamic analyses of the contact between the heel and the indenter were performed using commercial software ABAQUS ver. 6.10-3.

RESULTS: Figure 1 is obtained from the experiment and FEM of subject 1, and it represents relationships between the indentation test time and the reaction force of heel pad. As shown, the reaction force of experiment is slightly higher than that of FEM before 0.66 sec, but the results are a good agreement after that time. The maximum forces from the experiment and the analysis are 7.4112 N and 7.481 N, respectively, and the error is only 0.0698 N (around 1%).

DISCUSSION: The FEM predictions are in good agreement with the experimental results, suggesting the usability of the tissue model in simulating various tissue experiments.

SIGNIFICANCE/CLINICAL RELEVANCE: The use of heel pad FE model is high potential, which increases the possibility of tissue study on many crucial conditions. This will definitely enhance the understanding of ulceration mechanisms especially in the diseased population.

REFERENCES:

ACKNOWLEDGEMENTS: The study was funded by Korea National Research Funding (2-2016-1057-001-1).
Identifying variables which may affect an individual’s response to insole design

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INTRODUCTION: Elevated plantar peak pressures (PP) are a key risk factor for foot ulceration in people with diabetes. A number of studies have investigated whether structural or biomechanical factors predict PP patterns during walking and whether such factors may also dictate the response to a particular off loading insole design. Despite understanding responses to insole design being relevant to effective clinical decision making, no research has attempted to identify factors which predict individual response to insole designs. The aim of this study was to investigate foot characteristics that may influence changes in PP due to different insole designs.

METHODS: National Health Service (NHS) and University ethical approval were obtained and 60 participants (65.9 ± 12.6 years and BMI 29.4 ± 5.2) with diabetes and neuropathy were recruited. They attended two test sessions. On the first session, tissue stiffness at different areas of the foot sole, multi segment foot kinematics, and range of ankle, 1st metatarsophalangeal joint (MPJ) and subtalar joint range of motion were collected (ROM). Also, barefoot plantar pressures and a 3D foot scan were collected and used to design customized insoles. These insoles had a metatarsal bar and a void tailored to the participant’s plantar pressures. Nine insole designs were tested, including variations where the metatarsal bar was moved distally and proximally by 2% of foot length and using soft EVA or Poron as forefoot cushioning in the void. On the second visit, in-shoe plantar pressures were collected while walking with each of the 9 insole designs. Logistic regression tests were performed to investigate the relationship and influence of each foot variable (tissue stiffness, kinematics, ROM) with the changes in PP due to each insole collected at session 2. Accordingly, two sets of logistic regressions were made (1) to investigate the characteristics that may have influenced a PP reduction due to the insole, and (2) to investigate which characteristics may influence a PP increase due to the insole.

RESULTS: Associations were found between changes in PP and the following foot characteristics:

- **High 1st metatarsal head tissue stiffness (620 – 702º Shore O)** was associated with:
  - Being 5 times more likely to achieve a reduction in PP under the 1st metatarsal head (using either base or distal metatarsal bar position, with either Poron or soft EVA in the void (p = 0.024)).
  - Being 8 times less likely to get an increase in PP under the 5th metatarsal head (using a distal metatarsal bar position and Poron in the void (p = 0.048)).

- **Low 1st MPJ ROM (10 – 30º)** was associated with:
  - Being twice as likely to achieve a reduction in PP under the 1st metatarsal head (with a distal metatarsal bar position and Poron in the void (p = 0.047)).
  - Being 6 times more likely to experience an increase in PP under the 1st metatarsal head (with proximal metatarsal bar position and Poron in the void (note, p = 0.085)).

- **Low subtalar joint mobility**
  - **Low inversion (0 – 12º)** was associated with:
    - Being 3 times more likely to achieve a reduction in PP under the 1st metatarsal head (with a base metatarsal bar position with Poron in the void (p = 0.043)).
    - Being ~4 times more likely to achieve a reduction in PP with any condition under the central metatarsal heads (p all insole conditions <0.05).
  - **Low eversion (0 – 4º)** was associated with:
    - Being ~5 times less likely experience PP increase under all regions except 5th metatarsal head (p ≈ 0.03).

- **Low ankle joint velocity (48.8 – 108.4 radians)** was associated with:
  - Being 3 times more likely to achieve a reduction in PP under the central metatarsal heads (when using a base metatarsal bar position with no material in the void (note, p = 0.064)).
  - Being 4 times more likely to increase PP under the hallux (with a proximal metatarsal bar position and no material in the void (p = 0.041)).

DISCUSSION: In general, factors that might increase pressure, such as stiffer plantar tissues and lower joint motion (e.g. stiffer joints), were associated with increased likelihood of reducing PP. This suggests the insole designs compensate for the effect of these factors when the factors are present, but the same insoles may not have the same effect if those factors are absent.

CLINICAL RELEVANCE: We identified different individual characteristics that influence changes in PP due to different insoles. Factors such as tissue stiffness and joint range of motion can be measured in a clinic. This knowledge could help practitioners in their day-to-day decision making about insole prescriptions in those with diabetes.
Validation of the Foot Profile Score

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INTRODUCTION: In three-dimensional gait analysis, the Gait Profile Score (GPS) gives clinicians a single measurement of the quality of an individual’s gait pattern. The GPS is calculated as the average root mean square (RMS) difference between a patient’s data and normative data taken over 9 key kinematic variables for both legs [1]. The GPS includes the traditional measurement of the foot as a single rigid segment in two-dimensions. More recently, three-dimensional multi-segment foot models have been developed to improve our understanding of foot motion during gait. The Oxford Foot Model (OFM) was designed to measure tibia, hindfoot, forefoot and hallux motion in three dimensions [2]. To fully understand an individual’s gait pattern both the lower limb kinematics and the foot kinematics should be measured in three dimensions. The Foot Profile Score (FPS) is a single number, designed to be the equivalent of the GPS, but based on the OFM kinematics. The aim of this study is to validate the FPS against a global clinical assessment of foot deformity.

METHODS: In order to validate the FPS we sent sagittal and coronal close-up foot videos of 60 subjects to 10 clinicians from 4 countries. Each subject was scored by 5 clinicians. The subjects included a range of demographic characteristics and severity of foot deformity. There were 30 children and 30 adults, 21 subjects had cerebral palsy, 17 had orthopaedic diagnoses, 16 had neurological diagnoses and 6 had clubfoot. We used right leg data in 31 subjects/ and left leg data in 29 subjects, there were 36 males/ 24 females. There were no markers on the feet in the videos. We asked the clinicians to rate the severity of foot deformity using a scale from 0-3 which we termed the Clinical Foot Deformity Scale (CFDS) (0=normal, 1=mild, 2=moderate, 3=severe foot deformity). Mild, moderate and severe deformity could range from planovalgus to cavovarus foot deformities. All 60 video subjects also had the Oxford Foot Model kinematics collected using a Vicon T-series motion capture system (Vicon Motion Systems Ltd.) including 16 infra-red Vicon T-series cameras.

The CFDS was taken as the mean of all 5 clinician’s ratings for each subject. The FPS was calculated by the average RMS difference between a patient’s data and normative data taken over 6 key kinematic graphs for both the same leg as used for the CFDS scoring for each subject. The 6 variables are: sagittal - hindfoot, forefoot; coronal – hindfoot, forefoot; transverse – hindfoot, forefoot. The Spearman Rank Correlation was used to analyse the relationship between the FPS and the CFDS.

RESULTS: The mean of the FPS scores for all 60 subjects was 8.6 degrees (SD 3) indicating a range of deformity within the group (a normal FPS is 3 degrees or less). The mean of the CFDS subjects is presented in Table 1. This indicates there was a range of deformity within each group, and the amount of deformity was consistent across groups. The correlation between the FPS and the CFDS was significant at 0.69 with p<0.01 (Figure 1).

DISCUSSION: The CFDS is based on a visual impression of foot deformity that clinicians use in their daily practice. It is encouraging that the FPS correlates well with the CFDS. We wouldn’t expect a perfect correlation as the FPS gives quantitative information on all 3 planes of movement; in particular the transverse plane which is difficult to evaluate clinically or with two-dimensional video. The relationship between the FPS and GPS now needs further investigation to understand how to use these two measures together to inform optimal clinical decision making.

SIGNIFICANCE/CLINICAL RELEVANCE: The FPS will assist clinicians and researchers in quantifying foot abnormalities during gait to monitor change over time and to measure the outcome of intervention.

Muscle activation with two types of foot orthoses in participants with cavus feet during walking

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Disclosures: None to declare

INTRODUCTION: Foot orthoses (FOs) are regularly used to treat, with an overall good efficacy, many musculoskeletal pathologies such as posterior tibial tendon dysfunction [1] and patellofemoral pain syndrome [2]. Extrinsic modifications are regularly added to FOs in clinical practice to increase treatment specificity despite limited evidence supporting the treatment. These modifications can be described as any material you add to the FOs’ shell. Benefits of using these modifications are still poorly described in the scientific literature even if clinical results are generally good when they are utilized. One of these modifications is a lateral bar, which is a one-centimeter wide ethylene-vinyl-acetate bar glued under the lateral part of the FOs [3]. It has been shown that adding this bar to the FOs decreases the activity of the fibularis longus muscle during walking [3]. However, no study to date has quantified its effect on a population with cavus feet. The aim of this study was to quantify the effects of FOs with and without a lateral bar on muscle activity of participants with cavus feet.

METHODS: Fifteen participants with cavus feet were recruited from the Université du Québec à Trois-Rivières (UQTR) outpatients podiatry clinic. Muscle activity of tibialis anterior, fibularis longus, gastrocnemii, vastus medialis and lateralis, biceps femoris and gluteus medius were recorded with a wireless surface EMG system (Delsys Trigno, Boston, USA) when walking as fast as possible during two experimental conditions (FOs with and without a lateral bar) and a control condition (shoes). Experimentations were performed after one month of adaptation to each experimental condition (two testing sessions). The experimental protocol was approved by the UQTR Ethics Committee. All participants provided an informed written consent upon enrolment in the study. The analyses were performed on the Root Mean Square (RMS) of the data, calculated with a moving window of 50 milliseconds width with a 50% overlap. RMS data were normalized with the mean peak RMS of all trials of the control condition for each testing session. Each individual stride was normalized to 100%. To compare effects between conditions, a curve analysis was performed using one-dimensional statistical parametric mapping [4].

RESULTS: There was a decrease tibialis anterior muscle activity at 78% of the gait cycle with FOs with lateral bar (see Fig.1) and an increase lateral gastrocnemius muscle activity from 45 to 46% of the gait cycle with FOs (see Fig.2).

DISCUSSION: The increased activity of the lateral gastrocnemius muscle is not consistent with previous literature, which found no difference [5] or a decreased activity [3]. The effects of FOs on the activity of the tibialis anterior muscle are contradictory in previous studies in which no difference [6], a decreased [3] and an increased activity [5] were observed. Considering that the changes were significant for a very short period of time and that the effect size was under 0.5, this result should not be considered clinically significant. The different type of FOs used, the faster walking speed and participants’ foot type could perhaps explain the heterogeneity of these results compared to previous studies. One limitation of this study is that muscle activity was assessed on asymptomatic participants while FOs are usually prescribed for symptomatic patients.

SIGNIFICANCE/CLINICAL RELEVANCE: FOs with and without a lateral bar have minimal effects on muscle activity of participants with cavus feet during fast walking. Their beneficial effects when treating musculoskeletal pathologies could perhaps be explained by kinematics of kinetics changes.

REFERENCES:

FIGURES AND TABLES:

Caption: Blue= Experimental conditions, Black=Control condition (shoes), Shadows= Standard deviation clouds

Fig.1 Tibialis anterior muscle activity with shoes and FOs with lateral bar

Fig.2 Lateral gastrocnemius muscle activity with shoes and FOs
Comparison of 3 casting methods for custom foot orthoses

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DISCLOSURE STATEMENT: Performance Laboratories Inc. manufactured CFOs and Vertical Orthotics Pty Ltd provided supplies to fabricated CFO.

INTRODUCTION: Custom foot orthoses (CFOs) are commonly used to manage foot problems associated with aberrant foot biomechanics, including heel pain syndrome and diabetic foot ulcers. [1] CFOs are traditionally made from a cast captured in non-weight bearing condition using plaster of Paris. Recently, 3-dimensional laser scanners facilitate the registration of foot shape. In addition, Vertical Foot Alignment System (VFAS) developed a novel method to directly fabricate CFOs in weight bearing corrected alignment.[2] However, performances of these new casting techniques have not been objectively evaluated.

METHODs: Healthy asymptomatic subjects, between ages 18-65 years with moderate pes planus, provided consent and participated in the study. Two pairs of CFOs were casted by designated experienced clinicians using plaster of Paris (P) and a 3D laser scanner (Q). These CFOs were fabricated in standard manner by the same orthotic laboratory. In addition, a pair of VFAS was directly molded in weight bearing corrected position. Shoe comfort rating and dynamic in-shoe plantar pressure were measured (novel pedar-X, sampled at 100 Hz) during comfortable self-selected walking in a standard shoe (New Balance, #574), following 10-minutes of accommodation. The lower limb was used as the unit of observation instead of the individual. A Generalized Linear Model with an identity link function was used to test difference across 4 shod conditions while accounting for potential dependence in bilateral data. The Wald Chi-square was calculated for each dependent variable with significance set at p < 0.05. Post hoc pairwise comparisons for all pairs were performed using the Generalized Chi Square test at p < 0.05.

RESULTS: Participants consisted of 24 subjects (9 female) with mean age of 25.4 ± 3.86 years old and BMI of 25.6 ± 3.79 kg/m². A significantly reduced Resting calcaneal stance position (RCSP) was noted when participants stood over each of CFO (9° valgus), compared to barefoot posture (13° valgus). Overall shoe comfort was significantly higher in V condition than other 3 shod conditions. Sneaker only (C) yielded the largest force time integral (FTI, Ns) under the metatarsal 1 (p=0.013) and the lowest FPI under medial arch (p < 0.001). No significant difference is noted between P and Q conditions in any regions. CFO made from weight bearing corrected method (V) yielded a significantly lowered FTI under metatarsal 2-4, lateral arch, and heel regions than P and Q shod conditions.

DISCUSSION: Various theories and techniques exist for casting, fabrication, and utilization of CFOs. Two novel casting methods were compared to the standard of care. No significant differences were noted between the traditional (P) and a 3D laser scanning (Q) based on shoe comfort rating and in-shoe plantar pressure assessment in 24 healthy subjects with moderate pes planus. Weight bearing neutrally aligned direct molded CFO (V) yielded significantly greater shoe comfort and reduced total load in many regions of foot. Additional studies are needed to refine the methods and clinical efficacy of CFO.

CLINICAL SIGNIFICANCE: CFOs are commonly used to manage many foot pathologies. Biomechanical assay is needed to advance the casting, fabrication, and clinical utility of this common treatment modality.

REFERENCES
Transfer load strategies in typically developed children
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Disclosures: Alexis Brierty (N), Claudia Giacomozzi (N), David Bade (N), Sean Horan (N), Christopher Carty (N)

INTRODUCTION: Pressure measurement technology has progressed substantially since its inception in the late 1800’s, with many options now available (e.g. insole in-shoe systems, pressure mats/walkways). As a result there is a large body of literature related to foot pressure profiles experienced by typical and pathological individuals, the latter being most commonly reported for diabetic or obese populations. Many scientific reports have provided insight into the foot-ground surface interaction, progression of centre of pressure, effect of walking speed, and prediction and prevention of injuries and ulcers. A recent technological advancement in the study of plantar pressure has been the integration of 3D motion analysis [1], which allows the demarcation of anatomically correct regions of interest (ROI’s) of the foot – a noted limitation of traditional approaches. This is particularly important in children, where development of the foot and a typical gait cycle may cause a large amount of variance in plantar loading of specific regions, both within and between young participants [2]. Interestingly and somewhat surprisingly, there has been a scarcity of research on how load is transferred between consecutive steps in typically developed children. The aim of this study was to therefore present a method of quantifying regional load transfer between consecutive footsteps in typically developed children to determine the effect of the preceding step on foot ground pressures during the subsequent step.

METHODS: Six typically developed children between 4 and 16 years of age, without a history of lower limb injury were recruited. All participants provided informed consent, with all procedures approved by the Griffith University human ethics committee. An EMED-XL pressure platform (Novel GmbH; 88x188 capacitive sensors (4 sensors/cm²); 100Hz) was positioned in the centre of a 10-camera, motion capture laboratory (10x Vicon V16 Vantage cameras; 2000Hz). Reflective markers were attached in accordance with the Oxford Foot Model (OFM), following which participants performed 12-14 walking trials at a self-selected speed while synchronous marker trajectory and pressure data were collected. A dedicated Matlab code was developed to automatically re-align the reference systems [1, 3], superimpose markers onto the acquired footprints, identify five ROIs as reported by Stebbins et al [1], and process all the extracted parameters associated with both the single steps and the whole gait cycle. The procedure to optimize the matching between marker configuration at midstance and maximum pressure footprint was further developed from previous studies [1, 2], by taking into account the lowest marker motion along the three axes.

RESULTS: Only the load transfer phase of the gait cycle was analysed (i.e. both feet in contact with the ground). Eight typical load transfer phases were included for each participant (4 R-L and 4 L-R), and were normalised to body mass. For the rear foot, normalised medial forefoot forces (72.8% of body weight) at the start of the load transfer phase were consistently larger than lateral forefoot forces (28.5%) across all participants (see figure). For the front foot, medial and lateral hindfoot forces were similar at heel strike with the medial hindfoot forces being generally higher during the initial loading phase (36.1% medial and 30.1% lateral at 50% of load transfer) - although this varied across participants (see figure).

DISCUSSION: The present findings provide preliminary data regarding load transfer in young typically developed children. For the participants in the current study, normalised forces show a common load transfer pattern for selected walks in a typically developed cohort. Reviewing the load transfer phase of subsequent steps, will assist in determining if, and to what extent, plantar forces in one step influence the next step. Future work will analyse irregular load transfers from the current cohort to explore strategies used to correct abnormal or off-balance steps, and compare with gait kinematics and kinetics.

SIGNIFICANCE/CLINICAL RELEVANCE: Integrating plantar and kinematic data allows for an in depth analysis of the foot’s interaction with the ground. Analysing the load transfer phase also allows interpretation of the effect of previous and subsequent footsteps on plantar loading.

REFERENCES:
Session 10: Ankle

Co-Moderators: Uwe Kersting, PhD (Aalborg University) and Sorin Siegler, PhD (Drexel University)

2:00PM 10-1: Ding, Rui. & Derrick, Timothy R. Tibial Strain with and without the Fibula

2:10PM 10-2: Kersting, Uwe G., et al. Single-leg balance training – lifting ankle training beyond one single joint

2:20PM 10-3: Kurar, Langhit. & Charles, Loren. Ankle Fracture Management: A Unique Cross-Departmental Quality Improvement Project

2:30PM 10-4: Patel, Akshay, et al. Talocural manipulation effects walking speed in a young and older population


2:50PM 10-6: Siegler, Sorin, et al. Surface-to-surface Interaction at the Joints of the Ankle Complex and Foot in Varus and Valgus Deformities
INTRODUCTION: In many finite element analysis studies, the leg segment is composed of a tibia, without a fibula. However, researchers have found that the fibula is capable of some weight bearing, resulting in relief of the strain in tibia by 5%-15%\(^1\). It has also been found that the interosseous membrane plays a role in distributing the load between the tibia and fibula\(^2,3\). The purpose of this research is to find the difference in strain at the mid-shaft of tibia and fibula in three levels of finite element models, the models with tibia only, with tibia and fibula, and with tibia, fibula and interosseous membrane.

METHODS: 3D geometry and material properties of tibia and fibula were obtained from the VAKHUM project\(^4\). The model included 97809 nodes, and 87387 hex8 elements. The software suite FEBio\(^5\) was used for finite element analysis. The bones were modeled as a linear isotropic material with young’s moduli estimated from CT scans. The Poison ratio was set to 0.3. The properties of additional materials were obtained from literature. Articular cartilage was added to tibia and fibula at the proximal tibiofibular joint by extruding the elements of the joint contact area in a local-normal direction by 0.7mm to make the two bones “just touching”\(^6\). The cartilage was modeled as a nearly incompressible Mooney-Rivlin material\(^7\). Tension only linear springs were used to model the interosseous membrane\(^8\), the proximal anterior tibiofibular ligamentous complex \(^8\), proximal posterior tibiofibular ligamentous complex \(^9\), distal anterior tibiofibular ligament\(^10\) and distal posterior tibiofibular ligament\(^10\). The center of the distal surface of tibia was fixed in all three translation directions, and a node on the medial side of the proximal tibia was fixed in the anterior-posterior and medial-lateral directions. Another node on the medial malleolus was fixed in anterior-posterior direction, so the bone would not rotate in the transverse plane\(^11\). The fibula was fixed in a similar way with bone bottom node and other two nodes on the lateral side of the bone. But the distal bottom of fibula was not fixed in medial-lateral direction. A downward axial load of 60kg (588.6N) was applied to the point that was just anterior to the center of knee joint and center of ankle joint in the sagittal plane on the proximal surface of tibia\(^1\). The strain along the axial direction of three elements on medial, lateral and posterior sides of tibia, and three elements from the anterior, lateral and posterior side of the fibula were selected to be compared among each condition\(^3\).

RESULTS AND DISCUSSION: The strain values from the three models are listed in table 1. The strain on the mid-shaft of tibia and fibula was in the similar range of gauge study\(^3\). And similarly, the strain decreased a lot but did not turn to zero after removing the interosseous membrane. Interosseous membrane plays an important role in the weight bearing function of fibula. One study that found the strain at fibula was zero after incision of interosseous membrane\(^5\). This may be due to the difference of position/direction of loading and precision of instrument. The current models have the fibula fixed at the distal end. This may differ from the real condition with support of the lateral surface of talus. This might be the reason why the strain in tibia didn’t change much when the fibula was removed. The talus will be included in the model to further study the relationship of tibia, fibula and interosseous membrane in weight bearing.

SIGNIFICANCE/CLINICAL RELEVANCE: This study provided an initial picture of the function of fibula and interosseous membrane in weight bearing using finite element analysis, helping understand the repair of tibia and fibula fracture\(^12\).

REFERENCES:

FIGURES AND TABLES:
Table1: Strain in different conditions (µε)

<table>
<thead>
<tr>
<th>Models</th>
<th>Medial</th>
<th>Tibia Lateral</th>
<th>Posterior</th>
<th>Medial</th>
<th>Fibula Lateral</th>
<th>Posterior</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibia alone</td>
<td>-287.3</td>
<td>28.3</td>
<td>-50.2</td>
<td>N/A</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>Tibia with Fibula</td>
<td>-284.5</td>
<td>25.0</td>
<td>-52.8</td>
<td>23.7</td>
<td>-5.1</td>
<td>6.7</td>
</tr>
<tr>
<td>Tibia with Fibula and Membrane</td>
<td>-286.4</td>
<td>25.3</td>
<td>-52.0</td>
<td>69.7</td>
<td>-38.1</td>
<td>1.4</td>
</tr>
</tbody>
</table>

Negative value means compression, and positive value means tension.
**Single-leg balance training – lifting ankle training beyond one single joint**

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**Disclosures**: none

**INTRODUCTION**: Wobble board training is a well-known intervention for the prevention and rehabilitation of ankle injury, in particular in cases where a chronically unstable ankle (CAI) has developed. Epidemiological studies provide strong support of re-injury rates being reduced by this kind of training. The common belief is that the ankle joint is mainly affected in regard to strength, stiffness or reflex-mediated muscular activity. Recently, various groups have indicated that this kind of training involves more complex motor control mechanisms, in fact a variety of these [1], requiring a holistic view on ‘joint adaptation’. The purpose of this paper was to illustrate the multi-faceted effects of ankle training on a wobble board by reporting various aspects of whole body effects and putting these into perspective.

**METHODS**: All included studies were based on 4-6 weeks of single-leg wobble board training. Subjects were exposed to exercises in 30-min sessions, three times per week, with increasing difficulty. To assess training effects balance tests on firm surfaces, on the wobble board and sport-specific cutting and landing tasks were used. Outcome measures were based on inverse dynamics modelling and extensive electromyographical studies. For statistical analyses, repeated measures ANOVAs were applied as well as correlational analyses.

**RESULTS**: Results showed that wobble board training leads to significantly improved performance on the task itself (standing on the board) but no related improvements of standing on a firm surface. Training effects were a reduced and slower movement of the ankle but also the free leg, the upper body and arms [2]. Muscle activity was reduced with increased duration of training, but more so for the ankle muscles than for the thigh and hip muscles. Generally, these specific training effects were accompanied by adaptations to landing technique after jumps [3], reactive muscle activation in particular during perturbed cutting movements and a reduction of mechanical loading of the ankle and knee joints which are commonly interpreted as being protective [4]. Further, we showed a cross-educational effect of these adaptations to the untrained body side.

**DISCUSSION**: Based on these studies we suggest that isolated, single-leg wobble board training leads to overall motor control adaptations rather than a strengthening of the ankle joint alone. We propose that these adaptations of motor control system are suitable to explain the protective effect balance training being shown in epidemiological studies. We also suggest that single-leg balance tests on firm surfaces may not be the optimal tests to assess dynamic balancing performance. We therefore believe that tests on unstable surfaces should be included in risk factor assessments as they will likely have a greater predictive value. Future research should address this question.

**SIGNIFICANCE/CLINICAL RELEVANCE**: Balance training on unstable surfaces - not necessarily on a wobble board, despite the fact that we mainly used this mode of training - leads to whole body adaptations which partly explain the protective effect of ankle disk training. These exercises should be initiated with the opposite if the injured ankle is not fully functional again to further improve rehabilitation programs.

**REFERENCES**:  

**ACKNOWLEDGEMENTS**: CAPES foundation, Brazil; Aalborg University, Denmark.
Ankle Fracture Management: A Unique Cross-Departmental Quality Improvement Project

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Disclosures: Nothing to disclose

INTRODUCTION: In light of recent BOAST 12 (August 2016) published guidance on management of ankle fractures, the project aimed to highlight key discrepancies throughout the care trajectory from admission to point of discharge at a district general hospital. Wide breadth of data covering three key domains: accident and emergency, radiology, and orthopaedic surgery were subsequently stratified and recommendations on note documentation, and outpatient follow up were made.

METHODS: A retrospective twelve month audit was conducted reviewing results of ankle fracture management in 37 patients. Inclusion criterion involved all patients seen at Darent Valley Hospital (DVH) emergency department with radiographic evidence of an ankle fracture. Exclusion criterion involved all patients managed solely by nursing staff or having sustained purely ligamentous injury. Medical notes, including discharge summaries and the PACS online radiographic tool were used for data extraction.

RESULTS: Cross-examination of the A & E domain revealed limited awareness of the BOAST 12 recent publication including requirements to document skin integrity and neurovascular assessment. This had direct implications as this would have changed the surgical plan for acutely compromised patients. The majority of results obtained from the radiographic domain were satisfactory with appropriate xrays taken in over 95% of cases. However due to time pressures within A & E, patients were often left without a post manipulation XRAY in a backslab. Poorly reduced fractures were subsequently left for a long period resulting in swollen ankles and a time dependant lag to surgical intervention. This had knock on implications for prolonged inpatient stay resulting in hospital acquired co-morbidity including pressure sores.

DISCUSSION: The audit has highlighted several areas of improvement throughout the disease trajectory from review in the emergency department to follow up as an outpatient. This has prompted the creation of an algorithm to ensure patients with significant fractures presenting to the emergency department are seen promptly and treatment expedited as per recent guidance. This includes timing for xrays taken in A & E. Re-audit has shown significant improvement in both documentation at time of presentation and appropriate follow-up strategies. Within the orthopaedic domain we are in the process of creating an ankle fracture pathway to ensure imaging and weight bearing status are made clear to the consulting clinicians in an outpatient setting.

SIGNIFICANCE/CLINICAL RELEVANCE: As a result of the ankle fracture algorithm we have adapted the BOAST 12 guidance to shape an intrinsic pathway to not only improve patient management within the emergency department but also create a standardised format for follow up.
Talocural manipulation effects walking speed in a young and older population
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Disclosures: This work was a part of the PG dissertation completed by Akshay Patel; Helen Branthwaite and Nachiappan Chockalingam were involved in the design and supervised this study.

INTRODUCTION: The aging process has been shown to have a detrimental impact on both kinetic and kinematic variables of the talocural joint in older adults (Arnold et al., 2014). These changes can alter walking speed with slower speeds correlating to increased hospital admissions (Middleton et al 2015) and increased risk of falls associated with restriction in the talocural joint (Menz et al 2006). Ankle Manipulation is a common manual therapy performed to improve talocural range of motion (Loudon et al 2013). Therefore, the aim of this paper is to assess the effect of talocural manipulation on walking speed in a young and older population with the view for it to be used as an intervention to reduce risk of falls in the elderly.

METHODS: Ethical approval was granted from Staffordshire University. 10 young (mean age 35.8±11.1 years; sex ratio 5:5 male:female; mean height 170±8.9 centimetres; mean weight 74±12.1 kilograms) and 10 older (mean age 72.5±7.2 years; sex ratio 5:5 male:female; mean height 164±12.7 centimetres; mean weight 67.6±6.7 kilograms) were recruited from a local population, Hertfordshire, UK. A quasi-experimental pre-test post-test study was used to measure the immediate effects of talocural manipulation (Figure 1) on walking speed. A 10 metre walkway was used to assess walking speed prior to and immediately after talocural manipulation. This consisted of a straight, flat indoor space with markers to indicate the distance at each end. An acceleration phase of 2.5 metres was used before participants walked the 10 metres at a self-selected pace. Infrared timing gates, Brower Timing Systems (TC Timing System) were used to capture walking speed within the 10 metre distance, which was repeated 3 times to gain an overall average. A paired sample t-test, 95% confidence with bonferroni correction, was used to test significance of walking speed changes pre and post manual therapy intervention.

RESULTS: The younger group improved the walking speed after manipulation from mean 1.25 to 1.29 m/s (SD= 0.55, percentage change = ↓3.37%, p>0.01) The older group also improved with mean walking speed increasing from mean 0.94 to 1.02 m/s (SD= 0.83, percentage change= ↓8.22%, p>0.02).

DISCUSSION: There is an improvement of walking speed immediately after a Talocural manipulation in young and older adults. This improvement for the older adult cohort changes the risk stratification from ‘at risk of hospital admission in 1 year’ to ‘less likely to be hospitalised’ on the walking speed schematic (Middleton, et al., 2015). This could reduce the risk of falling. The results of this study encourages continued exploration of talocural manipulation as an intervention for falls management. Additionally further work is also required to explore the lasting effects of manipulation.

SIGNIFICANCE/CLINICAL RELEVANCE: Improved walking speed after talocural manipulation could reduce hospital admissions and reduce the risk of falling in older adults.

REFERENCES:

FIGURE 1 Talocural manipulation applied as the intervention.
The effect of ankle arthritis and total ankle arthroplasty on COP position at toe-off.

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Disclosures: None for all authors.

INTRODUCTION: End-stage ankle osteoarthritis (OA) leads to pain, disability, low self-report scores on physical and emotional well-being, and significant changes in ankle mechanics including changes in loading, ankle plantarflexion, and mechanical energy recovery.1-4 Few studies, however, have examined the center of pressure (COP) in ankle OA. Horisberger et al.5 reported reduced loading in specific regions but did not study the path of the COP. This study tests the hypothesis that patients with ankle OA will toe off from a more proximal position (COP will move less anterior) in the affected foot and that a distal COP position will be restored following surgery.

METHODS: Data were collected from 4 embedded force plates, during 7 self-selected speed level walking trials along a 10 meter walkway for 93 subjects (43 female and 50 males) with unilateral ankle OA who progressed to total ankle arthroplasty (TAA). The force plate data were used to calculate anterior-posterior and medial-lateral position of the COP relative to foot length and width based on caliper measurements for both the surgical (S) and nonsurgical (NS) limbs. Testing was completed within 2 weeks prior to TAA and again one and two years following TAA. Hindfoot alignment was characterized in each patient as valgus, varus, or neutral through radiographic assessment (32 neutral, 22 valgus, and 39 varus patients). This study was approved by the institutional review board and informed consent was obtained for all subjects.

RESULTS: There were no differences in walking speed by hindfoot alignment before or after surgery. The initial location of the COP and the mediolateral distribution of the COP as measured by the coronal index and the number of midline crosses did not differ between hindfoot alignment groups or across time. COP beginning on average at 15.8% ± 7.9 of foot length and CI at -0.005 ± 0.019, which indicates limited mediolateral translation. The final location of the COP differed between the NS and S limbs (p = 0.03) with the COP final position being 92.5% ± 0.06 and 88.7% ± 0.06 of foot length, respectively (Figure 1). Within hindfoot alignment categories this pattern held for neutral (92.8% vs. 87.8%; p =0.003) and varus subjects (91.5% vs. 87.6%; p = 0.003), but not for valgus subjects (93.6% vs. 91.9%; p=0.51). TAA restored hindfoot alignment to neutral and increased the final COP position for the S limb to 90% or higher at both post-operative time points (Figure 1).

DISCUSSION: As would be predicted by changes in toe-off mechanics, patients with severe ankle OA limit the extent of the anterior-posterior position of the COP and, therefore toe off in a relatively proximal position on digits I and II. This can lead to abnormal loading of the metacarpophalangeal and proximal interphalangeal joints and further joint damage.

SIGNIFICANCE/CLINICAL RELEVANCE: Surgical intervention shifts the final location of the COP to a more distal location and to a position closer to that of asymptomatic limb. COP position is an effective measure of disability and could be useful in identifying and understanding changes in function following surgical intervention.


FIGURES AND TABLES: Figure 1

Figure 1: Final COP location between the surgical and non-surgical side and across time. An asterisk (*) indicates a significant difference between the surgical and non-surgical side.
Surface-to-surface interaction at the joints of the ankle complex and foot in varus and valgus deformities
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INTRODUCTION: Hindfoot malalignment is recognized as a major cause of pathology in the foot and ankle and has considerable influence on its biomechanics. Current measurements based on 2D radiographs are subjected to anatomical and operator bias based on inaccuracies due to projections on a 2D plane. Recent developments including Weight Bearing Computed Tomography (WBCT) and semi-automated 3D measurements of hindfoot malalignment provided a significant improvement. However they do not provide information regarding the specific surface-to-surface interactions that occur in the various joints of the ankle complex and foot. In this study a 3D biometric tool based on distance mapping, which describes this interaction, is applied to WBCT data to characterize the effect of hindfoot valgus and varus deformities on the individual joints of the ankle complex and foot.

METHODS: In this retrospective cohort study, 30 bilateral data sets (10 normal feet, 10 varus, 10 valgus) from WBCT (PedCAT; CurveBeam LLC, Warrington, PA) were obtained by a specialized foot and ankle unit. The images were processed to produce three dimensional models of the tibia, fibula, talus, calcaneus, navicular, cuboid, cuneiforms and the metatarsals. The models were imported into a 3D CAD software (Geomagic Control, Morrisville, North Carolina, United States) in order to calculate the surface-to-surface interactions at the various joints. Color-coded distance maps, representing interarticular surface distances, were generated for the Ankle Joint, Subtalar Joint, Transverse Talar Joint (Chopart joint), Cuneonavicular joint, and the Metatarsocuneiform joint. The results from the deformed feet were compared to normal feet.

RESULTS: In HINDFOOT VALGUS (Fig. 1) the antero-medial side of the talus is closer to the medial malleolus as this bone moves into slight external rotation and plantarflexion. The calcaneus is externally rotated relative to the talus resulting in surface approximations on the lateral side of the posterior articular facet as well as an impingement of the sinus tarsi. In the Chopart joint a strong approximation on the superior side of the calcaneocuboid joint and the lateral side of the talonavicular joint occurs. At the cunei-navicular joint, similar to normal, there is a significant approximation of the lateral cuneiform to the navicular. At the metatarsals-cuneiform joints there is a strong approximation at the 2nd and 3rd metatarsals joint surfaces. In HINDFOOT VARUS (Fig. 1) the medial and lateral sides of the talus are closer to their respective malleoli and anteromedial side of the talonavicular joint moves into inversion and slight dorsiflexion. The calcaneus is internally rotated and inverted relative to the talus resulting in surface approximations on the medial side of the calcaneus bridging the gap between the posterior and middle articular facet and at the lateral side of the posterior articular facet of the calcaneus. The cuboid is displaced inferior to the navicular resulting in contact on the lateral facet of the calcaneocuboid joint. Also, a strong approximation at the talonavicular joint can be observed on the medial side. At the cunei-navicular joint, the lateral cuneiform surface is displaced away from the navicular surface compared to normal.

DISCUSSION: Current 2D1,2 and 3D measurements based on WBCT3 can provide accurate classification of hindfoot varus and valgus deformities. The application of distance mapping in this study was successful in characterizing the specific surface-to-surface interactions at the joints. The observations based on these biometric measurements show that there are significant differences in surface-to-surface interaction between the hindfoot deformities. These differences may help explain certain pathologies associated with hindfoot varus and valgus deformities such as sinus tarsi impingement, 2nd and 3rd metatarsalgia associated with valgus deformity, and the frequent association of hallux valgus with pes plano valgus.2

SIGNIFICANCE AND CLINICAL RELEVANCE: Color-coded distance maps applied to WBCT provide an effective tool to assess the effect of hindfoot deformities on articular joint surface interaction in the foot and ankle. The application of these maps may be used to enhance diagnostics and in the future help evaluate the effectiveness of therapy in restoring normal alignment.

Session 11: Hallux Rigidus and Hallux Valgus

Co-Moderators: Craig Payne, B. Pod (La Trobe University) and Don Anderson, PhD (University of Iowa)

3:00PM 11-1: Netto, Cesar de Cesar, et al. First Tarsometatarsal Joint Shape and Orientation: Can We Trust in Our Radiographic Findings?


3:30PM 11-4: Morgan, Oliver, et al. Cheilectomy and Moberg Osteotomy: A Biphasic Prediction of First Metatarsophalangeal Joint Stress

3:40PM 11-5: Kimura, Tadashi, et al. Evaluation of joint mobility around the cuneiform between hallux valgus & normal feet using 3D analysis system & weight-bearing CT

First Tarsometatarsal Joint Shape and Orientation: Can We Trust in Our Radiographic Findings?

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1 Hospital for Special Surgery, NY, 2 University of Alabama at Birmingham

NOTHING TO DISCLOSE

INTRODUCTION

Studies have demonstrated that patients with hallux valgus (HV) deformities have increased mobility in the first tarsometatarsal (TMT) joint. Anatomical factors widely considered to play a role in the instability are shape and frontal plane orientation of the joint. An oblique rather than horizontal orientation of the articular surfaces and a round shape, rather than a flat shape, are believed to predispose to the deformity. The purpose of this study was to assess whether the shape and angulation of the first TMT joint are affected by the positioning of the foot and orientation of the x-ray beam.

METHODS

Ten adult above knee fresh frozen cadaveric specimens were used, with a mean age of 79.9 (range, 54-88) years. There were no clinical forefoot deformities noted in any of the feet. One of the specimens had moderate ankle arthritis and one had a mild cavus-varus. A radiolucent loading apparatus was built that, allowing neutral positioning of a plantigrade foot and controlled angulation of 5°, 10°, 15° and 20° in dorsiflexion, plantarflexion, inversion and eversion. Fluoroscopic images were obtained of each cadaveric specimen in all seventeen different positions, with the x-ray beam perpendicular to the floor and aiming to the base of the 1st metatarsal. Two blinded orthopaedic surgeons independently measured the 1st tarsometatarsal (TMT) joint angle, where values above 90° represented increased valgus or abduction alignment of the 1st TMT and graded the distal articular cartilage of the medial cuneiform as flat or curved. Readers also graded the image quality into assessing the joint into “Low”, “Intermediate” and “Good”.

RESULTS

The mean value for 1st TMT joint angle was 112.92°± 6.89°. Values were significantly different between the cadaveric specimens (p<.0001) and ranged from 96.7° to 129.98°. There was a tendency for increased valgus angulation of the joint in images positioned in neutral, plantarflexion and inversion and decreased valgus angulation with dorsiflexion and eversion.

Regarding the shape of the distal articular cartilage of the medial cuneiform, joints with flat configuration showed significantly increased mean 1st TMT joint angle when compared to curved surfaces (115.9° vs. 110.7°, p<.0001). There was also a tendency for flattening of the joint in images positioned in dorsiflexion and inversion.

In 8 out of 10 of the cadaveric specimens (80%) the shape of the 1st TMT joint changed between curved or flat configuration depending on the positioning of the foot. In only 2/10 (20%) the joint configuration remained the same for all different positions (one flat and one curved)

Image quality for visualization of the 1st TMT joint was progressively better for increased angles of dorsiflexion and inversion and progressively worse for plantarflexion and eversion.

CONCLUSION

Our cadaveric study found that the shape and angulation of the first TMT joint is affected by the positioning of the foot and orientation of the x-ray beam.

CLINICAL RELEVANCE: Clinical usefulness of the 1st TMT radiographic anatomical characteristics is limited and should not influence in the treatment of patients with possible instability the first tarsometatarsal (TMT) joint.
Great Toe Adduction Decreases Blood Flow to Plantar Fascia: A Pilot Study

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**INTRODUCTION:** Plantar fasciitis has been reported to be the most common foot condition diagnosed by medical professionals. However, in spite of its prevalence, the cause of plantar fasciitis remains unclear. Most researchers agree that the etiology is multifactorial with mechanical and physiological components, but further investigation of the development of plantar fasciitis is needed in order to help clinicians develop effective treatment plans. The plantar fascia, along with several intrinsic foot muscles, receives blood supply from the medial and lateral plantar arteries. Physicians have theorized that an adducted great toe, as seen inside a narrow shoe, may put passive tension on the abductor hallucis. Since the medial and lateral plantar arteries run deep to the abductor hallucis between the muscle and the calcaneus, tensing of this muscle may put pressure on the plantar arteries, decreasing blood flow to the plantar fascia. The purpose of this study was to compare blood flow within the lateral plantar artery before and after passive great toe adduction.

**METHODS:** Ten subjects (5 male, 5 female, age: 29.2 ± 10.0 years, height: 177.5 ± 9.1 cm, weight: 74.8 ± 17.2 kg) volunteered to participate in this IRB approved study. Subjects were healthy and free of lower extremity injury. Blood flow was measured using pulse-wave ultrasound imaging (ML6-15 probe, GE Logiq S8). Subjects were seated with the ankle in 30° plantar flexion. The lateral plantar artery was imaged just deep to the abductor hallucis. Blood flow was first measured with the great toe in a resting neutral position. The great toe was then passively adducted with moderate pressure enough to cause visible tensing of the muscle, and blood flow was measured again. The diameter of the vessel was measured from the images, and the rate of blood flow was then calculated using the internal software of the ultrasound machine.

**RESULTS:** A paired t-test indicated the rate of blood flow was significantly lower post-adduction compared to the pre-adduction condition (Pre: 1.51 ± 0.76 ml/min, Post: 0.88 ± 0.61 ml/min, p=0.001) (Figure 1). There was a 42% decrease in the rate of blood flow after passive adduction.

**DISCUSSION:** Based on this preliminary data, it appears that passive adduction of the great toe may decrease the rate of blood flow within the plantar arteries. This decrease in blood flow is likely due to the tensed abductor hallucis putting pressure on the vessels. However, seeing as there were only ten subjects and was no control condition, further investigation is needed to confirm these findings. Diminished blood flow in combination with mechanical stress on the plantar fascia could lead to poor healing and contribute to the development of plantar fasciitis. Future research will aim to determine the effect of footwear on great toe position and plantar artery blood flow in order to draw conclusions of clinical significance.

**SIGNIFICANCE/CLINICAL RELEVANCE:** The results of this pilot study suggest there may be a relationship between an adducted great toe position and decreased blood flow to the plantar fascia. This decrease in blood flow, if prolonged, could ultimately result in the development of plantar fasciitis or the prevention of its healing.

**FIGURES AND TABLES:**

Figure 1:

![Mean Plantar Artery Blood Flow Before and After Passive Great Toe Adduction](image-url)
Effect of Plantar Fascia Stiffness upon First Metatarsophalangeal Joint Stress – A Biphasic Finite Element Study

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Disclosures: Rajshree Hillstrom (Moximed), Oliver J Morgan (N), Matt Koff (N), Scott J. Ellis (Wright Medical), Jonathan T Deland (N), Howard J Hillstrom (Moximed, APOS, JAK Tool)

INTRODUCTION: The plantar fascia is considered an important architectural element for maintaining the height of the arch.1 Excessive strain of this tissue promotes bony remodeling of the calcaneus (heel spur) and one of the most common pedal pathologies: plantar fasciitis. Prolonged stress will limit 1st metatarsophalangeal (MTP) dorsiflexion and may impact the magnitude and location of stress within the 1st MTP joint. Wright (1964) initially described the plantar fascia as a linearly elastic element with a 350MPa modulus.2 This study aims to model the 1st MTP joint with viscoelastic representations of cartilage to estimate joint contact stress with the plantar fascia moduli at 200%, 100%, 50%, and 25% of nominal (350MPa).

METHODS: Three-dimensional (3D) subject-specific finite element (FE) models of the 1st MTP joint were created from MRI-datasets of a cadaveric first ray (metatarsal, proximal phalanx, sesamoids, and hallux), without OA, and were previously developed and compared with in vitro testing to provide experimental validation. Tissues were segmented in Mimics (Materialise, Belgium), assembled in CATIA (Dassault Systèmes, France), meshed in Abaqus (Dassault Systèmes, France), and loaded with axial force and bending moments in FEBIO (Utah, USA). Bones were modelled as rigid bodies for this quasi-static analysis. Cartilage was modeled as viscoelastic (E=10MPa, v=0.07, permeability=0.002). Fourteen ligaments (tension-only ‘wires’: E=260MPa) and the plantar fascia (E=87.5 MPa - 700 MPa, v=0.4) were also modelled. Vertical forces of were applied to the hallux and sesamoid bones, respectively, to simulate a quarter of physiological loading conditions experienced during gait (Hillstrom, 2013). A horizontal force was applied to the plantar fascia to achieve static equilibrium. The plantar fascia modulus was evaluated at 200%, 100%, 50%, and 25% of nominal (350MPa).

RESULTS: Peak von Mises stress distributions on the proximal phalanx base cartilage surface (presented in Figure 1) for plantar fascia moduli of 700MPa, 350MPa, 175MPa, and 87.5MPa are 1.66 MPa, 1.67 MPa, 1.74 MPa and 1.90 MPa, respectively. The corresponding peak von Mises Stress at the cartilage-subchondral bone interface are 3.0 MPa, 3.0 MPa, 3.15 MPa and 3.44 MPa, respectively.

DISCUSSION: Arch height was inversely related to plantar fascia moduli.1 This investigational team has recently demonstrated that peak von Mises stress increases by 15% when plantar fascia modulus is 25% of nominal. Taken together with the results of this study it is logical that Peak stress would increase with decreasing plantar fascia moduli.

REFERENCES:
1. Tak-Man Cheung et al, Clinical Biomechanics 19, 2004

Figure 1: Effect of plantar fascia stiffness on first metatarsophalangeal joint von Mises stress
Cheilectomy and Moberg Osteotomy: A Biphasic Prediction of First Metatarsophalangeal Joint Stress

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Disclosures: Oliver J Morgan (N), Howard J Hillstrom (Moximed, Apos, Jak Tool), Matt Koff (N), Scott J. Ellis (Wright Medical), Jonathan T Deland (N), Rajshree Hillstrom (Moximed)

INTRODUCTION: First metatarsophalangeal (1st MTP) joint osteoarthritis (OA) is the most common degenerative joint disease in the foot (1). Cheilectomy combined with proximal phalangeal dorsiflexion (Moberg) osteotomy is a joint sparing procedure for mild to moderate disease. The purpose of combining a cheilectomy with a Moberg osteotomy is to shift the area of joint contact and thus, reduce edge-loading post-cheilectomy; however, 1st MTP joint contact mechanics after the preferential Moberg osteotomy wedge geometry (3 mm) is unknown. Therefore, the aim of this study was to predict 1st MTP joint loading following cheilectomy with a Moberg osteotomy titration of 3 mm using a biphasic finite element (FE) model of the first ray.

METHODS: A three-dimensional (3D) cadaver-specific FE model of a first ray was developed. The specimen was imaged using a 3T MRI-scanner (metatarsal, proximal phalanx, distal phalanx, sesamoids, plantar fascia, and cartilages), and was free from OA. Each tissue was segmented in Mimics (Materialise, Belgium), assembled in Catia V5 (Dassault Systèmes, France), meshed in Abaqus 6.14 (Dassault Systèmes, France), and simulated in FEBio 2.6.4 (University of Utah, USA). To reduce computational expense the bones and ground were assigned as rigid bodies. The material properties of cartilage (E=10MPa, ν=0.07, perm=0.002), plantar fascia (E=350MPa, ν=0.4), and ligaments (tension-only fiber models, E=260MPa) were used for the remaining soft-tissue structures. Dorsal metatarsal head geometry was reduced by approximately, 30% to simulate a cheilectomy, and the proximal phalanx modified with Moberg osteotomy titrated to 3 mm. The metatarsal bone was then rotated in a physiological manner to simulate a planus arch-alignment (10° metatarsal declination angle). Loading conditions were determined firstly, at a quarter scale of maximum force parameters reported beneath the metatarsal head (33 N) and hallux (34 N) for pes planus (2), and secondly, the resulting plantar fascia force (225 N) in order to achieve static equilibrium.

RESULTS: The computationally-predicted von Mises (MPa) distributions at the proximal phalanx cartilage surface for each simulation are shown in Fig 1. A cheilectomy and Moberg osteotomy titrated to 3 mm presented greater focal stress, as well as a more plantar distribution of cartilage loading for this specimen.

DISCUSSION: The analyses demonstrated a positive correlation between plantar shifting of joint contact and Moberg osteotomy titration as well as an increase in the magnitude of peak stress post-virtual surgery. There are several limitations with this experiment, 1) only one model was analyzed, 2) the material properties were not specimen-specific including an increased elastic modulus for cartilage and, 3) the plantar fascia force was assumed the same between each simulation and as such, may not reflect the concomitant changes to the plantar fascia load and likely, maximum force parameters following Moberg osteotomy.

SIGNIFICANCE/CLINICAL RELEVANCE: For this specimen, a Moberg osteotomy titration of 3 mm will plantarly load the cartilage. A larger dataset will be used in the future to better predict preferential loading of 1st MTP joint cartilage post-cheilectomy and Moberg Osteotomy.

REFERENCES:

ACKNOWLEDGEMENTS: Hospital for Special Surgery Foot and Ankle Fund.

Figure 1. von Mises distributions in the proximal phalanx base cartilage (posterior view) from left to right: intact, cheilectomy and no osteotomy, and cheilectomy and 3 mm osteotomy.
Evaluation of joint mobility around the cuneiform between hallux valgus and normal feet using 3D analysis system and weight-bearing CT.

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Disclosures: Tadashi Kimura (N), Makoto Kubota (N), Hidekazu Hattori (N), Kazuhiko Minagawa (N), Naoki Suzuki (N), Asaki Hattori (N), Keishi Marumo (N)

INTRODUCTION: An association has been reported between hallux valgus and hypermobility of the first ray, but hypermobility of the intercuneiform 1-2 joint was also suspected in some cases. However, dynamics of the intercuneiform 1-2 joint has seldom been investigated. This study used weightbearing computed tomography (CT) and a 3-dimensional (3D) analysis system to evaluate displacement of the intercuneiform 1-2 joint, intercuneiform 2-3 joint, and second cuneonavicular joint due to weightbearing in hallux valgus and normal feet.

METHODS: Patients were 11 women with hallux valgus (mean age, 56 years; mean hallux valgus angle, 43 degrees; mean first-second intermetatarsal angle, 22 degrees) and 11 women with normal feet (mean age, 57 years; mean hallux valgus angle, 14 degrees; mean first-second intermetatarsal angle, 9 degrees). Each patient was placed supine with the lower limbs extended, and CT was performed under nonweightbearing and weightbearing conditions (load equivalent to body weight)(Fig 1). 3D models reconstructed from CT images (Fig 2). Then, 3D analysis was performed to quantify the displacement of the middle cuneiform relative to the medial cuneiform and the displacement of the lateral cuneiform relative to the middle cuneiform under nonweightbearing and weightbearing conditions. We compared data between the control group and the hallux valgus group.

RESULTS: Relative to the medial cuneiform, the middle cuneiform was displaced by 0.1 and 0.8 degrees due to dorsiflexion, 0.2 and 1.0 degrees due to inversion, and 0.7 and 0.7 degrees due to abduction in normal feet and feet with hallux valgus, respectively, with the latter having significantly greater dorsiflexion (P = .0067) and inversion (P = .0019) (Fig 3a). There was no significant intergroup difference at the intercuneiform 2-3 joint and second cuneonavicular joint (Fig 3b,3c).

DISCUSSION: We conducted a detailed three-dimensional evaluation of cuneiform mobility, which has been difficult to evaluate by other methods such as plain radiography. The findings of this study showed that relative to the medial cuneiform, the middle cuneiform was significantly displaced due to dorsiflexion and inversion under weightbearing conditions in patients with hallux valgus, suggesting that hallux valgus also involves hypermobility at the intercuneiform 1-2 joint.

SIGNIFICANCE: It may be possible to further improve postoperative outcomes of the modified Lapidus procedure through arthrodesis of the intercuneiform 1-2 joint, especially in patients with severe hypermobility of this joint.

REFERENCES:
Comparing First Metatarsophalangeal Joint Flexibility Measurements in Hallux Rigidus Patients Pre- and Post-Cheilectomy Using a Novel Flexibility Device

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Disclosures: No disclosures.

INTRODUCTION: Few authors have studied flexibility of the first metatarsophalangeal joint (MTPJ), which has been shown to be decreased in patients with hallux rigidus (HR). The most widely used classification system for HR, which uses only range of motion and radiographic findings to guide treatment, may give an incomplete picture of the involved pathology.

In a previous study (Phase I), we described a method to measure first MTPJ flexibility using a custom jig and tested it in both HR patients and controls, with excellent intra- and interrater reliability. In this study (Phase II), our objective was to measure first MTPJ flexibility of the same HR patients post-cheilectomy to determine the effect of surgery on joint flexibility.

METHODS: This study was approved by the IRB at the authors’ institution and informed consent was obtained from all enrolled patients. Thirteen hallux rigidus patients from Phase I were contacted to participate. Ten patients were enrolled and underwent MTPJ flexibility testing. The operated first MTPJ was manually loaded to maximum dorsiflexion and then back to neutral cyclically five times for each trial. Prior to each trial, the joint was cyclically loaded ten times to pre-condition the soft tissues. Testing was performed both sitting and standing. Early and late flexibility [º/lb-inch], laxity angle [º], laxity torque [lb-inch], torque angle [º], and maximum dorsiflexion angle [º] were compared between the preoperative and postoperative datasets for six patients using paired-samples 1-tailed t-tests with significance set at alpha = .05. Only data for six patients were used due to a technical problem affecting four datasets.

RESULTS: Late flexibility (P < 0.022), maximum dorsiflexion (P < 0.010), and laxity angle (P < 0.006) were significantly improved postoperatively in the seated position (Table). Early flexibility and laxity torque did not differ significantly between the pre- and postoperative conditions in the seated position. No measurements were significantly different in the standing position.

DISCUSSION: Measures of flexibility were significantly improved after cheilectomy. This finding likely reflects improved function of the MTPJ after removal of dorsal impingement, but could also be due to the effects of releasing the joint capsule. The reason why improvement was noted only in the seated position is not clear, but may be related to increased soft tissue tension in the standing position. Future studies will be needed to confirm these results with more patients and to determine the effect of different procedures. In addition, it will be helpful to correlate other outcome measures with measures of laxity.

SIGNIFICANCE: This is the first study to demonstrate improvement in first MTPJ flexibility after cheilectomy. These measures may help surgeons better gauge function compared to traditional radiographic and motion parameters.

Table: P values for Comparisons between Pre- and Post-Operative Patients

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Pre-Op</th>
<th>Post-Op</th>
<th>Delta</th>
<th>p-value</th>
</tr>
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<tbody>
<tr>
<td>Early Flexibility</td>
<td>0.52 (0.24)</td>
<td>0.71 (0.61)</td>
<td>0.19 (0.61)</td>
<td>0.240</td>
</tr>
<tr>
<td>Late Flexibility</td>
<td>0.18 (0.09)</td>
<td>0.10 (0.04)</td>
<td>0.08 (0.07)</td>
<td>0.022</td>
</tr>
<tr>
<td>Max Dorsiflexion</td>
<td>32 (11)</td>
<td>51 (16)</td>
<td>19 (14)</td>
<td>0.010</td>
</tr>
<tr>
<td>Laxity Angle</td>
<td>20 (9)</td>
<td>33 (10)</td>
<td>13 (8)</td>
<td>0.006</td>
</tr>
<tr>
<td>Laxity Torque</td>
<td>72 (24)</td>
<td>94 (55)</td>
<td>22 (44)</td>
<td>0.136</td>
</tr>
<tr>
<td>Early Flexibility</td>
<td>0.16 (0.09)</td>
<td>0.16 (0.13)</td>
<td>0.00 (0.13)</td>
<td>0.466</td>
</tr>
<tr>
<td>Late Flexibility</td>
<td>0.14 (0.07)</td>
<td>0.11 (0.04)</td>
<td>0.03 (0.07)</td>
<td>0.180</td>
</tr>
<tr>
<td>Max Dorsiflexion</td>
<td>21 (10)</td>
<td>33 (18)</td>
<td>11 (15)</td>
<td>0.062</td>
</tr>
<tr>
<td>Laxity Angle</td>
<td>13 (6)</td>
<td>24 (16)</td>
<td>11 (15)</td>
<td>0.074</td>
</tr>
<tr>
<td>Laxity Torque</td>
<td>148 (36)</td>
<td>242 (161)</td>
<td>93 (165)</td>
<td>0.112</td>
</tr>
</tbody>
</table>

P < 0.05 are bolded.

Figure: Flexibility Device
Session 12a: Poster Teasers

4:00PM
Session 12b: Posters

4:15PM

Co-Moderators: Irene Davis, PT, PhD (Spaulding Rehab, Harvard University) and Rajshree Hillstrom, PhD, MBA (Anglia Ruskin University)

Poster No. 1: Roberts, Lauren, et al. Ankle Fusion Percutaneous Home Run Screw Fixation: technical aspects and soft tissue structures at risk


Poster No. 4: Siegler, Sorin, et al. Validation of a Subject-specific Computational Models of the Ankle Joint Complex

Poster No. 5: Sturnick, Daniel R., et al. The Function Axis of Rotation of the Ankle Joint during Simulated Gait

Poster No. 6: Hatton, Anna L., et al. Textured shoe insoles to improve balance & walking in adults with diabetic peripheral neuropathy: study protocol for a randomised controlled trial

Poster No. 7: Marc, Janin. & Philippe, Dupui. Foot Orthoses reduce repercussions of Ns4 Noxious stimulus

Poster No. 8: Marc, Janin. Foot Function & Sensorimotor Orthoses of Ns4 Noxious stimulus

Poster No. 9: Breard, Thomas. & Janin, Marc. Posturodynamic-6 Perform Sport relate influences performance and competitiveness in sports

Poster No. 10: McClymont, Julie. The magnitude and spatial distribution of variability in plantar pressure at a wide range of walking speeds

Poster No. 11: Oakman, Joyann, et al. Reliability of New Forefoot-Rearfoot Measurement


Poster No. 13: Telfer, Scott. & Bigham, J. The Effect of Age and Disease on Regional Plantar Loading: A Systematic Review and Meta Regression Analysis
Poster No. 14: Turner, Robert. Finding the Kinematic Driver


Poster No. 16: Chicoine, Dominic. Customized foot orthoses in the treatment of posterior tibialis tendon dysfunction

Poster No. 17: Tulchin-Francis, Kirsten, et al. Plantar pressures in patients with symptomatic flexible flatfoot: How are they different than adolescents with asymptomatic flatfoot?

Poster No. 18: Giacomozzi, Claudia. Testing Muscle Activation and In-shoe Foot Loading Under Repeatable Conditions: A Stepper-based Approach


Poster No. 20: Franzese, Richard C., et al. Velocity, footwear, and foot-strike angle effects on lower-leg muscle activity during running

Poster No. 21: Freedman, Happy. & Serotta, B. Intro to bike road fitting

Poster No. 22: Greene, Andrew J. & De Paula, A. Are you experienced yet?

Poster No. 23: Mahaffey, Ryan, et al. Effects of acute fatigue of the lower limb on running mechanics


Poster No. 26: Roney, Andrew R., et al. Knee Adduction Moments Associated with Knee Osteoarthritis are Increased by Medial Arch Supports

Poster No. 27: Cha, Seungwoo, et al. Weight bearing with standing position for tibiofibular clear space measurement; using 3D US

Ankle Fusion Percutaneous Home Run Screw Fixation: technical aspects and soft tissue structures at risk

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Disclosures: No relevant disclosures.

INTRODUCTION:
During internal fixation of ankle fusions, besides the standard crossed screw fixation pattern, the use of a percutaneously placed augmenting screw, directed from the posterolateral tibial metaphysis proximally across the ankle into the talar neck ("ankle fusion home run screw"), is a widely used technique. The placement of this screw is technically demanding and multiple attempts under fluoroscopy guidance are frequently needed to achieve a perfect positioning of the implant. Injuries to local neurovascular and tendinous structures may occur. The objective of this cadaver study was to identify the number of attempts necessary for a perfect positioning of the ankle fusion home run screw and to identify the neurovascular and tendinous structures at risk.

METHODS:
Eleven fresh frozen cadaver limbs were used. Guide wires (3.2mm) from the Stryker (Selzach, Switzerland) 7.0-mm headless cannulated set were percutaneously placed into the distal posterolateral aspect of the leg by a Fellowship Trained Foot and Ankle Surgeon, under fluoroscopic guidance, with the ankle in neutral position. Malpositioned pins were not removed and served as guidance for the following pins. The number of guide wires needed to achieve an acceptable positioning of the implant (pin centered on the axis of the talar neck) was noted. After a layered dissection from the skin to the tibia, we evaluated neurovascular and tendinous injuries, and measured the shortest distance between the closest guide pin and the soft tissue structures, using a precision digital caliper.

RESULTS:
The mean number of guide wires needed to achieve and acceptable positioning of the implant was 2.09 (SD 0.83, range 1 - 4). The mean distances between the closest guide pin and the soft tissue structures of interest were: Achilles tendon 6.90mm (SD 3.74mm); peroneal tendons 9.65mm (SD 3.99mm); sural neurovascular bundle 0.97mm (SD 1.93mm); posteromedial neurovascular bundle 14.26mm (SD 4.56mm). Sural bundle was in contact with the guide pin in 5/11 specimens (45.5%) and transected in 3/11 specimens (27.3%).

CONCLUSIONS:
The placement of percutaneous ankle fusion home run screws is technically demanding and multiple guide pins are needed. Our cadaveric study showed that important tendinous and neurovascular structures are in close proximity with the guide pins and that the sural bundle is injured in approximately 73% of the cases.

SIGNIFICANCE/CLINICAL RELEVANCE:
Neurovascular injury on any scale negatively affects outcomes. Thus, small open incisions should be made to dissect safely down to bone and protect surrounding neurovascular structures while placing screws in ankle fusion.

Figure 1:
A) Image on the left shows a radiograph of the trajectory of the home run screw.
B) Image on the right demonstrates the proximity of the guidewire to surrounding posterior lateral structures.
Intraoperative Syndesmotic Instability Test: An Alternative Technique

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Disclosures: No relevant disclosures.

INTRODUCTION:
Precise diagnosis of distal tibiofibular syndesmotic injury is challenging, and a distinction should be made between syndesmotic ligament disruption and real syndesmotic instability. Tibiofibular clear space identified on radiographic imaging is considered the most reliable indicator of the injury as it is not significantly influenced by tibial rotation. A clear space greater than 6 mm at a point 1 cm above the tibial plafond is suggestive of injury. The Cotton test, (or Hook test), is the most widely used intraoperative technique to evaluate the syndesmotic integrity. We advocate an alternative technique using a 3.5mm blunt cortical tap.

METHODS:
Nine fresh-frozen cadaveric specimens were used with mean age of 79 (range, 54-88) years. First, a 2.5mm hole was drilled percutaneously on the lateral aspect of the distal fibula, in position for possible placement of a syndesmotic screw or suture button. A 3.5mm cortical tap was then threaded in the hole. For each specimen, three sequential fluoroscopic mortise images were taken. The first image was with the syndesmotic ligaments intact and no force applied to the tap (intact, non-stressed). In the second, with the ligaments intact, the cortical tap was advanced until its blunt tip was pushing against the lateral tibial surface, thus providing a tibiofibular separation force (intact, stressed). The third one was acquired after the same stress was applied to the tibia through the tap, but all syndesmotic ligaments were released (injured, stressed). Tibiofibular clear space was measured twice, 1 cm above the tibial plafond, by two independent viewers. Non stressed and stressed measurements were compared by Student’s t-test. Intra and inter-observer agreements were evaluated by intra-class correlation coefficient (ICC). P-values <.05 were considered significant.

RESULTS:
We found excellent intra-observer (ICC 0.97) and inter-observer (0.98) agreement following the imaging assessment. The mean values for the tibiofibular clear space were: 4.21 ± 1.16mm in the intact, non-stressed ankles; 4.49 ± 1.25 mm in the intact, stressed ankles; and 7.10 ± 1.05 mm in the injured, stressed ankles. Significant differences were found in the paired comparison between the groups (p<.05). Our novel syndesmotic instability test has demonstrated a 67% sensitivity, 78% negative predictive value 100% specificity and 100% positive predictive value in diagnosing syndesmotic instability.

CONCLUSIONS:
Our cadaveric study showed that this novel syndesmotic instability test using a 3.5mm blunt cortical tap is a simple and reliable technique. The test was able to demonstrate significant differences in the tibiofibular clear space when comparing normal ankles without stress, normal ankles with stress, and complete injury of the syndesmotic ligaments with stress. It represents a viable, simple, quantitative, and low-cost alternative to the most used Cotton test. Furthermore, the hole that is made for this test can be used in the event that syndesmotic fixation is necessitated, or simply a bi-cortical fibular screw may be placed in this hole if the syndesmosis does not require fixation.

SIGNIFICANCE/CLINICAL RELEVANCE:
Accurately diagnosing syndesmotic instability remains challenging. This novel syndesmotic instability test shows improved sensitivity and reliability to optimize the diagnosis of instability intra-operatively thus accurately guiding intraoperative fixation of the syndesmosis to allow for best possible patient outcomes.

FIGURE 1. Fluoroscopic images of an ankle demonstrating an increase in tibia fibula clear space with the tap test. 3.5mm blunt cortical tap seen in the top right of each of the images entering through the fibula and against the tibia.
Accuracy of Transarticular Lateral Soft Tissue Release of the 1st Metatarsophalangeal Joint – A Cadaver Study

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Disclosures: No relevant disclosures.

INTRODUCTION:
The release of contracted lateral soft tissue structures of the first metatarsophalangeal joint is frequently part of the surgical treatment of hallux valgus deformity. The 1st intermetatarsal space open dorsal approach and the single medial incision transarticular approach represent possible options. Advantages of the transarticular approach include avoidance of a second incision and a theoretically lower risk of 1st metatarsal head AVN. However, the inherently limited visualization of the structures through this approach might limit its effectiveness. The objective of this study was to evaluate the accuracy of hallux valgus lateral soft tissue release through the transarticular approach.

METHODS:
Ten bellow-knee fresh-frozen cadaveric specimens were used (6 females/4 males, mean age, 73.4 years), including two specimens with moderate hallux valgus deformity. None of the specimens had considerable degenerative changes of the 1st MTP joint. Lateral release was performed by the same Fellowship Trained Foot and Ankle Surgeon through a single medial approach of 2.5cm with a no15 scalpel blade. Surgical aim was to release four 1st MTP joint complex structures: lateral collateral ligament, lateral capsule, adductor hallucis muscle tendon and lateral metatarsoesamoid suspensory ligament. Once completed, a lateral extended dissection of the 1st - space was performed. Accuracy was graded in accordance to the number of structures successfully released: 0% (no structures), 25% (1/4), 50% (2/4), 75% (3/4) and 100% (4/4). Inadvertent injuries to soft tissue structures (flexor hallucis brevis and longus tendons, deep transverse metatarsal ligament and first intermetatarsal neurovascular bundle) and articular cartilage of 1st metatarsal head and proximal phalanx were recorded.

RESULTS:
The surgical accuracy for lateral soft tissue release of the 1st MTP joint through the transarticular medial approach was 100% in 7 cadaveric specimens, and respectively 75%, 50% and 25% in the other 3 specimens. The lateral collateral ligament was successfully released in all cadavers. The lateral joint capsule, adductor hallucis muscle tendon and lateral metatarsoesamoid suspensory ligament were released in 80% of the specimens. Chondral damage of the 1st metatarsal head and unintended release of the lateral head of the flexor hallucis brevis occurred respectively in 40% and 50% of the procedures. No injuries to the flexor hallucis longus tendon, neurovascular bundle, deep transverse metatarsal ligament and chondral damage of the proximal phalanx were recorded.

CONCLUSIONS:
Our cadaveric anatomical study has shown a high accuracy in the release of specific lateral soft tissue structures of the 1st MTP joint through a medial transarticular approach. Inadvertent release of the lateral head of the flexor hallucis brevis and iatrogenic chondral damage of the 1st metatarsal head are complications to be considered. Limitations include a small sample size and inherent differences between live and cadaveric tissue.

SIGNIFICANCE/CLINICAL RELEVANCE:
The medial transarticular approach for distal soft tissue release in a bunion correction can accurately release the specific relevant lateral soft tissue structures of the 1st MTP joint. Performing the release in this way limits the number of incisions and removes the potential risks of a 1st web space incision thus potentially decreasing morbidity to the patient.
Validation of a Subject-specific Computational Models of the Ankle Joint Complex
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1Drexel University, Philadelphia, PA, USA, 2Movement Analysis Laboratory, Istituto Ortopedico Rizzoli, Bologna, Italy
Disclosures: Sorin Siegler, Vishnuvardhan Balakrishnan, Claudio Belvedere, Paolo Caravaggi, Alberto Learndini (N–all)

INTRODUCTION: Three dimensional (3D) image based, subject specific models of the ankle complex, can be a useful predictive and planning tool in a variety of clinical and biomechanical applications such as diagnosis of ligament injuries and evaluation of surgical reconstructive procedures. However, few computational models can produce the full 3D biomechanical properties of the ankle complex, particularly on a subject-specific basis. Our group introduced one such computational model in 2004 (1). This model was partially validated against experimental data. In the current study, we have improved this model and validated it, on a subject-by-subject basis, using a wide range of 3D biomechanical properties.

METHODS: Five cadaveric lower limb specimens were used in this study. Each specimen was CT-scanned and then tested on a special linkage where they were loaded in three dimensions and the applied loads and associated displacements were recorded (2). Using the imaging data, modified 3D models of the ankle complex used by us in the past (1) were then produced for each specimen. These models included the 3D bone morphology, attachment of the surrounding ligaments, and contact mechanics associated with cartilage mechanical properties. The main improvements in the current models included the specific distribution of cartilage thickness at the ankle rather than a uniform thickness distribution and pre-straining the ligaments to values reported in the literature (3). The evaluation consisted of a subject specific comparison using repeated measure analysis of the following biomechanical parameters: rotational range of motion (ROM) in 3D; three dimensional displacement-load curves and total laxity (ROM/Max torque); surface-to-surface interaction based on distance maps at the ankle and subtalar joints. The straining during 3D motion of the ATFL and CFL obtained through the models were compared against published literature (4).

RESULTS: The results show good agreement in ROM, limited to less than 2 degrees difference between simulation and experiment in dorsiflexion/plantarflexion, 0.3 degrees in inversion/eversion and less than 2.5 degrees in internal/external rotation. Total laxity showed good agreement between simulations and experiments limited to less than 10% while the displacement vs. load curves obtained from the simulations showed similar trends with typical stiffness increase towards the end of ROM and hysteresis (Figure 1). Distance maps produced in neutral and at the extreme of the ROM in all direction showed good agreement between the simulations and the experiments (Figure 2).

DISCUSSION: The data obtained from a subject-specific computational model was compared, on a subject specific basis, to experimental data from cadaver specimens and produced similar results over a wide range of biomechanical parameters, confirming the reliability of this model. Such model improves the state-of-the-art in foot and ankle modeling by enabling reliable simulations of a variety of 3D events.

SIGNIFICANCE AND CLINICAL RELEVANCE: This image-based, subject specific model can be used as a predictive tool to aid in diagnosing and surgical planning for a variety of ankle disorders such as the effect of various ligament injuries and the effect of variations in the surface geometries of total ankle replacements on biomechanical behavior.


Figure 1 – Displacement vs. load from one typical simulation dorsiflexion/plantarflexion

Figure 2 – Distance maps comparison simulation vs. experiment
The Function Axis of Rotation of the Ankle Joint during Simulated Gait

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INTRODUCTION: Implant component positioning is considered as an important factor in function and longevity in total ankle arthroplasty (TAA). However, accurate and repeatable positioning remains a limitation with current techniques and instrumentation. In addition, further investigation is needed to objectively define the optimum component positioning. Cadaveric gait simulation is a valuable tool for investigating foot and ankle joint mechanics during functional tasks such as the stance phase of gait. The objective of this study was to investigate the functional axis of rotation of the native ankle joint during simulated gait.

METHODS: The stance phase of healthy gait was simulated with six mid-tibia cadaveric specimens using a previously validated device and methodology. A robotic platform reproduced tibial-ground kinematics by moving a force plate relative to the stationary specimen while physiologic loads were applied to the extrinsic tendons to actuate the foot. (Figure 1A). Ankle kinematics were measured from reflective markers attached to the tibia and talus via surgical pins. The helical axes of rotation of the talus with respect to the tibia was calculated during three portions of stance: initial plantarflexion during earlier-stance after heal strike, dorsiflexion during mid-stance, and final plantarflexion during late-stance. The position and orientation of these kinematic-defined axes of rotation were compared to the transmalleolar axis and reduced to its anteroposterior position (Figure 1B).

RESULTS: Analyses revealed that ankle joint functional axis of rotation varied from the anatomic reference throughout stance. The kinematic center of rotation was located 16.4 ± 5.8 mm, 16.5 ± 6.6 mm, and 15.6 ± 6.5 mm anterior to the transmalleolar axis during early-, mid- and late-portions of stance, respectively.

DISCUSSION: This study revealed that the position of the flexion-extension axis varies greatly between specimens during simulated gait. While previous reports have suggested that the transmalleolar axis is an acceptable approximation for the ankle joint center, these findings suggest that further research in warranted to better describe the complex tibiotalar kinematics.

SIGNIFICANCE/CLINICAL RELEVANCE: This work may provide future insight to guide implant design and advance techniques, to better place articular constraints of a total ankle in the native center of rotation of the joint.

A) 

B) 

A) 

B) 

 Axes of Rotation 

Early-Stance 

Mid-Stance 

Late-Stance 

Tibia 

Talus 

Superior 

Medial 

Anterior
Textured shoe insoles to improve balance and walking in adults with diabetic peripheral neuropathy: study protocol for a randomised controlled trial

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INTRODUCTION: Peripheral neuropathy is a major risk factor for falls, affecting the lower limbs of up to 86% of fallers with diabetes [1]. Nerve damage can disrupt vital sensory cues about the supporting surface and position of body segments, to help people remain upright. Innovative footwear devices which artificially manipulate the sensory environment at the feet, such as textured insoles, are emerging as an attractive option to help mitigate balance and walking problems in clinical populations [2, 3], by way of a ‘sensory training’ effect [4]. However, the therapeutic effects of textured insoles for adults with peripheral neuropathy remain unknown. This study will explore whether wearing textured shoe insoles can improve balance, walking, foot sensation, physical activity and reduce the risk of falls, in adults with diabetic peripheral neuropathy.

METHODS: A prospective, single-blinded randomised controlled trial with two parallel groups will be conducted of 70 community-dwelling adults with diabetic peripheral neuropathy, across Brisbane, Australia. Men and women with a diagnosis of peripheral neuropathy (secondary to Type 2 Diabetes), who are aged over 18 years, ambulant over 20m (with or without an assistive device), and meet specific inclusion criteria, will be recruited. Participants will be randomised to a smooth control insole (N=35) or textured insole (N=35) group. The allocated insole will be worn for 4 weeks within participants’ own footwear, with self-report wear diaries and falls calendars being completed over this period. Blinded assessors will conduct one baseline assessment and one 4-week post-intervention assessment. Participants will complete surveys addressing their self-perceived foot health (Foot Health Status Questionnaire), fear of falling (falls Efficacy Scale-International) and will be asked to rate the comfort of wearing their allocated insole (100m visual analogue scale). Habitual activity levels will be assessed using a wireless activity monitor (activPAL), worn for 7 consecutive days (prior to baseline and at Week 3). Lower limb sensory function will be assessed bilaterally, including light-touch pressure (monofilaments), vibration perception (neurothesiometer), and ankle joint proprioeception (internet-based goniometer). Static, bilateral standing will be assessed (AMTI force plate) over 30 seconds, under two visual (eyes open, eyes closed) and two surface (firm, foam) conditions (randomly presented). Level-ground gait will be evaluated by completing a 12m walk over an instrumented walkway (GAITRite® CIR Systems Inc.). Balance and gait tasks will be completed barefoot, wearing standardised shoes only, and with two different shoe insoles (smooth, textured). Ethical approval has been obtained from the Human Research Ethics Committee at The University of Queensland. Participants will provide written informed consent prior to enrolment.

RESULTS: The primary outcome measure will be centre of pressure path velocity and excursion in anteroposterior and mediolateral directions. Secondary outcome measures include spatiotemporal gait parameters, physical activity levels, perception of foot sensation and proprioeception. To establish any differences between the intervention and control groups for all outcome measures, a repeated measures mixed models approach will be undertaken using data at baseline and 4-weeks. Non-parametric tests will be used where data is not normally distributed. Participant characteristics (e.g. age, gender) will be included as covariates. Multiple regression will be used to determine any relationships between foot sensation and proprioeception, balance and gait. Group allocation will be concealed and all analyses conducted on an intention-to-treat basis.

DISCUSSION: There is an urgent need to develop more effective and multi-faceted falls prevention strategies for adults with diabetes, to help reduce escalating health care costs. This study will be the first to explore whether artificially manipulating plantar sensory information, using novel shoe insoles, can address balance and mobility problems in people with diabetic peripheral neuropathy. The findings will be used to inform the development of new, affordable, non-invasive neuropathic treatments, which specifically target diabetic foot sensory complications that can contribute to falls. Importantly, wearing simple shoe insoles have the capacity to promote self-management by the user and enhance independent living.

SIGNIFICANCE/CLINICAL RELEVANCE: This study will determine the efficacy of an innovative footwear device in targeting deficient plantar sensation associated with neuropathic damage, to address falls risk factors in people with diabetes.

REFERENCES:
Foot Orthoses reduce repercussions of Ns4 Noxious stimulus.

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INTRODUCTION: Previous studies reported that heterotopically painful stimulation could both depress the nociceptive spinal reflex (1). This effect could represent the neural basis for counter-irritation phenomena, that is, the paradoxical pain relief produced by heterotopic painful stimulation (2, 3). Noxious stimuli of type 4 (Ns4), condition labelled épine irritative, represents one of several counter-irritation phenomena and designates a nociceptive capacity of plantar irritating stimulus (4, 5). In fact, NS4 is a specific heterotopic nociceptive stimulus produces pain, no expressed by the patient, when all of the four following criteria are met: 1) score variation of clinical PDN-6 into hard and foam ground (6-8); 2) asymmetrical perception of pain on the Ns4 area (5, 8); 3) loss of spatial discrimination (5) ; and 4) loss of perception of somesthesia (9). We evaluate the efficacy of sensorimotor orthoses/insoles (SO, 9, 10) to reduce the repercussion in patients presenting with Ns4.

METHODS: 20 males and 20 females (29-54 years) with Ns4 (single-point on the first metatarsal head) participated to the study; half of the patients (10 male and 10 female participants) were fitted with SO while the rest was the control group (C). We collected score of pain intensity (v.a.s, 0-100mm), and somesthesia by comparison into Ns4 levels and forehead reference in terms of 2-point stimuli, delivered with two distances (10-20 mm). These measures were taken at baseline and after a period of SO use: 1-2-3-6-12 week’s adaptation period. ANOVAs and paired t-tests were used for statistics analyses.

RESULTS: Reduction is observed for pain sensitivity (fig 1) and the number of errors of discrimination of 2-point stimuli (fig 2) was observed. The reduction of sensation of nociceptive signal became perceivable from the 3rd week of use, but proved only significantly beneficial after 4 weeks of use for pain and 20 mm and 4 weeks use from 10 mm.

DISCUSSION: The use of SO induced a new repartition of plantar sensory field information and reduced the heterotopic nociceptive perception of Ns4, mediated by a complex loop involving supraspinal structures. The change in pain intensity diminished the nociceptive withdrawal reflex field size considerably. This effect optimized the fidelity of sensory discrimination by controlling the extent of the neuronal receptor field on decreasing control inhibition/facilitation. When induced by SO, pain stimuli were subject to greater inhibition, therefore improving the 2-point stimuli discrimination. Improvement in integration occurred earlier for 20 mm soles than 10 mm ones, certainly due to the modification of the plantar field in regard to the NS4 localization. Saliency detection is considered to have a pivotal role in sensory integration. The use of SO has the potential to induce a reduction of the Ns4 and to restore foot sole to a functional level. While the perception and the integration of somesthesia improved. Results need further investigations.

SIGNIFICANCE/CLINICAL RELEVANCE: modulation of pain of Ns4, heterotopic nociceptive stimuli, pain perception and somesthesia discrimination by SO influence. SO influence of the repercussion of Ns4 in the sensory integration.

REFERENCES:

Key-words: Foot orthoses, Epine Irritative, Heterotopic Noxious Stimuli, Ns4, podiatry.
INTRODUCTION: The Foot Function Index (FFI) is one of the five health measurement scales most frequently used in podiatry (1). It was developed to measure the impact of foot pathology on function in terms of pain, disability and activity restriction (2, 3). The FFI was specifically designed to assess the effects of foot orthotics treatments on foot-related problems (e.g Gross et al. for patients with plantar fasciitis (4), Powell et al. for children with juvenile idiopathic arthritis (5) and Wrobel et al. for plantar heel pain (6)). One such foot pathology is an heterotopic nociceptive stimulus entitled Noxious stimuli type 4 (Ns4), *épine irritative*: a nociceptive capacity of plantar irritating stimulus (7,8) characterized by 4 sine qua non criteria: 1) score variation of the clinical PDN-6 into hard and foam ground (8, 9); 2) asymmetrical perception of pain on Ns4 area (7, 9-11); 3) loss of spatial discrimination (7) and 4) loss of feeling of somesthesia (10,11). For podiatrists, the most critical clinical issue is that pain cannot be expressed directly by the patients because it falls under the pain threshold (7, 9, 10). Sensorimotor Orthoses/Insoles (SO) already proved potentially effective in reducing the impact of Ns4 and in improving postural performance of patient (10). A systematic literature search found no study using an established quality-of-life instrument to measure the impact of the noxious stimuli Ns4 and the effects of SO on Ns4. This study aims at evaluating the impact and efficacy of SO on patients with Ns4 with regards to pain thresholds, disability, and activity limitation.

METHODS: 30 males and 30 females (aged 29-54 years), all presenting Ns4 (localized as a single-point on the first metatarsal head), participated in the study. Every study participant completed the 3 subscales pain, disability and activity restriction of the FFI at screening, baseline, after 7 and after 21 days (11, French validated translation, 13). Bilateral custom-made SO were given to every patient (4, 10-12). The use of SO was expected to 1) reduce neurogenic cues (14), 2) change the muscular tone distribution (postural mechanical expression of sensory disorders), 3) improve the muscular chain integration of the patient’s and 4) limit the mechanical constraints, by stimulating the proprioceptors of the sole (15).

RESULTS: SO induced a reduction of FFI scores at baseline, 7 and 21 days, for: a) pain: 40 28 13 (up to 90), b) distability 31 26 17 (up to 90), c) activity limitation 12 12 8 (up to 50) and d) total maximum score 83 65 38 (up to 230).

DISCUSSION: This study proves that Ns4 negatively impact foot function for the first time. The influence of Ns4 on the FFI score is comparable to the influence of other foot pathologies (previous reports): scores on each subsections were lesser when wearing SO for all subjects. The effect of SO is comparable to the literature: pain and disability readings were better with SO, after 7 days and only after 21 days on activity limitation. There could be two explanations: a) Ns4 cannot be expressed by subjects which makes it more difficult to report; b) SO reduced the noxious expression of Ns4 but 21 days may be too short a time to observe its field impact. Also most patients reported their ability to perform everyday activities with less pain and restriction of the FFI at screening, baseline, after 7 and after 21 days (11, French validated translation, 13). Bilateral custom-made SO were given to every patient (4, 10-12). The use of SO was expected to 1) reduce neurogenic cues (14), 2) change the muscular tone distribution (postural mechanical expression of sensory disorders), 3) improve the muscular chain integration of the patient’s and 4) limit the mechanical constraints, by stimulating the proprioceptors of the sole (15).

SIGNIFICANCE/Clinical Relevance: Ns4, heterotopic nociceptive stimuli, affect foot function. This study supports that sensorimotor orthoses improved foot function of the patients with Ns4. SO also improves patients’ quality of life.

REFERENCES:

Key-words: Foot orthoses, épine irritative, Heterotopic Noxious Stimuli, Ns4, podiatry.
INTRODUCTION: One of the fundamental and primary human principles is to control our environment, to achieve this, you must perform certain functions. Among the most important is the balance function: postural control (seating, standing) and the dynamic control named locomotion (walking, running or jumping; 1). Successful locomotion depends on postural control to establish and maintain appropriate postural orientation of body segments relative to one another and to the environment and to ensure dynamic stability of the moving body, principally sport practice (2). This process critically depends on integration of sensory inputs and must operate within the limits of biomechanical constraints inherent to the individual and the task. The integration of the information and the motor responses are very complex. It require network of neuronal connection between the different structures of the central and peripheral nervous system and also involve other body systems (1, 2).

The sport context, the position and the materials (like shoes) influence the balance function (3) by different constraints (i.e: inside outside ground (gymnasium or green and sand); fencing, tennis, golf, kayak for participation of the foot sole and legs information’s; running, climbing, football or soccer for different contains applied by shoes). In podiatry practice, we use the Posturodynamic-6 clinical test to evaluate the balance function performance (3), through the postural control (automatic control), the stability with the one legs stance test and the lateral bending movement to assess motor control (4). Each sports induce specifics constraints When athlete practice, the context require an efficient and adapted balance function performed by the central and peripheral nervous system with information’s of muscles, joints, and the skeleton, and their actions in the context of movement (5). In fact, posture and movement are not different entities, understanding standing as a posture movement and movement as a quick succession between different positions. Then to evaluate athlete, PDN-6 must be adapt.

AIM: The sport context are different in all sports then we named it PDN-6 Perform Sport s to differentiate it with the clinical practice (PDN-6).

RESULTS: The PDN6 PS is a new tool for evaluating the posture of the athlete. The aim of the PDN6 PS is to help evolve the limitations of performance of the sportsman due to his sport (sports posture, material, wedging, etc...). This test aims to help the athlete and his team to better choose equipment, stalls and athletic ergonomics in order to optimize his performance. The amendment between PDN-6 and PDN-6 Ps will give us the postural limitations present during the sport activity. Like the PDN-6 score, the PDN-6Ps score indicate the performance. Lower is the score, the less stress, postural dysfunction and best coordination and integration of body segment there will be. In the athletic setting, the athlete will gain the lowest score possible, which will increase his athletic performance and reduce the risk of injury.

But this score is not enough, because it does not take into account the specific constraints related to the sport itself. It is therefore necessary to correlate it with the PDN6 Ps, which is the same test but carried out in the sporting conditions. Ours proposition PDN-6 Ps could be used to evaluation tools in regard of sport context: posture and movement of practice, mechanics constraints, integration and material. With the same score of the PDN-6 we could find two principals orientations: 1) scores PDN-6 = PDN-6 Ps indicating that there are no restrictive limitations on the athlete in his or her sports practice ; 2) scores PDN-6 < PDN-6 Ps indicating the presence of a performance limitation factor due to the constraints of sport. In the latter case, it will be interesting to investigate the various elements that could cause this limitation (i.e: Shoes (football, rugby, racing, sport shod); wedging (kayaking, rowing, etc.); ergonomics (spacing and opening of the feet in shooting sports). The PDN6 Ps is therefore a diagnostic tool for the athlete and his team in order to objectify his postural abilities in his sporting situation, and to help him in the choice of equipment, stallion, and position. It could be used for evaluation of the performance and clinical monitoring.

SIGNIFICANCE/CLINICAL RELEVANCE: This paper provides a framework for considering balance function dynamic postural control and dynamic control, highlighting the importance of coordination, consistency, and challenges to evaluate the control processing in different conditions posed by various sports.

REFERENCES:
**Title:** The magnitude and spatial distribution of variability in plantar pressure at a wide range of walking speeds.

**Juliet McClymont**

**Introduction:** During walking, variability in step parameters allows the body to adapt to changes in substrate or unexpected perturbations that may occur as the feet interface with the environment. Despite a rich literature describing biomechanical variability in step-parameters, there are no studies that explore the habitual spatial distribution or magnitude of variability in plantar pressure in healthy humans.

**Method:** This experiment used pSPM and two standard measures of variability, the MSE and CV, to assess the magnitude and spatial distribution variability in plantar pressure across a range of controlled walking speeds.

**Results:** Reduced major axis, and pSPM regression, revealed no consistent linear relationship between MSE and speed or MSE and Froude number. A positive linear relationship however was found between CV and walking speed and CV and Froude number. The spatial distribution of variability was very different when assessed by MSE (Figure 1) and CV (Figure 2): relatively high variability was consistently confined to the medial and lateral forefoot when measured by MSE (Figure 1), but in the forefoot and heel when measured by CV. In absolute terms, variability by CV was universally low (<2.5%).

**Discussion:** From these results, it was determined that pressure variability assessed by MSE to be independent of changes in walking speeds, and that CV is not an accurate measure of peak pressure for this analysis of spatial variability.

**Clinical Relevance:** Clinical decisions about normal or otherwise behavior of plantar pressure, requires an understanding of how and why pressure distributions vary with speed, step-to-step. The unique combinations of units of the foot experiencing the highest variability that are confined to the lateral and medial borders of the forefoot, provide insights into the effect of speed on the variability in plantar pressure.

![Figure 1](image1.png)

**Figure 1:** MSE variation maps report the distribution and magnitude of the combined MSE in each pixel from the individual walking speed trial mean p-images in all subjects. Intra-subject spatial variability is highest, and confined almost exclusively to under MTH5 and MTH1, and no consistent increasing or decreasing relationship with walking speed.

![Figure 2](image2.png)

**Figure 2:** CV variation maps report the distribution and magnitude of the combined CV from each pixel from the individual walking speed trial mean p-images in all subjects. Intra-subject spatial variability is highest and localised under the hallux, medial phalanges, and midfoot. No consistent increasing or decreasing relationship is evident with walking speed.
INTRODUCTION: Many foot pathologies, such as heel pain and bunions, are theorized to be associated with aberrant foot biomechanics.[1] Forefoot-rearfoot (FFRF) angle is a biomechanical measurement often used to characterize gross foot architecture.[2-4] The FFRF remains a technically challenging measure with a weak inter-rater reliability.[5] The main challenges of this technique include keeping the hind foot and forefoot alignment in position while placing the goniometer appropriately and visualizing the angle. The PI has developed a new weight-bearing FFRF jig, which utilizes a simple platform with a separate forefoot plate that can rotate about the long axis of the foot in stance. A digital goniometer attached to the forefoot plate displays the resulting forefoot angle as the rater places the subtalar joint in neutral position. This device is solely intended to characterize gross foot architecture, and the objective of this pilot study is to test intra- and inter-rater reliability of the new weight-bearing FFRF measurement on 20 healthy subjects by 2 independent raters.

METHODS: Temple University’s Institutional Review Board approved the protocol, and all participants gave full verbal consent. Twenty healthy subjects (10 F, 10 M) reported to the Gait Study Center for participation in the study. RCSP (the position of the heel during relaxed stance), arch height index (AHI), malleolar values index (MVI), and barefoot walking plantar pressure were measured to characterize subject population. The two raters made two independent measures of the FFRF angle, both in the traditional open chain setting with the subject prone, and in closed chain using the new weight-bearing jig. The order of raters was randomly selected for each subject, and each rater was blinded to the obtained measurements of the other rater. Intraclass correlation coefficient (ICC) was calculated using an absolute agreement definition for intra-rater, inter-rater, and inter-method reliability.[6]

RESULTS: High intra-rater reliability exists for both the open chain method (Rater 1 ICC=0.904; Rater 2 ICC=0.829) and the new weight-bearing method (Rater 1 ICC=0.890; Rater 2 ICC=0.905). High inter-rater reliability exists for both the open chain method (ICC=0.751) and the new weight-bearing method (0.788). Inter-method reliability was not as high (ICC=0.444).

DISCUSSION: The biomechanical measurement of FFRF angle is critical in diagnosing and treating foot pathologies. For example, subjects with flatfoot often demonstrate everted heel and forefoot compensation in standing posture. The FFRF angle traditional measurement is still technically challenging to do, and there is a clinical need for an easier technique that is reproducible. Our study concluded that both the open chain traditional clinical measurement and our new closed chain weight-bearing measurement of FFRF angle are very reliable between multiple measurements by one rater (intra-rater), and between measurements taken by two different raters (inter-rater). In a study looking at intra-rater and inter-rater reliability for various foot and ankle measurements, it was found that extensive examiner training in these measures can improve reliability between testers.[5] Both raters in our study had thorough practice with each method beforehand to limit inconsistency. Our results also determined that there was not as much reliability between measurements obtained between the two methods. This difference may be attributed to more ease in simultaneously recording FFRF angle while keeping the subtalar joint in neutral in stance. More research is needed to determine if the new method is accepted and reproducible among clinicians and also applicable to foot function, considering that true biomechanical function occurs in closed chain.

SIGNIFICANCE/CLINICAL RELEVANCE: The measurement of FFRF angle is critical in diagnosing and treating foot pathologies, and the current standard for measuring it is technically difficult for the examiner, leaving room for human error. The PI’s newly developed closed chain method of measuring FFRF angle on a weight-bearing jig has very reliable intra- and inter-rater results, indicating it may be useful and preferred due to the simplicity of the device and technique.

REFERENCES:
**Comparison or Torsional Stiffness of Orthotics Made from Two Different Materials— A Pilot Study**

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**Disclosures:** none.

**INTRODUCTION:** Foot orthotics have been utilized for treating painful feet for over 100 years. The original Whitman plates were made of stainless steel. Other materials that have been utilized over the years include cork, various plastics, and foams.

Steindler (1929) advocated treating flat feet with a varus wedge under the heel of the shoe and a valgus wedge under the forefoot. Root (1971) took this concept inside the shoe by casting the foot with subtal joint neutral and maximally pronating the forefoot, then using a rigid material to resist the forefoot from inverting against the rearfoot.

A great many of the orthotic efficacy studies have utilized “rigid” or “semi-rigid” to describe the stiffness of the orthotic materials, yet no researcher or orthotic manufacturer has defined what the terms “rigid” or “semi-rigid” mean. The PI has found in clinical practice that changing orthotic stiffness properties can greatly affect the success of the orthotic. The goal of this research proposal, is to investigate torsional stiffness of orthotics made from two common materials, one considered to be a rigid material (acrylic) and one considered a semi-rigid material (polypropylene).

**METHODS:** Approval for this project given by Orlando VAMC Research Committee. Custom-made orthotics were tested for their torsional stiffness before being dispensed to patients. 12 acrylic and 11 polypropylene orthotics for the right foot were tested, with no knowledge of the identity or pathology of the patient. All orthotics had a noncompressible heel post. The following measurements were taken: “a” the maximum medial arch height, “b” the maximum lateral arch height, “c” the width of the orthotic in the center, “d” the thickness of the orthotic, and “e” the length of the orthotic from the anterior heel post to front clamp.

Testing was performed by two independent testers, 14 independent measures were taken at each torque and averaged. The orthotic was clamped to a solid table with the heel post set flat on the surface of the table, the anterior edge hanging off the edge of the table. A clamp was attached to the front edge of the orthotic, with a 7/16” bolt aligned with the midline of the orthotic. A digital angle finder was taped to the top of the clamp and set to 0° with the orthotic at rest. A clique torque wrench was used to invert the fore part of the orthotic against the rearfoot until the torque wrench cliqued and the distortion angle read (Fig 1) Torques from 5 in-lbs to 75 in-lbs were tested. The tests were then repeated in the eversion direction.

**RESULTS:** All orthotics had very close to linear torque/displacement slopes. Polar moment of inertia for each orthotic [J] was calculated by $J = cd(c^2+d^2)/12$. The slope of the torque/deflection curve per unit length is plotted against J in Fig 2. It is noted that the acrylic orthotic is much stiffer than the polypropylene orthotics in both direction, however the difference between the two materials greater in the eversion direction than in the inversion direction.

The Torsional Modulus [G] is calculated by $G = (torque/deflection) \cdot length / J$. Table 1 shows the comparison for G of the two materials. The G for polypropylene is only slightly less in the inversion direction than the eversion direction. On the other hand, while G for acrylic is greater than polypropylene, it is much stronger in the eversion direction than in the inversion direction.

Because orthotics are complex curves, an analysis of how the arch heights affected the torsional modulus is presented. The asymmetry is the medial arch height minus the lateral arch height. Figure 3 shows G plotted against the asymmetry. For acrylic, as asymmetry increases there is no change in stiffness in the inversion direction and an increase in the eversion direction. On the other hand, for polypropylene, as the asymmetry increases, the stiffness of the orthotic decreases.

**SIGNIFICANCE/CLINICAL RELEVANCE:** Because all metatarsal heads maintain contact with the ground in stance, orthotics try to resist pronation and supination of the rearfoot by resisting forefoot inversion and eversion of the forefoot. This is the first study that tries to measure how well different materials provide this resistance. It also shows that different foot types may need different materials to provide optimal clinical results. Much more research needs to be done on the torsional properties of materials and how to take advantage of those differences for different foot types.


Steindler. JBJS (1929)11: 272-276


<table>
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<th>Material</th>
<th>Avg J (cm³)</th>
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INTRODUCTION: The measurement of plantar pressure distributions during gait can provide a number of insights into foot function in health and disease [1, 2]. A range of tools and protocols are available for the collection of this data, with some conflicting results reported between individual studies. In this systematic review of the literature, we used a meta-regression approach to investigate the effects of age, BMI, disease, and the equipment used on the results reported by studies measuring regional plantar loading.

METHODS: Titles and abstracts containing the search term “plantar pressure” or “pedobarography” along with related Medical Subject Headings were identified from the Pubmed database. Original research studies reporting regional peak pressures during barefoot walking were eligible for inclusion. When a repeated measures analysis was carried out for an intervention, we used the control condition, which was the subject’s baseline walk. Abstracts were reviewed by two authors for suitability and data extracted from full texts as required. A mixed-effects modeling approach was used to analyze the data, with moderators including cohort type, age and BMI were included. The effect of different measurement systems was also assessed. Chi-square tests of the moderators were performed to determine significant effects.

RESULTS: From an original 1513 abstracts, 399 subject cohorts were found to meet the inclusion criteria. The measurement system was found to have a significant effect on the results, therefore these were assessed independently. Sufficient complete datasets (88 cohorts) were available to test groups made up of healthy individuals, those with diabetes, and those with diabetes and neuropathy (total n = 2954). Age was found to be significantly associated with increased peak pressures under the 1st metatarsal head (2kPa increase per year, $p < 0.001$, 95% CI [1.00, 3.01]), and BMI was found to be significantly related to peak pressures under the 1st metatarsal head (5.8kPa increase per unit rise in BMI, $p = 0.029$, 95% CI [0.6, 11.1], see Figure 1) and 4th metatarsal head (3.5kPa increase, $p = 0.0002$, 95% CI [1.6, 5.3]). Increased BMI was also associated with elevated peak pressures at the midfoot (4.9kPa rise in peak pressure per unit increase in BMI, $p < 0.001$, 95% CI [2.06, 7.59]). The presence of diabetic neuropathy was associated with elevated pressures under the 1st (119.4kPa, $p = 0.009$, 95% CI [48.7, 190.2]), 4th (29.0, $p = 0.0086$, 95% CI [7.4, 50.7]), and 5th metatarsal heads (58.7kPa, $p = 0.028$, 95% CI [6.1, 111.4]) compared to non-neuropathic subjects. The model was estimated to account for up to 70% of the heterogeneity in the data (1st metatarsal head).

DISCUSSION: We found significant changes in plantar loading associated with age, BMI and diabetic neuropathy. These effects were not universally apparent across all regions of the foot, with, in particular, the relationships between neuropathy and peak pressures limited to forefoot regions. While alternative methods of processing dynamic pressure data have been proposed [2], region loading variables, and primarily peak pressures remain the most commonly reported; therefore, our analysis was based around this measurement. In addition, we note that some ambiguity exists in the terms used to describe variables, with terms like peak pressure, mean peak pressure found to be used inconsistently, leading to some confusion in the reported results.

SIGNIFICANCE/CLINICAL RELEVANCE: We found evidence at the level of the literature that age, BMI, and diabetic neuropathy all significantly influenced regional plantar loading. It should be noted that regional peak pressure results from different collection equipment is not directly comparable.

REFERENCES:

Figure 1- Reported peak pressures below 1st metatarsal for healthy individuals and those with diabetes, with and without neuropathy.

The Effect of Age and Disease on Regional Plantar Loading: A Systematic Review and Meta Regression Analysis
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Finding the Kinematic Driver
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INTRODUCTION: When treating foot and ankle dysfunctions, it is important to look not only from the perspective of the foot, but also from the kinematics derived from dysfunctions above that may be influencing the foot to compensate sub-optimally. This presentation will discuss the influence a marked scoliosis may have on foot/ankle kinematics as the body weight shifts due to spinal deformity as well as how hip dysplasia may influence what we find below. This presentation will present alternative evaluation techniques to illustrate the importance of finding the major driver of dysfunction which can influence outcomes both pre and post operatively.

METHODS: Case presentations and videos will be used in this discussion to appropriately visualize dysfunctional movement patterns during functional movement tasks and how appropriately directed therapeutic interventions can be used to correct motor patterning and reduce or eliminate sub-optimal patterning. There will be time at the end of the session for discussion with the participants on what they see during the presentation and how they might utilize their expertise in similar cases.

DISCUSSION: It is not uncommon that as we specialize in a body region, that we develop monocular vision and look only at the area of interest. In this case, focusing only on the foot/ankle and not examining how the rest of the body moves over the foot/ankle can lead to poor surgical and rehabilitative outcomes. Successful treatment of the foot/ankle complex is dependent on a broader approach to how the structure and function of the foot/ankle can be appreciated with a more comprehensive physical examination. Conservative efforts to treat these dysfunctions prior to interventional or surgical procedures is critical to optimize procedural success.

SIGNIFICANCE/CLINICAL RELEVANCE: Prior to making orthotics, injecting tissues or performing surgery, it is imperative to put the foot/ankle complex into the context of the rest of the body and use provocative maneuvers to pinpoint areas of dysfunction that may not be within the foot complex itself. This information is useful especially if the practitioner is challenged with a non-progressing or difficult case.

REFERENCES:
Biomechanics of a Lateral Ankle Sprain – and the Effect of a Minimized Lateral Shoe-Surface Friction

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INTRODUCTION: Ankle sprains are the most frequent musculoskeletal injury. They account for up to 30% of all sports injuries and make up 1/6 of all time-loss absence due to injury. Up to 90% of all ankle sprains are ligamentous sprains to the lateral ligament complex making it the most frequently injured musculoskeletal structure in the body. Indoor sports are responsible for the highest prevalence of lateral ankle sprains. Sprain® is an adhesive polytetrafluoroethylene (PTFE) patch designed to prevent both primary and secondary acute non-contact ankle sprains in indoor sports (Figure 1). Sprain® attaches to the outside of the shoe thus not affecting ankle joint RoM. The intent is to minimize lateral shoe-surface friction whenever initial contact is made with the foot placed in a supinated position. This allows the foot to slide instead of “catching” the floor. Previous studies have shown no reduction in performance or safety when using Sprain® during typical indoor sports. Thus, the objective of present study was to test the effect in an injury situation.

METHODS: One male athlete (26 years, 1.74 m, 75.5 kg) participated in this preliminary pilot trial. The subject performed 66 lateral cutting movements onto a force platform (AMTI OPT464508HF-1000). In all cases the subject aimed to land with an initial plantar flexion and ankle inversion to force initial contact with the lateral edge of the shoe. Ground reaction forces were collected at a 1000 Hz sample rate and kinematic data were collected at 500 Hz using eight infrared Qualisys Oqus 300+ series cameras (Qualisys AB). Kinematic data were low-pass filtered using a 4th-order Butterworth filter with a cut-off frequency of 14 Hz. Inverse dynamics simulations were conducted using Visual 3D v6 (C-Motion Inc.). Ankle joint kinetics were analyzed between initial contact and impact peak.

RESULTS: The first 65 trials were all performed with Sprain® attached to the outside of the shoes and no lateral ankle sprains were sustained. The subject then removed Sprain from the shoes prior to the subsequent 66th trial. This resulted in an accidental grade 1 lateral ankle sprain. This led to immediate pain and swelling and the injury promoting control trial could not be repeated, thus ending the test session.

DISCUSSION: A complete change in ankle joint kinematics in the frontal plane were evident when comparing the control (injury) trial to a trial with Sprain® with identical pre-contact kinematics (Figure 2). The foot was further supinated after initial contact in the control trial, whereas the foot was realigned into proper position when using Sprain®. In addition, ankle joint kinetics revealed a complete lack of inversion moment at initial contact when using Sprain®. The non-existing horizontal components of the ground reaction force at initial contact and the absent inversion moment highlights the minimized friction. This minimized friction initiates a sliding mechanism in which the normal force acts to realign the foot through an eversion moment. This mechanism appears to be able to prevent acute non-contact lateral ankle sprains.

SIGNIFICANCE/CLINICAL RELEVANCE: The outcomes from this laboratory case-study were of huge clinical relevance prior to the ethical approval and registration of a randomized controlled trial (ClinicalTrials.gov ID: NCT03311490) in which 500 previously injured sub-elite athletes are randomized to use Sprain® for 52 weeks to prevent ankle sprains.

FIGURES AND TABLES:

Fig. 1 Sprain®

Fig. 2 Ankle joint inversion angle in degrees.
Customized foot orthoses in the treatment of posterior tibialis tendon dysfunction

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Introduction:
Posterior tibialis tendon dysfunction (PTTD) is characterized by a loss of function of the posterior tibialis muscle secondary to a tendon deterioration (tendonitis). This pathology shows up by a medial pain to the ankle present at each step. Progression of the pathology is frequently associated by a loss of the medial longitudinal arch, abduction of the forefoot and a valgus rear foot. The aim of this project is to do a narrative review of the literature about foot orthosis treatment for the PTTD.

Methods:
The research of scientific paper has been done by the database CINAHL and Pubmed. The inclusions criteria are to have customized foot orthotics for the experimental condition, the participant had to suffer of a PTTD of grade 1, 2 or 3 according to the classification of « Jonhson and Strom » and the variable had to be the pain. The following terms had been used for the research: adult acquired flatfoot, posterior tibial tendon dysfunction, orthoses, orthosis, insole and conservative treatment.

Results:
Following the reading of the title and the abstract, 18 articles had been kept. After take into account the inclusions and exclusions criteria, four had been kept. When the foot orthoses are used to treat the PTTD, the pain can drop by 62% to 100%. Each study had a higher decrease of pain when the subject wears foot orthoses compare to the subject who hadn’t wearing it.

Discussion:
The conservative treatment for the PTTD are not really count in the scientific literature. This is not surprising to find that they are even less present about the foot orthoses. On the other hand, it is demonstrated in the scientific literature that the customized foot orthoses is effective to diminish pain in PTTD of grade 1 and 2. The orthotic treatment is more effective when he’s combined with eccentric and concentric exercises. In fact, the orthotic treatment has never been isolated from other treatment because he’s always combined with strengthening or stretching exercises. Also, study with control group has never be undertaken to be compare with orthotic treatment. But, many authors said that the condition will never improve without treatment. In conclusion, a few studies have demonstrated the customized foot orthoses to be effective in the treatment of PTTD, the big methodological heterogeneity decreases the extern validity of the results. So, it will be important to do a good quality study like a randomized control trial to optimized the orthotic treatment because the PTTD can be very debilitating.

Clinical relevance:
This project will guide clinician in conservative treatment for the light to moderate PTTD by using foot orthoses combine with strengthening exercise instead of other conservative treatment like ankle foot orthoses.

Bibliography
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**Plantar pressures in patients with symptomatic flexible flatfoot: How are they different than adolescents with asymptomatic flatfoot?**

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**INTRODUCTION:** Flexible flatfoot deformity describes a condition characterized by a collapse of the medial longitudinal arch during weight-bearing which is constituted by unloading of the foot. Several studies have examined the differences in plantar pressures between symptomatic pediatric flexible flatfoot deformity (FF) and healthy age-matched controls [1-3]. The purpose of this study was to assess differences in plantar pressures and patient-reported outcomes in adolescents with symptomatic flexible flatfoot deformities and children with asymptomatic flatfoot.

**METHODS:** Thirty-two adolescents (57 feet) with painful FF underwent plantar pressure analysis through participation in this Institutional Review Board approved study following informed consent/assent. The foot was subdivided into medial/lateral hindfoot, midfoot, and forefoot regions. Contact area (% of total foot contact, CA%), contact time (% of roll-over process, CT%) and peak pressure (PP) were assessed for each region. Hindfoot to forefoot angle (HFA) was used to quantify forefoot abduction, and displacement of the center of pressure (COP) line relative to the bisection of the plantar angle was assessed. In those with painful FF, patient-reported outcomes were evaluated with the Pain Numeric Scale Rating (PNS), Oxford Ankle Foot questionnaire and Foot and Ankle Outcome Score (FAOS). One hundred and twenty-five age-matched painless feet were used for comparison. Within the control cohort, feet with increased medial midfoot CA% greater than 1 standard deviation from the group mean were identified as the asymptomatic flatfoot [AS] group (13 feet).

**RESULTS:** Comparing all symptomatic FF patients to the AS flatfeet, there were significant increases in CA% in the lateral hindfoot (p=0.009), medial midfoot (p=0.046) and 1st metatarsal (p=0.042) regions with decreased lateral midfoot area (p=0.001). In the medial midfoot, CT% was significantly increased in FF patients (p=0.040) with increased PP approaching significance (p=0.057). The COP was significantly more medial in FF patients than in AS flatfeet (p=0.023). Nine symptomatic patients went on to ultimate surgery and 48 were successfully managed nonoperatively. The surgery group had significantly increased PP in the medial hindfoot (p=0.035) and a tendency to have increased CT% and PP in the medial midfoot. The surgery group also reported significantly greater pain on the FAOS pain subscale (p=0.013), increased pain on the PNS (p<0.001) and scored lower on the Oxford School and Play domain (p=0.015).

**DISCUSSION:** There is a greater medial shift in the pressure variables within symptomatic flatfoot patients when compared to the asymptomatic flatfoot. CA%, CT% and PP were increased in the medial midfoot and CA% was increased in the 1st metatarsal region. COP was shifted even more medially in the flatfoot patients as well. Although small, the surgical cohort displayed a greater medial shift than the nonoperative patients and had significantly greater self-reported pain.

**SIGNIFICANCE/CLINICAL RELEVANCE:** Differences in plantar pressures are seen in patients with symptomatic flexible flatfeet when compared to those with asymptomatic flatfoot deformities.


**ACKNOWLEDGEMENTS:** The authors wish to acknowledge support from the TSRHC Research Program Fund.

| TABLE 1. Plantar Pressures Comparing Symptomatic Flatfoot (n=57) and Asymptomatic Flatfoot Controls (n=13) |
|-----------------|-----------------|-----------------|-----------------|-----------------|-----------------|-----------------|-----------------|-----------------|-----------------|-----------------|-----------------|-----------------|
|                 | Hindfoot        | Midfoot         | Forefoot        |                 |                 |                 |                 |                 |                 |                 |                 |                 |
|                 | Medial          | Lateral         | Medial          | Lateral         | 1st Met.        | 2nd Met.        | 3rd Met.        |                  |                  |                  |                  |                  |
|                 | Mean ± SD       | Mean ± SD       | Mean ± SD       | Mean ± SD       | Mean ± SD       | Mean ± SD       | Mean ± SD       | Mean ± SD       | Mean ± SD       | Mean ± SD       | Mean ± SD       | Mean ± SD       |
| Peak Pressure   |                 |                 |                 |                 |                 |                 |                 |                 |                 |                 |                 |                 |
| (N/cm²)         |                 |                 |                 |                 |                 |                 |                 |                 |                 |                 |                 |                 |
| FF              | 433.7 ± 166.8   | 344.1 ± 102.1   | 153.3 ± 77.3    | 145.4 ± 69.5    | 394.9 ± 248.6   | 436.1 ± 275.5   | 278.0 ± 162.2   |                  |                  |                  |                  |                  |
| AS              | 399.6 ± 194.6   | 396.6 ± 194.6   | 335.4 ± 134.5   | 127.7 ± 48.3    | 355.4 ± 225.6   | 441.5 ± 189.1   | 295.0 ± 90.9    |                  |                  |                  |                  |                  |
| P               | 0.567           | 0.830           | 0.057           | 0.201           | 0.814           | 0.385           | 0.042           |                  |                  |                  |                  |                  |
| Contact Area    |                 |                 |                 |                 |                 |                 |                 |                  |                  |                  |                  |                  |
| (% CA)          |                 |                 |                 |                 |                 |                 |                 |                  |                  |                  |                  |                  |
| FF              | 11.6 ± 1.69     | 11.7 ± 1.44     | 13.5 ± 6.21     | 14.5 ± 2.83     | 13.3 ± 2.3      | 9.3 ± 1.1       | 14.7 ± 2.6      |                  |                  |                  |                  |                  |
| AS              | 11.0 ± 0.82     | 10.9 ± 0.84     | 11.1 ± 3.04     | 16.6 ± 1.62     | 12.4 ± 1.1      | 8.8 ± 1.2       | 15.4 ± 1.8      |                  |                  |                  |                  |                  |
| P               | 0.119           | 0.099*          | 0.046*          | 0.001*          | 0.042*          | 0.201           | 0.220           |                  |                  |                  |                  |                  |
| Contact Time    |                 |                 |                 |                 |                 |                 |                 |                  |                  |                  |                  |                  |
| (% ROP)         |                 |                 |                 |                 |                 |                 |                 |                  |                  |                  |                  |                  |
| FF              | 66.0 ± 33.3     | 84.5 ± 138.9    | 63.5 ± 19.39    | 69.4 ± 20.5     | 77.5 ± 8.7      | 79.3 ± 8.7      | 78.9 ± 9.9      |                  |                  |                  |                  |                  |
| AS              | 58.2 ± 8.5      | 57.6 ± 9.49     | 54.8 ± 11.31    | 66.4 ± 6.9      | 77.4 ± 7.0      | 78.3 ± 6.8      | 79.4 ± 6.4      |                  |                  |                  |                  |                  |
| P               | 0.123           | 0.152           | 0.04*           | 0.385           | 0.940           | 0.651           | 0.814           |                  |                  |                  |                  |                  |

*Flatfoot patients (FF) different from Asymptomatic flatfoot controls (AS)
Testing muscle activation and in-shoe foot loading under repeatable conditions: a stepper-based approach

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INTRODUCTION: The in-vivo assessment of footwear, a key-point in several scenarios, is often performed through in-shoe pressure measurements during ground walking. However, even under controlled conditions, natural gait variability may mask relevant differences among tested items. Gait may be better controlled on treadmills, though progression against a counter-moving belt may sometimes differ too much from a real scenario ([1-2]). Based on previous experience on cyclic testing [4], this study aims at using a passive stepper to improve repeatability of foot function measurements during a load-bearing indoor motor task, slightly more demanding than natural gait but reasonably considered as part of daily motor activities.

METHODS: A commercial stepper was fixed to the floor lab to validate the testing protocol on a healthy volunteer (F, 49 years, 54kg, 1.58m), after approval of the Institutional ethic committee and signed consent. The protocol consisted of: 5 minutes for warm-up and familiarization; 7 randomly tested sport shoes with different sole flexibility and profile (4 flexible soles, 3 semi-rigid rocker soles); 10s of balance; 60 full steps (to the stroke end of the stepper), at a controlled low cadence (40-50spm); 5 minutes for recovery. In-shoe pressure was measured by the Pedar-X system (Novel GmbH, Germany; 50Hz; wide insoles). Tibialis Anterior and Medial Gastrocnemius (GA) activation of the dominant leg was acquired through a sEMG wireless system (OT Bioelectronics, Italy; 2048Hz), synchronized with Pedar. For the instrumented side (30 step cycles), averaging was applied to the EMG normalized envelopes (5Hz II order Butterworth) and to Pedar parameters of whole foot, hindfoot, midfoot, forefoot and toes ([4], cubic-spline interpolation, force normalization to body weight). ANOVA (p<0.05) with post-hoc Bonferroni-Holm correction (p<0.002) was applied to all extracted parameters.

RESULTS: High consistency was found between (trials duration: 81.8±0.9s) and within (stance phase: 1.64±0.04s) trials (Figure 1). Despite the residual intrinsic variability, interesting differences emerged: the double rocker flexible shoe significantly changed (p<0.002) all foot parameters in all regions, with a much greater midfoot involvement (i.e. force integral: 150.5±16.7Ns) with respect to flexible (74.7±9.9Ns) and to semi-rigid soles (33.8±1.5Ns); GA (relative units, r.u.) modified as well, significantly reducing (p<0.002) both peak value (0.34±0.18r.u.; 0.59±0.07r.u.; 0.59±0.04r.u.) and integral (0.18±0.06r.u.*s; 0.23±0.04r.u.*s; 0.23±0.01r.u.*s); semi-rigid rocker soles, when compared with flexible soles, entailed a higher force integral at hindfoot (215.1±10.9Ns; 188.9±4.9Ns) and a lower value at forefoot (187.6±10.8Ns; 202.7±5.5Ns).

DISCUSSION: The use of a passive stepper under controlled conditions seems feasible and effective to improve repeatability during an indoor motor task with full load bearing (max force 98.5±3.4% b.w.). 30 consistent steps per trial seemed adequate to characterize the tested shoes and detect changes at all foot regions without raising any fatigue effect.

SIGNIFICANCE/CLINICAL RELEVANCE: The proposed approach may contribute understanding the impact of shoes and orthoses on muscle activation and foot function. The protocol, reasonably helpful in a clinical scenario, is currently only intended for healthy adults. However, a dedicated re-engineering process may render it safe and suitable for clinical settings.

REFERENCES:

Figure 1. Averaged processes (mean±sd) of A) normalized sEMG envelope from medial gastrocnemius, B) total foot vertical force and C) forefoot peak pressure acquired during 30 step cycles for each of the 7 tested pair of shoes.
Velocity, footwear, and foot-strike angle effects on lower-leg muscle activity during running

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INTRODUCTION: Increased plantarflexion during barefoot (BF) running may be caused by higher pre-activation of the tibialis anterior [2]. This is characteristic of an 'impact reduction' style [3], which lowers or even completely removes the vertical ground reaction force impact peak and reduces the loading rate [4]. Discerning population effects on EMG signals is difficult and signals are typically normalized in some fashion before statistical analysis. Here we use a generalized linear mixed-effects model (GLMM) to avoid such manipulations and retain the original scale of the data, whilst also accounting for between-subject variability and heteroskedasticity. We account for stride-to-stride foot strike angle (FSA) and study the effects of velocity and condition on the root-mean-squared (RMS) EMG signal for tibialis anterior, gastrocnemius medialis, and gastrocnemius lateralis. The applied statistical technique provides less biased results and allows for the measurement and interpretation of between-subject variability. We hypothesize that in general there is more muscle activity during BF running than in SD running, increased muscle activation with increased velocity, and potentially different behaviours of the two heads of the gastrocnemius muscle.

METHODS: Informed consent was obtained from the University of Oxford Research Ethics Committee. Nineteen habitually shod (SD) male distance runners (1500 m personal record: mean (st. dev.) = 3:59.8 (10.0); age: median = 21, min = 19, max = 31) were recruited from a University cross country club. Participants ran BF and then SD on a treadmill (Ultim8 Fitness Ltd, UK) for at least 3 minutes per trial at 12.9 km h⁻¹, 14.3 km h⁻¹, and 16.0 km h⁻¹ in a randomized order for BF and SD conditions. All subjects used their personal running shoes. Twelve MX cameras (Vicon, Oxford, UK) captured heel and toe marker data at 200 Hz and muscle activity was measured at 1000 Hz using a wireless surface EMG system with Ag/AgCl electrodes (Cometa Wave Wireless EMG, Milan, Italy). Between 15 and 27 consecutive gait cycles were analyzed for each trial. Foot-strike angle was defined relative the treadmill, and a shoe-specific incline was calculated to ensure FSA in the SD electrodes (Cometa Wave Wireless EMG, Milan, Italy). Between 15 and 27 consecutive gait cycles were analyzed for each trial. Foot-strike angle was defined relative the treadmill, and a shoe-specific incline was calculated to ensure FSA in the SD condition still reflected the orientation of the foot relative the treadmill. The time of foot strike was ascertained using kinematic methods [5].

A series of GLMMs were developed to test the effects of velocity, shoes, and FSA on tibialis anterior, gastrocnemius medialis, and gastrocnemius lateralis RMS EMG voltage during five time periods: pre-activation, impact phase, braking phase, stance phase, and the entire stride cycle. A Bonferroni correction was applied for statistical tests to account for multiple testing. Slow BF running was set as the reference condition and the fixed effects were saturated to avoid bias associated with stepwise regression.

RESULTS: The EMG activity reduced for SD versus BF running and EMG activity increased for the highest velocity. The tibialis anterior activity increased whilst the gastrocnemius medialis and gastrocnemius lateralis activities reduced with more foot dorsiflexion at foot strike. Changes in activity for medium versus slow running were statistically significant, but small in magnitude when interaction effects were accounted for.

Random intercepts at the subject level were appropriate for all models, which may be interpreted to mean that each subject had an independent EMG activity for the reference trial. In addition, there was a random effect of condition for gastrocnemius medialis pre-activation.

Between-subject variability was typically larger than the largest statistically significant fixed effect, up to a factor of 2.

DISCUSSION: With the use of GLMMs, statistically significant differences may be observed in the function of the gastrocnemius heads during steady-state endurance running. From a clinical perspective, understanding the individual is of the utmost importance, evidenced by the between-subject variability explaining most of the change in EMG signal. To understand population behaviour, this variability must be appropriately accounted for in any statistical analysis.

The random effect of condition implies that there is unexplained variability in how gastrocnemius medialis activity is affected by footwear. For other muscles a more systematic, population-wide effect, may exist.

SIGNIFICANCE: Between-subject variability is large compared to experimental effects, which underlines the importance of considering EMG activity in the context of an individual. For BF running, the gastrocnemius medialis may be more susceptible to fatigue due to the increased activation when compared to the gastrocnemius lateralis, possibly because the gastrocnemius medialis plays a role in foot stability. In addition, there may be subject-level attributes which affect the response of gastrocnemius medialis activity to the use of footwear.

REFERENCES
INTRODUCTION: Bikefit is the art and science of determining a cyclist’s optimal position on a bicycle. It has evolved over the past 40 years to address physical conditions and cyclists’ efficiencies for a great diversity of riders, from professional to recreational, young, old, male, female, typical to atypical and everyone in between. Currently, bikefit is approached as a collection of measurements and data points while losing sight of the cyclist’s body as a whole. In our current work, we are exploring a return to a holistic approach to bikefit, wherein individual attributes are never considered in isolation, but rather as an integral aspect to be considered while examining the whole cyclist. Our goal is to find the cyclist’s optimal fit zone – a position where the cyclist can move around depending on terrain, weather, and type of event, all while being comfortable and efficient. In addition, we address changes that may occur in the cyclist’s body over time and the changes in position that may result.

METHODS: At Hospital for Special Surgery (HSS) our bike fitting protocol starts with a detailed patient history, a musculo-skeletal evaluation, measuring the existing bicycle’s geometry, followed by an on-bike evaluation observing the cyclist’s form, technique, areas of strength and weakness, breathing, cadence, and power output, as well as center of mass. We employ visual and auditory observations, motion capture, 3D imaging, and video, as appropriate.

RESULTS: Anecdotal observations and comments collected from our pool of cyclists at various intervals post bikefit include: reports of reductions of muscular-skeletal pain in neck, shoulders, back, and knees; reduction or elimination of tingling/burning sensations in hands and feet; increased efficiency in breathing; improvements in bike handling, such as balance, cadence, cornering, acceleration, climbing, descending.

DISCUSSION: The standard practice for bike-fit technicians is to begin with the foot, identifying foot type, evaluating stability and reviewing the shoe and pedal selection as well as the shoe cleat placement (the cleat creates a fixed or semi-fixed connection with the pedal). Often, corrections are made to accommodate limb length inequalities, arch collapse, and varus and valgus angles of the feet. Then, the process works up from the foot interface to adjust saddle position and finally back profile and hand position.

At HSS, our approach departs from this standard by starting with an on-bike evaluation (as opposed to the foot first) focusing first on the cyclist’s airway, as diaphragmatic breathing is the first landmark to look for in improving performance. We prepare the cyclist to bring the intercostals into play and throughout the process, special attention is given to the cyclist’s form and posture for maintaining optimal breathing. Along the way, we focus on core stabilization and the use of gluteus medius and gluteus maximus as counterbalancing muscles for the quadriceps. In addition to stabilizing the quads, we try to bring the rhomboids into play to help keep the chest open so that we can increase the volume of air taken in per breath. In addition, we work on form and technique to find the optimal position for arms so as to maintain optimized breathing. During these processes, the saddle and handlebar placement and as well as other “cockpit (any component that the cyclist is directly or indirectly connected to)” component selection may be altered to enable the cyclist to maintain this most effective posture. Occasionally the client’s existing bicycle is beyond the range of required adjustments at which time the fit process is transferred to a more adaptive stationary bicycle fitting device. After adjustments have been completed, the cyclist’s position is re-evaluated for comfort as well as sustainability. This process may take several hours for the bikefit, and is followed up with a visual examination or verbal report. Data from the current semi-quantitative/qualitative bike fitting process is being used to design a quantitative study from which objective evidence will be evaluated on an appropriate sample size of riders.

SIGNIFICANCE/CLINICAL RELEVANCE: At HSS, we see injured cyclists, cyclists with joint replacements, as well as recreational and pro athletes, and the adjustments that we make on and off the bike are anchored in a clinical evaluation ahead of the start of the fit. This protocol allows us to alter the cyclist’s position through counterbalancing exercise, stretching and other modalities, allowing for changes over time, in the cyclist’s body.

REFERENCES:
Are you experienced yet?

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INTRODUCTION: When assessing the foot and ankle joint, the literature is in agreement that the foot should be modelled as number of segments as opposed to a single, rigid segment as was traditionally used. As such, the use of multi segment foot models have been given significant attention in the literature and their application in gait laboratories has become increasingly widespread. The Oxford Foot Model (OFM) [1] is one such model that has been shown to provide valid and reliable multi segment foot assessment. With a large number of markers being placed upon a relatively small segment, correct marker placement and its repeatability is vital to ensure consistent and comparable measures in both an applied and clinical setting. Recent studies [2] have stressed that experience of marker application is vital for consistent marker placement within gait laboratories and so the training of clinicians and practitioners should be given special attention. As such, the aim of the study was to look at the marker placement repeatability of a novice practitioner before and after a targeted training program, and to compare their findings to that of an experienced marker placement practitioner.

METHODS: One adult (Female, 52, 163cm, 61kg) and one child (Female, 12, 126.5 cm, 26kg) volunteered and provided informed consent for the study which was approved for ethics by the University of Roehampton. Each participant had markers placed by both an experienced (400+ marker placements) and a novice practitioner (MSc Biomechanics student, > 5 marker placements) three times within a 6 week period. Markers were placed according to the OFM documentation [1]. Data was collected for 5 walking trials on each occasion using a 12 camera Vicon Motion Analysis System (Vicon, Oxford, UK) recording at 200Hz. The novice practitioner then undertook a ‘training program’ during which they palpated and placed markers using the OFM landmarks on 20 adult and 20 child feet in an attempt to improve marker placement and foot structure familiarity (motion data was not collected). Experienced and novice practitioners then placed markers on the original adult and child participants on a further three occasions and collected walking data to assess whether the marker placement repeatability had improved as a result of training. Within Assessor and Between Assessor variability was calculated for both practitioners and results compared with respect to the guidelines outlined in the literature [3].

RESULTS: The mean within-assessor variability of the relative 3D rotations ranged from 0.45–4.83° (mean = 1.98°) and 1.48-5° (mean = 2.47°) for the experienced practitioner for adult and child assessments respectively. Within-assessor variability for the novice practitioner ranged from 2.25-13.04° (mean = 7.07°) pre training and 0.55-1.85° (mean = 1.32°) post training for adult placement, and from 1.81-11.06° (mean = 4.86°) pre training and 0.84-15.6° (mean = 4.35°) post training for the child placement. Between-assessor variability for the adult participant ranged from 2.92-13.49° (mean = 8.03°) pre training and between 3.3-9.78° (mean = 4.94°). Between-assessor variability for the child participant ranged from 3.2-13.56° (mean = 7.25°) pre training and 3-11.6° (mean = 5.09°) post training. For the novice practitioner, during both pre and post testing trials of the child participant, Ankle Inversion / Eversion angle variability was greater than 5°.

DISCUSSION: After a targeted period of training, the novice assessor was able to improve their within-assessor variability to enable all OFM joints two be assessed within the 5° acceptable limit for the adult participant. Subsequently, post training between-assessor variability for the same participant fell below 10° (acceptable limit as stipulated for clinical gait laboratory repeatability standards) for all variables, showing comparable data was collected by both assessors post training. Similar findings were seen for all variables in the child assessment except for Ankle Inversion / Eversion, where variability was greater than 5° in both pre and post training testing for the novice practitioner. This subsequently led to larger ranges of within-assessor and between-assessor variability for this specific variable when examining the child’s foot, and therefore a lack of agreement between assessors as to the frontal plane motion at the ankle joint.

SIGNIFICANCE/CLINICAL RELEVANCE: This study shows that a targeted training protocol can enable novice practitioners to develop greater repeatability and comparable marker placement skills to an experienced practitioner for the OFM with adult participants. When placing markers on children’s feet, given the greater susceptibility for marker placement error due to the smaller foot structure, novice assessors may need further training and exposure to the developing foot to ensure that repeatable and comparable data is able to be collected.

REFERENCES:
Effects of acute fatigue of the lower limb on running mechanics

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Disclosures: None.

INTRODUCTION
There is a general understanding that running creates a higher incidence of overuse injuries, in particular at or below the knee, which may be related to fatigue of control mechanisms of the lower limb. Research has examined aspects of proximal and distal control of the limb and ground contact, but little has compared distinct muscle groups above or below the knee. This research seeks to identify if acute local muscular fatigue of knee or ankle controlling musculature affects mechanical variables of running gait.

METHODS
Four, injury free male amateur 5/10 km runners (Age 32 ± 17yrs, Height 174.1cm ± 6.7cm, Body Mass 64.0 ± 6.4) consented to participate. Testing was approved by the University Ethics Committee.

A randomized crossover design was employed to test the effects of two local muscular fatigue conditions; (1) of the ankle dorsiflexors and, (2) of the knee extensors, on running mechanics. Pre and post fatigue running mechanics were captured at a controlled pace, on an indoor 15m runway before and after the fatiguing protocols.

A 12-camera system recorded 3D foot segments (shank, calcaneus, midfoot and metatarsals). Medial-lateral Centre of Pressure (MLCoP) excursion and plantar pressures were collected from a pedobarograph. Additionally, electromyography of tibialis anterior, gastrocnemius medialis, vastus medialis, and biceps femoris was collected.

Principle component analysis used employed to reduce the data sets to components of the stance phase with shared variance. The components and peak EMG where entered into a 2-way repeated measure ANOVA. Effect sizes were calculated to compare variables between the groups before 90% confidence intervals and magnitude-based inferences were derived.

RESULTS
There was a significant effect of the type of fatiguing protocol on foot segment angles and MLCoP during the stance phase (Figure 1). Compared to the effects of fatigue of the ankle dorsiflexor, knee extensor fatigue resulted in greater dorsiflexion of the midfoot (effect size 0.55 to 1.3, \( p = .01 \)), less dorsiflexion of the metatarsals (EF 0.58 to 1.3, \( p = .008 \)), and greater medial position of the center of pressure (EF 0.59 to 1.3, \( p = .008 \)).

DISCUSSION
Local muscle fatigue affects running mechanics in a specific manor. Fatigue of the ankle dorsiflexors had a greater effect on metatarsal dorsiflexion and greater lateral movement of the center of pressure possibly due to a reduction in control of foot. These findings are relevant to athletes and coaches looking to avoid fatigue related injuries and to shoe manufactures interested in supporting the foot during long distance running.

Figure 1. (A) Metatarsal dorsiflexion (B) midfoot dorsiflexion (C) medial lateral CoP excursion. Solid line knee extensor fatigue, dash line ankle dorsiflexor fatigue
Load-limiting Sports Shoe Sole to Reduce Injuries

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Disclosure: Christopher A. Brown and Worcester Polytechnic Institute are stockholders in (Sports Engineering Inc.)

INTRODUCTION: The conceptual design and prototype of a load-limiting shoe sole for field, court and floor sports, which is intended to reduce ankle and knee injuries is developed. This design would test the hypothesis that limiting the transmission of loads from the playing surface to the foot can reduce ankle and knee injuries. Ground reaction forces transmitted through the shoe contribute to ankle and knee injuries in floor, court and field sports. Injuries occur during a cutting movement. It is hypothesized that at least some significant part of this injurious load is transmitted through shoe sole and that controlling this transmission and limiting the transmitted loads can reduce injuries. Similar device in ski bindings with adjustable release loads, which thereby limit the load transmissions, practically eliminated tibia fractures in alpine skiing during the 1970s. The most widely used strategy for adjusting the release load in ski bindings is based on transmitting the loads just greater than those required for ordinary, and some extraordinary, skiing maneuvers, and then releasing the ski from the boot at higher loads. The approach with designing the response loads for the load-limiting shoe soles is similar in regard to limiting transmitted loads to those required for ordinary play and then responding at higher loads. The sole, however, does not affect some kind of release, like a ski binding, which would require some kind of manual reset. With ski bindings, a release takes a racer out of the competition. The response by release leads to the use of high release adjustment settings by racers in order to avoid inadvertent releases. The shoe sole, instead responds with controlled displacement under loads just above ordinary playing loads, absorbing the energy that could otherwise do work on the ankle or knee and maybe cause injury. After displacement, while the foot is off the playing surface before being planted again, the sole recovers to its normal configuration. The player might not even know that there has been a response, because it occurs at a constant, high and safe, load. An important constraint in the design of the sole is to maintain current performance characteristics of the shoe.

METHODS: An axiom-based, engineering design method, a scientific approach, is used for this technical, computational study (Suh 1990). Suh’s axiomatic design postulates that the best design solutions maintain the independence of the functional elements and minimizes the information content. Functional requirements (FRs) are developed for the shoe to reduce injuries. Candidate design parameters (DPs) are studied to find the best physical solutions, according to Suh’s two axioms. The design solution includes a collectively exhaustive and mutually exclusive hierarchy of FR-DP pairs, which shows abstractions from high-level concepts to specific detail. At each level, candidate DPs are evaluated with respect to the constraints and the axioms. This is essentially a specialized approach of functional modeling for device designs.

RESULTS: The FR-DP decomposition has been completed at the highest levels with detail sufficient to verify feasibility of the key concepts. The design includes a special system of springs, retaining elements, and sliding interfaces that are carefully integrated into the sole of an otherwise normal sports shoe. This multi-component sole modification allows displacement, under approximately constant load, in three directions. Rotation can take place about any vertical axis, simultaneous with lateral displacement, avoiding the problem with a single vertical axis of rotation in most ski bindings that leads to ACL injuries. During sole displacement, a constant-force spring system absorbs potentially injurious loads when the shoe is in contact with the playing surface. The stored elastic energy in the spring is used for the recovery of the sole to its normal, unloaded configuration. Currently, work is being done on detailed designs of components for a prototype. Final component designs and manufacture are expected in the next two months, with assembly, and some testing, before the conference date.

DISCUSSION: This displacement design provides an important advance over systems that release. Every doubling of the displacement can reduce the peak loads by a similar amount. This system is limited in that is cannot influence loads unless they are transmitted through the sole. The potentially injurious loads must exceed ordinary playing. It cannot protect against loads due to muscular activity, e.g. contraction of the quads, which can apply significant loads across the knee without loading the sole.

SIGNIFICANCE/CLINICAL RELEVANCE: This should keep players out of the clinic and reduce the incidence of ankle and knee injuries in many sports.

REFERENCES:

ACKNOWLEDGEMENTS: The generous support of Sports Engineering Inc. and Ed Cowle are gratefully acknowledged.
Percutaneous Posterior to Anterior Screw Fixation of the Talar Neck: Soft Tissue Structures at Risk

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NOTHING TO DISCLOSE

INTRODUCTION:
Fractures of the talar neck and body can be fixed with percutaneously placed screws directed from anterior to posterior or posterior to anterior. The latter has been found to be biomechanically and anatomically superior. Percutaneous pin and screw placement poses anatomic risks for posterolateral and posteromedial neurovascular and tendinous structures. The objective of this study was to enumerate the number of trials for proper placement of two parallel screws and to determine the injury rate to neurovascular and tendinous structures.

METHODS:
Eleven fresh frozen cadaver limbs were used. 2.0mm guide wires from the Stryker (Selzach, Switzerland) 5.0-mm headless cannulated set were percutaneously placed (under fluoroscopic guidance) into the distal posterolateral aspect of the ankle. All surgical procedures were performed by a fellowship-trained foot and ankle surgeon. Malpositioned pins were left intact to allow later assessment of soft tissue injury. The number of guide wires needed to achieve an acceptable positioning of the implant was noted. Acceptable positioning was defined as in line with the talar neck axis in both AP and lateral fluoroscopic views. After a layered dissection from the skin to the tibia, we evaluated neurovascular and tendinous injuries, and measured the shortest distance between the closest guide pin and the soft tissue structures, using a precision digital caliper.

RESULTS:
The mean number of guide wires needed to achieve acceptable positioning for 2 parallel screws was 2.91 ± 0.70 (range, 2 - 5). The mean distances between the closest guide pin and the soft tissue structures of interest were: Achilles tendon, 0.53 ± 0.94mm; flexor hallucis longus tendon, 6.62 ± 3.24mm; peroneal tendons, 7.51 ± 2.92mm; and posteromedial neurovascular bundle, 11.73 ± 3.48mm. The sural bundle was injured in all the specimens, with 8/11 (72.7%) in direct contact with the guide pin and 3/11 (17.3%) having been transected. The peroneal tendons were transected in 1/11 (9%) of the specimens. The Achilles tendon was in contact with the guide pin in 6/11 (54.5%) specimens and transected in 2/11 (18.2%) specimens.

CONCLUSION:
The placement of posterior to anterior percutaneous screws for talar neck fixation is technically demanding and multiple guide pins are needed. Our cadaveric study showed that important tendinous and neurovascular structures are in close proximity with the guide pins and that the sural bundle was injured in 100% of the cases.

CLINICAL RELEVANCE:
We advise performing a formal small posterolateral approach for proper visualization and retraction of structures at risk. Regardless, adequate patient education about the high risk of injury from this procedure is crucial.
Knee Adduction Moments Associated with Knee Osteoarthritis are Increased by Medial Arch Supports

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Disclosures: None

INTRODUCTION: During walking and running an external adduction moment acts on the knee as a result of a medially acting ground reaction force (GRF) vector (Fig. 1). The knee adduction moment (KAM) magnitude is determined by the GRF magnitude and the moment arm of the GRF about the knee. Although KAMs are not a direct measure of site-specific loading in the knee, they are a useful proxy for the dynamic loads sustained by the medial tibiofemoral joint and have been linked to the incidence and progression of knee osteoarthritis (OA). Certain types of footwear have been shown to increase KAMs, but the specific features of shoe design responsible are unknown. One potentially important feature is the medial arch support, which can enhance comfort and limit foot pronation but might increase KAMs by causing a medial shift in the foot’s center of pressure (COP). Medially shifting the COP ought to also direct the GRF more medially and thus lengthen the GRF moment arm about the knee (Fig. 1). This study sought to determine if medial arch supports indeed increase KAMs, and if so, whether it is due to medial translation of the COP.

METHODS: Walking and running gaits were analyzed in 18 healthy male participants under three footwear conditions: (1) barefoot, (2) sandals without arch supports (‘shod’ condition), and (3) sandals with prefabricated medial arch supports (‘arch’ condition) (Fig. 2). Limb kinematic and kinetic data were measured by tracking reflective body surface markers using a motion capture system while participants walked (0.2 Froude) and ran (1.0 Froude) on a treadmill instrumented with a force plate. Average peak KAMs for each participant were calculated for walking and running using inverse dynamics. The mediolateral position of the foot’s COP was measured at the moment of peak the KAM. Effects of footwear condition and COP location on peak KAMs were analyzed using repeated-measures ANOVAs. All methods were approved by the IRB of Harvard University and written informed consent was obtained from each subject.

RESULTS: The peak KAMs generated while walking in the arch condition were, on average, 9% greater than those generated during barefoot walking (p = 0.013), but not significantly different than those generated during the shod condition (p=0.467). Shod walking generated 6% greater peak KAMs than barefoot (p=0.176) (Fig. 3). Average peak KAMs generated during arch running were not significantly greater than those generated when running barefoot (p=0.214) or in the shod condition (p=0.896), respectively. The peak KAM was 8% greater during shod running compared to barefoot, although this difference only approached significance (p=0.092) . Center of pressure position at peak KAM was not different in any of the three footwear conditions for walking or running (p>0.295 for all comparisons).

DISCUSSION: The results indicate that neither the medial arch support nor the minimalist sandal directly increases the first peak KAM during walking. However, the sandal-arch combination does increase the first peak KAM compared to barefoot, suggesting that minimalist footwear may be the best option for reducing harmful forces associated with OA. Despite previous research relying on the COP theory, there was no causal relationship found between the COP and the KAM.SIGNIFICANCE: The findings suggest that footwear with arch supports should be used with some caution as they may increase KAMs that potentially contribute to OA, additionally, the often cited COP link to KAM theory should be reviewed.

FIGURES:

Fig. 1. Knee adduction moment. Red GRF vector has longer moment arm than blue, due to medial shift in COP.

Fig. 2. Three footwear conditions.

Fig. 3. Average KAMs during walking across all participants. Circled 1st peak, arch significantly greater than barefoot.
Weight bearing with standing position for tibiofibular clear space measurement; using 3D US

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Disclosures: There are no financial conflicts of interest to disclose.

INTRODUCTION:
The ankle syndesmotic ligamentous complex plays a significant role in the stabilization of the ankle joint [1]. Classically, tibiofibular clear space was measured for syndesmotic ligamentous injuries on radiographs [2, 3]. But the current radiologic parameters in use seems to be inconsistent and of minimal diagnostic value [4].

Recently 3D-ultrasonography has been developed and it can make consistent location for the tibiofibular clear space measurement (figure 1). The objective of this study was to understand the possibility of 3D-ultrasonography for evaluation of tibiofibular clear space. And to clarify the changes of tibiofibular clear space against the weight bearing.

METHODS:
The study was approved by the IRB at Seoul National University. Five consecutive subjects who met the following inclusion criteria were included: (a) without ankle injury or pain and negative findings on routine ankle ultrasonography protocol; (b) who agree to this clinical trial; (c) who can stand up alone.

Neutral images were acquired with the patient sitting on the bed and bending knees about 90 degrees (figure 2A). Weight bearing images were acquired with the patient standing on the bed (figure 2B). After searching the tibiofibular clear space, 3D scan mode was initiated, and volume data were acquired. Then we measured tibiofibular clear space 1cm above the tibial plafond.

RESULTS: Five subjects (age 22.8±1.3, all males) volunteered for the study. Mean values for the syndesmosis clear space on ultrasound examination were determined as 3.44mm in neutral and 4.00mm in standing position.

DISCUSSION: We used weight bearing with standing position as a stress test for the clear space measurement. The result was resemble in the previous study by Mei-Dan et al. who used external rotation as a stress test (Mean ; 4.08mm) [5].

CLINICAL RELEVANCE: Standing position would be the most similar to ankle kinematics. 3D US can make consistent location for the tibiofibular clear space measurement.

REFERENCES:

FIGURES AND TABLES:
Figure 1. 3D US can make cross sectional images on accurate level we want to evaluate.

Figure 2A. Neutral images with the patient sitting on the bed and bending knees about 90 degrees.
Figure 2B. Weight bearing images with the patient standing on the bed.
An explanatory model of risk factors for foot ulcers in patients with diabetes

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Disclosures: Roozbeh Naemi (N), Nachiappan Chockalingam1 (N), Janet K. Lutale (N), Zulfiqarali G. Abbas (N),

INTRODUCTION: The annual incidence rate of Diabetic Foot Ulcers is estimated to be between 15 to 25 % with foot ulceration is the main cause of lower limb amputation in patients with diabetes worldwide. The presence of foot ulcers in a diabetic patient increases the risk of death at 5 years by 2.5 times [Walsh et al, 2016]. Understanding and extending the knowledge of characteristics of patients with diabetic foot ulcer is essential and will be useful in developing clinical management protocols for this group of patients. The aim of this study is to identify the biomechanical, neurological and clinical parameters along with other demographics and life style risk factors for foot ulcers in diabetic patients that can explain the presence of diabetic foot ulcers.

METHODS: A total of 1219 (M/F: 666/553) diabetic patients who attended the diabetic foot clinic in Tanzania between Jan 2011 and Dec 2015 were included in this study. Foot ulcer was defined as a full-thickness wound involving the foot or ankle and was observed in 75 patients, whereas 1144 patients had no foot ulceration. A combination of generic and specific categorical and continuous parameters were collected from the patients during a single visit. The general categorical parameters were: diabetes type, assistive device usage, smoking habit, alcohol habits, physical activity status, previous amputation, and history of ulceration. The general continuous parameters included: age, body mass, height, shoe size, duration of diabetes, and body mass index. The specific continuous parameters included: ankle brachial pressure index, vibration perception threshold, temperature sensation and tolerance thresholds and barefoot plantar pressure during walking. The specific categorical parameters included: neuropathy based on impaired skin sensation to monofilament, arterial pulse, foot deformity, Charcot foot, skin status, Mycosis, nail ingrowth, swelling, and presence of callus. The specific categorical parameters for each participant were defined as if these occurred on either (or both) feet for the participant. The specific continuous parameters were averaged between the left and right feet. Multivariate logistic regression was utilised to develop the explanatory model for foot ulceration based on Backward stepwise selection that involves removal of the risk factors on the probability of the Wald statistic (p>0.1) and retaining variables with (p<0.05).

RESULTS: The proposed model was able to adequately (with 95.3% diagnostic accuracy) justify the presence of foot ulceration based on the common risk factors. While the model’s specificity (as the percentage on of participants with no ulceration incidence that are correctly diagnosed) was 99.1 %, the sensitivity (as the percentage of participants with ulceration that are correctly diagnosed) was only 37.3%. Fasting blood sugar level, foot swelling and ankle mobility were among the significant contributors to diagnosing foot ulcers. Also, lower average foot Temperature Tolerance and Temperature Sensation thresholds to cold probe were observed to be significant (P<0.05) contributor to diagnosing foot ulceration. The addition of foot specific parameters to the model improved the diagnostic strength. The model as a whole could predict between 17.4 % (Cox and Snell R Square) and 47.1 % (Nagelkerke R Square) of the variation in ulceration status.

DISCUSSION: The model was adequately specific in identifying the factors that protect the patients against ulceration. However, the ability of model in justifying the characteristics of patients with ulcerated foot is currently relatively limited. With just over 1 out of three patients with ulcerated foot showing common characteristics that were investigated in this study.

SIGNIFICANCE/CLINICAL RELEVANCE: This study indicates that specific clinical characteristics of the foot that were investigated in this study can offer further explanation to diabetic foot ulceration. This justifies the need to include further foot-specific parameters like skin perfusion and soft tissue mechanical characteristics in the model that can potentially improve the diagnostic accuracy of the model and can provide further explanation for diabetic foot ulceration.

REFERENCES:

ACKNOWLEDGEMENTS: Financial support from Higher Education Funding Council of England under quality-related (QR) research funding is acknowledged.
Session 13: Healthy Locomotion

Co-Moderators: William Ledoux, PhD (University of Washington; Seattle VA) and Julie Stebbins, PhD (Oxford University)

9:00AM  13-1: Deschamps, Kevin, et al. Multi-segment foot joint kinetics associated to running with a rearfoot striking pattern


Multi-segment foot joint kinetics associated to running with a rearfoot striking pattern

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INTRODUCTION: The prevalence of running related injuries has increased the interest in the assessment of running biomechanics (1). Special attention has been given to the foot-strike pattern during running as evolving evidence suggests a considerable impact on lower limb kinetics and associated injuries. Though, current biomechanical studies lack adequate measurements of foot joint kinetics since typically single-segment foot models are used. Therefore, the purpose of the present study was to compare joint kinetics of the rearfoot, Chopart, Lisfranc and the first metatarsophalangeal joint (hallux) joints while running barefoot with a heel-striking pattern with those provided by a single-segment foot model.

METHODS: Kinematic and kinetic data of seven active university students (4 men/ 3 women) were collected by a Vicon® 3D motion analysis system (100 Hz) while running barefoot at an imposed constant speed of 3.3 m s−1 ±10%. Running sessions were performed on a 10-m walkway, with a plantar pressure platform (RSscan international, 200 Hz) placed on top of a force platform (AMTI®, 1000Hz); a dynamic calibration procedure, a so-called 3D synchronization box, was used for these two instruments. During each test session, the striking pattern was analysed through the feedback provided by the pressure platform. Multi-segment foot kinematics (Multi-Segment Ankle=Shank-Calcaneus, Chopart=Calcaneus-Midfoot, Lisfranc=Midfoot-Metatarsus, Hallux=first metatarsophalangeal joint) and single segment foot kinematics (Single Segment Ankle=Foot –Shank) were measured using the Rizzoli Foot Model (RFM) (2). The center of pressure and resultant ground reaction force was analysed for each RFM segment and for each collected frame by estimating subarea shear forces and normal moments as a proportion of the measured forces (so-called proportionality scheme described in Saraswat et al. (3)). Inertial parameters of foot segments were based on their mass and geometric volumes, whereas the mass of the foot was distributed at 30/30/30/10 percent (from proximal to distal). Inverse dynamic calculations were according to the IOR-4Segment-model1 as in Deschamps et al. (4). All participants provided and signed an informed consent, and the ethics committee of the University Hospitals Leuven approved the protocol.

RESULTS: Considerable overestimation of the ankle power generation and absorption was observed in the single-segment foot model. The Chopart joint had an average peak power generation of 3.19 Watt/kg and power absorption of 2.88 Watt/kg. The Lisfranc joint showed considerable power generation whereas the Hallux mainly contributed to power absorption.

DISCUSSION: The current study provides new insight into the compliant behavior of the foot joints during running. An overestimation of power absorption at the ankle of 40% with respect to the single segment model is demonstrated. Cross-validation of the current findings is recommended since a proportionality scheme was used, which was be validated only in walking.

SIGNIFICANCE/CLINICAL RELEVANCE: The findings of the current study may help in unraveling and managing the etiology of certain foot and lower limb pathologies such as stress fractures and tendon lesions.


Table 1. Summary table (group mean and standard deviation (SD) representing peak negative and positive power estimated with the single segment model and the IOR-4Segment Model.

<table>
<thead>
<tr>
<th></th>
<th>Single Segment Model</th>
<th>Mean (SD)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle + Peak Power (Watt/kg)</td>
<td>17.04 (1.58)</td>
<td></td>
</tr>
<tr>
<td>Ankle - Peak Power (Watt/kg)</td>
<td>-7.63 (1.35)</td>
<td></td>
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<tr>
<td>Chopart + Peak Power (Watt/kg)</td>
<td>3.19 (1.01)</td>
<td></td>
</tr>
<tr>
<td>Chopart - Peak Power (Watt/kg)</td>
<td>-2.88 (1.01)</td>
<td></td>
</tr>
<tr>
<td>Lisfranc + Peak Power (Watt/kg)</td>
<td>1.57 (0.58)</td>
<td></td>
</tr>
<tr>
<td>Lisfranc - Peak Power (Watt/kg)</td>
<td>-0.24 (0.14)</td>
<td></td>
</tr>
<tr>
<td>Hallux + Peak Power (Watt/kg)</td>
<td>0.34 (0.27)</td>
<td></td>
</tr>
<tr>
<td>Hallux - Peak Power (Watt/kg)</td>
<td>-2.17 (0.92)</td>
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The Effects of Subtalar Axis Orientation on Kinematics and Kinetics During Walking and Running

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INTRODUCTION: The ankle and subtalar joint (STJ) provide primary dorsiflexion/plantarflexion and inversion/eversion during gait, respectively. Biomechanical models frequently keep the STJ locked. Further, when it is included, like in the commonly used OpenSim model Gait2392 [1], the orientation angles of the STJ axes are set at the low range as reported by Inman [2]. Inman reported average inclination and deviation angles as 42° from the horizontal plane and 23° from the midline of the foot. Gait2392 STJ orientation angles are set at 37.2° and 8.7°, respectively. Therefore, the purpose of this study was to determine how the variation to Inman’s axes affect kinematics and kinetics during dynamic motion in OpenSim (SimTK).

METHODS: Three-dimensional marker position and ground reaction force (GRF) data of walking and running were exported from Visual3D (C-Motion) for five subjects as part of an IRB approved study after informed consent was obtained. The data were imported into OpenSim v3.3 and used to scale the default Gait2392 for each subject. Inverse kinematics and dynamics were computed using OpenSim in the standard model and after changing the STJ orientation in the scaled model to match Inman’s average inclination and deviation angles. A two-tailed paired t-test was used to compare kinematics and kinetics of the original Gait2392 STJ axis orientation and Inman’s mean axis. Significance was set at p<0.05 for all analyses.

RESULTS: Changing the STJ orientation to match Inman’s mean inclination and deviation angles resulted in a significant increase (p=0.0384) in the ankle range of motion during walking. The range of ankle joint moments calculated using Inman’s mean axes during running was significantly different (p=.0074) as compared to Gait2392 STJ axes. Significant differences were also found in peak STJ joint moments in both walking (p=.0002) and running (p=.0007). Changing the STJ orientation also resulted in large variation of STJ moment patterns during walking, with the average Inman’s mean axes producing two large peaks while the axis used by Gait2392 showed little to no second peak at all (Figure).

DISCUSSION: The mean ankle and STJ kinematics patterns for both Inman and default Gait2392 STJ models were similar to in vivo bone pin studies for walking [3]. The ankle kinematics calculated for both models were similar to the in vivo bone pin study for running [4], however there were no STJ kinematics. The STJ moments during walking show variability, which is consistent with prior studies [3]. The model with Inman’s mean axes has a double-peak curve for all 5 subjects while the model with Gait2392 STJ angles shows a slight double-peak for only 2/5 subjects, similarly to the literature that showed a double-peak for 1/3 subjects. A limitation of this study is that due to lack of studies and the variability in subject specific differences, we are not able to verify which model is correct. Future work will aim to determine if a generic STJ axis is sufficient or if subject specific STJ axes should be used.

SIGNIFICANCE/CLINICAL RELEVANCE: Biomechanical models may help answer questions about STJ kinematics and kinetics in vivo, if the validity of the models is established.


Fig. Mean subtalar joint moment of one stance cycle calculated using original Gait2392 subtalar axes vs Inman’s axes for five subjects i) walking and ii) running. Dotted line indicates Gait2392 subtalar axes while solid line is for Inman’s.
Kinematic mechanisms controlling heel velocity in late swing in healthy human walking

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Disclosure: C Pongmala was supported on a scholarship from King Chulalongkorn Memorial Hospital, Bangkok, Thailand while conducting this research.

INTRODUCTION: Winter first measured the velocity of a marker placed on the distal heel at foot contact during walking and concluded that this was virtually zero in the vertical direction and low in the horizontal direction (1) questioning the relevance of the term 'heel strike'. He suggested a number of late swing mechanisms which have the potential to achieve this low velocity, but did not specify which were actually responsible. This study has thus sought to explore these mechanisms.

METHODS: 10 healthy volunteers (7 female, 31 [SD 4] years old) consented to participate in the study which had been reviewed by the institutional ethics committee of the University. Kinematic data from 6 trials each of walking barefoot at self-selected speed, and 25% slower and faster, was captured with an optoelectronic tracking system using a CAST protocol. One marker was placed as distally as possible on the heels of participants without it being displaced during walking. Foot contact was detected from four force plates with a 10 N threshold. Horizontal and vertical components of the heel marker velocity and ankle, knee and hip joint centers were recorded. The recorded linear velocities being within the kinematic chain enabled the velocity at the heel (compared to the hip/centre of mass) to be attributed to kinematic mechanisms (joint angular velocities were also recorded for reference). Data was not filtered and smoothed only by averaging.

RESULTS: Horizontal heel velocity at foot contact was measured as 0.04 m.s⁻¹ (95% CI: -0.03, 0.11) at self-selected walking speed, 0.00 m.s⁻¹ (95% CI: -0.08, 0.07) at slow speed and 0.20 m.s⁻¹ (95% CI: 0.09, 0.31) at fast speed. Vertical heel velocity varied less with speed and at foot contact was -0.13 m.s⁻¹ (95% CI: -0.19, -0.07) at self-selected speed (Figure 1). Velocities of the joint centers of the hip (1.32-1.99 m.s⁻¹), knee (1.14-1.84 m.s⁻¹) and ankle (0.41-0.77 m.s⁻¹) in the horizontal direction also demonstrated similar values across walking velocities.

DISCUSSION: The heel horizontal velocities are lower than reported by Winter (0.87 m.s⁻¹) and the vertical velocities towards the floor a little higher (slow walking -0.17 95% CI: -0.24, -0.10) than Winter’s virtually zero report (1). Both differences can be explained by variations in experimental protocols; Winter (1) utilized digitized video data and did not report the exact placement position of the heel marker. The values in the current study are closer aligned to others reported in more recent literature with similar cohorts (vertical 0.19 m.s⁻¹ (2); horizontal 0.27 m.s⁻¹ (3)). The mechanisms accounting for the difference in horizontal velocity between the hip and heel, and therefore the nature of the foot-floor contact, were knee flexion (55%) and ankle plantarflexion (30%) with a limited amount of hip extension also making a smaller contribution (11%). These mechanisms were consistent across velocities and highlight the importance of both knee flexion and ankle plantarflexion commencing in late swing to ensure smooth contact of the foot with the ground.

SIGNIFICANCE/CLINICAL RELEVANCE: The work establishes, in agreement with Winter, that heel velocity is extremely low at the end of swing and this may be a requirement for normal walking. Extending this methodology to explore whether and how patients with different conditions achieve this may be insightful.

REFERENCES:

Figure 1. Horizontal (a) and vertical (b) velocity of the heel marker across the gait cycle (from foot off to foot off) at self-selected walking speed. The grey areas represent ± one standard deviation from the mean. The red area denotes the last third of swing to foot contact (dashed vertical line).
Kinematic comparison of multi-segment foot models in healthy adults

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INTRODUCTION: Three dimensional gait analysis is widely used to analyze altered gait patterns in pathologies (e.g. cerebral palsy, Charcot-Marie-Tooth), to identify problems and evaluate treatment. Conventionally, the foot has been modeled as one rigid segment. To comply with foot complexity, especially in the case of foot deformity, several multi-segment foot models have been developed, including the Oxford Foot Model (OFM) and Rizzoli Foot Model (RFM).

These multi-segment foot models have different characteristics (e.g. amount of segments, marker placement and axis definition), which makes their output difficult to compare. Insight into how these differences affect ankle and foot kinematics will assist in comparing studies that use different models, however, the output of these models has not yet been thoroughly compared. Therefore, the aim of this study was to compare the kinematic output of OFM and RFM during normal walking.

METHODS: Ten healthy adults (26.6±2.6 years, 4 male), participated in this study. Markers were placed on both lower extremities according to the Newington marker model. On the right foot, additional markers were placed to define both OFM and RFM. A static standing trial was performed to calculate offset angles and to define the segment axes. Subjects walked barefoot at their comfortable walking speed on a 10m walkway, while markers were captured by a 12-camera Vicon system. Five force plates (AMTI) were used to determine gait events. Each trial was time-normalized to 100% of the gait cycle and the average over 3 trials was used. Statistical parametric mapping paired t-tests were used to compare joint angles between the models, including the hindfoot-shank (HF-SH), forefoot-hindfoot (FF-HF) and hallux-forefoot angle (HX-FF).

RESULTS: Differences between OFM and RFM are presented in Table 1 and the FF-HF dorsiflexion angle is shown in Figure 1. The HF-SH waveforms did not differ in plantar/dorsal flexion. However, more inversion was shown for RFM for almost the entire gait cycle and more internal rotation was shown in the OFM waveform for periods during the early stance and swing phase. The FF-HF angle of RFM compared to the OFM showed more plantar flexion around toe-off, more eversion during periods in the stance phase and almost no differences (< 3% of the gait cycle) in abduction/adduction angle. The HX-FF angle of RFM was in more dorsal flexion compared to the OFM during the period around toe-off.

DISCUSSION: This study shows the differences in kinematic output between OFM and RFM during gait in healthy subjects. These differences are mainly a result of different axes definitions between the models. For example, the coordinate system of the hindfoot in RFM is more inverted compared to OFM, which not only yields an systematic difference ("offset"), but also means that movement of the foot segments will be decomposed differently ("crosstalk"). Different marker locations will yield different sensitivity to skin movement artefacts, that might further explain differences found. Our next step will be to analyze voluntary modified gait patterns by healthy subjects to challenge the robustness of the models and their differences.

SIGNIFICANCE/CLINICAL RELEVANCE: This study gives insight in the differences in kinematic output between OFM and RFM during gait in healthy subjects. These differences are mainly a result of different axes definitions between the models. For example, the coordinate system of the hindfoot in RFM is more inverted compared to OFM, which not only yields an systematic difference ("offset"), but also means that movement of the foot segments will be decomposed differently ("crosstalk"). Different marker locations will yield different sensitivity to skin movement artefacts, that might further explain differences found. Our next step will be to analyze voluntary modified gait patterns by healthy subjects to challenge the robustness of the models and their differences.

REFERENCES:

Table 1. Statistical comparison between OFM and RFM

<table>
<thead>
<tr>
<th></th>
<th>HF-SH</th>
<th>FF-HF</th>
<th>HX-FF</th>
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<tbody>
<tr>
<td>% within the gait cycle with a significant difference (p&lt;0.05)</td>
<td>0.63</td>
<td>50.73</td>
<td>51.53</td>
</tr>
<tr>
<td>Average maximal difference for each significant period (°)</td>
<td>12.2</td>
<td>13.0</td>
<td>4.4</td>
</tr>
</tbody>
</table>

DIFP=Dorsal/Plantar flexion, InEv=Inversion/Eversion, InExt=Internal/External rotation; AdAb=Adduction/Abduction

Figure 1. The dorsiflexion angle of the FF-HF. The shaded areas are SD around the group mean for OFM and RFM. The significant part during the gait cycle is marked at the top of the graph.
Magnetic Resonance Assessment of Lower Leg Muscle Activation After Blood Flow Restricted Exercises: A Pilot Study

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Disclosures: Victoria V. Shumway (N), A. Wayne Johnson (N), Neal Bangerter (N), Grayson Tarbox (N)

INTRODUCTION: Blood flow restriction (BFR) exercise is a form of exercise intervention that has been growing in popularity for both athletes and recovery patients. Studies show that BFR exercise leads to an increase in protein synthesis, hormone response, and muscle activation resulting in increased muscle strength and hypertrophy1. The leading way to measure muscle activation is surface EMG, and only one study has investigated BFR exercise muscle activation using EMG2. In this pilot study, we propose to (1) assess the effectiveness of MRI T2 mapping in showing muscle activation and (2) to see whether BFR exercises show a greater increase in muscle activation compared to non BFR exercises.

METHODS: Three healthy male 23-year-old subjects (ht:74.3±2.08cm, wt:77.86±9.45kg) were recruited according to established IRB procedures and gave informed consent before participating. The subjects had their lower leg scanned using a 3Tesla magnet (Siemens TIM-Trio 3.0T MRI) prior to exercise. A 2D Multi-Echo Spin Echo sequence was used to construct T2 maps with a matrix of 256x256 (ROxPE) with 8 echoes. An 8-channel foot/ankle coil was used to obtain a total of 20 10mm slices. The subjects then exercised according to the following protocol: 3 sets of 25 calf raises followed by 3 sets of 15 10mm jumps with 30-60 seconds of rest between sets. If the subjects were not able to do the full 15 or 25 on the last set, they were told to go to failure. Another MRI was performed immediately following their exercise regimen (<5 minutes) so that muscle activation before and after could be compared.

One week later another exercise session was performed with KAATSU bands around their proximal thighs. The pressures ranged from 280-310mmHg, all within the ideal 50-85% occlusion range which was measured via pulse wave ultrasound (GE Logiq P6, 9L probe). The same procedure was followed as in the control condition, with a T2 map created for pre and post exercise. The T2 maps were segmented and compared manually with Osirix software. Paired t-test were used to determine differences between and within conditions. To assess within condition change in activation, all segmented muscles were evaluated as whole. To assess differences between conditions, muscle segmentation was done by lower leg compartments.

RESULTS: There was no significant difference between the pre-exercise activation levels between the control (C) and KAATSU (K) groups (C=38.9±2.69, K=39.1±2.31, p=0.776).

Within each group, there was an overall significant increase in activation in both the C (p=.001) and the K (p<.001) when comparing their pre and post T2 maps of the entire lower leg. C showed a 5.9% increase and K showed a 11.5% increase.

Between the conditions, the K posterior compartment produced significantly increased activation levels compared to the C posterior compartment values (p=.001). C increased 7.1% while K increased 12.2%. In the lateral compartment, there was also significant increase in activation (p<.05). C increased 7.5% and K increased 14.7%. The anterior compartment showed no significant difference between C and K conditions (p=.30). C decreased 3.9% and K increased 4.2%.

DISCUSSION: Our results showed that T2 mapping was an effective way of assessing muscle activation after exercise. A benefit of MRI assessment of activation is that it shows the entirety of the muscle body rather than just small sample region of the muscle. It also allows for accurate measurement of deeper muscles that previously could not have been evaluated with surface EMG alone, like the tibialis posterior. One limitation of this method is that MRI is a post measurement, and cannot be measured while exercise is going on, like EMG allows. We observed that BFR exercise showed a greater increase in muscle activation than exercise alone. This supports the limited research that exists regarding muscle activation with BFR.

SIGNIFICANCE/CLINICAL RELEVANCE: Noninvasively identifying muscle activation could be used to diagnose metabolic muscle disease, identify and pinpoint muscular dysfunction, observe muscle deterioration in aging individuals, and help researchers better understand the biological foundation of muscle chemistry.

BFR allows for an increased muscle activation with lower resistance exercises, which helps retard muscle atrophy in recovery patients.

REFERENCES:
How do the hindfoot axes of a multi-segment foot model and the underlying bony anatomy compare?

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Disclosures: None.

INTRODUCTION: Musculoskeletal models used in gait analysis require coordinate systems to be identified for the body segments of interest. These are necessary for tracking segment motion and calculating the angles at the joints. Bone-embedded axis systems are typically specified based on palpable anatomical landmarks or identifiable features and are assumed to represent the underlying bony anatomy. In most multi-segment foot models, skin-mounted markers placed on the calcaneus are used to define the medio-lateral, antero-posterior (AP), and proximal-distal axes of the hindfoot (or rearfoot) segment and, with tibia-based axes, to calculate the overall motion of the ankle joint complex, a combination of the relative movements of the tibia, talus, and calcaneus. Since the specified marker positions on the calcaneus may be difficult to identify and the talus is entirely inaccessible, it is not clear exactly how the hindfoot axes defined by gait markers relate to the anatomy of these two bones. Therefore, the aim of this study was to compare the marker-based axes to the bone morphology.

METHODS: Twenty adult females (age 26 ± 6 yrs, height 1.68 ± 0.07 m, weight 57.0 ± 7.0 kg) with no known foot deformities participated in the study, which was approved by National Health Service (England) Research Ethics Committee. Radio-opaque monitoring electrodes (Type 2223, 3M Healthcare) were placed on the participants’ feet and ankles at the marker locations specified for a multi-segment foot model used widely in clinical gait analysis [1]. CT images (GE 64-slice Light-speed VCT scanner, 100 kV, 110 mA) were acquired as the participants lay supine with their feet in a semi-weight bearing posture, achieved using a custom-built rig (based on [2]) that applied 40% body-weight to each foot. Standard bone and soft tissue scan algorithms (slice thickness and distance both 0.625 mm) were used to scan each foot separately.

The spatial coordinates of the electrodes were obtained from the CT images using Mimics software (Materialise). These were used to define the AP-axis of the hindfoot [1] and the long-axis of the foot (Heel-Toe or HT axis, taken to lie along a line connecting the marker on the posterior distal end of the calcaneus to one on the dorsum of the foot between the heads of the second and third metatarsals). Segmented masks of the tali and calcanei were created using a thresholding algorithm supplemented with manual editing and exported in STL format. SolidWorks (Dassault Systèmes SolidWorks Corp.) was used to create 3D bone models. The unit vectors of the principal axes were obtained using the mass properties tool. The first principal (FP) axis was taken to represent the orientation of each bone (t – talus, c – calcaneus). The projections of the four axes (AP, HT, FPt, FPc) in the plantar plane were compared.

RESULTS: Displacement of at least one marker, usually on the heel, in the CT loading rig meant that only the FPt and FPc axes could be calculated for some subjects. Results are summarized in Table 1. The angles between the AP axis and the HT and FPc axes were highly variable. The FPc axis aligned more closely and consistently with the HT axis than did the AP axis. A plot of FPc – FPt versus AP – FPc showed no apparent relationship between the two quantities. The difference between the AP and FPc axes increased the further away the two axes were from the HT axis.

DISCUSSION: The FPc – FPt angle on average fell within the normal range for Kite’s angle (15-30º) [3]. Even for these normal, healthy feet, the marker-based AP axis was highly variable in relation to the underlying bony anatomy. The difference between the AP and FPc axes is likely to be greater in feet with deformity.

SIGNIFICANCE/CLINICAL RELEVANCE: The hindfoot AP axis considered here does not consistently represent the long axis of either the calcaneus or the foot. This implies that the third hindfoot rotation in the Euler angle sequence for the ankle (inversion-eversion or abduction-adduction, depending on the multi-segment foot model) takes place about an axis whose orientation relative to the standard planes of the foot or to the bony anatomy of the calcaneus is likely to be subject-specific.


<table>
<thead>
<tr>
<th></th>
<th>AP – HT (º)</th>
<th>FPt – HT (º)</th>
<th>AP – FPc (º)</th>
<th>FPc – FPt (º)</th>
</tr>
</thead>
<tbody>
<tr>
<td>n = 18 feet</td>
<td>2.5 ± 11.9</td>
<td>1.7 ± 2.8</td>
<td>0.9 ± 12.2</td>
<td>21.5 ± 5.0</td>
</tr>
<tr>
<td>Mean ± Std Dev</td>
<td>2.8</td>
<td>0.7</td>
<td>2.8</td>
<td>1.2</td>
</tr>
</tbody>
</table>

Table 1. A positive angle indicates that the AP or FPc axis points more laterally than the HT axis and that the AP axis points more laterally than the FPc axis.
Session 14: Good Vibrations (SS)

10:30AM - Noon

Program Chairs: Leah Bent, PhD (University of Guelph) and Howard Hillstrom, PhD (HSS)

Conceptual, experimental and computational foot models rarely consider sensorimotor components beyond EMG and predictions of muscle force. An inability to stimulate and record responses from individual skin mechanoreceptors has significantly restricted our understanding of the link between foot sensation and foot biomechanics. However, recent developments in the motor control community demonstrate important contributions from skin sensation to gait and the potential to modify gait through sensory (e.g. vibration) rather than, or in addition to, mechanical means (e.g. orthoses). In parallel, there has been rapid growth in technologies that claim to stimulate feet for therapeutic benefit, often with minimal empirical support. It is therefore timely for the iFAB community to consider: (1) evidence to support a role for foot skin sensation in the control of foot biomechanics, balance and gait, and (2) opportunities to develop foot stimulation based interventions.

Speaker 1: Dr Leah Bent (University of Guelph, Canada) will address the feasibility of foot sole skin to contribute to gait and balance; particularly the distribution and density of plantar mechanoreceptors. The physical location of mechanoreceptors and the physical characteristics of plantar skin (hardness, thickness) have implications on perceptual threshold and clinical testing techniques. The use of microneurography in parallel with other techniques weaves together the link between clinical measures of tactile and vibration thresholds and actual cutaneous afferent firing thresholds. She will aim to convince the audience that activation of foot sole mechanoreceptors really can play a role in whole body movement and control.

Speaker 2: Dr. Ryan Peters (University of Calgary, Canada) will review the relationship between plantar sensitivity, lower-limb cutaneous reflex strength, and standing balance stability in healthy adults and Parkinson’s disease patients. Given the importance of cutaneous feedback in standing balance control, a critical assessment of the effectiveness, feasibility, and physiological underpinnings of proposed therapeutic interventions is warranted. One approach are insoles that vibrate and supposedly enhance afferent sensitivity via the stochastic resonance phenomenon. Dr. Peters will finish his talk by discussing recent work examining the effect of stochastic resonance on plantar perceptual sensitivity and lower-limb cutaneous reflexes.

Speaker 3: Dr Paul Zehr (University of Victoria, Canada) will discuss evidence supporting the idea that feedback from plantar mechanoreceptors strongly influences gait parameters. In particular, cutaneous feedback from the plantar surface is an important contributor to neurorehabilitation and sensation from the dorsum is crucial to the stumbling corrective reaction. Stimulation of cutaneous nerves innervating the foot results in functionally relevant, phase- and
nerve-dependent neuromechanical changes in walking parameters. Stimulation of discrete skin regions has revealed that neuromechanical responses are also topographically organized. Feedback from discrete regions of the feet contribute to ‘sensory steering’, which has implications in rehabilitation and athletic training.

**Speaker 4: Dr Kristen Hollands** (University of Salford, UK) will discuss her systematic review of the effects of sensory stimulating insoles in healthy older adults and neurologic patient groups. She will also present evidence for the way in which foot sensation is affected by brain injury and how impaired sensation relates to balance deficits and differences in foot biomechanics (e.g. plantar pressure, foot motion, muscle action). She will also consider how "sensory stimulating" insoles may (or may not) meet the needs of people with stroke and whether or not these devices meet the wishes of patients (gathered through qualitative patient studies).
Session 15: Sports Injuries

Co-Moderators: Joseph Hamill, PhD (University of Massachusetts) and Robert Turner, PT, DPT (HSS)


1:40PM  15-2: Becker, James, et al. Metatarsal Loading In Runners Who Habitually Use Rearfoot or Mid/Forefoot Strikes


2:00PM  15-4: Casillas, Christopher. & Becker, James. Relationship Between Arch Height and Metatarsal Loading in Runners

2:10PM  15-5: Deeble, John, et al. Mechanical analysis of barefoot and shod treadmill running using a smartphone application

2:20PM  15-6: Henderson, Adrienne, et al. Midfoot Angle Changes During Running After an 8-Week Intervention Program

2:30PM  15-7: Houston, Megan, et al. Association Between Concussion and Ankle Sprain History in Collegiate Athletes

2:40PM  15-8: Johnson, Wayne, et al. The differences in time to stability, foot muscle size and toe flexor strength between gymnast, cheerleaders and non-athletes


3:00PM  15-10: Matias, Alessandra B.et al. Foot kinematics of forefoot and rearfoot strikers on recreational runners using the Rizzoli’s foot model

3:10PM  15-11: Veloso, Antonia, et al. Joint moment contributions to forward and upward acceleration of body CG are affected by the foot to ground contact model
Development and evaluation of the Running Shoe Comfort Assessment Tool (RUN-CAT)

John Arnold¹, Jonathan D. Buckley¹, Adrian E. Esterman²,³, Christopher Bishop¹

¹Alliance for Research in Exercise, Nutrition and Activity (ARENA), Sansom Institute for Health Research, University of South Australia, Adelaide ²School of Nursing and Midwifery, University of South Australia, ³Australian Institute of Tropical Health and Medicine, James Cook University, Townsville, Australia

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INTRODUCTION: Footwear comfort is an important aspect of running shoe prescription which can improve running performance [1], and reduce the incidence of foot-related complaints or injury [2,3]. Although existing tools are useful for evaluating footwear comfort [4-6], to the authors’ knowledge, none have been developed in a systematic fashion that incorporates items deemed meaningful by runners, is reliable and sensitive to differences between footwear models, and has established thresholds for clinically meaningful changes. Therefore, the aim of this study was to develop a new running footwear comfort assessment tool for use in clinical and research settings addressing these requirements.

METHODS: A four-phase development process was used: (i) a survey of 282 runners to identify meaningful items of footwear comfort, (ii) field testing of runners (n=100) who assessed the comfort of three different shoes (ASICS Nimbus 18, Contend 4 and Onitsuka Tiger), (iii) item reduction using bootstrap aggregation and weightings from multiple regressions relative to overall comfort to identify a final set of items, and (iv) a reliability study (n=30) to establish standard error of measurement (SEM), minimal detectable difference (MDD₉₀) and minimal clinically important difference (MCID) for the component items and final comfort index. Ethical approval for this research was granted by the UniSA HREC (ID 35827).

RESULTS: Of 19 initial items relating to different aspects of footwear comfort, after item reduction, four weighted items were included in the final comfort assessment tool: heel cushioning, shoe stability, forefoot cushioning and forefoot flexibility. Reliability of the overall comfort index (within-day and between-day) was excellent (ICC 0.98 and 0.95, respectively) with all four component items also displaying good reliability (ICC >0.70). The SEM of the component items ranged from 4.3-4.9 mm, and subject nominated MCID values ranged from 9.3-9.9 mm on a 100 mm VAS. The SEM and MDD₉₀ for the final comfort score was 2.0 and 4.6 points, respectively. The overall comfort index was different between two of the three shoes based on the mean score (Nimbus: 82, Contend: 78, Tiger: 46, p<0.001 – 0.053).

DISCUSSION: This study presents the development and initial evaluation of a new assessment tool (RUN-CAT) for evaluating the comfort of running footwear. The RUN-CAT demonstrates excellent reliability, acceptable measurement error and good discriminative ability between footwear models. Comfort perception is presented as a composite score for overall rating using weightings derived from four individual items (heel cushioning, shoe stability, forefoot cushioning and forefoot flexibility). The identification of these comfort items as predictors of overall comfort is largely in agreement with other recent findings [6], although features in this study were more specific to different aspects of the footwear. Based on the data from this study, group changes in running footwear comfort exceeding 11 mm were both larger than error thresholds and deemed to be clinically important to runners. The RUN-CAT will benefit from further evaluation and expanded testing with different cohorts and footwear models to fully elucidate its clinical utility.

SIGNIFICANCE/CLINICAL RELEVANCE: (1-2 sentences): The selection of running footwear may be improved using the RUN-CAT, with group changes greater than 5 points above measurement error and 11 points considered clinically meaningful. Footwear design features may also be altered to optimize comfort which can be measured with the RUN-CAT.

REFERENCES:

ACKNOWLEDGEMENTS: ASICS Oceania donated all footwear used in this study. JBA is currently supported by a National Health & Medical Research Council Early Career Research Fellowship (ID: 1120560). CB is currently an ASICS industry funded research fellow.
Metatarsal Loading In Runners Who Habitually Use Rearfoot or Mid/Forefoot Strikes

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Disclosures: James Becker (N), Christopher Casillas (N)

INTRODUCTION: There currently is a debate in the scientific community regarding whether switching from a rearfoot strike (RFS) or a mid/forefoot (FFS) pattern might be an appropriate strategy for preventing running injuries [1,2]. Making a recommendation either way requires a detailed understanding of the differences between these foot strike patterns. While numerous studies have compared ground reaction forces, joint kinematics, and joint kinetics [3], to date relatively little is known about how metatarsal loading varies with foot strike pattern. Thus, the purpose of this study was to compare metatarsal loading in habitual RFS and FFS runners during treadmill running. We hypothesized that peak loads applied to the metatarsals would be similar between foot strikes but that loading rates would be higher in runners who use a FFS.

METHODS: 30 habitual RFS runners (sex: 19M, 11F, age: 33.1 ± 12.5 years, weekly mileage: 44.8 ± 17.1 miles) and 20 habitual FFS runners (sex: 14M, 6F, age: 27.3 ± 10.4 years, weekly mileage: 48.9 ± 13.6 miles) participated in this study. All participants were injury free at time of testing. The procedures were approved by the IRB and all participants provided informed consent. In-shoe plantar pressure was collected at 100 Hz while participants ran on a treadmill. Prior to data collection, a pressure insole was trimmed to fit the insole from each participants’ shoes and calibrated using step calibration. Participants wore their own shoes. Plantar pressure software was used to identify six regions including all the metatarsals (AllMets) and each individual metatarsal (M1, M2, M3, M4, and M5, respectively). Peak force, percent stance when peak force occurred, maximal loading rate, and impulses were calculated for each region. All forces were normalized to body mass. Differences between RFS and FFS groups were compared using t-tests and effect sizes (Cohen’s d).

RESULTS: Peak forces were higher in the FFS group than the RFS group in the AllMets (p = .021, d= 0.729, Figure 1A), M1 (FFS: 0.44 ± 0.19 BW, RFS: 0.34 ± 0.10, p = .039, d= 0.658), and M3 (FFS: 0.36 ± 0.11 BW, RFS: 0.29 ± 0.08, p = .034, d= 0.214) regions. The percent stance when peak force occurred was not different between groups for any of the six regions. Loading rates were higher in the FFS group for the AllMets (p = .029, d= 0.704), M1 (p=.042, d=0.652), and M5 (p = .034, d=0.64) regions (Figure 1B) while impulses were higher in the FFS group for the AllMets (p = .006, d = .801), M1 (p = .035, d = 0.639), M3 (p = .019, d = 0.710), and M5 (p = .017, d = 0.694) regions (Figure 1C).

DISCUSSION: Peak plantar forces, loading rates, and impulses are higher in the metatarsals in runners who habitually FFS than those who habitually RFS. It is likely that habitual FFS runners have adapted to these higher loads. Thus, if differences persist when a habitual RFS runner switches foot strike pattern then these individuals would be exposed to higher loads. This might help explain studies reporting metatarsal stress injuries after individuals switch to a FFS pattern [4,5].

SIGNIFICANCE/CLINICAL RELEVANCE: Recommendations regarding switching from a RFS to a FFS pattern should be made cautiously as habitual FFS runners experience higher loads in the metatarsals.

REFERENCES:

FIGURES: Vertical force in the AllMets region (A), peak loading rate in the AllMets, M1, and M5 regions (B), and impulse in the AllMets, M1, M3, and M5 regions (C).
Foot Mechanics in Drop Landings
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INTRODUCTION: Tools to quantify foot mechanics have increased in recent years, providing much needed insight into foot and ankle function during walking and running. A non-locomotor task, such as a drop landing, may provide a unique perspective to further study foot function. Landing mechanics are sequentially reversed from locomotion, with ground contact moving from distal to proximal. Landing forces are also higher, potentially amplifying stresses to foot structures such as the medial longitudinal arch. The purpose of this study was to investigate the mechanics of the foot during drop landings using two recent models: a multi-rigid segment foot model [1] and a deformable foot model [2].

METHODS: Forty-eight females (age = 20±2 yr, ht = 1.6±0.1 m, wt = 57±6 kg) participated after signing ethics committee approved consent forms. Twenty-eight markers were attached to the dominant leg and foot according to a three segment kinetic foot model [1]. Subjects performed one legged barefoot drop landings from a height of 0.4 m. Hanging from wooden rings, subjects dropped onto two adjacent force platforms, so that the hindfoot and forefoot contacted separate plates. Angles, power, and work were calculated for the modeled foot joints [1] while total power distal to the rearfoot segment was calculated from a deformable foot model [2].

RESULTS & DISCUSSION: The midtarsal joint was plantarflexed at the start of the drop, with a slight increase in plantarflexion just before contact (Left Figure). Upon contact, it quickly underwent 27° of dorsiflexion excursion, much greater than has been found in walking and running [2]. The metatarsophalangeal (MTP) joint extended approximately 20° during the drop and an additional 10° just after contact. MTP motion just prior to impact corresponded with Midtarsal plantarflexion, suggesting some engagement of the windlass mechanism to tension the plantar fascia in preparation for landing. During landing, the MTP joint returned to a neutral position as the Midtarsal joint (and ankle) dorsiflexed. Across the impact phase, the Midtarsal joint performed -0.42 J/kg of work, 56% as much as the ankle (-0.75 J/kg), while the total deformable structures distal to the rearfoot performed work similar in magnitude to the ankle (-0.74 J/kg) (Right Figure). These structures include the Midtarsal and MTP joints, as well as elastic and viscoelastic soft tissues. A comparison of Midtarsal joint and Deformable Foot power (Middle Figure) show in particular the influence of the heel fat pad, as a second power peak is visible in the Deformable power when the rearfoot contacts the ground.

SIGNIFICANCE: A dynamic drop landing task amplifies joint motion and loading in the foot when compared to locomotion. Various foot tissues made substantial contributions to impact absorption comparable in magnitude to the ankle joint. The role of these tissues may have implications in athlete and pathology treatments that limit foot mobility, such as footwear or taping.

Figures: Left = Representative Midtarsal and MTP joint angles, plotted from start of drop to lowest position of the center of mass (vertical line indicates ground contact). Middle = Midtarsal joint power and Deformable Foot power, plotted from initial contact to lowest position of the center of mass. Right = Group mean Ankle, Midtarsal, and Deformable foot work done during the same time period.

**Relationship Between Arch Height and Metatarsal Loading in Runners**

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**INTRODUCTION**: Metatarsal stress fractures are common injuries among runners. While the cause of these injuries are multifactorial, arch height has been suggested to play a role in which metatarsal gets injured. Previous authors have reported that high arched individuals experience more 5th metatarsal stress fractures while low arched individuals experience more second and third metatarsal stress fractures [1]. However, studies on plantar loading have reported that low arched individuals have higher forces on the lateral forefoot compared to the medial forefoot, suggesting that low arched individuals would be at higher risk for 5th metatarsal stress fractures [2]. One explanation for this discrepancy is that there currently are few studies in the literature evaluating relationships between arch height and plantar loading during running, and those that do exist typically lump multiple metatarsals into one region. Therefore objective of this study was to determine whether there is a relationship between arch height and loading among each of the metatarsals.

**METHODS**: 30 runners (sex: 23M, 7F, foot strike: 14 RFS, 16 FFS, weekly mileage: 48.3 ± 16.3 miles) participated in this study. All participants were injury free at time of testing. The procedures were approved by the IRB and all participants provided informed consent. Arch height index assessment was performed on each participant. [3] Participants then ran on a treadmill with an in-shoe plantar pressure system sampling at 100 Hz. Prior to data collection, a pressure insole was trimmed to fit the insole from each participants’ shoes and calibrated using a step calibration. Participants wore their own shoes. Plantar pressure software was used to identify each individual metatarsal region (M1, M2, M3, M4, and M5, respectively). Maximal force in each metatarsal was calculated in each region for ten gait cycles. All forces were normalized to body mass and averaged. Both their left and right feet were analyzed (total n=60 feet analyzed). Linear regressions were used to determine the relationship between arch height index and peak force for each metatarsal region.

**RESULTS**: There was a weak relationship between higher arches and increased force under M1 ($R^2 = 0.166, p = 0.001$). However, there was not a relationship between arch height and peak force under M2 ($R^2 = 0.042, p = 0.114$), M3 ($R^2 = 0.017, p = 0.317$), M4 ($R^2 = 0.002, p = 0.739$), or M5 ($R^2 = 0.001, p = 0.845$).

**DISCUSSION**: Our results show no relationship between arch height and plantar forces under the second, third, or fifth metatarsals, the most common locations for metatarsal stress fractures. Since increased plantar loads are thought to be related to metatarsal stress fracture development [2], this calls into question whether there is a relationship between arch height and metatarsal stress fracture location. Future prospective studies on plantar loading in individuals who sustain metatarsal stress fractures are required to address relationships between plantar loading and injury.

**SIGNIFICANCE/CLINICAL RELEVANCE**: (1-2 sentences): The use of arch height to assess risk of stress fractures to specific metatarsals is not supported by these results.

**REFERENCES**:

**FIGURE 1**: Linear regressions comparing arch height to peak plantar forces under M1 (A), M2 (B), and M5 (C).
Mechanical analysis of barefoot and shod treadmill running using a smartphone application

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Disclosures: None.

INTRODUCTION: There is much deliberation in literature over the possible conditioning benefits of barefoot running activity¹-³ with equivocal findings. Furthermore, if footwear conditions affect optimal movement patterns of the lower limb they may affect magnitudes of between-leg symmetry and potentially injury incidence⁴-⁵. The increase of smart phone applications is a relatively new, but potentially valid method for analysing the mechanical variation between footwear conditions⁶. The aim of this study was to employ a smart phone app to analyse between-leg asymmetry and mechanical variables between shod conditions.

METHODS: Twenty-five undergraduate sports students (Age, 20 ± 0.8 yrs; Height, 179 ± 7 cm; Body Mass, 79 ± 12kg) volunteered based on regular physical activity (5 ± 2hrs per week) which included running and were injury free and had no previous experience of barefoot running. All participants completed PAR-Q and informed consent documents and all testing was approved by the St Mary’s University Ethics Committee. Participants performed two, randomised 1-minute bouts of treadmill running at 4 m·s⁻¹ in two conditions, shod and barefoot separated by 5 minutes rest. A smart phone application (Runmatic⁶ at 240fps) was used to calculate (based on 8 consecutive steps) multiple biomechanical measures after recording the final 15 seconds of each bout from the rear of the treadmill; these were contact time (s), flight time (s), step frequency (Hz), relative max force (BW) and leg stiffness (kN/m), furthermore to compare a percentage of between-leg asymmetry. Paired samples T-Test and Wilcoxon tests were used to examine for significant difference (p set at 0.05) between footwear conditions, based on whether the data was normally distributed via Shapiro-Wilk analysis.

RESULTS: Contact time significantly decreased in the barefoot condition (from 0.22 to 0.21s), while step frequency (from 2.92 to 3.03Hz) and leg stiffness significantly increased (15.4 to 18.2kN/m) during the barefoot condition. Flight time, and relative max force significantly increased in the barefoot condition. There was a greater, but insignificant, score of asymmetry in the barefoot condition (4.5 ± 3.85% vs 3.3 ± 3.4%).

DISCUSSION: Contact time decrease and step frequency and leg stiffness increase during the barefoot condition follow that seen in other literature. Flight time and relative max force significantly increased in the barefoot condition which could be an artefact of suboptimal acute changes to the condition, though much research highlights an increase in force during initial barefoot contact though flight time would expect to reduce in line with increased step frequency. Asymmetrical gait has multiple research publications and suggestions that footwear may reduce asymmetry⁷ is a tentative finding from this study since research also highlights the subject specific nature of lower limb mechanics across footwear conditions⁸. The immediate response to the footwear condition after 30 seconds of treadmill acclimatization may not be a true representation of adaptation to the given condition; furthermore the use of low homogeneity in the participants may have increased variation in acute response to barefoot running. Additionally, grouping participants into ‘foot type’ categories may provide more interesting results particularly with proxies of force and stiffness that the application calculates.

SIGNIFICANCE/CLINICAL RELEVANCE: (1-2 sentences): The Runmatic smartphone application seems a valid tool to measure acute changes in footwear conditions or monitor adaptations in mechanical variables and/ or symmetry in treadmill running following intervention with a clinical or performance focus.

REFERENCES:
INTRODUCTION: The medial longitudinal arch has been called the central core of the foot. Its structure, movement, and integrity during running gait largely depends on the function of intrinsic and extrinsic foot muscles. There are many injuries that can be associated with a dysfunctional medial longitudinal arch. Improving the strength of the foot muscles could help improve arch function during running. The purpose of this study was to observe changes in arch deformation after eight weeks in a foot strengthening exercise group (FSG), a group walking in minimalist shoes (MSG), and a control group (CG).

METHODS: 24 healthy college age subjects (age 22.6±2.6 years, height 174.4±10.3cm, weight 69.0±13.0kg) were recruited and randomly assigned to either FSG (n=9), MSG (n=7) or CG (n=8) and monitored over 8 weeks. All subjects were recreational runners with an average weekly running mileage of 15 to 30 miles for the last 6 months prior to participation. Exclusion criteria included any lower extremity injury within the last 3 months or if they had run in barefoot or minimalist shoes at least 3 times within the previous 3 months. All runners, regardless of group assignment, maintained their pre-study mileage in traditional running shoes throughout their participation in the study. FSG subjects followed a series of progressive exercises designed to strengthen the intrinsic muscles of the foot while MSG subjects were given a pair of minimalist shoes to use while walking. These subjects started walking 2,500 steps daily, increasing their daily walking step count to 7,000 steps a day by the end of the 8-week study. Biomechanical data was collected at the beginning of the study and again at week 8. To do this passive-reflective motion capture markers were placed on participants’ right and left feet and ankles according to the Oxford Foot Model. Subjects ran at a self-selected pace on a treadmill while data was collected for at least 10 strides. Data was processed within Visual 3D where peak midfoot angles were extracted and averaged within each trial. Paired t-tests were used to compare group means at week 0 and week 8 with alpha set at 0.05.

RESULTS: While all groups experienced a decrease in midfoot angle, only the FSG group experienced a significant change. See table below.

DISCUSSION: The results suggests that the foot exercise intervention resulted in more arch control during running than the MSG or CG. The FSG may have been the only group to experience a significant reduction in arch drop because the exercises were targeted for the muscles that control the medial longitudinal arch whereas the minimalist shoes were less targeted. The MSG group also started at a daily step count that is roughly equivalent to walking 1 mile which could be considered a small distance when compared their daily running mileage. A limitation of this study was the small sample size which would help strengthen the significance in the FSG.

SIGNIFICANCE/CLINICAL RELEVANCE: It is possible for patients to reduce the amount of arch deformation during running by using targeted foot exercises. This could help with patients who have over-use injuries associated with dropped arches.

REFERENCES:

ACKNOWLEDGEMENTS:

FIGURES AND TABLES:

<table>
<thead>
<tr>
<th></th>
<th>Week 1</th>
<th>Week 8</th>
<th>p-Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>CG</td>
<td>10.8401</td>
<td>9.9399</td>
<td>0.332</td>
</tr>
<tr>
<td>FSG</td>
<td>11.4583</td>
<td>7.5893</td>
<td>0.045*</td>
</tr>
<tr>
<td>MSG</td>
<td>9.0624</td>
<td>8.7442</td>
<td>0.401</td>
</tr>
</tbody>
</table>

Average midfoot angles for weeks 1 and 8. Significance shown by *.
Association Between Concussion and Ankle Sprain History in Collegiate Athletes

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Disclosures: The authors have no conflicts of interest to disclose.

INTRODUCTION: Ankle sprains and concussions are common injuries sustained by collegiate athletes. Multiple studies have determined that athletes who sustain a concussion are at greater risk for lower extremity musculoskeletal injuries. A positive association between concussion and lower extremity musculoskeletal injury has also been identified when examined retrospectively over the course of a collegiate athletic career. Despite these recent findings, few studies have specifically examined the association between concussion and ankle sprain injury history or factors which may impact this relationship such as number of concussions or gender. Therefore, the purpose of this study was to examine the association between concussion and ankle sprain history; as well as, the influence of number of concussions and gender on this relationship.

METHODS: A sample of 468 National Collegiate Athletic Association student-athletes (200 Males, 268 Females; 19.5±1.3y, 173.9±10.5cm, 71.9±13.6kg) were recruited from 17 sports at either the Division I or III level. Institutional Review Board approval was attained and voluntary completion of the study packet was deemed consent to participate. Participants provided demographic, injury history, and athletic participation information through a survey. Participants were asked to describe all injuries that they could recall over their lifespan. For this study, analysis was limited to reported ankle sprains and concussions. Chi-square analyses with corresponding odds ratios examined the relationship between concussion and ankle sprain history. Additional analyses were completed by stratifying athletes with a single or multiple concussion history and within each gender. For all analyses, Fisher's exact tests were used to determine statistical significance (p<0.05).

RESULTS: Athletes with a history of concussion reported a greater ankle sprain history rate compared to athletes with no history of concussion (OR=2.07, p<0.001). Further examination determined that athletes with a history of multiple concussions (OR=2.56, p=0.003) and those who reported a single concussion (OR=1.78, p=0.040) had a greater association with ankle sprain history compared to athletes with no concussion history. Females with a history of concussion were more likely to report a history of ankle sprain compared to females with no concussion history (OR=2.54, p=0.001). However, males with a history of concussion were not more likely to report a history of ankle sprain when compared to males with no history of concussion (OR=1.41, p=0.360). A summary of the data is reported in Table 1.

DISCUSSION: Overall, there was a positive association between concussion history and ankle sprain history among collegiate athletes. Athletes with a concussion history were twice as likely to report an ankle sprain history compared to athletes with no history of concussion. Athletes who reported more than one concussion or females that reported a history of concussion were 2.5 times more likely to report an ankle sprain history.

SIGNIFICANCE/CLINICAL RELEVANCE: Collegiate athletes who reported a history of concussion were up to 2.5 times more likely to have a history of ankle sprain. Although the underlying mechanism(s) for the relationship between concussion and ankle sprain has not been established, this association should be taken into consideration during the management of these injuries. Concussion and ankle sprain injuries may share common sensorimotor or neurocognitive impairments which increase susceptibility to additional injuries over time. Future research should consider examining the effects of sensorimotor training following concussion and neurocognitive training following ankle sprain.

Table 1. Association between concussion history and ankle sprain history in collegiate athletes

<table>
<thead>
<tr>
<th>Groups</th>
<th>n</th>
<th>Rate of Ankle Sprain History</th>
<th>Odds</th>
<th>Odds Ratio</th>
<th>Odds Ratio 95% CI</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Entire Sample</td>
<td></td>
<td></td>
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<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>History of Concussion</td>
<td>115</td>
<td>58.0%</td>
<td>1.40</td>
<td>2.07</td>
<td>1.35, 3.18</td>
<td>&lt;0.001*</td>
</tr>
<tr>
<td>No History</td>
<td>353</td>
<td>40.0%</td>
<td>0.67</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>History of 1 Concussion</td>
<td>66</td>
<td>54.5%</td>
<td>1.20</td>
<td>1.78</td>
<td>1.05, 3.02</td>
<td>0.040*</td>
</tr>
<tr>
<td>No History</td>
<td>353</td>
<td>40.0%</td>
<td>0.67</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>History of &gt;1 Concussion</td>
<td>49</td>
<td>63.0%</td>
<td>1.72</td>
<td>2.56</td>
<td>1.38, 4.75</td>
<td>0.003*</td>
</tr>
<tr>
<td>No History</td>
<td>353</td>
<td>40.0%</td>
<td>0.67</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Males</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>History of Concussion</td>
<td>36</td>
<td>50.0%</td>
<td>1.00</td>
<td>1.41</td>
<td>0.68, 2.91</td>
<td>0.360</td>
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<tr>
<td>No History</td>
<td>164</td>
<td>41.0%</td>
<td>0.70</td>
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<tr>
<td>Females</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>History of Concussion</td>
<td>79</td>
<td>62.0%</td>
<td>1.63</td>
<td>2.54</td>
<td>1.48, 4.35</td>
<td>0.001*</td>
</tr>
<tr>
<td>No History</td>
<td>189</td>
<td>39.0%</td>
<td>0.64</td>
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</tbody>
</table>
The differences in time to stability, foot muscle size and toe flexor strength between gymnast, cheerleaders and non-athletes
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INTRODUCTION: There has been recent speculation that the intrinsic foot muscles may play a larger role in lower extremity control and injury than previously believed. It has been suggested that training in less supportive footwear or barefoot may lead to an increase in intrinsic and extrinsic foot muscle size and strength. Our purposes were: 1) to compare intrinsic and extrinsic foot muscle size and strength in gymnasts (who predominately train barefoot), cheerleaders (who perform similar movements to gymnasts and predominately train shod), and weight matched non-athletes; and 2) to measure time to stability (TTS) after drop landings among the groups.

METHODS: Forty-eight women participated and signed IRB approved consent forms. Three groups were recruited: collegiate gymnasts (Gym), (n=16, age= 19.4 yr, ht= 159.3 ± 4.9 cm, wt= 56.7 ± 4.3 kg), collegiate cheerleaders (Cheer), (n=16, age= 20.3 yr, ht= 161.9 ± 5.4 cm, wt= 58.7 ± 7.1 kg), and non-athletes (CON) (n=16, age= 21.5 yr, ht= 167.2 ± 5.8 cm, wt=57.2 ± 5.7 kg). Sizes of 3 intrinsic (abductor hallucis, flexor digitorum brevis, quadratus plantae) and 3 extrinsic (tibialis posterior, fibularis longus and fibularis brevis) foot muscles were measured using ultrasound imaging (GE LogiqP6 with ML6-15 linear probe). The muscles were measured using protocols previously used and shown reliable. Toe flexor strength was assessed from the first toe (GT) individually, and the lateral toes (LT) together using a customized dynamometer. The subject completed three trials of a maximal contraction for 3 seconds. Subjects then performed 3 drop landing trials of each of 4 conditions – double and single leg, shod and unshod. We defined TTS as the time when the sequentially averaged medial-lateral ground reaction force during landing remained within one-quarter standard deviation of the overall series mean. The mean of the 3 trials for each condition were compared using ANOVA comparisons with alpha set at 0.05, weight was a significant co-variate.

RESULTS: Gym had significantly larger muscles in all 6 muscles compared to the CON group (p<0.002) and significantly larger fibularis longus (p<0.001) and fibularis brevis (p=0.048) muscles compared to the Cheer. Cheer had significantly larger muscles than the CON group in the abductor hallucis (p=0.028), quadratus plantae (p<0.001), tibialis posterior (p<0.001), and near significance in the flexor digitorum brevis (p=0.055). Gym had significantly greater GT strength compared to CON (p<0.001) and Cheer (p=0.018) and LT strength significantly greater than CON (p<0.001), but not Cheer (p=0.286). Cheer also had significantly greater strength compared to CON in both toe flexion tests (p<0.001). CON had significantly longer times to stability than both Gym and Cheer in all conditions (p<0.006). Gym and Cheer did not differ in time to stability in any of the conditions (p>0.08).

DISCUSSION: Athletes performed significantly better in the strength and TTS tests than CON. Gym had the greatest GT flexion strength and overall largest muscle sizes normalized to weight. Many factors likely affect drop landings to achieve shorter TTS beyond the size and strength of foot muscles that are worth considering. These athletes routinely perform tumbling and landing maneuvers that require high levels of motor control, hours of practice, and fitness. There does seem to be significant training effects in these athletes’ foot muscle development compared to the CON group. Training barefoot may have a larger influence on strength and muscle size increase compared to training or performing activities of daily living shod. Limitations to our study may be small sample sizes and similarity in the training of between the athletes.

SIGNIFICANCE/CLINICAL RELEVANCE: Stronger and larger foot muscles appear to contribute to decreased TTS. Our data supports strengthening of foot muscles to improve performance.

Table 1. Between-group comparison (mean±SD) of intrinsic and extrinsic foot muscle size, strength and time-to-stability.

<table>
<thead>
<tr>
<th>Strength (kg)</th>
<th>Muscle Size (cm^2)</th>
<th>Time to Stability (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>GT</td>
<td>LT</td>
</tr>
<tr>
<td><strong>CON</strong></td>
<td>3.3±.8</td>
<td>2.8±.9</td>
</tr>
<tr>
<td><strong>Cheer</strong></td>
<td>4.5±1</td>
<td>3.9±1</td>
</tr>
<tr>
<td><strong>Gym</strong></td>
<td>5.2±3</td>
<td>4.0±2</td>
</tr>
</tbody>
</table>

CON=non-athlete control, Cheer=Cheerleader, Gym=Gymnast; GT=great toe, LT=lateral toes; AH=Abductor Hallucis, FDB=Flexor Digitorum Brevis, QP=quadratus plantae, TP=Tibialis Posterior, FL=fibularis longus, FB=fibularis brevis; BD= barefoot double foot landing, BS=Barefoot single foot landing, SD=shod double foot landing, SS=shod single foot landing.
Do Achilles Tendon Properties differ with Habitual Foot–strike Running Patterns?

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Disclosures: SC Wearing (N), IS Davis (N), SL Hooper (N), T Horstmann (N), T Brauner (N)

INTRODUCTION: There is a 52% lifetime incidence of Achilles tendinitis in runners.1 It has been suggested that a stiffer tendon is a healthier tendon. The capacity of foot–strike running patterns to influence the mechanical properties of the Achilles tendon is controversial.2,3 This study used transmission–mode ultrasound to investigate the influence of habitual running foot–strike patterns on Achilles tendon properties during walking and running.

METHODS: Axial transmission velocity of ultrasound, a measure of the instantaneous elastic modulus of tendon, was measured in the right Achilles tendon of 15 runners with an habitual rearfoot foot–strike (RFS) and 10 with an habitual forefoot–strike (FFS) running pattern during barefoot walking (1.1 ± 0.1 ms⁻¹) and running (2.0 ± 0.1 ms⁻¹). Basic gait parameters, ankle motion and vertical ground reaction force were simultaneously recorded at 120Hz. Statistical comparisons between habitual foot–strike patterns were made using repeated measure ANOVAs.

RESULTS: Mean (±SD) gait characteristics and ultrasound velocity in the Achilles tendon during walking and running are summarized in Table 1. FFS was characterized by significantly shorter stance duration (−4%), greater ankle dorsiflexion (2°), and higher peak vertical ground reaction force (+20% bodyweight) than RFS during treadmill running (P<.05). Both groups adopted a RFS pattern during barefoot walking, with only the relative timing of peak dorsiflexion (3%) and ground reaction forces (1–2%) differing between groups (P<.05). Peak force loading rates were 22–23% lower in FFS than RFS during walking and running (P<.05). Peak ultrasound transmission velocity in the Achilles tendon was significantly higher during walking (≈100 ms⁻¹) and running (≈130 ms⁻¹) in FFS than RFS (P<.05).

DISCUSSION: Despite adopting a similar heel-toe gait pattern during walking, ultrasound transmission velocity in the Achilles tendon was systematically higher during both walking and running in habitual FFS than RFS; indicating the Achilles tendon had a higher material stiffness in those with a habitual FFS running pattern.

SIGNIFICANCE/CLINICAL RELEVANCE: Habitual footfall patterns during running may influence the mechanical properties of the Achilles tendon in recreational runners. A stiffer tendon in FFS may aid rapid force development as well as offer protection against injury.

REFERENCES:

Table 1. Mean (±SD) gait characteristics and ultrasound velocity in the Achilles tendon during walking and running

<table>
<thead>
<tr>
<th>Gait Condition</th>
<th>Walk</th>
<th>Run</th>
</tr>
</thead>
<tbody>
<tr>
<td>Habitual foot-strike pattern when running</td>
<td></td>
<td></td>
</tr>
<tr>
<td>n</td>
<td>RFS</td>
<td>FFS</td>
</tr>
<tr>
<td>15</td>
<td>10</td>
<td>15</td>
</tr>
<tr>
<td>Stance Phase Duration (%GC)</td>
<td>63 ± 1</td>
<td>64 ± 2</td>
</tr>
<tr>
<td>Swing Phase Duration (%GC)</td>
<td>37 ± 1</td>
<td>36 ± 2</td>
</tr>
<tr>
<td>First Ground Reaction Force Peak (BW)</td>
<td>1.10 ± 0.06</td>
<td>1.17 ± 0.07</td>
</tr>
<tr>
<td>Ground Reaction Force Minimum (BW)</td>
<td>0.84 ± 0.09</td>
<td>0.85 ± 0.09</td>
</tr>
<tr>
<td>Second Ground Reaction Force Peak (BW)</td>
<td>1.13 ± 0.08</td>
<td>1.14 ± 0.08</td>
</tr>
<tr>
<td>Peak Loading Rate (BW s⁻¹)</td>
<td>33.5 ± 9.2</td>
<td>26.2 ± 6.2*</td>
</tr>
<tr>
<td>Peak Ankle Dorsiflexion (°)</td>
<td>5 ± 1</td>
<td>6 ± 2</td>
</tr>
<tr>
<td>Peak Ankle Plantaflexion (°)</td>
<td>-6 ± 2</td>
<td>-7 ± 4</td>
</tr>
<tr>
<td>Minimum Ultrasound Velocity 1 (ms⁻¹)</td>
<td>1972 ± 111</td>
<td>2085 ± 112</td>
</tr>
<tr>
<td>Peak Ultrasound Velocity 1 (ms⁻¹)</td>
<td>2189 ± 100</td>
<td>2311 ± 70 *</td>
</tr>
<tr>
<td>Minimum Ultrasound Velocity 2 (ms⁻¹)</td>
<td>1908 ± 143</td>
<td>1994 ± 137</td>
</tr>
<tr>
<td>Peak Ultrasound Velocity 2 (ms⁻¹)</td>
<td>2091 ± 122</td>
<td>2181 ± 71*</td>
</tr>
</tbody>
</table>

* Statistically significant difference between RFS and FFS groups (P < .05); RFS, Rearfoot foot–strike running pattern; FFS, Forefoot foot–strike running pattern; %GC, percentage of gait cycle; BW, normalized to body weight
Joint moment contributions to forward and upward acceleration of body CG are affected by the foot to ground contact model

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INTRODUCTION: Induced acceleration analysis (IAA) is based in the dynamic coupling effect caused by the multiarticulated nature of the body, meaning that when a muscle contracts it produces acceleration, not only in those segments that are spanned by that muscle but on all body segments of the chain, due to the intersegmental forces. IAA has being used to ascertain the contribution of the different muscle groups to the forward progression and vertical acceleration applied to the body center of gravity (CG) in gait and more recently in sports activities where maximal effort is demanded from the athletes. Nevertheless the IAA results are sensitive to the type of constraints and particularly to foot to ground modeling. The purpose of our work was to explore the different contribution and synergist’s action of the ankle plantar flexors moment in relation to a free or a fixed foot contact simulation in hopping and sprinting.

METHODS: For the sprinting data 1 male national elite sprinter performed a sprint start from blocks and for the hopping data 1 national elite male triathlete performed a series of 10 unilateral hopping’s. Motion and ground reaction forces of the first step after leaving the blocks in sprinting, and the 10 hopping’s were captured at a sampling frequency of 200Hz using an optoelectronic system of 8 infrared cameras (Qualisys Oqus 300 & QTM software), synchronized in time and space with a force plate. The best trial for each movement was selected for analysis in Visual 3D (C-Motion software). A biomechanical model, composed by 8 rigid segments (HAT, pelvis, bilateral thighs, shanks and feet) was built, and optimized through inverse kinematics. The contribution of all joint moments and gravity to the horizontal and vertical acceleration of the participant’s center of mass was computed through an induced acceleration analysis, the dynamic equations of motion can be expressed in the following form (eq.1):

\[ \ddot{\theta} = M^{-1}(\theta)\tau + M^{-1}(\theta)C(\theta, \dot{\theta}) - M^{-1}(\theta)G(\theta) \]  (eq.1)

Where \( \ddot{\theta} \) is the joint acceleration term, \( M^{-1} \) is the inverse inertia matrix (where the segments inertial parameters and CM positions are taken into account), \( \tau \) is the joint moment’s term, \( C \) and \( G \) are the Coriolis and Gravitational terms, respectively. Given equation 1, \( C \) and \( G \) terms are set to zero allowing us to obtain the accelerations produced by each of the joint moments and the generated GRF. In this study two different contact models were used to establish the connection between the foot and the environment. In the first one, here designated as free-foot system, the foot–floor interface was modeled using a hinge joint with the axis of the hinge passing through the center of pressure in a direction parallel to the medio-lateral axis of the foot. In the second model, called fixed-foot system, the foot was constrained to the floor so that no motion of the foot was permitted. The free foot model was used when the foot to ground sagittal angle was changing and the fixed foot was used when this angle deed not varied.

RESULTS: As expected, different contributions of the lower legs joint moments of force were perceived when using the two types of foot contact model. In hopping, when free foot was used the main contributor to vertical acceleration were the ankle plantar flexors nevertheless when the fixed foot model was used, the foot/ground angle was fixed, the contribution of knee extensors was predominant and a synergistic action of ankle and knee extensors was reveled, with the ankle plantar flexors being responsible for fixing the foot to ground interaction and allowing the transfer of knee extensors action to the body CG upward acceleration. In sprinting a comparable synergistic action was observed when using a fixed foot model when no movement between foot and ground was computed but in this case between the hip extensors and plantar flexors, allowing the hip extensors to propel the CG forward.

DISCUSSION: The results we obtained showed that the IAA is sensitive to the foot to ground contact modeling solution and this was especially relevant to understand the major role of the ankle plantar flexors of guaranteeing the rigidity of the foot to ground contact probably due to a quasi-isometric behavior of these muscles and so contributing to the transfer of mechanical power developed by hip and/or knee extensors to the ground during an important phase of stance in explosive sports movements.

REFERENCES:

ACKNOWLEDGEMENTS: to Tom Kepple for the interpretation of IAA results.
Foot kinematics of forefoot and rearfoot strikers on recreational runners using the Rizzoli’s foot model

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INTRODUCTION: Several studies investigated and compared biomechanical patterns of different foot strikes, mostly regarding kinetic and kinematic of the lower limbs; even though those studies rarely report foot kinematics either for not using a multi-segment foot model or for analyzing shod running [1]. The few biomechanics-related variables described in running focuses on planter arch dynamics and rearfoot kinematics. The literature has showed lower limb kinematic differences when comparing rearfoot (RS) and forefoot strikers (FS) for knee and ankle’s range of motion (ROM) and their angles at the initial contact, vertical loading rates, running economy [1,2]. However, the literature lacks information about how the foot moves during running, and which adjustments the foot joints performs during each foot strike technique. Therefore, our aim was to investigate forefoot, midfoot and rearfoot kinematics and compare them between naturally RS and FS strikers in recreational long-distance runners.

METHODS: Kinematic data of twenty healthy long-distance recreational runners of both sexes was assessed using 8-camera motion capture system (Vero - VICOM) at 200 Hz, while they ran barefoot on an AMTI™ force-sensing tandem treadmill at self-selected speed at 1kHz. The Rizzoli’s Foot Model (IOR foot) marker protocol was used [2]. Runners’ foot strike pattern was identified using both kinematic and kinetic online data and slow-motion sagittal plane video recorded of each runner [3]. The maximum angle and the ROM of 12 foot-related variables were analyzed across 10 steps for each 10 runners from each group (RS and FS). Groups were not different in body mass (64.5±12.6kg), height (168.5±7.7cm), age (41.0±5.9yr.), self-selected running velocity (9.8±1.9km/h), and foot posture index (2.0±4.4). Groups were compared using t-test for each kinematic variable (p<0.05). (Ethic committee HC-FMUSP 031/15).

RESULTS: Differences between groups were found for the range of the angle between the first metatarsal to the ground (p=0.006), the second metatarsal to the ground (p=0.001) (Figure 1). Difference between groups was also found between the midfoot and metatarsus in the sagittal plane (p=0.05). No other differences were found.

DISCUSSION: Results revealed smaller range of motion of toes joints in the FS runners compared to RS runners. Possibly a greater foot strength developed by FS strikers enables a better control and damping of the body weight during the stance phase leading to a more controlled foot motion during running. Modelling foot kinetics using foot kinematic and ground reaction forces might help understanding the mechanisms that generated such foot mechanics differences. Increasing the sample size would reveal more differences because we had some results borderline to significance. Other analysis would also be possible, such as measuring the stance in two phases.

SIGNIFICANCE/CLINICAL RELEVANCE: This study proposes a better comprehension of the foot segments motion and its variability among runners with different strike strategies. Investigating normal foot kinematic patterns, variability and differences between FS and RS strikers are the basis for many clinical and surgical interventions in this population, and they might also be used as controlling variables of effectiveness.

REFERENCES:

ACKNOWLEDGEMENTS: The authors acknowledge FAPES (2015/14810-0; 2016/17077-4 ;2016/21304-6) and CAPES for the scholarships and grant.

Figure 1 Temporal patterns of toes angles during stance phase of treadmill running. F2G is the sagittal plane angle of the 1st metatarsal to the ground. S2G is the sagittal plane angle of the 2nd metatarsal to the ground. Mean (solid line), plus and minus a standard deviation (band) were calculated.
Session 16: Minimalist Shoes (SS)

3:40PM – 5:10PM

Chair: Irene Davis, PhD, PT Director of the Spaulding National Running Center and Professor of Physical Medicine and Rehabilitation, Harvard Medical School

Co-Moderators: Isabel Sacco, PhD Physical Therapy at the University of São Paulo

Our foot, with its 26 bones, 33 articulations, and 4 layers of arch muscles is well-adapted to both walk and run without the aid of footwear. The first forms of footwear were designed simply to protect the sole of the foot from injury. The running shoes of just 50 years ago were generally constructed with a soft upper and a sole and would be considered minimal shoes by today’s standards. Through the following years, running shoes have become increasingly more supportive and cushioned. There was then a swing towards more minimal running shoes such as the Nike Frees and the Vibram Five Finger shoes. Studies have indicated softer landings, and greater muscle strength when using shoes without cushioning and support. However, reports of transition injuries began to appear which has encouraged the proliferation of safe transitioning programs.

Walking shoes have also become progressively more cushioned and supportive over the years. Elderly, and those with pathological conditions such as knee OA, are often recommended to wear cushioned and supportive footwear. However recent studies by a number of groups have suggested that these individuals actually exhibit improved mechanics, as well as improved pain and function with minimal footwear….suggesting less may be more.

In this symposium, we will examine the evidence related to how minimal footwear influences mechanics, and benefits the musculoskeletal system in a varied population, suggesting this footwear may have a broad application. In addition, we will discuss safe transition, clinical applications, identify current barriers, and future areas for research.

10 min  Irene S Davis, PhD, PT  Symposium introduction

15 min  Irene S Davis, PhD, PT  Biomechanical comparison of minimal shoes to conventional shoes

15 min  Sarah T Ridge, PhD (Brigham Young University)  Effect of minimal footwear & foot strengthening on the MSK system of the foot
15 min  Isabel CN Sacco, PhD (Physical Therapy at the University of Sao Paulo)  Can minimal shoes improve knee osteoarthritis?

15 min  Jacob Hofer, MD (Esopus Medical)  How minimalist shoes have transformed one doctor’s practice

20 min  Irene S Davis, PhD, PT  Interactive exchange
i-FAB2018 Wrap-up

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Meeting Adjournment