2010 ESM

art science

proceedings

Providence, Rhode Island
August 14-17, 2010

Brown University
The Center for Restorative
and Regenerative Medicine
in cooperation with
University of Rhode Island

IN DEO SPERAMUS

WWW.ESM2010.COM

Content

ESM 2010 Program

Welcome

Welcome to Providence
Rhode Island's Renaissance City

It is my pleasure to welcome you to the Center for Restorative and Regenerative Medicine for the 2010 ESM Conference. The Center is a VA Center of Excellence and is comprised of researchers from the Providence VA Medical Center, Brown University and Massachusetts Institute of Technology. It is an exciting time for our center as we continue to grow and add new research programs. We have recently moved to a new building which houses clinical and translational rehabilitation research programs and regenerative medicine laboratories all in one facility. This includes the Gait and Motion Analysis Laboratory, the Neurorehabilitation Lab and facilities for outcome studies.

The meeting that we have planned for you should be a particularly exciting one both scientifically and socially. The scientific program will combine 39 oral presentations from over 60 submitted abstracts, a poster session, a workshop for novel users and two interesting keynote lectures.
In addition to the outstanding scientific program, you will have the opportunity to explore the art and cultural diversity of the city and beyond. You will have the opportunity to attend WaterFire, a powerful work of art and a moving symbol of Providence's renaissance and enjoy the conference venues chosen to highlight the ambiance of the city. The activity day on Block Island will reveal the beauty of an island, which offers exploration, relaxation and adventure.

On behalf of the Center of Restorative and Regenerative Medicine and Brown University, it gives me great pleasure to host this event. Enjoy the meeting and your stay in Rhode Island!

Best regards,

Susan E. D'Andrea
Director, Gait and Motion Analysis Laboratory
The Center for Restorative and Regenerative Medicine
Assistant Professor, Department of Orthopaedics, Brown University

Host

The Center for Restorative and Regenerative Medicine is a collaboration between the Providence VA Medical Center and Brown University. The mission of the Center is to improve function for individuals with limb trauma by developing technologically advanced solutions for the restoration of limb function. To achieve this goal, the Center brings to bear state-of-the-art techniques in tissue engineering, orthopaedics, neurotechnology, prosthetic design, and rehabilitation. These are complementary techniques and they converge in the concept of the biohybrid limb – composed of both biological and non-biological materials – enabling us to envision solutions that transcend the limitations of biological tissue or prosthetic materials alone. Biohybrid structures are composed of both biological tissue and non-biological components. Examples in current clinical use are joint replacements, in which metal implants are integrated directly into bone. Biohybrid structures often have unique physical and physiological properties resulting from the integration of tissues and materials that require full understanding before they can be most effectively utilized in the clinical setting. Biohybrid limb research integrates independent developments in regenerative medicine, neurotechnology, prosthetics, and orthopedics to maximize limb function.

THE CENTER FOR RESTORATIVE AND REGENERATIVE MEDICINE

The biohybrid limb is conceptualized as consisting of biological tissues and non-biological materials. Conceptualizing a limb as a biohybrid organ frees researchers and clinicians from constraints imposed by biological tissue and biomaterials, respectively.

Venue

Brown University

- Workshops

The Center for Restorative and Regenerative Medicine
830 Chalkstone Ave.
Providence R.I. 02908
www.biomed.brown.edu/orthopaedics/CCRM/newsletter_jan09.html

Hotel Providence

- Welcome Reception
- Conference/Presentations
- Banquet Dinner

139 Mathewson Street
Providence RI 02903
tel 401-861-8000
www.hotelprovidence.com

Host and Conference Chair

Susan E. D'Andrea, Ph.D.
Director, The Gait and Motion Analysis Laboratory

The Center for Restorative and Regenerative Medicine
Providence VA Medical Center

Assistant Professor of Orthopaedics (Research)
Department of Orthopaedics

The Warren Alpert Medical School of Brown University
Providence, RI 02906

401.273.7100 x 6232 (Office)
susan.dandrea@gmail.com

Organizing Committee

Susan Diekrager, Maria Pasquale, Susan D'Andrea, Daniela Jírová-Enzmann, Michael Morlock, Domenic Bahmann

Scientific and art in science award Committee

Michael M. Morlock, PhD
Director, Institute of Biomechanics
TUHH Hamburg University of Technology
Denickestrasse 15
D-21073 Hamburg, Germany
e-mail: morlock@tuhh.de

Rami J. Abboud, PhD
University of Dundee
Institute of Motion Analysis & Research (IMAR)
University Dept. of Orthopaedic & Trauma Surgery
Ninewells Hospital & Medical School
DD1 9SY, Dundee, UK
e-mail: r.j.abboud@dundee.ac.uk

Gert-Peter Brüggemann, PhD
Institute of Biomechanics and Orthopaedics,
German Sport University Cologne
Am Sportpark Müngersdorf 6
50933 Colone, Germany
e-mail: brueggemann@dshs-koeln.de

Thomas G. McPoil, PhD
Regents Professor
Department of Physical Therapy
P.O. Box 15105
Northern Arizona University, Flagstaff, AZ 86011, USA
e-mail: tom.mcpoil@nau.edu

Dieter Rosenbaum, PhD
Motion Analysis Lab
Orthopaedic Department
University Hospital Muenster
Domagkstr. 3
48149 Muenster, Germany
e-mail: diro@uni-muenster.de

Julie Steele, PhD
University of Wollongong
Biomechanics Research Laboratory
University of Wollongong
Northfields Ave
Wollongong, NSW, Australia
e-mail: julie_steele@uow.edu.au

Thomas Mittlmeier, PhD
University of Rostock
Dept. Orthopaedic Trauma
Schillingallee 35
18055 Rostock, Germany
e-mail: thomas.mittlmeier@med.uni-rostock.de

Marc Libotte, PhD
Cabinet d'Orthopédie
Av. des Héliotropes 13
1030 Brussels, Belgium
e-mail: marc.libotte@skynet.be

Susan E. D'Andrea, PhD
Director, The Gait and Motion Analysis Laboratory
The Center for Restorative and Regenerative Medicine
Providence, RI 02906, USA
e-mail: susan.dandrea@gmail.com

Jeff Pisciotta
Nike Inc
Sport Research Laboratory
One Bowerman Drive,
MH-1 Beaverton, OR 97005.6453, USA
e-mail: jeff.pisciotta@nike.com

Peter Cavanagh, PhD
University of Washington
Dept of Orthopaedics and Sports Medicine
1959 N.E. Pacific St.
Box 356500
Seattle, WA 98195, USA
e-mail: cavanagh@u.washington.edu

Helen Woo, PhD
USA
e-mail: h_woo@ymail.com

Workshops & Sponsor

Workshops

Axel Kalpen	Natalie Wilhelm
Maria Pasquale	Andrew Burke
Susan D'Andrea	Julie Stebbins
Helen Huang	Claudia Giacomozzi

Sponsor

novel gmbh
Munich, Germany
www.novel.de

www.novel.de

novel electronics inc
St. Paul MN, USA
www.novelusa.com

www.novelusa.com

2008 Meeting Review: Dundee, the City of Discovery

The 11th emed scientific meeting in 2008 was hosted by Professor Rami Abboud and his team at the Institute of Motion Analysis & Research (IMAR) at the University of Dundee in bony Scotland. The meeting was an excellent event shared by over 100 scientists who attended from 17 different countries to make the 11th user meeting a great success.

The novel award 2008 was presented to Scott Wearing by Michael Morlock, Scientific Chair

ESM 2008 host, Rami Abboud, presented the poster award 2008 to Claudia Giacomozzi

WRITTEN BY PROF. RAMI ABBOUD

On the 1st day, three workshops were delivered at IMAR for Novel users. The 1st two workshops highlighted the recent advances in the Novel arsenal of new technology and an up to date to the recent software development for the evaluation of pedographic data in clinical and scientific settings. The 3rd workshop was an excellent display of the power of synchronisation of various systems with the Novel systems for clinical assessment, e.g. Vicon and Pedar. The day ended with a wine reception at the Bonar Hall at the University.

Science took centre stage on day 2 and 4 of the meeting with 30 oral presentations, 20 posters and 4 keynote speakers displaying a diverse theme of Art in Science and creating a healthy debate. The oral presentation and poster Novel awards winners were Scott Wearing, UK (novel award 2008), Claudia Giacomozzi, Italy (poster award 2008) and Tobias Mayer, Germany (presentation award 2008).

Delegates were welcomed to the RRS Discovery ship after an intense scientific day only to be bombarded by another keynote lecture, but this time about the history of Dundee by one of the University most eminent historians, Professor Charles McKean. The lecture was followed by a Civic reception, co-sponsored by Dundee City Council and Dundee & Angus Convention Bureau, and tour of the ship. Unbelievably the weather was still good with hardly any rain!

Day 3, the most important day: the activity day! We were promised a miserable rainy Scottish day; Group 1 did not care as they were going to be inside Blair Castle and then sampling the Edradour whiskey. Group 2 did not care either as they chose Water sports; does it matter if water is above or under then! The day turned out to be one of the warmest and sunniest day in Scotland that was enjoyed by all and finished with a lovely dinner at the Cairngorm Mountain, albeit a bit of a delay to oil the trains!

After a hectic 4th day of scientific presentations and more debate, everybody looked tired and was looking for a relaxing dinner during the Novel award Ceremony. Relaxing dinner in Scotland...never. A Scottish-theme Tartan presentation and most importantly Ceilidh were on hand to make sure that everybody has the opportunity to never forget Scotland....

The team at IMAR worked closely with the team at Novel to organise what was perceived by many as one of the best ESM meeting in the hope that this tradition carries on with many years to come thanks to one man's vision, Peter Seitz!

novel award
art in science award
Poster & Presentation awards

In 1991, the first novel award was presented in recognition of excellence in pressure distribution research. The novel award recipient was determined by an international review committee from the fields of biomechanics and medicine. The novel award for pressure distribution measurement research continued since then to be endowed by novel and the best scientific manuscript in the field of load distribution measurement will receive the 2010 art in science award with a prize of $5,000.00. The paper must be entirely original, not published at the time of the meeting in any journal nor submitted for publication to any journal or book. The paper must describe a scientific study including pressure distribution measurement. Seven abstracts were nominated for the art in science award.

From 2010 on, the "novel award" will be named the "art in science award". In the future, the prize will be awarded also to scientists presenting papers not only associated with pedography or foot biomechanics, therefore the award going forward will be the "art in science award".

The nominated authors were requested to send a full length paper by the June 15, 2010 date and will present the paper at the meeting during a 25 minute talk. The review of the papers was conducted by the scientific review committee. The art in science award will be presented to the winner at the final reception on Tuesday evening, August 17, 2010.

Presentation award

Will be presented during the final reception on Tuesday August 17, 2010.
The award is based on voting of the scientific committee. The award winner will receive a prize of $500.

Poster award

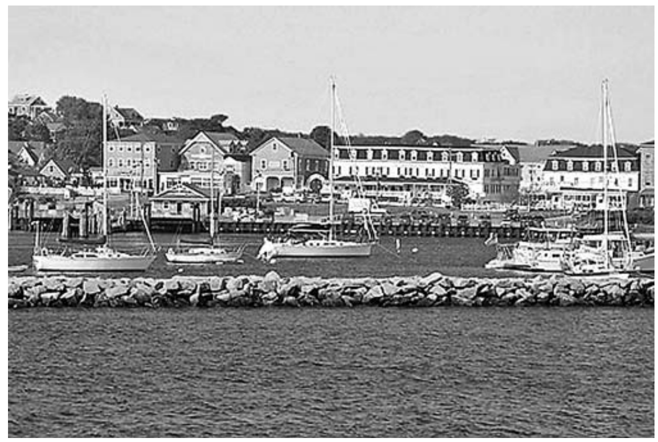
Will be presented during the final reception on Tuesday August 17, 2010.
The award is based on voting of the scientific committee. The award winner will receive a prize of $500.

Previous Winners

Dundee, 2008, Scott Wearing UK
Spitzingsee, 2006 Wolfgang Potthast Germany
Leeds, 2004, Joshua Burns Australia
Leeds, 2004, Mark Thomson Germany
Kananaskis, 2002, Katrina S. Maluf USA
Munich, 2000, Matthew Nurse Canada
Calgary, 1999, Brian Davis USA
Brisbane, 1998, Margret Hodge Australia
Tokyo, 1997, Erez Morag USA
Pennstate, 1996, Dieter Rosenbaum Germany
Ulm, 1994 Michael Morlock Germany
Vienna, 1991 Benno Nigg Canada

Scott Wearing, UK, winner of the novel award 2008

Activity Day
Monday
August 16, 2010



The traditional ESM Activity Day will be held on the beautiful Block Island www.blockislandinfo.com. Transportation from the hotel to Block Island will be provided via a Ferry ride. Once on Block Island guests may enjoy hiking, biking, kayaking, sailing, swimming and much more. After a full day of activities The Atlantic Inn will host an early evening New England Clam Bake. www.atlanticinn.com

Sailing – Ruling Passion 2.0 hours
A family run company that has shown off spectacular Block Island sunsets for more than 10 years. They sail their 45-foot trimaran from New Harbor three times a day. The spacious, comfortable vessel is USCG certified for 29 passengers. Ruling passion is owned and operated by the Puckett family. Fun outing for all ages. www.rulingpassion.com

Guided Hiking – Hodge Family Preserve 1.5 – 2.0 hours
At the north end of Corn Neck Road is this picturesque 25-acre field sloping to Middle Pond and West Beach. The property was acquired for conservation in 2002 and is jointly owned and managed by the town, The Nature Conservancy, the BI Land Trust and the Block Island Conservancy. Walkers and others engaging in passive recreation are welcome.

Guided Hiking – Fresh Pond Greenway & Rodman's Hollow – 4.0 – 5.0 hours
This area is called "Smilin' Thru" and reportedly inspired the 1920s song composed by Arthur Penn, who stayed on the island. Walkers should be prepared for steep slopes and some tricky footing. The trails starts on Lakeside Drive near the intersection with Cooneymus Road, and heads west along the shore of Fresh Pond. This trail links to Rodman's Hollow. Rodman's Hollow is a dramatic hollow that drops below sea level and features panoramic views of the Atlantic Ocean as well as access to the beach. Owned and protected by the Block Island Conservancy, Rodman's Hollow is a memorable hike and the keystone to the island's conservation efforts.

Guided Kayaking – 1.5 hours
View wildlife and the scenic shoreline of the fascinating Great Salt Pond. Suitable for beginners and experienced kayakers.

Guided Biking – 3.0 hours
All Block Island biking is on road.

For those attendees choosing the shorter activities, biking, beach, and Old Harbor shopping will also be available. Destination information will be provided at the ESM 2010 conference site.

Telephone Numbers and Events

Providence

- Hotel Providence (rooms, conference site) – 401-861-8000
- Center for Restorative and Regenerative Medicine (Workshops) – 401-273-7100 ext 6234
- Susan D'Andrea office – 401-330-1477
- On-site help via novel cell (mobile) phone – 612-221-0505
- Providence Police Department 401-272-3121
- Emergency Service: 911
- Providence Convention and Visitors Bureau – 401-456-0200, 401-751-1177 www.goprovidence.com
- Yellow taxi service 401-941-1122 (other taxi services are available)
- Amtrak Train (train code - PVD): 401-727-7388 www.amtrak.com

Events

Additional information can be found in packets distributed within the conference bag

- WaterFire
- Burnside Park
- Trolley Tours
- Newport Tours
- Lighthouse Tours
- Roger Williams Park Zoo
- Federal Hill
- Providence children's museum
- John Brown House
- RISD Museum

Meeting Overview

Saturday, August 14

Site:
Hotel Providence – Lobby
Registration for Workshop attendees **10:30 – 11:30**
Transportation to Workshops from Hotel Providence Lobby **11:30**

Site:
Center for Restorative and Regenerative Medicine
novel Workshops **12:00 – 17:00**
Tour of the Neurorehabilitation Lab **15:00 – 15:30**
Transportation from the Workshops to Hotel Providence **17:00**

Site:
Hotel Providence – Lobby and Blackstone Terrace
Registration **17:30 – 19:30**
Opening Reception in Blackstone Terrace **17:30 – 19:30**

Site:
Memorial Boulevard Providence
Providence WaterFire throughout
downtown Providence **19:30**

Sunday, August 15

Site:
Hotel Providence
Breakfast in Lobby **7:30**
Scientific presentations in Ballroom **8:15**
Lunch in Lobby **12:00**
Scientific presentations in Ballroom **13:00**
Posters and Break in RISD Room/Johnson & Wales Room **15:00**

Site:
RISD Museum of Art
Hors d'oeuvres, drinks and dinner **18:00**
Music by Henriette Gaertner
Tour Museum until **21:00**

Monday, August 16

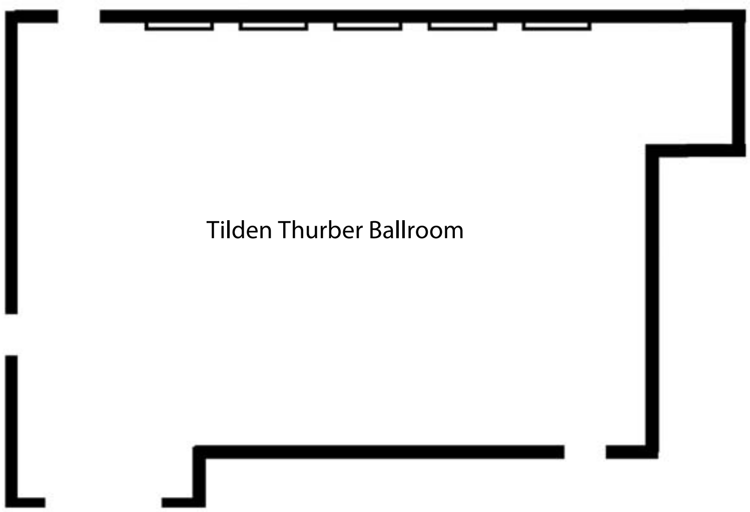
Site:
Block Island Activity Day
Breakfast pickup **6:00**
Buses depart **6:30 MUST BE in lobby by 6:15**
Ferry Ride from Point Judith to Block Island **8:00**
Arrival on Block Island, distribution of bikes and lunches **9:30**
Clam Bake at The Atlantic Inn **16:00**
Walk back to Old Harbor **19:00**
Ferry Departure **19:30**

Tuesday, August 17

Site:
Hotel Providence
Breakfast in Lobby **8:00**
Scientific presentations in Ballroom **9:00**
Lunch in Lobby **12:00**
Scientific presentations in Ballroom **13:00**

Site:
Aspire Restaurant (Hotel Providence)
Banquet Dinner, Music and Award Presentations **19:00**

Hotel Providence Floor Plans



Enter through the Hotel
Lobby located on 1st Floor

HOTEL
PROVIDENCE

139 Mathewson Street Providence, RI 02903
401.861.8000 | www.hotelprovidence.com

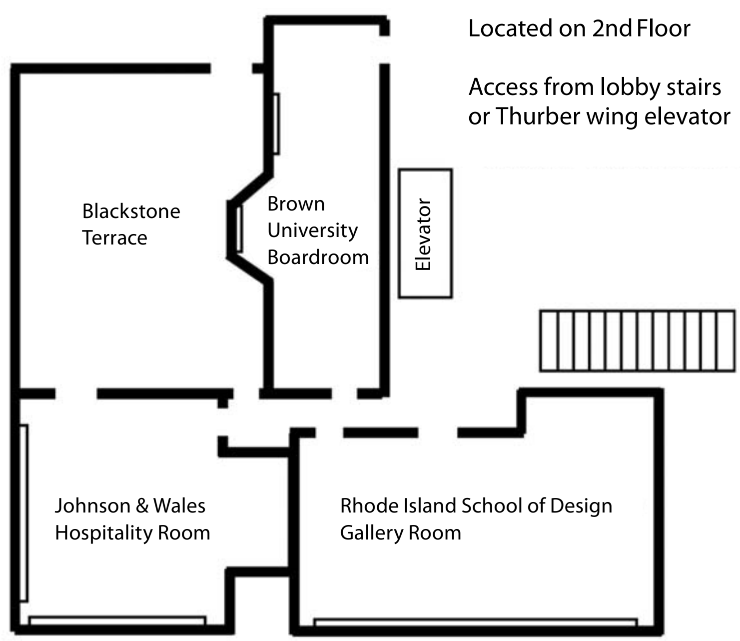
Opening Reception located on the Blackstone Terrace

Poster session located within the Johnson & Wales and RISD Gallery Rooms

Speaker Ready Room located in the Brown University Boardroom

All **scientific talks** will occur within the Ballroom

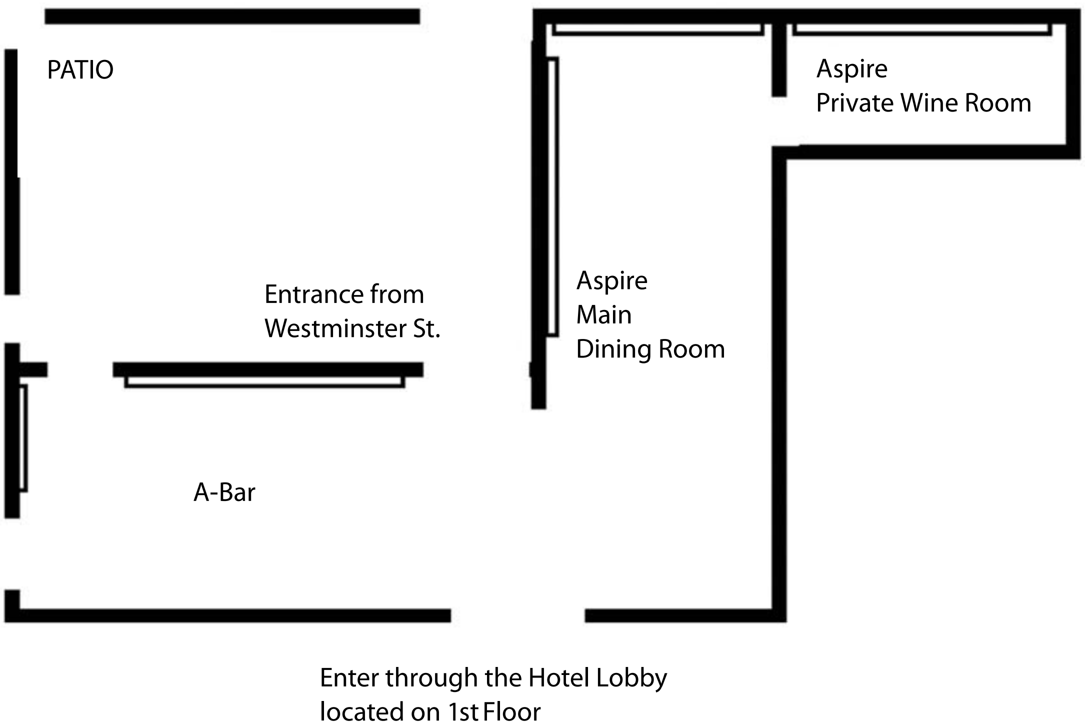
Registration will occur within the Hotel Lobby



Located on 2nd Floor

Access from lobby stairs
or Thurber wing elevator

Hotel Providence Floor Plans



Tuesday Evening Banquet will be located at Aspire Restaurant, Wine Room, Bar and Patio.

Program ESM 2010 Providence

Saturday, August 14th 2010

Welcome to ESM 2010

10:30 – 11:30	**Registration for the Workshop attendees in the Hotel Providence Lobby**
11:30	**Transportation to Workshops from Hotel Providence Lobby**
	novel Workshops at Center for Restorative and Regenerative Medicine and tour of Neurorehabilitation Lab Each session will be approximately 45 minutes, with 15 minutes for questions. There will be a 10-15 minute break between each session.
12:00 – 13:00	**Workshop I: novel hardware, data collection and analysis software overview** · Which system is most appropriate for which application · Parameters which are calculated with each system · New parameters available in novel software · Meaning of calculated parameters
13:10 – 14:10	**Workshop II: Project overview including setup of mock data collection and analysis** · Come up with new study idea · Decide how to collect data · Data collection · Data analysis (projects, database) · Generation of reports · Export of results to be used with other programs · "Pitfalls" in planning data collection or comparing results
14:20 – 16:00	**Workshop III: pliance sensor applications** **(Tour of the Neurorehabilitation Lab 15:00 – 15:30)** · Types of sensors and how they differ · Which sensor properties need to be taken into consideration for each application · Measurement of hands, sockets, chairs, garments, etc. · Methodologies and data collection protocols for pliance
16:00 – 17:00	**Workshop IV: 3D motion capture systems and pedography** · Overview on synchronization modes for emed/pedar/pliance for combination of kinematic and kinetic data · Purpose of using markers to help identify anatomical areas of feet · Foot models used for masking of emed data · Examples of this method for healthy and pathological subjects · Results of types of novel standard masks compared to marker based masks · Usage of this methodology in daily clinical routine in general
17:00	**Transportation from the Workshops to Hotel Providence**
17:30 – 19:30	**Registration in the Hotel Providence Lobby** **Opening Reception in Hotel Providence, Blackstone Terrace**
19:30	**WaterFire Providence Memorial Boulevard**

Sunday, August 15th 2010

7:30 – 8:15	**Breakfast, Registration** Hotel Providence Lobby
8:15 – 8:30	**Welcome** *Susan E. D'Andrea, ESM 2010 Host* *Peter Seitz, novel gmbh* *Leslie Klenerman*
8:30 – 10:00	**SESSION 1 Foot Deformities** Chair: *McPoil T.*
8:30 – 8:55	**art in science award finalist** **First ray instability in hallux valgus deformity –** **a kinematic radiographic and pedobarographic analysis** *Dietze A., Bahlke U., Martin H., Mittlmeier T.*
8:55 – 9:20	**art in science award finalist** **Is hallux valgus associated with different peak plantar pressure and pressure-time integrals?** *Galica A., Dufour A.B., Hillstrom H.J., Lenhoff M. W., Frey J. C., Casey V. A. , Hannan M. T.*
9:20 – 9:32	**Comparison of geometry hallux angles with radiographic hallux abductus angles** **for predicting hallux abductovalgus** *Furmato J. A., Song J.*
9:32 – 9:44	**Plantar loading in the cavus foot** *Kraszewski A. P. , Chow B., Backus S.I., Deland J.T., Demp P.H., Song J., Heilman B. , Rajan S. , Woodley A., Hillstrom H.J.*
9:44 – 9:56	**Foot structure is related to foot function** *Mootanah R., Song J., Backus S., Deland J., Demp P., Heilman B., Lenhoff M., Frey J., Kraszewski A., Hillstrom H. J.*
9:56 – 10:25	**Break**
10:25 – 12:00	**SESSION 2 The Evolution of Gait** Chair: *Pisciotta J.*
10:25 – 11:10	**Keynote lecture 1** **Human Walking: Comparative and evolutionary approaches** *Wunderlich R., James Madison University*
11:10 – 11:35	**art in science award finalist** **Effect of shoe flexibility on plantar loading in children who are learning to walk** *Hillstrom H., Buckland J.M., McCarthy C., Scher D., Root L., Backus S.I., Kraszewski A.P., Frey J., Song J., Whitney K., Choate C., Scherer P.*
11:35 – 11:47	**Dynamic plantar pressure changes during loaded gait** *Teyhen D.S., Goffar S.L., Reber R.J., Christiansen B.C., Miller R.B., Naylor J.A., Rodriguez B.M., Walker M.J.*
11:47 – 11:59	**'Kidfoot Muenster' – Nine-year results of plantar pressure measurements** **in childrens' foot development** *Rosenbaum D., Bosch K., Gerss J.*
12:00 – 13:00	**Lunch, Hotel Providence Lobby**

13:00 – 15:00	**SESSION 3 Pressure Distribution Measurements as a Diagnostic Tool** Chair: *Ford K.*
13:00 – 13:12	**Dynamic foot loading patterns in children with juvenile idiopathic arthritis (JIA)** *Rosenbaum D., Michels H., Hartmann M.*
13:12 – 13:24	**Plantar fasciitis and pain symptom are related to the longitudinal arch shape and not to the plantar pressure during running** *Sacco I. C.N., Ribeiro A.P., Trombini-Souza F., Tessutti V., João S.M.A.*
13:24 – 13:36	**Foot pressure distribution following operative reduction of high grade intra-articular fractures of the calcaneus** *Ayalon M., Ben-Sira D., Nyska M., Hetsroni I.*
13:36 – 13:48	**Plantar pressure distribution following operative treatment for proximal femur fractures** *Brunk M., Emmerich J., Mittlmeier T.*
13:48 – 14:00	**The Māori foot: static morphology and dynamic function in healthy and diabetic populations** *Gurney J., Kuch C., Rosenbaum D., Kersting U.*
14:00 – 14:12	**Asymmetry in plantar loading during gait in Native Americans with and without diabetes and with and without neuropathy** *Kernozek T.W., Heizler C., Greany J.F.*
14:12 – 14:24	**Plantar pressure distribution patterns in multiple sclerosis patients with different neurological status** *Tsvetkova T.L., Stolyarov I. D., Pichugina O.L., Petrov A.M., Ilves A.G. , Prakhova L.N. , Nikiforova I.G., Lebedev V.V. , Yakushev F.V. , Tartakovskiy V.N.*
14:24 – 14.36	**Can gait initiation process be evaluated with pressure platforms?** *Rosenbaum D., Lobo da Costa P.H., Bosch K.*
14.36 – 14.48	**The effects of footwear, learning, and fatigue on center of pressure excursion during single limb balance** *Zaferiou A.M., Zifchock R.A., Brown A.M., Frey J., Hillstrom H.J.*
14.48 – 15.00	**Relationship between foot range of movement and plantar pressure distribution in diabetic neuropathic patients** *Sacco I., Sartor C.D., Picon A.P., Roveri M.I., Dinato R.C.*
15:00 – 16:00	**Posters & Break** RISD Room/Johnson & Wales Room, Hotel Providence
	The effect of special shoe insert designed for diabetic patients on plantar foot pressure distribution *Ayalon M., Hetsroni I., Nyska M.*
	Effect of gait speed changes on foot loading characteristics in children *Bosch K., Westhues M., Rosenbaum D.*
	Effects of intense running to exhaustion on the in-shoe plantar pressure patterns in young middle-distance athletes *Fourchet F., Kelly L., Horobeanu C., Millet G.P.*
	Clinical applications of a forefoot conic curve model *Hillstrom H., Kraszewski A., Demp P., Chow B., Lenhoff M., Song J., Heilman B., Rajan S., Woodley A.*

	Variation of tactile cues reduce nociceptive capacity of plantar irritating stimulus impact on walking gait *Janin M., Dupui P.* **Effect of interlocking pattern in electrical bed on the prevention of bedsore** *Lim D., Choi H., Kim J.H., Hong J.S., Chun K.J.* **Biomechanical factor to be considered during power-lift design to reduce the risk of musculoskeletal disorders** *Lim D., Choi H., Chun K.J.* **Variations in plantar loading patterns in individuals with soft tissue versus boney rearfoot trauma: a preliminary study** *McPoil T.G., Albin S., Cornwall M.W.* **Effects of sub-hallucial wedge and medial arch support on dynamic plantar pressure** *Song J., Nimick C., Banks J., Marroquin R., Hood C., Tango D., Kane R., Furmato J., Whitney K.* **The effects of slippers and lower limb positioning on plantar pressures in selected ballet balances** *Valenti É.E., Lobo da Costa P.H., Bosch K., Rosenbaum D.* **Dynamic effects of different alternating cycle time conditions on soft tissue perfusion recovery** *Won B.H., Choi Y.J., Chun K.J., Song C.* **A novel approach using a FE-foot model for clinical applications** *Wyss C.* **Reliability of in-sole plantar pressure using simple and detailed masks of the forefoot, midfoot, and hindfoot** *Akins J.S., Keenan K.A., Dugan B. P., Francis M. F., Abt J.P., Sell T.C., Lephart S.M.* **The influence of fatigue, ligament laxity and hormonal fluctuation on plantar pressure in female athletes** *Silke K.T., Wunderlich R. E. , Cebulski S. F.* **Comparison of plantar pressure measurements obtained during barefoot and shod conditions** *Akins J.S., Keenan K.A., Dugan B.P., Francis M., Abt J.P., Sell T.C., Lephart S.M.*
16:00 – 17:00	**Podium discussion I** **Pedography as a Diagnostic Method for Determining Foot Function: Technical, Methodological and Other Requirements** *Hillstrom H., Kalpen A., Giacomozzi C., McPoil T., Rosenbaum D.* Moderator: *Bus S.*
18:00 – 21:00	**Evening event** **RISD Museum of Art** Hors d'oeuvres, drinks and dinner Music by Henriette Gaertner Museum Tour

Monday, August 16th 2010

	Block Island Activity Day
6:00	Breakfast pickup
6:30 (be in the lobby by 6:15)	Buses depart MUST BE in lobby by 6:15
8:00 – 9:30	Ferry Ride from Point Judith to Block Island
9:30	Arrival on Block Island, distribution of bikes and lunches, activities begin
16:00 – 19:00	Clam Bake at The Atlantic Inn
19:00	Walk back to Old Harbor
19:30	Ferry Departure

Tuesday, August 17th 2010

8:00 - 9:00	**Breakfast, Hotel Providence Lobby**
9:00 – 10:25	**SESSION 4 Gait and Orthoses** Chair: *Stine R.*
9:00 – 9:25	**art in science award finalist** **Cast-boot ankle-foot orthoses yield greater forefoot load reduction than total contact casts** *Gutekunst D.J., Bohnert K.L., Hastings M.K., Sinacore D.R.*
9:25 – 9:50	**art in science award finalist** **Preservation of the first rocker is related to increases in gait speed in individuals with hemiplegia and AFO** *Nolan K.J., Yarossi M.*
9:50 – 10:02	**Effect of CAD designed pedorthosis with build-in wedge for children with clubfoot** *Liu X.C., Rizza R., Thometz J., Lyon R. , Tassone C. , Tarima S.*
10:02 – 10:14	**Intraarticular and muscle force reactions of the leg using different insoles** *Wyss C., Roethlisberger M., Gerber H.*
10:14 – 10:26	**Ratios of lateral to medial patellofemoral forces and pressures in a simulated operative environment of total knee arthroplasty** *King J.J., Cerynik D.L., Peretz J.I. , Black A., Johanson N.A.*
10:26 – 10:59	**Break**
11:00 – 12:00	**SESSION 5 Analysis Methods** Chair: *Nurse M.*
10:59 – 11:24	**art in science award finalist** **Calculation of pressure time integral, a different approach** *Melai T., IJzerman H., de Lange T.LH. , Willems P. B., Meijer K., Schaper N. C., Savelberg H.HCM.*

11:24 – 11:36	**Footprint masks: reproducibility of the Oxford anatomically-based selection by means of novel geometrical masks** *Giacomozzi C., Stebbins J., Kalpen A.*
11:36 – 11:48	**A technique to assess plantar soft tissue stiffness during the stance phase of gait** *Roposh C., Tucker C., Orlin M., Seliktar R.*
11:48 – 12:00	**Reliability of dynamic foot geometry assessment using a pedobarographic platform and a two-step approach** *Akins J.S., Keenan K.A., Dugan B.P., Francis M., Abt J.P., Sell T.C., Lephart S.M.*
12:00 – 13:00	**Lunch, Hotel Providence Lobby**
13:00 – 14:00	**SESSION 6 Data Collection Methods and Gait** Chair: *Rosenbaum D.*
13:00 – 13:12	**Comparison of a six degree-of-freedom force sensor and pressure insole measurements in Selected skiing Manoeuvres** *Kersting U.G., Kurpiers N. , Hild E. , Kiefmann A., Senner V.*
13:12 – 13:24	**Data collection from two different pressure platforms using standardised methodology produces comparable results** *Stebbins J., Giacomozzi C.*
13:24 – 13:36	**Do motion control running shoes decrease sagittal plane movement of the medial longitudinal arch?** *McPoil T. G., Cornwall M.W., McNulty M. R.*
13:36 – 13:48	**Lateral plantar loading during cutting correlates to decreased footwear rating** *Ford K.R., Manson N. A., Myer G. D., Gwin R. C., Heidt R. S. Jr, Hewett T. E.*
13:48 – 14:00	**Crossover second toe deformity: A robotic cadaveric study of increased second metatarsal length and plantar pressure** *Whittaker E. C., Weber J. R., Aubin P.M., Sangeorzan B.J., Ledoux W. R.*
14:00 – 14:30	**Break**
14:30 – 16:30	**SESSION 7 Dynamic Load Distribution** Chair: *Abboud R.*
14:30 – 14:55	**art in science award finalist** **Hypothenar pressure mapping provides insight for reducing ulnar nerve compression in cyclists** *Slane J., Timmerman M., Ploeg H., Thelen D.*
14:55 – 15:07	**Using centre of pressure (COP) data to identify golf putting techniques** *McLaughlin P., Best R.*
15:07 – 15:19	**Effect of mattress for prevention of decubitus, bed position and subjects' bmi on interface pressure** *Bae T.S., Mun M.S.*
15:19 – 15:31	**A study of the effects of gel liner thickness on in-socket residual limb pressures in trans-tibial prosthesis users** *Stine R., Hansen A., Tucker K., Boutwell E., Edwards M., Gard S.*

15:31 – 15:43	**Force and impulse of keystroke during piano playing. Differences among professional players in classical passages** *Gaertner H., Pozzo R., Riehle H.*
15:43 – 16:28	**Keynote lecture 2** **The evolution of human running** *Lieberman D. E., Harvard University*
16:30 – 17:00	**Podium discussion II** **The Perfect Running Shoe is the Bare Foot?** *Pisciotta J., Lieberman D., Davis I., D'Andrea S. E., Wunderlich R., Nurse M.* Moderator: *Morlock M.*
17:00 – 17:15	**Closing remarks** *Peter Seitz* *Susan D'Andrea*
19:00	**Evening event** **Aspire Restaurant (Hotel Providence)** Banquet Dinner Music by Joe Potenza Award Presentations

Presentations

Presentations

Presentations

Poster Presentations

Abstracts

The following contains the abstracts for each platform and poster presentation throughout the conference.

FIRST RAY INSTABILITY IN HALLUX VALGUS DEFORMATY - A KINEMATIC RADIOGRAPHIC AND PEDOBAROGRAPHIC ANALYSIS

Dietze Albrecht[1], Ulf Bahlke[2], Heiner Martin[2], Mittlmeier Thomas[1]

Chirurgische Universitätsklinik Rostock, Abteilung für Unfall- und Wiederherstellungs-chirurgie[1], Rostock, Germany
Institut für Biomechanik Universität Rostock[2]

INTRODUCTION

Manifest hallux valgus deformity remains an abundant fore foot condition, whose entire or definite cause remains controversial. The extent of first ray instability and in particular of the first tarso-metatarsal (TMT-I) joint of the affected foot remains a key argument in the debate for optimal surgical treatment in hallux valgus conditions (Coughlin 2003). Numerous attempts have been undertaken to more precisely investigate and quantify first ray instability. Klaue (1994) designed an apparatus to evaluate static TMT-I instability and combined a clinical study with a cadaver radiograph analysis for validation of the device. Glasoe et al. (2000) modified the Rodgers apparatus in order to respect dynamic stabilization aspects of the foot. Additionally in the pedobarographic foot print of hallux valgus patients a deformity related reduced weight bearing of the first ray has been described (Lorei, 2006).

In our study, pedobarographic findings were correlated to the radiographically determined degree of first ray mobility of the foot.

PATIENTS AND METHOD

For our study 8 patients presenting a hallux valgus deformity on one foot or both feet, who were registered for corrective foot surgery out with our hospital, were enrolled. Seven were female one was male. Mean age was 44 (range 15-65). Clinical symptoms like foot pain, AAOF-Score and hallux valgus angle and intermetarsal angle were recorded. For simultaneous radiographic and pedobarographic measurement a mobile C-arm fluoroscope (ziehm, Nuremburg) and a plantar pedobarographic measurement platform (emed novel, Munich) were combined in an experimental setting to allow dynamic gait analysis in the entire stance phase of walking (including heel set down and fore foot take off). Pulsing radiograph recording (frequency 4 pulses/s) allowed digital analysis in selected frames and determination of motion among the joints of the first ray of the foot in particular the TMT-I joint (CAD program Solid Works, 2007). Simultaneously the pedobarographic footprint was recorded by the platform and peak pressure as well as maximum force were analyzed. Statistical analysis of hallux valgus radiological characteristics and pedographic findings of the first ray was by Pearson´s correlation.

RESULTS

All enrolled patients presented a manifest hallux valgus deformity and met requirements for operative treatment by an experienced foot surgeon blinded to the study set up. The mean intermetatarsal angle was 15°(range 8°-28°) and the mean hallux valgus angle was 45° (range 12°-80°). A large intermetatarsal angle correlated significantly with an increase of peak pressure under the first metatarsal (p=0,014). Furthermore the radiological computer-assisted determined maximal dorsiflexion of the first ray (from talus till metatarsal head) significantly correlated with the extent of the intermetatarsal angle (p=0,012). Mobility of the first tarso-metatarsal (TMT-I) joint however showed a significant correlation to an increase of maximum force of the mid-forefoot region.

DISCUSSION

Mobility of the first metatarsal ray and in particular of the first tarso-metatarsal joint has been the subject of extensive research. Various devices had been developed to quantify first ray mobility of the foot. As in our study first ray deviation was mainly analyzed in the sagital plane. In our study we can support the notion of an enlarged intermetatarsal angle being associated with an increased dorsiflexion of the first ray of the foot during gait. Further more we found instability of the TMT-I –joint to have an increasing affect on mid-forefoot maximum force that may be relevant for pathological correlates.

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IS HALLUX VALGUS ASSOCIATED WITH DIFFERENT PEAK PLANTAR PRESSURE AND PRESSURE-TIME INTEGRALS?

Andrew M. Galica (1), Alyssa B. Dufour (1, 4), Howard J. Hillstrom (2), Mark W. Lenhoff (2), Jocelyn C. Frey (2), Virginia A. Casey (1), Marian T. Hannan (1, 3)

1. Institute for Aging Research, Hebrew SeniorLife, Boston, MA, USA
2. Hospital for Special Surgery, New York City, NY, USA
3. Harvard Medical School, Boston, MA, USA
4. School of Public Health, Boston University, Boston, MA USA

INTRODUCTION

While there are many clinical and case reported studies of hallux valgus (HV), the etiology and biomechanics of the pathology remain poorly understood. Although previous studies have noted differences in peak pressures in various regions of the foot between individuals with and without HV (Kernozek 2003, Yamamoto 1996, Yavuz 2009), results are inconsistent and have not been confirmed in larger studies. The purpose of this research is to describe peak plantar pressures and pressure-time integrals in an epidemiological population-based study and to investigate whether these measures differ between those with and without HV as defined by standardized foot examinations. We hypothesize that it is possible to distinguish individuals with and without HV based on differences in peak pressure and pressure-time integral measures.

METHODS

Data were obtained from a subset of participants enrolled in the Framingham Heart Study (N = 464; 57% female; mean age, 65 years; mean BMI, 28). Between 2002 and 2005, plantar pressure values were collected using a Tekscan Matscan system (model 3150, resolution of 1.4 sensels/cm^2) while participants walked at a comfortable pace barefoot across the mat. Data was imported into Novel software and masked into 12 segments (toes, submetatarsal heads 1-5, medial arch and heel, lateral arch and heel). Peak pressure and pressure time integral values were then calculated. Two-tailed Students T-tests assessed differences between those with and without HV.

TABLES

Table 1: Peak Pressure Differences

Measure	Peak Pressure, N/cm^2		
	No Hallux Valgus	Hallux Valgus	P Value
FEMALE	n = 172	n = 87	
Hallux	18.5 ± 6.1	18.4 ± 6.1	0.84
Submetatarsal head 1	17.7 ± 5.8	17.1 ± 5.8	0.46
Submetatarsal head 2	22.2 ± 4.8	21.9 ± 5.1	0.67
MALE	n = 155	n = 45	
Hallux	19.3 ± 7.0	18.0 ± 7.3	0.27
Submetatarsal head 1	18.9 ± 5.9	19.3 ± 7.6	0.80
Submetatarsal head 2	23.0 ± 3.9	22.8 ± 5.2	0.82

Table 2: Pressure Time Integral Differences

Measure	Pressure Time Integral, (N/cm^2)×sec		
	No Hallux Valgus	Hallux Valgus	P Value
FEMALE	n = 171	n = 87	
Hallux	7.9 ± 3.9	8.0 ± 4.6	0.90
Submetatarsal head 1	8.4 ± 3.7	8.0 ± 4.2	0.50
Submetatarsal head 2	10.7 ± 3.1	10.9 ± 5.1	0.72
MALE	n = 155	n = 45	
Hallux	7.7 ± 4.5	8.2 ± 5.8	0.61
Submetatarsal head 1	8.7 ± 3.7	10.5 ± 7.9	0.15
Submetatarsal head 2	10.9 ± 3.1	11.9 ± 7.0	0.38

RESULTS

Preliminary analysis revealed significant sex differences in plantar pressures. Therefore, subsequent analyses were stratified by gender. Since the right foot is considered more dominant, only these results are reported. Tables 1 and 2 present a subset of the 12 analyzed foot areas. Female subjects with HV exhibited greater peak pressure values under the lateral arch than those without HV (p = 0.046). No significant differences were found in pressure-time integrals for females with and without HV, regardless of foot area. Male subjects with HV exhibited greater peak pressure (p=0.007) and pressure-time integral (p=0.015) values than those without HV under groupings of the 3rd, 4th, and 5th toes.

DISCUSSION

Significant gender differences in peak pressures and pressure time integrals were noted. The data did not support our hypothesis as analyses did not distinguish between those with and without HV based on pressure-related measures. It is possible that the accuracy and resolution of Tekscan, in comparison to other foot mat systems, may have affected our results. It is also possible, since both groups contained concomitant pathologies, the plantar loading effects for HV washed out. Future work may investigate whether consideration of additional foot disorders or deformities may better use the plantar pressure measures to distinguish foot pathologies.

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COMPARISON OF GEOMETRY HALLUX ANGLES WITH RADIOGRAPHIC HALLUX ABDUCTUS ANGLES FOR PREDICTING HALLUX ABDUCTOVALGUS

James A. Furmato, Jinsup Song

Gait Study Center, Temple University School of Podiatric Medicine, Philadelphia, PA

INTRODUCTION

Several criteria determine the presence and severity of the bunion deformity (hallux abductovalgus; HAV), a common and disabling foot disorder. Radiographic relationships in the weight-bearing dorso-plantar view include the intermetatarsal angle, tibial sesamoid position, and hallux abductus angle (HAA). The HAA determines the amount of hallux deviation in HAV. While the radiographic technique is well accepted, it requires a small dose of radiation. Evaluating foot structure without ionizing radiation may be clinically useful.

The emed™ Geometry software (novel GmbH, Munich) calculates two hallux angles based on dynamic footprint: HA1 based on the relationship of the medial hallux to the medial foot border and HA2 relating the center of the Hallux to the estimated bisection of the angle between the first and second metatarsals, see Figure 1(left) and 1 (center). Alternatively, investigators wanted to examine the utility a graphical analog of HA1, constructed from a scan of the plantar weight bearing foot (MVIa), see Figure 1 (right).[1] However, there has been no comparative study between radiographic versus the previously mentioned alternates.

Investigators examined the utility of a using the angles form the Geometry software or a graphical analog of HA1 or HA2 based on a flatbed scan of the plantar weight bearing foot collected to determine the Malleolar Valgus Index (MVI)[1]. When HAV is present, the Hallux Abductus angle is greater than 15 degrees[2].

METHODS

This is a retrospective study. Twenty eight (28) feet with clinical diagnosis of HAV and 10 feet without were included. Clinical assessment of HAV, radiological measure of HAA, MVI foot scans and EMED-X dynamic footprints were available for each subject. Bivariate linear fit of HA1, HA2 and MVIa against HAA was evaluated using JMP 8 (SAS Institute, Cary, NC)

RESULTS

Correlation coefficients and root mean square errors of HA1, HA2, and MVIa to corresponding HAA measurements were calculated (0.707 and 4.17, 0.334 and 14.8, and 0.822 and 6.17, respectively (all p-values <0.01). Descriptive statistics are shown in Table 1. However, none of parameters was able to distinguish between those feet with HAV and those without when one way Analysis of Variance was performed .

Figure 1: Radiographic angle (left), emed Geometry angles (center), HA1 MVI analog (right).

Table 1: Mean and standard deviation (parentheses) of four measurements with (HAV+; n=28) and without (HAV-; n=10) clinical finding of hallux abductovalgus.

	HAA	HA1	HA2	MVIa
HAV+	25.8 (7.2)	16.1 (5.8)	28.6 (18.0)	19.2 (9.0)
HAV-	18.0 (8.9)	13.1 (5.7)	27.9 (15.4)	13.2 (9.2)

CONCLUSIONS

When a plantar weight-bearing scan of the foot is available, a reasonable estimate of the presence of bunion can be determined based on the relationship of the medial hallux to the medial foot border. However, none of these variable by itself was able discern feet with HAV when evaluated via univariate analysis. Additional work is needed.

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PLANTAR LOADING IN THE CAVUS FOOT

Andrew P. Kraszewski[1], Betty Chow[1], Sherry I. Backus[1], Jonathan T. Deland[1], Philip H. Demp[2], Jinsup Song[2], Benjamin Heilman[2], Sonali Rajan[1], Andrea Woodley[1], Howard J. Hillstrom[1,2]

1. Leon Root MD, Motion Analysis Laboratory, Hospital For Special Surgery, New York, NY
2. Temple University School of Podiatric Medicine, Temple University, Philadelphia, PA

INTRODUCTION

Data are available describing plantar loading planus and rectus foot types[1], but little is available describing the loading of the healthy cavus foot. The specific aim of this project was to compare plantar loading during gait for asymptomatic healthy individuals as a function of foot type (pes planus, rectus, and cavus).

METHODS

We hypothesized asymptomatic healthy individuals with pes planus, rectus, and pes cavus foot types will show differences in plantar loading. Sixty-one subjects were stratified according to the resting calcaneal stance position and the forefoot to rearfoot relationship into pes planus, rectus, and cavus groups. Plantar pressures were recorded with an Emed-X system (Novel, Munich, Germany) while subjects walked barefoot at their self-selected comfortable speed. Five steps per side were analyzed with Novel masking software. Peak pressure (PP), maximum force (MF), pressure-time integral (PTI), force-time integral (FTI), and contact area were calculated over ten masked regions. Data were analyzed with a univariate mixed-effect analysis of variance (ANOVA) model, followed by Bonferroni post-hoc t tests where significance was found. Foot type and replication were modeled as fixed and random effects, respectively. ANOVA significance was set at $p \leq 0.05$; Bonferroni post-hoc significance was set at $p \leq 0.017$.

RESULTS

Table 1 tabulates the results for PP and MF. Hallux PP and MF values were significantly higher for planus foot types. PP and MF beneath the 1st metatarsal head (MTH1) in planus feet were significantly lower than both rectus and cavus feet. The 5th MTH PP and MF values were significantly higher for rectus and cavus as compared to planus feet. PP and MF was lowest under the lateral arch for cavus feet. Cavus feet showed significantly higher PP at the lateral heel than rectus or planus. MF beneath the medial heel was significantly different between rectus and cavus feet. The planus medial arch had twice the contact area of rectus, which had twice the area of cavus.

Region	**Peak Pressure, N/cm²** Mean(SD)			
	Planus *N=27*	**Rectus** *N=22*	**Cavus** *N=12*	***P-hoc**
Hallux	43.9(6.9)	37.1(5.6)	32.5(5.8)	1,2
MTH1	28.1(8.0)	35.8(6.6)	37.3(6.8)	1,2
MTH2	51.1(5.8)	37.7(4.8)	38.8(4.9)	1,2
MTH3	40.4(4.3)	34.2(3.6)	34.8(3.6)	1,2
MTH4	27.8(3.6)	26(3.0)	25.6(3.1)	†
MTH5	20.1(5.8)	26.3(5.2)	25.3(5.3)	1,2
LatHeel	33.8(4.0)	33.6(3.3)	38.4(3.4)	1,3
MedHeel	37.2(4.8)	36.5(4.0)	38.8(4.1)	
LatArch	11.5(1.7)	10.9(1.4)	7.7(1.5)	1,3
MedArc	20.1(5.8)	26.3(5.2)	25.3(5.3)	1,3
	Maximum Force, N Mean(SD)			
Hallux	135(17.1)	108.8(14.1)	91.5(14.5)	1,2,3
MTH1	134.9(23.4)	148.2(19.2)	157.2(19.7)	1
MTH2	177.1(16.2)	151.1(13.3)	152.1(13.6)	1,2
MTH3	172.6(17.8)	150.7(14.6)	152.3(14.9)	1,2
MTH4	104.6(14.9)	98.9(12.3)	100.8(12.6)	
MTH5	41(10.7)	51.6(8.8)	58.4(9.0)	1,2
LatHeel	221.2(19.4)	215(15.9)	207.1(16.3)	
MedHeel	265.4(21.6)	255.6(17.7)	241.7(18.2)	1
LatArch	107(25.8)	79.7(20.0)	52.7(20.5)	1,2,3
MedArc	26.8(7.7)	13.7(6.3)	6.2(6.5)	1,2,3

*Table 1. Plantar loading measurements across foot types. *P-hoc: 1=Cavus vs. Planus; 2= Rectus vs. Planus; 3=Cavus vs. Rectus ; †blank cells indicate a non-significant ANOVA result*

DISCUSSION

In most plantar regions several measures of plantar loading were sensitive to foot type. The lower 1st MTH loads support the "hyper-mobile 1st-ray theory" where load is supported by the 2nd and 3rd MTH in many planus feet. PTI and FTI corroborate this observation. Higher 5th MTH cavus loads showed, as expected, differences to those of planus feet. These data will serve as a reference for future investigations of pedal pathology as related to foot types, and for designing and evaluating treatments.

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FOOT STRUCTURE IS RELATED TO FOOT FUNCTION

Rajshree Mootanah (1), Jinsup Song (2), Sherry Backus (3), Jonathan Deland (3), Phillip Demp (2), Benjamin Heilman (3), Mark Lenhoff (3), Jocelyn Frey(3), Andrew Kraszewski (3), Howard J Hillstrom (3)

1. Anglia Ruskin University, Medical Engineering Research Group, Essex, UK.
2. Temple University, School of Podiatric medicine, PA, USA.
3. Hospital for Special Surgery, Rehabilitation Department, NY, USA.

INTRODUCTION

It has been suggested that foot structure may influence foot function (Song et al., 1996). The aim of this study is to investigate the relationship between foot structure and function as a predictive tool that could be used to plan more effective conservative and surgical treatments of pedal pathologies. We hypothesize that measures of foot structure (hindfoot alignment and arch height) are associated with biomechanical measures of foot function in asymptomatic healthy individuals.

METHODS

Foot structure was characterized by computing: (1) the maleolar valgus index (MVI) while standing, (2) arch height index (AHI) while sitting, and (3) AHI standing. MVI is a measure of static hindfoot alignment (Song et al, 1996). The subject's plantar foot is scanned while standing on a plexiglass platform over a flatbed scanner with a custom-made jig to register the lateral and medial malleolus. The deviation from the midpoint of the transmalleolar axis to the midpoint of the hindfoot, normalized to the foot width in this region, comprised the MVI. Note that AHI is the arch height at one-half of foot length normalized by the truncated foot length (Zifchock et al, 2006).

Foot function was characterized by calculating: (1) The center of pressure excursion index (CPEI - a measure of dynamic foot function), which is the lateral displacement of center of the pressure curve from the line constructed between the initial and the final center of pressure values, normalized by the foot width at the anterior one third of foot. (Song et al., 1996) The emed X system (Novel gmbh, Germany) and custom software, developed in C++, were employed to calculate the CPEI. Peak pressure (PP) and maximum force (MF) were calculated for the total plantar foot and each masked anatomical region using Novel software. Temporal-distance foot-fall parameters (e.g. step length, stride length, velocity, etc) were obtained with the GaitMatII (EQ systems, Glenside, PA). Each of 61 asymptomatic healthy adult test subjects walked at their comfortable self-selected speed across both the Emed-X and the GaitmatII systems to obtain the foot function data. Each subject also was structurally evaluated with MVI and AHI (sitting and standing).

RESULTS

Pearson correlation coefficients were calculated for each combination of structural and functional parameter for the entire cohort. The results are summarized in Table 1. MVI was significantly correlated with PP and MF at the hallux and negatively correlated with MF at the 1st MTPJ. AHI standing was correlated with PP and MF at the 1st MTPJ and negatively correlated with the PP at the 2nd MTPJ. AHI sitting was correlated with double support time and velocity. Step and stride lengths were negatively correlated with AHI sitting. AHI sitting was correlated with MF at the 5th MTPJ and negatively correlated with 2nd MTPJ PP and MF at the 1st MTPJ.

DISCUSSION

MVI was correlated with hallucial loading and negatively correlated with 1st MTPJ loading. This finding is consistent with the overpronation that accompanies valgus hindfeet and a hypermobile first ray. AHI was correlated with medial column loading as well as temporal-distance footfall parameters. Foot structure is correlated with foot function and one's basic gait pattern.

Table 1: correlation between foot structure & function

	AHI sit	AHI stand	MVI (%)
CPEI (%)			
Double support time	R=0.316, p=0.017		
Step Length	R=-0.354, p=0.007		
Stride Length	R=-0.340, p=0.010		
Velocity	R=0.339, p=0.010		
Peak Pressure-Hallux			R=0.380, p=0.004
Peak Pressure-1st MTPJ		R=0.320, p=0.015	
Peak Pressure-2nd MTPJ	R=-0.375, p=0.004	R=-0.316, p=0.017	
Maximum force Hallux			R=0.354, p=0.007
Maximum force 1st MTPJ	R=-0.344, p=0.009	R=0.265, p=0.046	R=-0.294, p=0.026
Maximum force 5th MTPJ	R=0.266, p=0.045		

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HUMAN WALKING: COMPARATIVE AND EVOLUTIONARY APPROACHES

Roshna E. Wunderlich

Department of Biology, James Madison University, Virginia

Human bipedal walking has a long and interesting evolutionary history. Fossilized feet and footprints have played a prominent role in the interpretation of the nature of bipedalism in early hominins. Recent discoveries of the fossil pedal remains of the Flores hominid (Figure 1), *Ardipithecus*, and ancient trackways have highlighted the diversity of early hominin bipeds and the mosaic of human and nonhuman skeletal features they possess. Proper interpretation of pedal function requires knowledge of form and function in living humans as well as our closest nonhuman primate relatives.

Figure 1: Late Pleistocene hominin foot from Flores, Indonesia (Jungers *et al*, 2009)

Human upright striding bipedal gait is unique among primates and is associated with a distinct foot roll-over pattern involving heel-strike, lateral midfoot pressure, lateral to medial pressure transfer across metatarsal heads, and a toe-off with high loads on metatarsals 1-2 and toe 1. Other primates exhibit higher medial midfoot pressures, higher lateral forefoot pressures, and lower toe pressure.

The human footfall pattern is associated with specialized anatomical features including broad calcanei and distinctly robust fifth metatarsals. The robust first metatarsal is broad dorsally, allowing a "close-packed" position during toe-off dorsiflexion rather than during grasping, and relatively short toes, decreasing mechanical work for toe flexors (Rolian *et al*, 2009).

Fossil hominids exhibit a unique combination of features. The Flores foot possesses a modern lateral column but a short hallux, long lateral toes and a weight bearing navicular (Figure 1). Our laboratory-based studies of functional anatomy in humans and other primates suggest that this hominin likely used a roll-over pattern resembling humans in early stance but similar to apes in later stance with medial midfoot weight bearing, a laterally-placed toe-off, and lack of full extension or high loads on the toes during toe-off (Figure 2). Analysis of fossil footprints provides additional insight into pedal function in early hominins when examined in the context of modern unshod human foot pressure. Taken together, paleontological and neontological studies of foot function provide deeper insights into modern human foot form, function, and pathology.

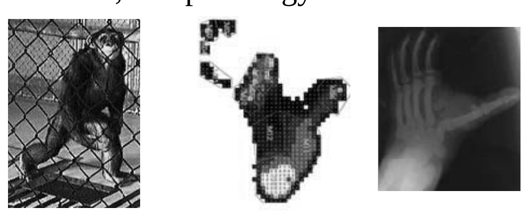

Figure 2: Chimpanzee walking bipedally Note the presence of pressure in the midfoot, the high pressures on metatarsals 2-3, and low toe pressures.

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EFFECT OF SHOE FLEXIBILITY ON PLANTAR LOADING IN CHILDREN WHO ARE LEARNING TO WALK

Howard J. Hillstrom (1,2), Melanie Buckland (1), Corinne McCarthy (1), David Scher (1), Leon Root (1), Sherry I. Backus (1), Andrew P. Kraszewski (1), Jocelyn Frey (1), Jinsup Song (2), Kendrick Whitney (2), Cherri Choate (2), Paul Scherer (3)

1. Leon Root MD, Motion Analysis Laboratory, Hospital For Special Surgery, New York, NY
2. Temple University School of Podiatric Medicine, Temple University, Philadelphia, PA
3. California College of Podiatric Medicine, Oakland, CA

INTRODUCTION

In a previous pilot study of 'cruisers' (non-independent ambulation), 'early walkers' (independent ambulation for 0 – 5 months), and 'experienced walkers' (independent ambulation for 6 – 12 months) developmental age significantly affected the children's stability when walking and performing functional activities.[1] The purpose of this investigation is to examine how shoe structural characteristics affect plantar pressure distribution in early walkers.

METHODS

We hypothesized that torsional flexibility of children's shoes affects plantar loading. Twenty-six children were evaluated in barefoot and each of four shoes that stratified a range of torsional flexibilities. The children were early walkers. Plantar pressures were recorded barefoot and shod with Emed-X and Pedar-X systems (Novel, Munich, Germany) respectively, at self-selected comfortable walking speeds. A minimum of five steps per side were analyzed using Novel masking software. Peak pressure (PP), maximum force (MF), pressure-time integral (PTI), force-time integral (FTI), and contact area were calculated over ten masked regions. Data were analyzed using a univariate mixed-effect ANOVA, followed by Bonferroni post-hoc t tests where significance was found. Footwear and replication were modeled as fixed and random effects, respectively. Significance was set at $p \leq 0.05$ for ANOVA and $p \leq 0.005$ for the post-hoc tests.

RESULTS

Peak pressure (Table 1, N/cm^2) for the right foot was significantly different across shoes for masked regions except the 5th MTPJ. In general, the stiffest shoe (ST) had the lowest PP while the most flexible shoe (UF) had the highest; ST and UF were also the most dissimilar and similar, respectively, to the barefoot pressures. Torsional flexibility (Figure 1, deg/Nm) showed a decreasing trend with increasing torsion angle across all footwear types. Highest flexibility was observed in the Ultraflex (pink line) footwear.

Peak Pressure (N/cm^2)	Barefoot (BF)	UltraFlex (UF)	MidFlex (MF)	LowFlex (LF)	Stiff (ST)	$p^{\dagger}$	*Post Hocs
Total	13.2(1.7)	13.3(2.2)	10.7(0.0)	11.8(1.9)	9.5(1.7)	**	2-5, 7,10
Hallux	11.2(1.5)	10.8(0.0)	10.8(0.0)	11.7(1.6)	8.3(0.0)	**	2,5, 8,10
1st MTPJ	8.1(1.1)	7.9(1.4)	7.5(0.0)	7.1(1.2)	6.2(0.0)	**	2-4, 7,9,10
2nd MTPJ	7.1(0.9)	8.1(1.1)	7.2(0.9)	7.4(1.0)	5.9(0.0)	**	1,4,5, 7,9,10
3rd MTPJ	6.0(0.6)	6.4(0.7)	5.3(0.0)	6.1(0.5)	5.5(0.0)	**	2,4,5, 7,9
4th MTPJ	4.9(0.8)	6.2(1.0)	5.8(0.0)	5.6(0.8)	5.6(0.0)	**	1,2,3, 4,7,9
5th MTPJ	3.9(0.8)	3.7(1.0)	3.5(0.9)	3.8(0.5)	3.9(1.0)	--	
LatHeel	8.2(1.2)	7.9(1.6)	6.0(0.0)	6.4(1.4)	5.2(0.0)	**	2-7, 9,10
MedHeel	8.9(1.4)	9.3(1.8)	6.5(0.0)	6.8(1.4)	5.3(0.0)	**	2-7, 9,10
LatArch	6.3(0.8)	7.2(0.9)	6.2(0.0)	6.7(0.8)	6.5(0.4)	**	1,5,8
MedArch	6.3(0.8)	7.1(1.0)	5.6(0.0)	5.8(0.8)	5.5(0.0)	**	1,2,4-6,7,9

*Table 1: Peak pressure values per plantar region across footwear type. *Post-hoc comparisons: 1=BFvsUF; 2=BFvsMF; 3=BFvsLF; 4=BFvsST; 5=UFvsMF; 6=UFvsLF; 7=UFvsST; 8=MFvsLF; 9= MFvsST; 10=LFvsST. †: --(p>0.05); **(p<0.001);*

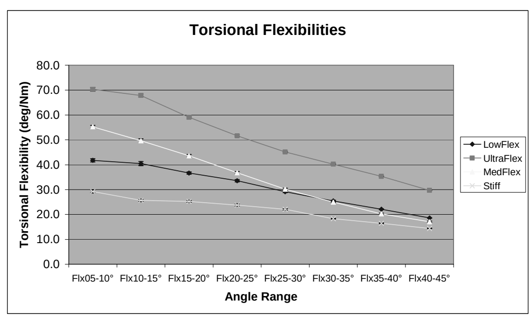

Figure 1: Plot of torsional flexibilities per shoe.

DISCUSSION

These data have implications for footwear design aiming to control plantar loading conditions that may also bear influence on a child's proprioception when learning to walk.

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T7

DYNAMIC PLANTAR PRESSURE CHANGES DURING LOADED GAIT

Stephen L. Goffar, Rett J. Reber, Bryan C. Christiansen, R. Benjamin Miller, Jacob A. Naylor, Brittany M. Rodriguez, Michael J. Walker, Deydre S. Teyhen

Doctoral Program in Physical Therapy, U.S. Army-Baylor University, Texas, USA

BACKGROUND

Lower extremity overuse injuries are reportedly the most common injuries in the military. Extreme values of static arch height (Williams, 2001) and the heavy loads commonly carried by military personnel in training and combat environments are associated with an increased risk of lower extremity overuse injuries (Bisiaux, 2008). While associations between static arch height and plantar pressure distributions have been demonstrated (Teyhen, 2009), limited knowledge exists regarding the impact of load carriage on plantar pressure distributions in the shod foot across arch types as delineated by AHI. The purpose of this study was to determine how load carriage affects dynamic plantar pressure distributions during gait in individuals with varying arch types.

METHODS

One hundred and fifteen healthy service members (97 males, 18 females, 31.3 ± 5.6 years, 177.1 ± 7.0 cm, 86.0 ± 11.0 kg) were enrolled in this study. They had no current medical condition which would preclude them from carrying up to a 40 kg load. Static measurements of heel to toe length (HTL) and arch height (AH) at 50% HTL were obtained with the Foot Assessment Platform System (FAPS) (McPoil, 2008). Arch Height Index (AHI) was calculated by dividing AH by HTL. Dynamic plantar pressure measurements were obtained using an in-shoe pressure measurement system (Pedar-x, Novel Electronics, Inc., St. Paul, MN, USA) while the subjects wore their own combat boots. Subjects walked for approximately 30 seconds at 3.0 mph on a treadmill under each of three levels of load: uniform without additional load (WB), 20 kg load including weapon, helmet, and body armor (20kg), 40 kg load adding weighted ruck sack (40kg). Load carriage sequence was counterbalanced.

DATA ANALYSIS

Participants were categorized by arch type based upon accepted cutoff values for AHI resulting in 28 high (AHI > .267), 61 normal, and 26 low (AHI < .229) arched right feet. An average of 9.8 ± .6 consecutive error-free steps were analyzed for each load condition. Maximum force (MaxF), force time integral (FTI), and pressure time integral (PTI), were calculated for regions of the plantar foot using a nine sector mask. Changes in each were analyzed with a 3×3 repeated measures ANOVA across the levels of load carriage and arch type.

RESULTS

There was a significant interaction between arch type and load for the MaxF ($p=.001$) and FTI ($p\leq.005$) in the medial midfoot. Although MaxF and FTI increased in all regions of the foot with load ($p<.001$) regardless of foot type, the forces in the medial midfoot were greater in those with low arches.

There was a significant interaction between arch type and load for the MaxF ($p=.004$) in the medial forefoot. MaxF was greater in the high arched feet relative to normal and low arched feet (<.001) across all loads. The reverse was true at the great toe region, in which low and normal arched feet demonstrated greater MaxF ($p \leq .004$) compared to high arched feet.

The relative distribution of PTI in the nine regions of the plantar foot increased proportionately regardless of foot type under all load conditions.

DISCUSSION & CONCLUSIONS

Higher forces in the medial midfoot in low arched feet may be related to the increased surface area in this region or may represent increased pronation. However, the relative increases in medial midfoot forces in low arch feet did not increase disproportionately with increases in load compared to normal or high arched feet.

Force distributions in the 1st ray differed based on foot type. Those with high arched feet had greater forces in the medial forefoot region, while those with normal or low arched feet had greater forces in the great toe region, regardless of load. These differences in force distributions may demonstrate different strategies to generate a rigid lever during toe-off.

Regardless of foot type, increases in load did not alter the relative distribution of pressure over the plantar foot. These findings possibly indicate a negligible impact of loads ≤ 40kg on footwear and orthoses prescription. However, differences in dynamic plantar pressure during gait based on AHI were supported.

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'KIDFOOT MUENSTER' – NINE-YEAR RESULTS OF PLANTAR PRESSURE MEASUREMENTS IN CHILDRENS' FOOT DEVELOPMENT

Dieter Rosenbaum (1), Kerstin Bosch (1), Joachim Gerss (2)

(1) Movement Analysis Lab, Department of General Orthopaedics and Tumor Orthopaedic
(2) Dept. of Medical Informatics and Biomathematics, University Hospital Münster, Germany

INTRODUCTION

In 1999, the 'Kidfoot Münster' project was initiated. The main scope was to investigate the individual development of the child's growing foot by assessing foot loading patterns in a longitudinal project. Over 100 children were recruited for this purpose. In previous reports we described preliminary results (Bertsch 2004, Bosch 2007; 2009, Unger 2004). Now the first group of children has successfully completed their nine years of study involvement so that we can present the changes in foot loading characteristics of healthy children from the onset of independent walking to the end of their elementary school age.

MATERIAL & METHODS

As soon as the children were able to walk without support for several meters they were invited to the lab for participation in this long-term investigation. Between 1 and 10 years of age, they were asked to visit the lab on 17 occasions. Initial visits were every 3 months during the first year, twice a year until the age of 6 and final visits were once per year. By the end of 2009, complete data sets of 36 children were available.

Dynamic foot loading patterns were measured during free walking across a capacitive platform (emed ST or X, 4 sensors/cm^2). For each foot, 5 trials were recorded and stored in a database (Medical Professional 13.3.30, Novel GmbH Munich). Five regions of interest (H=Heel, MF=midfoot, FF=forefoot, HX=hallux, LT=lateral toes) were analyzed with standard pressure parameters (PP=peak pressure, MF=maximum force, CA=contact area, AI=arch index).

RESULTS

Average peak pressure values of the total foot increased over time from 141 kPa to 409 kPa. The highest values were initially located under the hallux but moved towards the heel and forefoot. The relative maximum force increased especially in the heel (71%) and forefoot (87%) but decreased in the midfoot by 63%. Relative contact area decreased markedly under the midfoot. The arch index gradually decreased by 44% up to an age of six to seven years and then leveled off. The range of values indicates pronounced inter-individual differences (Fig. 1).

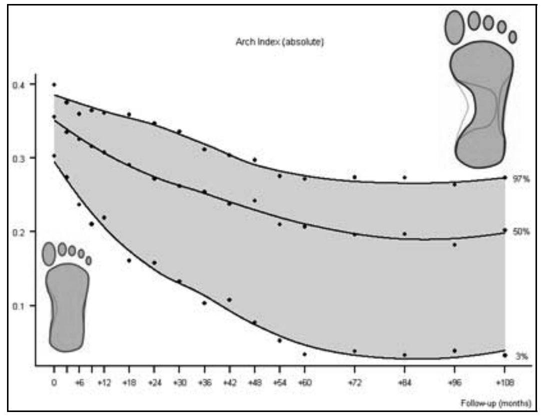

Fig. 1: Development of the arch index between the onset of walking and 9 years of walking experience (i.e. 1 to 10 years of age); 3rd, 50th & 97th percentile.

DISCUSSION

With the presented data we are able to describe – in more detail than before – the development of the child's healthy foot with respect to dynamic loading parameters. The results provide a range of normal values for the observed age range from one to ten years. These data may be used for clinical applications when potential pathologies in pediatric orthopedics shall be evaluated with comparable means.

While not overthrowing previous knowledge these data provide a more detailed insight into the individual foot development, the range of acceptable foot characteristics in certain ages, and finally the time course for the developmental changes.

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Contact: diro@uni-muenster.de, www.kidfoot.de

T9

DYNAMIC FOOT LOADING PATTERNS IN CHILDREN WITH JUVENILE IDIOPATHIC ARTHRITIS (JIA)

Dieter Rosenbaum (1), Hartmut Michels (2), Matthias Hartmann (2)

(1) Movement Analysis Lab, Department of General Orthopaedics and Tumor Orthopaedics University Hospital Münster, Germany
(2) German Centre for Paediatric and Adolescent Rheumatology, Garmisch-

Introduction

Juvenile idiopathic arthritis (JIA is an auto-immune disease that may affect various joints leading to specific malpositions with compensatory movements. Foot involvement is frequently encountered (Spraul & Koenning). A pronounced pain stimulus causes the children to respond with a pain-relieving position thus causing muscular dysbalance. Children with oligoarthritis can partially compensate the joint problems in the neighbouring joints maintaining an asymmetric but usually fairly smooth gait pattern. In children with polyarticular arthritis the adjacent joints are also affected and compensation is hardly possible. Therefore, these patients may develop gait pattern changes. The present study evaluated characteristic gait patterns in these patients in order to evaluate the need for specific treatment.

Material & Methods

Forty-two children, 20 oligoarthritis patients (**OA**, 15 girls/5 boys; 11.0±3.5 years) and 22 polyarthritis patients (**PA**, 15 girls/7 boys; 14.2 ±3.6 years) were examined. Clinical and pedobarographic data were collected during inpatient stay. Every participant passed a clinical examination and the visual analogue scale was used to assess the current pain intensity. Plantar pressure measurements (emed ST 4, Novel; 50 Hz) were carried out with the instruction to walk normal at self-selected speed and recording a step in full gait. Five valid trials of each foot were stored for subsequent analyses. Dynamic foot loading parameters (**PP**=peak pressure, **FTI**=force time integral) were evaluated in ten plantar regions of interest. Forces were normalized to body mass. The averaged data of right and left foot were used for comparisons. The Mann-Whitney U-test was used for statistical analysis.

Results

While the maximum PP of the total foot did not show statistical differences, particularly higher PP were found under the hindfoot in PA children (Tab. 1). Furthermore, significantly higher PP in the first, second and third to fifth metatarsal head region were seen in PA children. The FTI showed higher values in the lateral and medial hindfoot and also in the third to fifth metatarsal in PA patients. The clinical findings indicated restricted dorsiflexion and plantarflexion in PA patients more frequently than in OA patients.

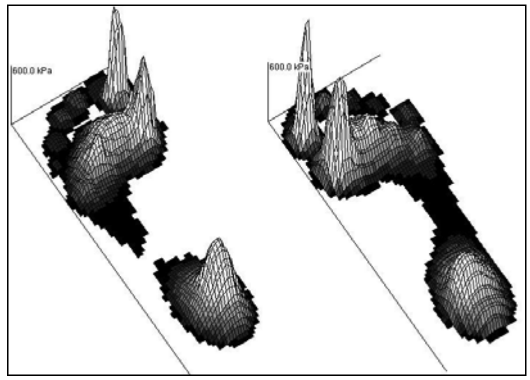

Fig. 1: Plantar pressure pattern of a selected PA patient #38 with elevated pressures under the first ray.

PP [kPa]	OA patients (n=20) Mean±SD	PA patients (n=22) Mean±SD	P-level
lat. heel	270±60	324±85	.016
med. heel	311±85	379±119	.037
MT-1	202±122	264±182	.049
MT-2	272±114	353±149	.022
MT-3-5	254±94	355±122	.008
FTI [Ns]			
lat. heel	29.9±14.0	41.4±18.4	.030
med. heel	36.8±15.2	50.8±22.6	.044
MT-3-5	49.7±20.9	68.3±33.2	.034

Tab. 1: Mean ± SD values of selected gait parameters of the two patient groups.

Discussion

The present clinical data in PA patients indicates changes in the loading response and terminal stance phase. This can be considered as a potential reason for higher PP and FTI values under the hindfoot and higher PP under the metatarsals in PA patients. Polyarticular arthritis may cause higher hindfoot and metatarsal loading as compared to oligoarthritis. This shows the need for preventive measures in these patients.

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acknowledgement

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Contact: diro@uni-muenster.de

PLANTAR FASCIITIS AND PAIN SYMPTOM ARE RELATED TO THE LONGITUDINAL ARCH SHAPE AND NOT TO THE PLANTAR PRESSURE DURING RUNNING

Ana Paula Ribeiro, Francis Trombini-Souza, Vitor Tessutti, Silvia M.A. João, Isabel C.N. Sacco

Physical Therapy, Speech and Occupational Therapy Dept., School of Medicine, University of São Paulo, Brazil.

INTRODUTION

Plantar fasciitis has been the third most common disease in runners, but its pathogenesis is still inconclusive. Changes in the longitudinal plantar arch and mechanical overload on the feet, have been described as risk factors for developing plantar fasciitis[1,2,3,4]. However, there are few studies investigating these factors during running in individuals with and without plantar fasciitis. Most of the literature investigated biomechanical parameters of the plantar fasciitis during gait and the results are still controversial, mainly because it is unclear the effect of pain associated with the disease over those parameters. Wearing *et al.* (2007)[3] discussed that symptomatic individuals make some adaptations during gait to reduce forces over the rearfoot and, consequently, increase loads over adjacent foot areas, such as over the mid and forefoot, as also observed by some authors[1,3]. According to biomechanical studies, it is possible that the plantar pressure distribution in individuals with plantar fasciitis would be different during the symptomatic and asymptomatic phases of the disease, and the pattern could be even more different during running. The purpose of this study was to investigate the association between plantar fasciitis and its pain symptom with the longitudinal plantar arch (LPA) shape and plantar pressure distribution during running.

METHODS

Ninety recreational runners were studied: 45 had plantar fasciitis (PF): symptomatic 30-SPF (45.4±8.1 yr, 69.6±14.0 kg, 1.68±9.2 m) and asymptomatic 15-APF (38.3±3.3 yr, 72.3±10.0 kg, 1.76±7.8 m) and 60 were healthy controls (CG) (35.0±9.0 yr, 66.8±12.0 kg, 1.71±9.0 m). Pain was assessed by a visual analogue scale (VAS). The LPA was evaluated by digital photogrammetry during weight bearing static posture[5]. The index between 0.22 – 0.25 was used to classified the LPA as normal; < 0.21 cavus; > 0.26 valgus[5]. The plantar pressure was evaluated by Pedar X system during running of 40 meters at a speed of 12±5% km/h. Runners used a common standardized sport footwear. Contact area, contact time and pressure peak were evaluated in 5 areas: rearfoot (medial and lateral), midfoot and forefoot (medial and lateral). Groups were compared using ANOVAs for repeated measures, followed by Tukey *post-hoc* tests ($p<0.05$).

RESULTS AND DISCUSSION

SPF group reported mean time since onset of pain of 7±2 months and pain levels of 5±2 cm. In APF and CG groups, the pain level was 0 cm. The LPA was more elevated in both groups with plantar fasciitis: symptomatic (0.17±0.08; cavus LPA; p=0.009) and asymptomatic (0.17±0.07; cavus LPA; p=0.008), compared to controls (0.22±0.05; normal LPA).

Table 1- Mean, standard deviation and comparison among groups with plantar fasciitis (symptomatic-SPF and asymptomatic-APF) and the control (CG) of plantar pressure variables during running.

Variable	*Group*	***Contact area*** (cm^2)	***Contact Time*** (ms)	***Peak Pressure*** (kPa)
Rearfoot medial	SPF (1)	12.2±1.6	134.4±23.9	337±84.8
	APF (2)	11.4± 3.0	135.1±28.6	322±124.1
	CG (3)	12.5± 1.4	147.3±32.9	306.2±61
	p (tukey)	0.902 (1-2)	0.958 (1-2)	0.985 (1-2)
		0.998 (1-3)	0.998 (1-3)	0.588 (1-3)
		0.520 (2-3)	0.850 (2-3)	0.982 (2-3)
Rearfoot lateral	SPF (1)	10.4 ± 2.5	135.4± 36.2	346.1± 97.1
	APF (2)	9.8± 2.7	137.1± 46.0	291.5± 99.4
	CG (3)	10.8± 2.4	149.3± 38.8	331.1± 91.2
	p (tukey)	0.601 (1-2)	0.229 (1-2)	0.834 (1-2)
		0.809 (1-3)	0.085 (1-3)	0.892 (1-3)
		0.364 (2-3)	0.968 (2-3)	0.542 (2-3)
Midfoot	SPF (1)	39.5± 5.1	182.8± 37.1	129.0± 29.0
	APF (2)	38.3± 6.6	179.2± 38.2	106.7±20.9
	CG (3)	41.1± 5.4	198.0± 32.3	124.1± 30.6
	p (tukey)	0.109 (1-2)	0.822 (1-2)	0.998 (1-2)
		0.271 (1-3)	0.998 (1-3)	0.995 (1-3)
		0.202 (2-3)	0.916 (2-3)	0.991 (2-3)
Forefoot medial	SPF (1)	33.0± 2.6	207.6± 25.7	346.6±99.9
	APF (2)	32.1± 3.0	216.8± 28.1	312.4±99.2
	CG (3)	33.8± 2.6	217.6± 28.0	374.4± 96.4
	p (tukey)	0.883 (1-2)	0.461 (1-2)	0.818 (1-2)
		0.702 (1-3)	0.355 (1-3)	0.995 (1-3)
		0.883 (2-3)	0.493 (2-3)	0.884 (2-3)
Forefoot lateral	SPF (1)	37.0± 3.5	218.7± 25.5	284.3± 58.9
	APF (2)	36.5± 4.0	221.5± 27.9	242.4± 66.1
	CG (3)	37.8± 3.6	226.1± 26.4	266.5± 77.6
	p (tukey)	0.800 (1-2)	0.995 (1-2)	0.565 (1-2)
		0.734 (1-3)	0.619 (1-3)	0.924 (1-3)
		0.987 (2-3)	0.923 (2-3)	0.419 (2-3)

The cavus architecture of the LPA would lead to greater strain in the plantar fascia during static and, mostly, during dynamic activities, such as running. Chronically, these stresses will cause micro traumas in the plantar fascia and, probably, will lead to the progression of PF symptoms or even to the onset of PF. However, the plantar pressure distribution during running did not demonstrate association to the PF or its pain symptom.

CONCLUSION

The plantar fasciitis and the pain symptom are not associated to the plantar pressure distribution patterns during running. However, runners with plantar fasciitis with or without pain symptom present cavus LPA and more elevated arch compared to controls.

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FOOT PRESSURE DISTRIBUTION FOLLOWING OPERATIVE REDUCTION OF HIGH GRADE INTRA-ARTICULAR FRACTURES OF THE CALCANEUS

Moshe Ayalon (1), David Ben-Sira (1), Meir Nyska (2), Iftach Hetsroni (2)

1. The Zinman College of Physical Education, Wingate Institute, Israel
2. Meir Hospital, Sapir Medical Center, Kfar Saba, Israel

INTRODUCTION

Displaced intra-articular fractures of the calcaneus represent a complex injury typically affecting middle-aged active population. The ideal goal of open reduction with internal fixation in these fractures is to recreate the calcaneus width, height and length, the subtalar joint congruency, the soft tissue balancing, and consequently the foot and ankle kinematics. It is not known whether operative intervention for these injuries provide any advantage to non-operative management in terms of recreating the normal foot and ankle kinematics during the gait cycle. The current study was designed to assess the plantar pressure profiles of the foot following open reduction with internal fixation of high grade intra-articular fractures of the calcaneus.

METHODS

The study sample included 14 operated patients and 8 control healthy subjects. Plantar pressure distribution was collected during walking on a level floor at a natural preferred cadence with Pedar-x system (Novel gmbh, Munich). 10 zones of the plantar area of the foot were defined. The subjects walked using the same shoe model on a level floor at natural preferred cadence. Plantar pressure data were collected over 4 cycles. For each step the pressure time integral (PTI) [Ns/cm^2] was calculated for each zone. In the injured group, the sound limb was compared with the injured limb. In addition, the sound limb and the injured limb were separately compared with the control group (16 feet). Because some of the variables did not fulfill the assumptions of normality that underlie parametric statistics, a non-parametric approach was applied using the Mann-Whitney test for between samples comparisons. A p value of 0.05 was considered significant.

RESULTS

In eight of the ten masks PTI was significantly lower in the involved legs than in the healthy sample. PTI of the injured group was expressed as percentage of the healthy group. The relative means of the PTIIs of the involved legs were lesser in the medial aspect than the lateral aspect of the foot (MH-42% vs. LH-80%, MMF- 35% vs. LMF-103% ns, 1MTT-51% & 2MMT- 67% vs. LMTT-81% ns, 1TOE-23% & 2TOE-15% vs. LTOE-51%). Similar trends characterize the uninvolved leg. Thus, injured patients adopt a modified gait pattern typified by reduced plantar pressure, particularly on the medial aspect of the foot.

DISCUSION

Both limbs of operated patients demonstrated a similar PTI pattern of relative reduction, particularly in the medial aspect, compared to the healthy feet. The cause for this pattern remains unclear at this point. A potential explanation for the similar pattern of the injured and sound limbs is that over the years since the operative intervention and the rehabilitation process, patients may have implemented compensatory mechanisms in their non-operated limbs to prevent chronic over-loading of one side caused by gait asymmetry. The observed symmetry in gait adjustment may be related to the self-selected walking velocity chosen by each patient, and could indicate good functional recovery as related to walking. However, measuring at self selected gait velocities could potentially mask kinematical differences between operated and contra-lateral non-operated limbs that could emerge in higher locomotive velocities or during perturbations while running, walking on uneven terrains and side sloped surfaces.

PLANTAR PRESSURE DISTRIBUTION FOLLOWING OPERATIVE TREATMENT FOR PROXIMAL FEMUR FRACTURES

Brunk, M. ; Emmerich, J.; Mittlmeier; T

1. Dept. of Trauma and Reconstructive Surgery, University of Rostock, Germany

INTRODUCTION:

Hip fractures are regarded as the most common severe type of fall-related injury concerning elderly people and the most serious of the osteoporotic fractures because of their high morbidity, mortality and impairment in quality of life. (2-5).
Pertrochanteric and femoral neck fractures of the femur, with respect to healing, may be regarded as fracture-types with long rehabilitation time. (1)
Current rehabilitation protocols of patients following operative treatment recommend full weight-bearing on the operated extremity as early as possible. The present study seeks to highlight the reactions to early loading under the aspect of pain-control. In-shoe pressure redistribution to provide relief of postoperative mobilisation is based on assumed links between pressure and pain. However, little is known about pain-associated loading after operative treatment of hip fractures.

MATERIAL AND METHODS:

29 patients, who had an operative treatment of a fracture of the femoral neck or a pertrochanteric fracture, were allowed to full weight-bearing as tolerated on the injured limb. 12 man and 17 women, ranging in age from 45 to 96 years, took part in this study at our center. During postoperative mobilization elderly patients were allowed to practise full weight-bearing while loading of the injured limb was mostly limited due to pain in the early postoperative period.
Gait analysis was performed by using the novel PEDAR in-shoe plantar pressure measurement system (novel GmbH, Munich, Germany). Computerized gait-testing was performed at one, seven and twelve days postoperatively to quantify weight-bearing in association with pain.
Visual analog scale (VAS) score of pain were obtained from all subjects before and after testing with the PEDAR-system. Statistical analysis was used to analyze the relationship between the plantar pressure parameters and VAS scores in the period of hospitalisation. Pearson's correlation was applied to analyze the correlation between the changes in plantar pressure parameters and VAS scores. Statistical significance was set as $p < 0.05$.

RESULTS:

The average amount of weight these patients placed on the injured limb increased during the time of hospitalisation. During mobilisation maximum peak pressure (MPP), maximum pressure-time integral (PTI) increased at the seventh and twelfth day after surgery. In analogy, subjective pain scores decreased significantly during hospitalisation. 6 of 29 patients were measured at a fixed follow-up. For these 6 patients the increase in the PTI and MPP values were statistically correlated with the improvement in VAS scores ($r = 0.8$) in the course of follow up.
The average load supported by the injured limb was 70.4 (43.8-83.1) % of the uninjured limb after three days, and gradually increased to 91.1 (75.4-96.2) % at twelve days.

DISCUSSION:

This study investigated plantar pressure related to pain during gait in subjects at different time points after operative treatment of hip fractures. Plantar pressure came up to 40 to 90 % of the uninjured limb. Furthermore, a positive correlation of gait parameters with the improvement in VAS exists. Correspondingly, gait and balance disturbances caused by hip pain have an influence on dynamic plantar pressure distribution. Our subjects presented a worse load distribution pattern during gait at the beginning of hospitalisation overloading the midfoot and the rearfoot. These changes could be associated with pain pathogenesis that worsens the biomechanical condition of these patients and their clinical consequences

CONCLUSION:

Following operative treatment we recommend pain-associated full weight-bearing on the operated extremity as early as possible to avoid specific consequences on postural balance control; the reactions to early loading under the aspect of pain-control may provide a faster mobility after operative treatment.

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THE MĀORI FOOT: STATIC MORPHOLOGY AND DYNAMIC FUNCTION IN HEALTHY AND DIABETIC POPULATIONS

Jason Gurney (1), Celia Kuch (1), Dieter Rosenbaum (2), Uwe Kersting (3)

1. Department of Sport and Exercise Science, University of Auckland, New Zealand
2. Motion Analysis Lab, University Hospital Muenster, Germany
3. Center for Sensory-Motor Interaction, Aalborg University, Denmark

INTRODUCTION

Foot morphology and plantar loading may differ between ethnicities (Veves, 1995). Our previous work suggested altered loading strategies may exist between elite athletes of Māori and New Zealand Caucasian ('NZC') ethnicity (Gurney, 2009).

Māori suffer disproportionately from diabetes and particularly its complications, including foot ulceration (Joshy, 2006). If Māori exhibit altered foot morphology and function, perhaps different preventative strategies are required for reducing diabetic foot complications in Māori, such as footwear modifications.

Therefore the purpose of this study was to investigate the static morphology and dynamic function of the Māori foot in both healthy and diabetic populations compared to NZC controls. It was hypothesised that based on previous findings significant differences would be observed between ethnicities.

METHODS

A total of 40 participants were further divided into 10 Māori (8f; 42±11yrs; BMI 27±5) and 10 NZC (8f; 43±11yrs; BMI 26±4) who had no neuromusculoskeletal injury, plus 10 Māori (5f; 58±11yrs; BMI 32±5) and 10 NZC (5f; 60±10yrs; BMI 30±3) diagnosed with Type-2 diabetes.

Plantar pressure was evaluated using a Novel EMED-AT system (Novel GmBh, Munich, Germany) at a frequency of 50Hz during walking at a self-selected speed. Following familiarisation five trials were collected for each foot (Hughes et al., 1991).

Static morphology was measured using Harris mat techniques. Foot length, heel width, forefoot width, and arch index were determined. The latter was calculated as the width of the narrowest part of the midfoot divided by toeless foot length.

Plantar pressure data were analysed using Novel software, in which the foot was divided into 10 regions. Multiple pressure parameters were analysed.

Kolmogorov-Smirnov tests confirmed data normality. Independent T-tests were then used to test for significant differences between ethnic groups in both the healthy and diabetic participants (p = <0.05).

RESULTS

Arch index was found to be significantly greater in the healthy Māori compared to the healthy NZC, but this difference was not observed when comparing diabetic groups. No other static morphological differences were found between ethnicities.

Few significant differences were found between ethnicities, healthy or diabetic, in terms of dynamic plantar loading. Peak pressures under the central forefoot were found to be significantly greater in diabetic Māori compared to diabetic NZC (Fig. 1).

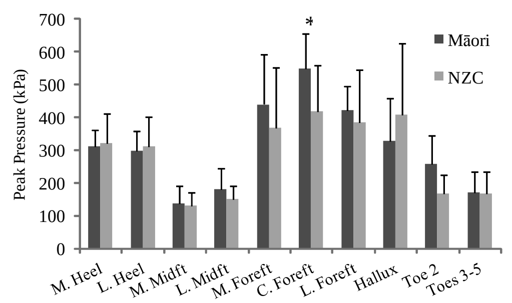

Figure 1: Peak pressure (kPa) data from Type-2 diabetic Māori and NZ Caucasians (=p<0.05).*

DISCUSSION AND CONCLUSIONS

The hypothesis that significant morphological and functional differences would be found between Māori and NZC feet was mostly proven incorrect. Since the compared groups were largely homogenous, we can conclude that Māori ethnicity alone may not affect foot morphology and function in healthy and diabetic populations. This may suggest that no special considerations are required when clinically treating or designing preventative care for diabetic Māori at risk of plantar ulceration. However further research is required on larger groups and non-urbanised Māori.

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ASYMMETRY IN PLANTAR LOADING DURING GAIT IN NATIVE AMERICANS WITH AND WITHOUT DIABETES AND WITH AND WITHOUT NEUROPATHY

Thomas W. Kernozek, Cassandra Heizler, John F. Greany

Health Professions Department, University of Wisconsin – La Crosse, WI USA

INTRODUCTION

The adjusted prevalence of diabetes among Native Americans has increased by 26.9%, from 6.7% to 8.5%, between 1996 and 2006 (Jernigan et al., 2010). On average, Native Americans are 2.3 times as likely to have diabetes as non-Hispanic whites of similar age (Burrows, 2000). Plantar ulceration is a common problem in individuals with diabetes. Among those with diabetes, 24% will require an amputation of the foot and/or leg (Lott et al., 2008).

Peak plantar pressure and pressure-time integral are two variables used to screen individuals with diabetes and neuropathy for the development of foot ulcers (Brown et al., 2004, Sauseng et al., 1999, Stess et al., 1997). The role asymmetry in plantar pressures has not investigated in Native Americans.

Our aim was to investigate plantar loading asymmetry during gait in Native Americans with no diabetes (ND), diabetes (D) and diabetes with neuropathy (D-NP).

METHODS

Ninety-eight Native American's volunteered to participate in the study (mean age 50.3 yrs; 19-86). Twenty three individuals had diabetes (D), 14 had diabetes with peripheral neuropathy (D-PN) and 61 did not have diabetes (ND). Neuropathy status was determined with a biothesiometer (> 25 V threshold). Twenty-four percent of participants were overweight (>25-29.9 BMI) and 59% were obese (>30 BMI).

Plantar pressure data were collected using an EMED-AT floor mounted capacitance based platform sampling at 50 Hz. The 2-step method was used to obtain plantar loading data from five trials of each foot. Each plantar loading trial was subdivided into ten specific plantar regions for analysis. Differences in the absolute value of peak pressure and peak pressure-time integral of feet were examined across metatarsal regions using an Analysis of Variance (ANOVA). Separate ANOVA's were run on peak pressure and pressure time integral for each of the metatarsal regions (alpha = 0.05).

RESULTS

Peak pressure asymmetry was different between feet across each the metatarsal regions 1-3 of the plantar surface during gait ($p < 0.05$). Table 1 depicts the asymmetry in pressure-time integrals between feet across metatarsal regions of the plantar surface ($p < 0.05$). Post hoc comparisons indicated that individuals without diabetes (ND) displayed less asymmetric loading described by peak pressure in metatarsal 1-3 regions and pressure-time integral across all metatarsal regions than those with diabetes (D) and diabetes with peripheral neuropathy (D-PN).

	MTH 1	MTH 2	MTH 3	MTH 4	MTH 5
ND	28.6 ± 31.4	27.1 ± 23.3	20.8 ± 17.4	15.4 ± 13.7	16.1 ± 13.5
D	49.5 ± 46.1	49.4 ± 50.6	33.7 ± 37.5	17.8 ± 11.9	21.2 ± 30.8
D-PN	58.8 ± 37.5	57.0 ± 60.2	41.9 ± 24.7	39.4 ± 40.1	37.3 ± 32.9

*Table 1: Mean (± Standard Deviation) Peak Pressure- time integral (kPA*s) for each group [Non-diabetic (ND), Diabetic (D) and Diabetic with Peripheral Neuropathy (D-PN)] for each Metatarsal Region (MTH1-MTH5).*

CONCLUSIONS

Native Americans with diabetes appeared to show greater asymmetry in plantar loading variables across the forefoot region compared to controls. Individuals with neuropathy and diabetes had the greatest amount of asymmetry with pressure time integral across these regions. Loading asymmetry may pose a role in the development of wounds in Native Americans with peripheral neuropathy and diabetes. Further research is warranted.

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PLANTAR PRESSURE DISTRIBUTION PATTERNS IN MULTIPLE SCLEROSIS PATIENTS WITH DIFFERENT NEUROLOGICAL STATUS

Tatiana L. Tsvetkova (1), Igor D. Stolyarov (2), Olga L. Pichugina (2), Andrey M. Petrov (2), Alexander G. Ilves (2), Lidia N. Prakhova (2), Irina G. Nikiforova (2), Viatcheslav V. Lebedev (3), Fedor V. Yakushev (1), Vadim N. Tartakovskiy (1)

1. novel SPb, St.-Petersburg, Russia
2. Institute of Human Brain RAS, St.-Petersburg, Russia
3. novel, Munich, Germany

INTRODUCTION

Expanded Disability Status Scale (EDSS) is widely used to assess the disability of patients with multiple sclerosis (MS). The EDSS quantifies disability in eight functional systems (FS).

The aim of this study was to obtain the plantar pressure distribution images in MS patients with different neurological status (EDSS) and to estimate the plantar pressure distribution parameters changes.

METHODS

106 patients (33m/73f, age 39±10 years), diagnosed with relapsing-remitting MS (duration of MS 8±6 years) according to McDonald's criteria, were examined. Neurological status of examined patients was characterized with EDSS 3.4±1.2. Patients were divided into five groups in relation to estimated EDSS: EDSS [1,1.5] (no disability), EDSS [2,2.5] (minimal disability), EDSS [3,3.5] (disability in mild to moderate), EDSS [4,4.5] (severe disability), and EDSS [5,6.5] (increasing limitation in ability to walk, walking assistance is needed). All patients received permanent pathogenetic therapy.

Plantar pressure measurements were performed with emed-AT 25 system (novel, Munich, Germany). Five dynamic records of each foot were made with first step protocol. novel database medical was used to collect clinical and pressure measurement data. Peak pressure (PP), mean pressure (MP), maximum force (MF), pressure-time integrals (PTI), force-time integrals (FTI), contact time (CT) and arch index (AI) were calculated in novel-projects with novel automask for hindfoot, midfoot, five metatarsal heads (MH1-MH5), big toe, second toe and lateral toes. Parameters were calculated for each subject and for five groups. The difference in pressure distribution parameters under the foot areas was checked with ANOVA. Significance level was set as $p<0.001$.

RESULTS AND DISCUSSION

Each group had a specific plantar pressure distribution pattern that varied with a degree of disability status. The patients from Gr.2 (vs. Gr.1) were characterized with decreased MP under the hindfoot, increased MF and FTI under MH5 and FTI under MH4; increased CT under the foot and foot regions. Significantly decreased PP, MP, MF, FTI, and PTI under the hindfoot, MH4 and MH5, increased FTI, PTI, and contact time (% roll over process) under MH1, increased MF under the midfoot and AI were found in patients from Gr.3 (vs. Gr.2). Decreased PP, MF and FTI under MH2 were found in Gr.4 (vs. Gr.3). At the same time PP, MP, MF, PTI and FTI were significantly increased under MH4 and MH5 and decreased under MH1; MF and FTI were decreased under the midfoot together with AI and reached the values comparable with parameters from Gr.2. Significant decrease of PP, MP, MF under MH2, MH3, MH4 and midfoot, decreased AI, increased FTI and PTI under MH1 were found in Gr.5 (vs. Gr.4) in concert with significant increase of CT under the foot and all foot regions.

Medial shift of loading is comprehensible because the first ray is the most appropriate structure used in weight bearing to provide an additional security. Reduced parameters under the second and third metatarsal heads can be explained with the development of the transverse arch in the forefoot with an increase of spasticity. Lower loading of midfoot (with decreased arch index) is supposed to be because of an increase of longitudinal arch height. The hindfoot off-loading could be a sequence of the foot circumduction and inversion during the gait. First signs of significant changes of all parameters occurred in MS patients with moderate disability can be regarded as a result of disability in several function systems.

CONCLUSION

These results confirmed that patients with minimal disability have minimal gait impairments. Patients with mild and moderate disability reveal the changes in decreasing hindfoot and lateral forefoot loading with medial shift of loading. MS patients with severe disability showed decreased parameters in central forefoot. Patients with limited ability to walk have stable pathologically changed plantar pressure distribution pattern characterized with low values of all parameters and significantly decreased loading of hindfoot, midfoot, forefoot with medial shift of loading.

CAN GAIT INITIATION PROCESS BE EVALUATED WITH PRESSURE PLATFORMS?

Paula H. Lobo da Costa (1), Kerstin Bosch (2), Dieter Rosenbaum (2)

1. Department of Physical Education, Federal University of Sao Carlos, Sao Carlos, Brazil.
2. Movement Analysis Laboratory, Orthopaedic Department, University Hospital, Münster, Germany.

INTRODUCTION

The transition from standing posture to cyclic walking is a special challenge to be mastered by young children. It is the result of anticipatory postural adjustments before stepping. Studies on this matter with children have focused on force plate data and electromyographic responses (Stackhouse *et al.*, 2007; Wicart *et al.*, 2006) in clinical contexts. The aim of this study is to verify whether gait initiation can be measured with the EMED system in order to understand the relative contribution of foot structure in this process. Besides, as the medial longitudinal arch of young children is in progress, the role of the development of foot functions on this process could also be assessed.

METHODOLOGY

Antero-posterior and medio-lateral displacements of the center of pressure at the foot/floor interface and the Chippaux-Smirak Index (CSI) were measured during gait initiation in 20 Brazilian school children aged 3.7 (±0.7) years and with body mass of 17.3 (±2.3) kg. The dynamic data were obtained with the EMED-ST System (NOVEL, Germany), sampled at 50Hz and foot indexes were measured with foot prints in bipedal stance. Three to five valid trials of gait initiation were collected. The children stood still over the plate and after a "go" signal they performed the gait initiation and continued to walk for about 6 m. The walking speed was freely selected by each child and all were barefoot.

RESULTS & DISCUSSION

The CSI classified of 80% of the children as flat-footed (CSI ranged from 49% to 69% for right and left feet). The average displacements of the center of pressure in antero-posterior and medio-lateral directions relative to the plate system were analysed. The results of one child are presented (Fig.1 & Tab.1). At 70% of the stance phase the anticipatory behavior to the step begins. The range of COP_ML motion toward the stance foot is 9.2 (±1.44) cm, which represents 1.3 times the forefoot width, whereas the COP_AP range of motion is 12.6 (±0.,7) cm, equivalent to 0.80 times the foot length of this child. The maximum COP_ML and the minimum COP_AP motions represent the displacements toward the stepping foot and the backward direction respectively. It is interesting to note that these very small motions in the directions opposite to the step are insufficient to identify anticipatory behavior.

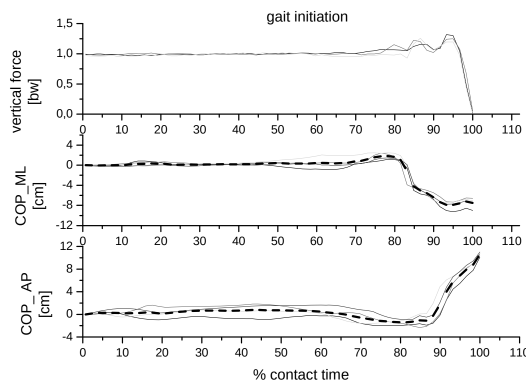

Fig. 1: Anterior-posterior (AP) and medio-lateral (ML) COP displacements and vertical force (body weight-BW) during gait initiation (one child, BR19, four trials).

GI	COP- AP [cm]			COP- ML [cm]		
	Min	max	range	min	max	range
Mean	-1.90	10.67	12.57	-7.70	1.50	9.2
SD	±0.62	±0.41	±0.72	±0.94	±1.01	±1,44

Tab. 1: Minimum, maximum and range of COP AP and ML displacements (one child, BR19, four trials).

Ledebt *et al.* (1998), on the other hand, found significant COP displacements for children of the same age. Since both feet were in contact with the platform, the system could not quantify the loading shift from one foot to the other as gait starts, but it was possible to measure the dynamic behavior of the COP before stepping, which may be of importance to follow individuals with gait and balance disorders.

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THE EFFECTS OF FOOTWEAR, LEARNING, AND FATIGUE ON CENTER OF PRESSURE EXCURSION DURING SINGLE LIMB BALANCE

Antonia M. Zaferiou, BE, Rebecca A. Zifchock, PhD, Allison M. Brown, PT, MA, Jocelyn C. Frey, BS, Howard J. Hillstrom, PhD
Hospital for Special Surgery, Leon Root, MD, Motion Analysis Laboratory, New York, NY

INTRODUCTION

Stability during single limb balance activities is important for functional tasks ranging from basketball lay-ups to putting on pants. During rehabilitation, single limb balance on an uneven surface is a common evaluation and neuromuscular treatment technique. Previous work has studied the relationship between postural sway during single limb posture and vision, varied support surfaces, and different athletic training backgrounds. However, little is known about the affect of footwear conditions on single limb stance stability.

Therefore, the purposes of this study were to investigate (1) how different footwear conditions affect stability, (2) how task learning and fatigue affects stability (by comparing the order of testing), and (3) if there is a coupling between AP and ML sway. No differences in stability were expected between footwear conditions. Anticipating a strong learning effect and minimal fatigue, stability was expected to increase with successive trials. Finally, it was predicted that the ratio of AP sway to ML sway will increase as an effect of learning.

METHODS

Five healthy male subjects (22±2.5yrs, 1.79±0.08m, 74.7±3.1kg) who participated in cutting sports were included in this study. Subjects balanced on their right legs on a 3" thick Airex® foam surface of a BioSway™ (Biodex Medical Systems, NY) for 30 seconds while center of pressure (COP) coordinates were recorded at 20Hz. Despite the challenge the compliant foam surface presented, participants maintained a posture with approximately 20° arm abduction and 45° left knee flexion. This study was part of a larger investigation comparing the following footwear conditions: barefoot (BARE), standard basketball sneaker (Nike Air Max Go™) (STD), the standard basketball sneaker with ankle taping (TAPE), and an experimental sneaker (EXP) designed to reduce ankle inversion. Participants ran and performed cutting maneuvers for about 30 minutes between footwear conditions, allowing for fatigue effects. The TAPE condition was used first for all subjects with the other conditions randomized.

The COP excursion was characterized by analyzing sway indices, defined as the standard deviation of the COP coordinates. COP sway indices were calculated independently in the AP and ML directions (denoted APSI and MLSI respectively) as well as in both directions combined into an overall sway index (OSI). One-way ANOVAs were used to test the effects of footwear condition and task order on OSI, APSI, and MLSI. Two-way ANOVAs were used to test the effects of footwear condition and task order separately as they relate to sway "direction" (APSI vs. MLSI). Due to the preliminary nature of the data, $P \leq 0.10$ was considered significant.

RESULTS AND DISCUSSION

Results are displayed in Table 1. No significant effect was found from footwear condition on OSI or APSI. However, there was a significant difference in MLSI detected between TAPE and BARE that may be caused by more sensory feedback in the BARE condition than in the TAPE condition. There was also a significant effect of sway direction throughout footwear conditions, showing greater sway in the AP direction than in the ML direction. This can be attributed to the greater lever arm of the foot in the AP direction. The comparison of the task order showed a significant effect of task order on OSI, suggesting task learning followed by fatigue. Also, a significant interaction between task order and direction of sway indicated that sway in the AP direction was consistently greater than ML direction sway. Further, the interaction also showed that AP sway may have been less affected by task learning than by fatigue, as compared to the ML sway (Fig 1).

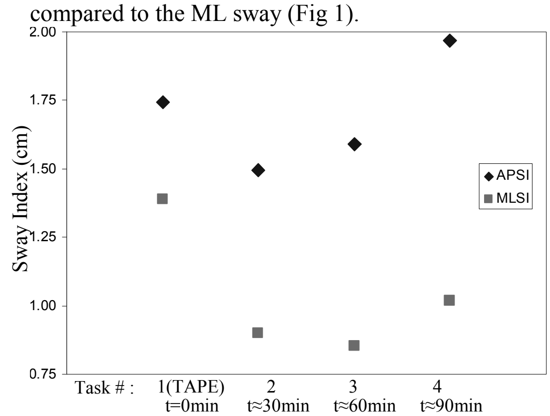

Figure 1: Average MLSI and APSI vs. task order.

While the study is underpowered and footwear condition comparison methods were not ideal, the results may indicate an interesting interplay between task learning and fatigue that overpowers footwear condition effects. This is important to consider during rehabilitation because once a patient has become fatigued, continuing to implement single limb balance tasks into a treatment session may increase injury risk.

Footwear	OSI	MLSI	APSI	Post-hoc relationship with	Task #	OSI	MLSI	APSI	Post-hoc relationship with
BARE	1.95	0.93	1.71	MLSI: TAPE	**1 (TAPE)**	2.25	1.39	1.74	OSI: 3 MLSI: 3
STD	1.96	0.89	1.73		**2**	1.75	0.90	1.50	OSI: 4 APSI: 4
TAPE	2.25	1.39	1.74	MLSI: BARE	**3**	1.81	0.85	1.59	OSI: 1, 4 MLSI: 1
EXP	1.89	0.95	1.62		**4**	2.22	1.02	1.97	OSI: 2, 3 APSI: 2
1 way ANOVA p-values	0.44	**0.05**	0.93			**0.04**	**0.04**	**0.06**	
2 way ANOVA p-values		**0.01**					**+ 0.10**		

Table 1: Average sway indices [cm] and ANOVA and Post-hoc results (+indicates statistical interaction).

RELATIONSHIP BETWEEN FOOT RANGE OF MOVEMENT AND PLANTAR PRESSURE DISTRIBUTION IN DIABETIC NEUROPATHIC PATIENTS

Cristina D. Sartor, Andreja P. Picon, Maria I. Roveri, Roberto C. Dinato, Isabel C. N. Sacco

Physical Therapy, Speech and Occupational Therapy Dept., School of Medicine, University of Sao Paulo, Brazil

INTRODUCTION

It is not widely known if foot static range of movement restriction influences abnormal plantar pressure distribution in patients with diabetic neuropathy, and thus be associated with higher chances of plantar ulcers occurrences. The purpose of this study was to investigate the static active range of motion (ROM) of the ankle (AJ) and the first metatarsophalangeal joints (1st MTF), plantar pressure distribution during gait, and the relationship between these pressure and ROM variables in diabetic neuropathic patients and non-diabetic individuals.

METHODS

Thirty six young adults participated in this study: control group (CG, n=18, 43±9yrs) and diabetic neuropathic group (NG, n=18, 57±4yrs), matched in body mass (p=.10), height (p=.98), but not in age (p<.01). Static active ROM of the 1st MTF and AJ joints were measured in sagittal plane using a manual goniometer and an electrogoniometer, respectively. Plantar pressure (Pedar X system, Novel) was analyzed during barefoot walking with antiskid socks, in 5 areas: rearfoot, midfoot, lateral and medial forefoot and hallux. General linear model for repeated measures ANOVAs were used to compare groups in the studied areas. ROM of both joints were compared between groups using t test. Pearson coefficients were calculated to test correlation between ROM and pressure variables in both groups.

RESULTS

NG presented smaller 1st MTF and AJ ROMs (table 1), smaller contact area at the rearfoot, higher pressure-time integral at the rearfoot, and smaller peak pressure at the hallux (table 2). There was a moderate and significant correlation between pressure-time integral at the hallux and 1st MTF ROM (r=.44, p=.06), and between pressure-time integral at the rearfoot and 1st MTF ROM (r=.48, p=.04) in the NG, but not in CG.

Table 1. Mean 1ST MTF and AJ ROM of CG and NG.

	1ST MTF ROM (º)	AJ ROM (º)
CG	80 ± 16	60 ± 10
NG	67 ± 18	46 ± 9
p	<.01	.029

DISCUSSION

The results demonstrated that the 1st MTP motion restriction play an important role in NG patients by changing the foot rollover mechanism during gait, and consequently the plantar pressure, particularly at the hallux. This may explain the lower pressures at the hallux and the increased pressure-time integral at rearfoot in diabetic subjects. These findings are in accordance with other authors that had also described the same trend and suggested that the neuropathy influences the loading and patterns of walking (Turner *et al*, 2007). Nurse and Nigg (2001) found that peak pressure and pressure-time integral were significantly higher in areas of normal sensitivity and lower at the insensate areas, and the center of pressure (COP) under the foot shifted away from areas of decreased sensitivity when sensory input is reduced from a portion of the foot. Sacco et al (2009a) found reduction of peak pressure at the hallux during gait in neuropathic patients wearing shoes. Giacomozzi and Caselli (2002) found that, in this population, the COP excursion was shorter longitudinally, so the metatarsal heads left the floor earlier in the stance phase. Sacco et al (2009b) studied this altered foot rollover and argued that it was associated with less dynamic mobility and altered plantar pressure distribution. Our results confirm that an altered ROM, especially at the hallux, can produce alterations in pressures, potentially increasing the risk of ulcer formation.

CONCLUSION

In clinical practice, we could use this non-sophisticated measuring tool (1st MTF ROM by manual goniometer) to predict a potential pressure alteration in neuropathic patients during walking. This ROM alteration would indicate a condition more susceptible to injuries and plantar ulcerations.

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Table 2. Mean values of Pressure-time Integrals, Contact Area and Peak Pressure statistically different between CG and NG

	Pressure-Time Integral (kPa.s)– heel	Contact Area rearfoot (cm^2)	Peak Pressure hallux (kPa)
CG	83.6 ± 10.3	30.8 ±2.2	263.6 ±37.2
NG	127.7 ± 102.7	29.2 ±4.9	227.3 ±91.3
p	.04	.01	.02

CAST-BOOT ANKLE-FOOT ORTHOSES YIELD GREATER FOREFOOT LOAD REDUCTION THAN TOTAL CONTACT CASTS

David J. Gutekunst, Kathryn L. Bohnert, Mary K. Hastings, and David R. Sinacore

Applied Kinesiology Laboratory, Washington University in St. Louis, MO USA

INTRODUCTION

Total contact casting (TCC) is a common off-loading strategy in individuals with plantar ulcers secondary to diabetes mellitus (DM) and peripheral neuropathy (PN). Cast-boot ankle-foot orthoses (AFO) may offer a lower-cost alternative to TCC.

The purpose of this analysis was to assess the off-loading capabilities of the cast-boot AFO by comparing the plantar loading patterns in TCC and AFO for subjects with DM, PN, and plantar ulcers. Specifically, we assessed the reduction in plantar loads in the hindfoot, midfoot, and forefoot in each off-loading device compared to unshod walking.

METHODS

Twenty-three subjects with DM, PN, and a plantar ulcer gave informed consent and were randomly assigned to off-loading treatment with either TCC (n=11) or AFO (n=12). Subjects first walked unshod across an EMED pressure platform (Novel Inc., St. Paul, MN). Next, subjects walked across a 6-meter walkway at preferred walking speed with a Novel Pedar pressure insole (Novel Inc., St. Paul, MN) placed in the off-loading device (TCC or AFO) on the ulcerated foot. The EMED trials effectively created an individualized "baseline" value for comparison with each subject's TCC or AFO trial.

For both Pedar and EMED trials, the plantar pressure map was divided into 3 masks reflecting the hindfoot, midfoot, and forefoot. Multiple steps were averaged on each subject's ulcerated side for contact area [cm^2], contact time [ms], maximum force [N], peak pressure [N/cm^2], force-time integral [N*s], and pressure-time integral [$N{*}s/cm^2$]. The effects of TCC and AFO were assessed using analysis of covariance, with walking speed and the EMED value of the outcome variable included as covariates. Mean values for TCC and AFO are reported as estimated marginal means in order to account for variability due to walking speed and EMED values.

RESULTS

No differences in estimated marginal means were found in the hindfoot mask, though there was a trend towards higher force-time integral in the AFO condition compared to TCC (p=0.071). Similarly, no differences were found in the midfoot mask, though there was a trend towards higher peak pressure in the TCC condition compared to AFO (p=0.052).

In the forefoot mask, the AFO showed greater off-loading capability for force and pressure-related outcomes. The TCC condition had significantly higher estimated marginal means than the AFO condition for maximum force (p=0.012) and force-time integral (p=0.010). Similarly, the TCC condition had significantly higher values compared to the AFO condition for peak pressure (p=0.011) and pressure-time integral (p=0.007).

Results are summarized in Table 1 below.

CONCLUSIONS

The cast-boot AFO yielded significantly greater reductions in force and pressure variables in the forefoot region of individuals with DM, PN, and plantar ulcers, suggesting that AFO may provide a viable, low-cost off-loading strategy for forefoot ulcers. Further research is needed to determine whether the load reductions seen in AFO correspond to improved healing rates.

ACKNOWLEDGMENTS

We acknowledge our funding sources: **R01 DK59224, NICHD T32 HD007434, R21 DK079457.**

TABLE 1: Comparison of forefoot force and pressure variables in EMED, TCC, and AFO conditions

	Estimated Marginal Means (Mean ± SE)			
Forefoot outcome variable	**EMED**	**TCC**	**AFO**	**TCC vs. AFO**
Maximum Force [N]	723.3 ± 27.9	182.5 ± 20.2	99.8 ± 19.3	p = 0.012
Force-Time Integral [N*s]	201.0 ± 21.2	66.1 ± 9.4	26.4 ± 9.0	p = 0.010
Peak Pressure [N/cm^2]	85.3 ± 32.9	13.4 ± 6.0	6.6 ± 5.0	p = 0.011
Pressure-Time Integral [$N{*}s/cm^2$]	35.9 ± 4.4	5.4 ± 0.8	2.0 ± 0.8	p = 0.007

PRESERVATION OF THE FIRST ROCKER IS RELATED TO INCREASES IN GAIT SPEED IN INDIVIDUALS WITH HEMIPLEGIA AND AFO

Karen J. Nolan, PhD (1,2), Mathew Yarossi, BS (1)

1. Kessler Foundation Research Center, Human Performance and Movement Analysis Laboratory, West Orange, NJ,USA, www.kesslerfoundation,org.
2. UMDNJ – New Jersey Medical School, Department of Physical Medicine and Rehabilitation, Newark, NJ, USA

BACKGROUND AND PURPOSE

Ankle foot orthotics are often prescribed to individuals with hemiplegia to assist with ambulation.[1] Gait speed has been used as a primary indicator of orthotic effectiveness, and improved functional ambulation.[2] Limited research has been conducted to understand the specific mechanisms leading to improved gait speed after orthotic intervention. Changes in impulse during the first rocker (braking force) and third rocker (propulsion force) may directly affect changes in gait speed after othotic intervention.[3] Therefore the purpose of this investigation was to objectively measure changes in impulse during double support and correlate those finding to changes in gait speed with and without AFO in individuals with hemiplegia.

METHODS

SUBJECTS: Fifteen individuals with hemiplegia (age 51 ± 12 y, height 1.72 ± 12.2 m, mass 85 ± 21 kg) greater than six months currently using an AFO during ambulation.

PROCEDURES: Subjects performed 10 walking trials at a self-selected pace in two conditions, with and without AFO. During all walking trials, foot pressure data was collected using the Pedar®-x Expert System (Novel Electronics Inc., St Paul, MN, USA). The main outcome measures were gait cycle time (sec), mean force (bodyweights), and impulse (mean force x time) in the, wholefoot, hindfoot (heel and arch) and forefoot (metatarsal heads, toes, and hallux), during initial double support (IDS) and terminal double support (TDS).

ANALYSIS: Wholefoot, hindfoot, and forefoot impulse were calculated in custom Matlab programs by integrating the local forces under specific anatomical regions during IDS and TDS. Independent sample t-tests were used to test for significant differences with and without AFO ($p \leq 0.05$).

RESULTS

Gait cycle timing significantly decreased with the AFO during the entire gait cycle (p=0.034), initial double support (IDS; p=0.034), single support (SS; p=.030), and terminal double support (TDS; p=0.048). During IDS, impulse on the affected limb significantly decreased in the wholefoot (with 0.052±0.016, without 0.086±0.57; p=0.016) and hindfoot (with 0.037±0.012, without 0.060±0.032; p=0.006), and remained relatively unchanged in the forefoot (with 0.016±0.007, without 0.025±0.027; p=0.14). Mean force during IDS on the affected limb also significantly decreased in the wholefoot (p=0.0029) and hindfoot (p=0.0069). During IDS, hindfoot impulse % change and velocity % change during a two-minute walk were significantly correlated (R^2=0.44, p=0.007, figure 1). During TDS, impulse on the affected limb was not significantly different in the wholefoot, hindfoot or forefoot (with 0.062±0.020, without 0.069±0.026; p=0.37). Mean force during TDS on the affected limb significantly increased in the wholefoot (p=0.0002) and remained relatively unchanged in the forefoot (with 0.30±0.11, without 0.27±0.10; p=0.28).

DISCUSSION AND CONCLUSION

Researchers have shown orthotics increase gait speed; this research suggests that the increase in speed is not due to increased propulsive forces at the end of TDS, but due to decreased breaking forces during IDS. The AFO provides increased dorsiflexion at footstrike creating a decreased impulse (breaking force) in the hindfoot thereby preserving the first ankle rocker and providing a more efficient weight acceptance and positively affected gait speed. Future research is required to fully understand the mechanisms underlying the increases in gait speed associated with orthotic intervention in adults with hemiplegia.

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ACKNOWLEDGEMENTS

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EFFECT OF CAD DESIGNED PEDORTHOSIS WITH BUILD-IN WEDGE FOR CHILDREN WITH CLUBFOOT

X.C. Liu (1), R. Rizza (2), J. Thometz (1), R. Lyon (1), C. Tassone (1), S. Tarima (3)

1. Dept. of Orthopedic Surgery, Children's Hospital of WI, Medical College of WI, WI
2. Dept. of Mechanical Engineering, Milwaukee School of Engineering, WI
3. Division of Biostatistics, Medical College of WI, WI

INTRODUCTION:

A customized dynamic pedorthosis has been developed to prevent recurrence of the treated clubfoot. This new pedorthosis was developed using our OrthoticPro™, a software package that utilizes dynamic plantar pressure data, and computer aided engineering tools. The pedorthosis is constructed using rapid prototyping technologies (RP). The purpose of this research was to: 1) compare the pressure metrics between barefoot in regular shoes and the pedorthosis, 2) quantify the deviation of the center of pressure (COP) trajectory from the normal trajectory with and without the use of the pedorthosis, and 3) develop a FEA model to predict this effectiveness of wedges.

METHODS:

Five typically developing children (average age 7.2 years, 2 girls and 3 boys) and five clubfoot patients with (average age 6 years 1 girl and 4 boys) were recruited. The finite element model was generated using CT based geometry and plantar pressure (Emed, Novel Inc., MN). The pedorthosis was constructed from the CAD model and manufactured using the RP (Stereolitography). Eight-pedorthtoses were fitted to the children with clubfoot and the children were measured during walking with and without orthotics using the Pedar insole pressure system (Novel Inc., MN).

RESULTS:

There was significant reduction of the average COP deviation following the use of the pedorthosis. The maximum reduction of the average COP deviation occurred in the forefoot (7.87%) and then the midfoot (4.00%). There are no significant differences of any pressure measurements at the midfoot, medial forefoot, and entire toes. Significant reduction of maximal force, peak pressure, and loading at the heel and the lateral forefoot are identified following the use of the new pedorthosis ($P<0.05$) (Figure 1).

DISCUSSION:

This kinetic change may imply a reduced supination of the forefoot in children with residual clubfoot. Our short-term follow-up demonstrates that the pedorthosis improves the dynamic misalignments in the residual clubfoot. A customized wedge as predicted by FEA indicates a corrective magnitude of the wedge angle which varies along the forefoot and midfoot regions.

FIGURES AND TABLES

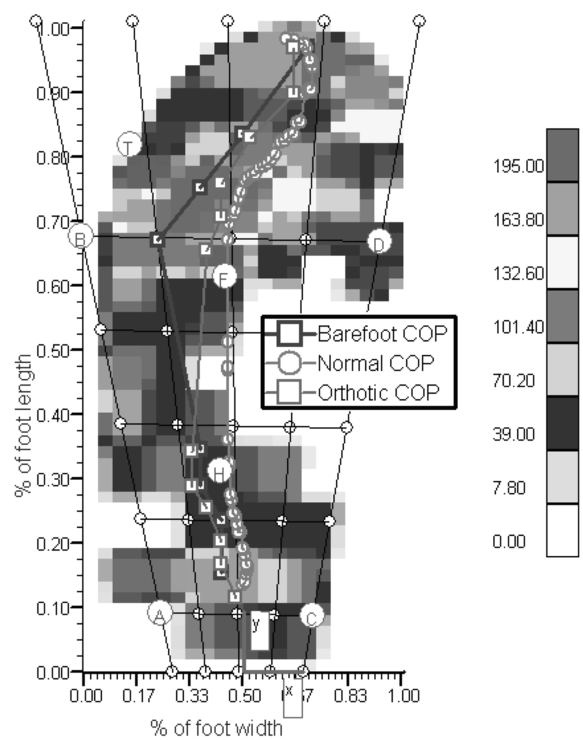


Figure 1: Average deviation of the COP trajectory from the normal trajectory in the hindfoot, midfoot and forefoot regions following the use of the new pedorthosis.

T22

INTRAARTICULAR AND MUSCLE FORCE REACTIONS OF THE LEG USING DIFFERENT INSOLES

Wyss, Ch.[1,2], Roethlisberger, M[3], Gerber, H[3]

1. Laboratory of Movement analysis, Childrens University Hospital Basel, Switzerland
2. Laboratory of Movement analysis, Cantonal Hospital Aarau, Switzerland
3. Institute for Biomechanics, ETH Zurich, Switzerland

Introduction

In the literature the effect of insoles is discussed controversial [1, 2, 3]. The aim of our study was firstly to find a correlation between kinematic, kinetic and electromyographic changes during gait caused by the application of different insoles. Secondly we were interested in the change of the intraarticular force in the knee and the change of muscle forces in the leg.

Methods

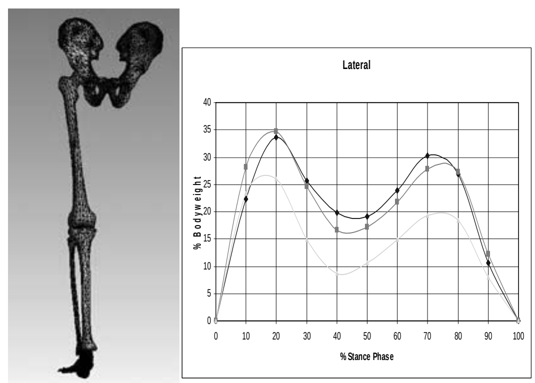
We examined 10 healthy subjects using 5 different insoles alternatively with a neutral insole. The kinematic measurements were performed with a Vicon MX System (6 Cameras). Additionally we used an AMTI force platform and a Novel SF pressure measurement System, even though a Pedar (Novel) insole pressure measurement system. The EMG was measured with a MegaWin system, synchronized with the Vicon system. The intraarticular reaction forces in the medial and the lateral compartment oft the knee were calculated with the help of a finite element model (ANSYS) of the leg which is considered as a rough estimate based on CT Scans. The muscle forces were determined using a slightly modified model from the repository AMMRV1.1 of ANYBODY TECHNOLOGY (Vaughan).

Results

Opposite to the literature we found that there is, for some kinematic parameters, a more or less unique change due to different insoles. The insoles with a medial wedge, for example, showed a significant decrease of eversion of the hindfoot and the insoles with a lateral wedge showed a significant increase of eversion. A more interesting result was the calculation of the intraarticular reaction force in the medial and lateral compartment of our knee model. In figure 1 the pink curve shows the reaction force in the lateral knee compartment in % Bodyweight with an insole with a medial wedge. Nearly the same curve (blue line) shows the measurement with a neutral insole. A significant decreased reaction force could only be observed when we simulated a stiff ankle joint that means inversion and eversion of the ankle joint was excluded in our simulation.

Using our muscle model we found no significant change of muscle forces in the whole leg. Opposite to that we found a significant increase of all thigh extensors comparing barefoot gait with gait in shoes with neutral insoles.

Fig. 1: Reaction force in the lateral knee compartment



DISCUSSSION & CONCLUSIONS

The findings concerning the kinematic parameters of the ankle joint confirm more or less the clinical assumptions. But the clinical conclusion, that the changing of these parameters influence the reaction forces in the knee, could not be confirmed without reservations. The results of our simulations shows clearly that the effect of an insole will be completely compensated by the ankle joint. There is no effect on muscle forces of the leg or joint reaction forces in the knee if the ankle is enough mobile.

Because the muscle forces are significantly increased, when gait analysis is performed with shoes, in the future it will be necessary to do the clinical gait analysis not only barefoot as hitherto.

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T23

RATIOS OF LATERAL TO MEDIAL PATELLOFEMORAL FORCES AND PRESSURES IN A SIMULATED OPERATIVE ENVIRONMENT OF TOTAL KNEE ARTHROPLASTY

Joseph J. King (1), Douglas L. Cerynik (1), Jeffrey I. Peretz (1), Aaron Black (2), Norman A. Johanson (1)

1. Department of Orthopaedic Surgery, Drexel University College of Medicine, Philadelphia, PA
2. Drexel University College of Medicine, Philadelphia, PA

INTRODUCTION

The differences of lateral and medial patellofemoral forces in different total knee arthroplasty (TKA) designs have not been extensively studied. The purpose of this study was to measure the distribution of patellofemoral forces in native and TKA cadaver knees utilizing sensor technology performed in a simulated operative environment. Currently the criteria for evaluating the success of soft tissue balancing in TKA involves only the surgeon's observation of intraoperative patellar tracking during a passive range of motion, but no good objective measure of patellar tracking has been described. We propose that the ratio of lateral to medial maximum force and peak pressure may be used as a surrogate marker for patellar tracking.

METHODS

The patellofemoral forces of 6 knees from 3 fresh-frozen half-body female cadavers (mean age 82 years) were evaluated with a capacitive sensor placed on the patellofemoral articulation in staged clinical scenarios. The half-body cadavers were placed supine with the pelvis stabilized such that all study aspects were similar to an actual operative procedure. The medial parapatellar approach was used and the patella was everted in the standard fashion. The sensor device, capacitive novel pliance system (Novel Electronics, St. Paul, MN), was sutured over the patellar surface and the joint capsule closed using 3 sutures.

Staged clinical scenarios tested were native knees (NKNP, 6 knees), total knee replacement without patellar resurfacing (RKNP, 6 knees), resurfaced knee and patella (RKRP, 6 knees), resurfaced knee and patella with lateral release (RKRP-LR, 3 knees), gender-specific knee with patella resurfacing (GKRP, 3 knees), and gender-specific knee with lateral release (GKRP-LR, 3 knees). Maximum force (N) and peak pressure (kPa) were simultaneously recorded over a set of 3-4 ranges of motion. Average values were compared for the medial and lateral patella compartments and for different clinical settings.

Component alignment, stability, and patellar tracking were clinically acceptable by direct visualization in all scenarios.

A student t-test with a pooled sample variance was used in comparisons, and the assumption of equal sample variances between lateral and medial sides was assessed using a F-test. P values of <0.05 were considered significant.

RESULTS

Significant differences in lateral and medial force and pressure differentials were seen in most scenarios despite clinically normal patellar tracking (Figures 1 and 2).

For the native knee (NKNP), a statistically significant difference was seen between the lateral and medial patellofemoral maximum force means ($p=0.04$) at a ratio between the lateral and medial sides of 1.63:1 with a trend ($p=0.11$) in the ratio of lateral to medial peak pressure (1.80:1). For RKNP scenario, an increase in the ratio of lateral to medial maximum force and peak pressure means was seen to 2.86:1 and 1.99:1 which was significant ($p<0.01$ and $p=0.04$).

For the resurfaced knee and resurfaced patella scenario (RKRP), statistically significant differences were seen in lateral to medial maximum force means and peak pressure means ($p<0.01$ and $p<0.01$) at the ratios of 2.75:1 and 2.57:1. The addition of a lateral release in this scenario (RKRP-LR) reduced the lateral to medial ratios of maximum force and peak pressure means to insignificant ratios (1.46:1 and 1.11:1, respectively).

For the gender-specific knee (GKRP), a statistically significant difference was seen in lateral to medial maximum force means ($p<0.01$) at the ratio of 1.96:1, but not in the lateral to medial peak pressure means (1.33:1). Addition of a lateral release to the gender-specific knees did not significantly alter the ratios of lateral to medial differentials.

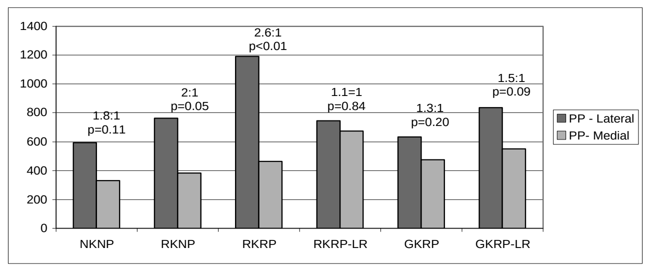

Figure 1: Peak pressure (PP) in the lateral and medial compartments including the ratio between the compartments

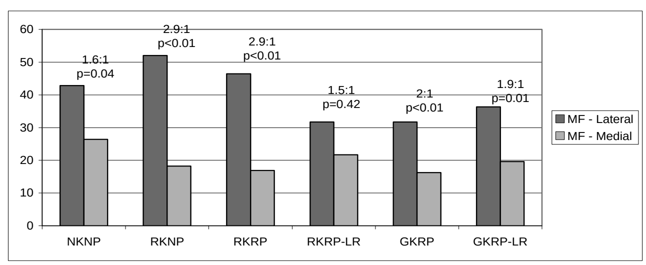

Figure 2: Maximum force (MF) in the lateral and medial compartments including the ratio between the compartments

DISCUSSION

Ratios of peak pressure and maximum force were increased in the lateral patellofemoral compartment on resurfaced knees which were not seen in gender knees. RKRP-LR and GKRP demonstrated lateral to medial patellofemoral force and pressure ratios most similar to the native patella (NKNP).

The addition of a lateral release in conventional total knee arthroplasty can equalize the medial-lateral force distributions.

The decreasing lateral to medial maximum force and peak pressure ratios in the lateral release group (RKRP-LR) after TKA and with gender-specific knees (GKRP) compared to the standard TKA (RKRP) demonstrates that these ratios are surrogate markers for patella tracking due to the known effects of lateral retinacular release on patella tracking and the design of gender-specific knees to improve tracking.

Intraoperative quantification of force and pressure differentials would enable the surgeon to assess proper patellar tracking, giving guidance for using lateral retinacular releases.

T24

CALCULATION OF PRESSURE TIME INTEGRAL, A DIFFERENT APPROACH.

Tom. Melai (1,2), Herman. IJzerman (2), Ton. LH. de Lange (1), Paul. B. Willems (2), Kenneth. Meijer (2), Nicolaas. C. Schaper (3), Hans. HCM. Savelberg (2)

1. Chair of Health Care & Technology for Quality of Life, Fontys University, The Netherlands
2. NUTRIM, Maastricht University Medical Centre, The Netherlands
3. CARIM, Maastricht University Medical Centre, The Netherlands.

INTRODUCTION

High plantar peak pressure values (PP) in people with diabetic polyneuropathy are correlated with ulceration (Boulton, 1983). An additional parameter that is commonly used to asses loading of the foot is the Pressure Time Integral (PTI) (Sauseng, 1999). However, a recent study (Waaijman, 2009) found high correlations between PP and PTI; and suggested that PTI would be of limited additional value.

Novel software calculates PTI as the product of contact time with PP in one mask (PTI_novel) instead of using pressure per sample. This is expected to give an overestimation of the true PTI. The Force Time Integral (FTI) is calculated using individual sensor pressures per sample. Therefore, a more accurate approach to obtain PTI per mask is dividing FTI by contact area (PTI_F).

We sought to test the differences between these two calculation methods in different populations with and without diabetes and diabetic polyneuropathy.

METHODS

Subjects were divided in a group of healthy elderly (HE), a group with diabetes type 2 (DM), and a group with diabetes and polyneuropathy (DPN). Subjects were asked to walk barefooted over a pressure platform at standardized walking speed (1.2±0.1m/s).

	N	Age	Length	Weight
HE	19	72.6±1.9	1.73±0.017	68.1±1.2
DM	33	76.5±3.4	1.69±0.017	74.7±2.7
DPN	76	78.0±2.0	1.74±0.009	82.4±2.7

Table 1: Subject Characteristics. (mean±std. Error).

Data was masked using the Novel 10 mask division (1=heel, 2=mid foot, 3-7=metatarsal region, 8= hallux, 9-10= toes).

RESULTS

Overall PTI values were lower for PTI_F compared to PTI_novel (PTI_novel= 9.1±5.4 Ns/cm^2, PTI_F= 2.6±1.1 Ns/cm^2). PP and PTI_novel showed a similarity of pattern, but PTI_F pattern differed from PTI_novel, especially in the region of the hallux. Figure 1 shows these differences for the DPN group. To operationalize the difference between both PTI calculations, PTI of the hallux was divided by PTI of the heel. These ratios differed significantly (p=0.000) between both calculations in each group (PTI_novel - PTI_F: HE 1.17 - 0.81; DM 1.34 - 0.79; DPN 1.19 - 0.80).

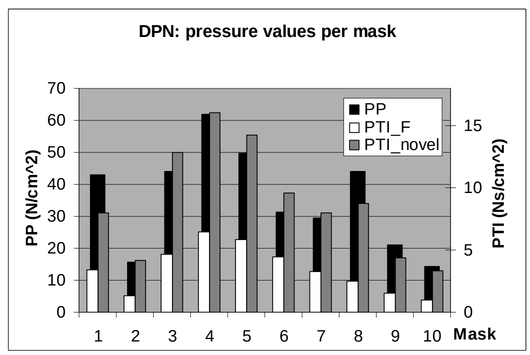

Figure 1: Pressure variables per mask in DPN group: PP (on the left axis), PTI_novel and PTI_F (on the right axis)

DISCUSSION

These results show that there is a difference in PTI calculation based on PP and FTI, especially in the hallux. Because the PTI_novel is based on PP, the additional value as described by Waaijman et al. (2009) may be limited.

In addition, when comparing groups, there is a similarity of differences between PP and PTI_novel (not shown here). However, the differences between PP and PTI_F show an inconsistency when comparing the different groups per mask.

Based on these results it is concluded that the more accurate calculation, as suggested here, may contribute to better understand foot sole loading.

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FOOTPRINT MASKS: REPRODUCIBILITY OF THE OXFORD ANATOMICALLY-BASED SELECTION BY MEANS OF NOVEL GEOMETRICAL MASKS.

Claudia. Giacomozzi (1), Julie. Stebbins (2,3), Axel. Kalpen (4)

1. Dept of Technology and Health, Istituto Superiore di Sanità, Rome, Italy
2. Oxford Gait Laboratory, Nuffield Orthopaedic Center, Oxford, UK
3. NDORMS, Oxford University, Oxford, UK
4. novel biomechanics lab, novel gmbh, Munich, Germany

INTRODUCTION

In the field of foot biomechanics, the clinical relevance of the integration of pressure, kinematics and force measurements has already been demonstrated (Giacomozzi, 2000). Previous work conducted by the authors in this field has been focused on two key areas: First, to produce a reliable combination of pressure instrumentation, a kinematic foot model (in this case the Oxford Foot Model (OFM) (Stebbins, 2006), procedures and software to provide automated masks for accurate pathologic footprint selection. Second to identify the most appropriate geometrical footprint selections which, in presence of regular footprints, compares reliably to a selection based on markers from the OFM. Since kinematic measurements are not always available, it is useful to know which geometric mask should be used to allow comparable results. The present study deals with this latter aim.

MATERIALS AND METHODS

Markers from the OFM were used to define the following 5 anatomically-based foot areas (Stebbins, 2006): 1) medial hindfoot (MH); 2) lateral hindfoot (LH); 3) midfoot (MI); 4) medial forefoot, toes included (MF); 5) lateral forefoot, toes included (LF). novel gmbh purposely modified the multimask software to integrate kinematic data from a VICON system acquired simultaneously with an emed pressure platform. As a result, the above areas were identified by vertically projecting the position of the relevant anatomical markers at midstance. Two automated masks (available in software) were found to be the most similar to the OFM based mask: i) BIS mask, based on the longitudinal bisecting line of the foot and on the two lines perpendicular to it placed at default distances from the bottom of the footprint (27% and 55% respectively); ii) HT2 mask: similar to the BIS mask, but based on the longitudinal line going from the middle of the hindfoot to 2nd toe (HT2 line). Since the OFM-based selection uses two different longitudinal axes for the hindfoot and the forefoot, two further geometrical masks were created based on a mixed use of the previous two masks: i) MIX: same hindfoot selection as BIS, same forefoot as HT2; ii) MIX1: same hindfoot as HT2, same forefoot as BIS. Thus, 5 masks were applied in all to 200 footprints acquired from a young healthy population of 19 girls and boys (38 feet). Up to now, 25 different, randomly selected footprints have been processed. %RMSEs were calculated along the whole loading process for vertical force (F), contact area (A), peak pressure (PP) and mean pressure (MP).

RESULTS

Very good matching was found in all the 4 geometric masks with the OFM-based mask – RMSE<5% for F, A and PP; RMSE≈6% for MP. BIS and MIX1 showed better results than HT2 and MIX, MIX1 giving the best results. MF was the most sensitive area as for F; MI was the most sensitive as for A, PP and MP. Table 1 reports the %RMSEs for MIX1 mask. Fig.1 shows the 5 masks on a regular footprint and the corresponding %RMSE averaged over F, A, PP and MP.

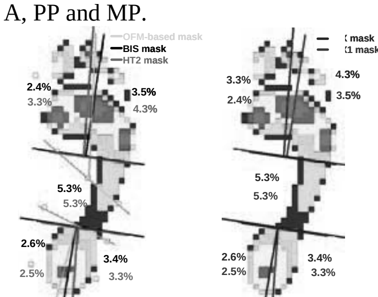

Figure 1: Masks on a regular footprint (left:OFM-based, BIS and HT2; right: MIX and MIX1). %RMSEs averaged over F, A, PP and MP.

Table 1. %RMSE (sd) of MIX1 with respect to OFM-based mask.

	F	A	PP	MP
MH	3.7(3.0)	2.0(1.5)	0.5(0.7)	3.8(2.7)
LH	3.6(3.4)	2.1(1.7)	2.8(3.9)	4.7(3.4)
MI	2.5(1.4)	4.1(1.8)	5.5(3.0)	9.0(4.6)
MF	3.5(2.5)	2.8(2.3)	1.2(2.7)	1.9(1.5)
LF	3.8(2.4)	3.7(2.2)	2.9(3.8)	3.8(2.1)
Total foot	3.9(2.0)	3.4(1.2)	3.8(2.1)	5.7(1.9)

DISCUSSION AND CONCLUSIONS

For regular footprints, the MIX1 geometric mask produced highly comparable results compared to the OFM-based anatomical mask, with %RMSE only slightly higher than intra-subject variability.

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A TECHNIQUE TO ASSESS PLANTAR SOFT TISSUE STIFFNESS DURING THE STANCE PHASE OF GAIT

Carolyn Roposh (1), Carole Tucker (2), Margo Orlin (1), Rami Seliktar (1)

1. Drexel University, Philadelphia, Pa
2. Temple University, Philadelphia, Pa

INTRODUCTION

The bones of the foot are largely supported by the soft tissue on the plantar surface, which acts as the interface with the ground (Tachdjian, 1985). Soft tissue stiffness could affect pressure distribution, shock absorption, or stability during gait. Assessment of soft tissue is important for foot structures that may be prone to increased stiffness such as in high arch structures that are commonly more rigid and less adaptable to the ground for impact cushioning (Van Boerum, 2003). Even though plantar stiffness has been evaluated using a variety of methods, there are no standardized methods to measure plantar stiffness that take into account the real-time, dynamic, weight bearing condition of stance phase of gait (Cavanaugh, 1999 & Rome, 2000). Therefore, we hypothesize that this technique will be able to describe the changes in plantar stiffness during stance based on differences in arch structures.

METHODS

Reflective markers were placed on the feet of 13 subjects (age range 18-21) with no foot pathologies. Trajectory data were collected with Vicon motion capture cameras during sitting trials and while walking over a Novel EMED-ST/E pedobarograph (Munich, Germany). To model the stiffness of the soft tissue, force and deformation had to be isolated. The deformation was the difference in distance between the markers on the foot and the ground during sitting unloaded trials and during walking. Deformation data were limited to the time each segment were in contact with the ground and decimated due differences in collection frequencies. Force data for each segment were matched to the corresponding deformation data, and all data were inputted into the equation for a simple spring to model stiffness.

RESULTS

When compared to the low arch subjects, the mean stiffness values for the high arched subjects were greatest in the heel and lowest in the hallux (Table 1). The real time stiffness plots show the changing stiffness during the weight bearing condition of stance phase (Figure 1).

FIGURES AND TABLES

	Low (N=4)	**Typical (N=3)**	**High (N=6)**
Heel	9.47 ± 3.6	16.19 ± 8.0	20.41 ± 6.3
Midfoot	2.01 ± 0.8	1.08 ± 0.3	0.90 ± 0.6
1st Met	1.51 ± 0.7	2.17 ± 1.8	5.78 ± 1.2
2nd Met	1.48 ± 0.6	2.50 ± 1.0	2.87 ± 0.9
5th Met	0.42 ± 0.1	2.35 ± 1.1	4.21 ± 1.0
Hallux	2.05 ± 0.9	3.69 ± 1.1	2.21 ± 0.9

Table 1: Mean Stiffness(N/mm) ± Standard Deviations

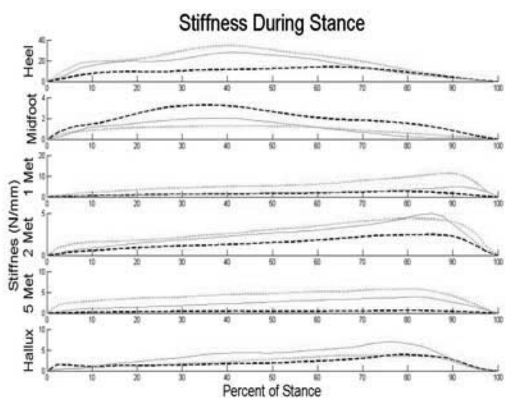

Figure 1: Changing stiffness during stance for high (dot), low (dashed), and typical (solid) arch structures

CONCLUSION

This work demonstrates the feasibility and utility of combining motion capture data with pedobarograph data to characterize plantar stiffness during stance for comparison between arch structures. Variations in plantar stiffness during loading and unloading could impact foot function and alter gait patterns, and therefore should be examined more thoroughly.

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RELIABILITY OF DYNAMIC FOOT GEOMETRY ASSESSMENT USING A PEDOBAROGRAPHIC PLATFORM AND A TWO-STEP APPROACH

Karen A. Keenan, Jonathan S. Akins, Brian P. Dugan, Melanie Francis, John P. Abt, Timothy C. Sell, Scott M. Lephart

Neuromuscular Research Laboratory, University of Pittsburgh, PA

BACKGROUND

Static measurements of foot structure have previously been used in both the clinical and research setting to describe and classify foot structure (Redmond, 2008). Attempts have been made to relate these different classifications to lower extremity injuries and to identify appropriate athletic footwear in an attempt to mitigate injury (Jenkins, 2007; Knapik, 2010). Such attempts, however, have yet to yield consistent results (Barnes, 2008; Burns, 2005). Since the foot is dynamically loaded during gait and sport activities, it may be more appropriate to classify foot structure based on dynamic geometric variables. Prior to this application, the reliability of these variables must be established. The purpose of this study was to determine the reliability of geometric variables obtained during gait at a self-selected speed.

METHODS

Ten healthy males (n=8) and females (n=2) participated in this study (age: 27.7 ± 4.1 years, mass: 77.6 ± 10.7 kg, height: 174.3 ± 7.0 cm). Data were collected on two different days using the EMED-X® system (Novel GmbH, Munich, Germany), with a sampling frequency of 100Hz. A two-step approach at a self-selected speed was utilized for all trials, which previously has been demonstrated to be reliable in gait analysis (McPoil, 1999). In order for a trial to be included the following criteria were met: only one foot contacted the platform, contact was made on the second step, subjects did not "target" the platform, and subjects appeared to walk with their normal gait and cadence. After familiarization of the task, subjects performed 5 right and 5 left trials.

Geometric variables were then obtained through the Novel software package. Intraclass correlation coefficients (ICC) were calculated using a two-way random effects model (ICC [2, k]) and means and standard errors were calculated for each foot for 21 geometric variables.

RESULTS

Means, standard errors of the measurement (SEMs), and ICCs for both the left and right feet are presented in Table 1. Excellent reliability (ICC>0.90) was demonstrated in 15 of the 21 geometric variables for the left foot and 16 of the 21 variables for the right foot. Good reliability (ICC>0.70) was demonstrated in 20 of the 21 variable for both the left and right feet.

	Mean ± SEM		ICC	
	Left	Right	Left	Right
Anterior plantar angle [°]	28.10 ± 0.79	28.46 ± 0.29	0.979	0.848
Posterior plantar angle [°]	28.03 ± 0.97	28.41 ± 0.32	0.975	0.775
Lateral tarsal angle [°]	154.02 ± 1.82	153.38 ± 1.62	0.708	0.629
Medial tarsal angle [°]	149.83 ± 1.49	149.76 ± 0.73	0.965	0.853
Lateral plantar angle [°]	7.32 ± 0.14	7.49 ± 0.08	0.991	0.971
Medial plantar angle [°]	7.32 ± 0.14	7.49 ± 0.08	0.991	0.971
Long plantar angle [°]	14.65 ± 0.28	14.98 ± 0.15	0.992	0.971
Transverse plantar angle [°]	16.61 ± 5.57	14.44 ± 5.00	0.785	0.733
Hallux angle [°]	4.42 ± 0.82	4.28 ± 0.75	0.986	0.983
Hallux angle (2) [°]	6.73 ± 1.46	5.80 ± 0.96	0.992	0.981
Forefoot angle [°]	113.94 ± 1.19	114.75 ± 2.17	0.736	0.921
Subarch angle [°]	114.73 ± 2.58	108.08 ± 2.73	0.972	0.975
Heel angle [°]	9.20 ± 1.15	10.06 ± 2.61	0.656	0.933
Foot progression angle [°]	7.46 ± 0.59	10.01 ± 0.54	0.990	0.988
Foot length [cm]	27.31 ± 0.15	27.43 ± 0.14	0.993	0.991
Forefoot width [cm]	9.75 ± 0.08	9.85 ± 0.13	0.970	0.989
Heel width [cm]	5.62 ± 0.05	5.63 ± 0.03	0.995	0.989
Coefficient of spreading	0.36 ± 0.00	0.36 ± 0.01	0.871	0.974
Arch index	0.24 ± 0.01	0.24 ± 0.01	0.965	0.985
Forefoot and heel coefficient	0.58 ± 0.01	0.57 ± 0.01	0.977	0.965
Forefoot coefficient	1.08 ± 0.01	1.09 ± 0.01	0.780	0.922

Table 1: Geometric variables: mean, SEM, and ICCs for left and right feet.

DISCUSSION

Reliable dynamic assessment of foot geometry can be obtained using the EMED-X pedobarography platform. These findings support the use of dynamic foot geometry assessment in future research to classify foot structure/type and to determine the relationship between foot geometry and lower extremity injuries.

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COMPARISON OF A SIX DEGREE-OF-FREEDOM FORCE SENSOR AND PRESSURE INSOLE MEASUREMENTS IN SELECTED SKIING MANOEUVRES

U.G. Kersting (1), N. Kurpiers (2), E. Hild (3), A. Kiefmann (4), V. Senner (4)

1. Center for Sensory-Motor Interaction, Aalborg University, Denmark
2. Dept. of Sport and Exercise Science, The University of Auckland, New Zealand
3. Novel GmbH, Munich, Germany
4. Dept. Sport Equipment and Materials, University of Technology Munich, Germany

INTRODUCTION

Alpine skiing has been characterized as particularly dangerous to the knee joint (Heir et al., 2005). Many of these may be avoidable with a greater understanding of biomechanical parameters related to skiing technique and equipment design. Attempts to measure the dynamic forces transferred to the skier have been described previously. Only few have successfully realized a full six degree-of-freedom measurement system due to the high technical challenges implied (Kiefmann et al., 2006). Alternatively, pressure insole systems have been used to collect data during skiing with different applications and studies (Schaff et al., 1989).

The purpose of this investigation was to compare a pressure insole in combination with a pressure sensor placed inside the shaft at the anterior aspect of the tibia to the data collected from a full six degree-of-freedom force sensor mounted between shoe and binding. It was hypothesized that vertical reaction forces and the moment about the ankle joint can be estimated from pressure measurements.

METHODS

Tests were carried out on two occasions on slopes of appr. 20° inclination. One subject performed various reference movements and relevant skiing motions including straight runs, short and long turns on a groomed slope. Secondly, six subjects performed runs on a mogul course. Skill levels ranged from advanced to elite. Two mobile 6 DoF force plates were mounted between ski boot and binding sampling to a biovision mini computer (500 Hz). A Pedar insole system (novel GmbH) consisting of an insole and a dorsal pad, sampling at 125 Hz was worn inside one of the boots. A synchronized video was recorded for all trials.

RESULTS

During both, the reference trials as well as skiing on the slope and moguls, anterior shaft contact forces of over 80% body weight were recorded. During single legged turns this was up to 100% body weight indicating a substantial load being transferred through the shaft of modern ski boots. For a relatively upright posture vertical forces of up to 80% of the reading of the force sensor were registered by the plantar insole.

DISCUSSION AND CONCLUSIONS

Results indicate a highly complex interaction between movement at the ankle joint, positioning of the body with respect to the ski and the transfer of loads through the boot-shaft system. It was thus not possible to get a reasonable representation of the vertical force by the pressure system. The combination of shaft and plantar pressure readings was used to estimate the anterior-posterior moment between leg and ski. In some relatively static riding situation the two systems were reasonably similar. It is thus indicated that the use of pressure inside a ski boot is not sufficient for characterizing the loads experienced by the body system during alpine skiing. However, for the assessment of boot fit, balancing tasks or biofeedback studies the use of pressure insoles will be valuable. Further, it might be possible to improve the pressure system setup by covering more area of the foot- and shank-boot interface. More studies are required to explore these relationships.

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DATA COLLECTION FROM TWO DIFFERENT PRESSURE PLATFORMS USING STANDARDISED METHODOLOGY PRODUCES COMPARABLE RESULTS.

Julie. Stebbins (1,2), Claudia. Giacomozzi (3)

1. Oxford Gait Laboratory, Nuffield Orthopaedic Center, Oxford, UK
2. NDORMS, Oxford University, Oxford, UK
3. Dept of Technology and Health, Istituto Superiore di Sanità, Rome, Italy

INTRODUCTION

It is widely believed that plantar pressure data is susceptible to a large degree of intra- and inter-subject variability. This reduces confidence in interpreting results, particularly following clinical intervention. There are now a range of pressure platforms available for collecting data, utilising a broad spectrum of technology. Along with this, there is currently a lack of standardization in methodology for collecting plantar pressure data, and reporting of results.

It is unclear how much of the variability reported in plantar pressure measurement is due to varying methods of data collection and instrumentation, and how much is attributable to real variation in the data.

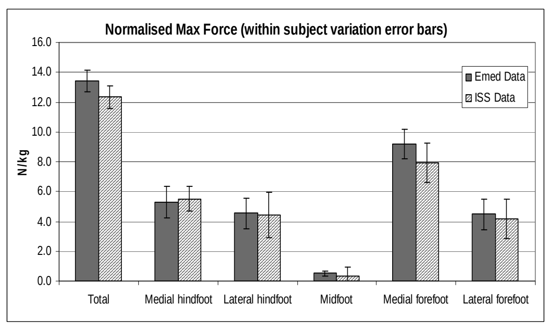
The aim of this study was to determine the amount of variability in plantar pressure assessment that is due to differences in the types of equipment used to collect the data. This was achieved by using a standardised protocol, and comparing the results and the intra- and inter-subject variability from two different pressure platforms.

METHODS

A Novel Emed-M capacitive pressure platform was used to collect data from 19 healthy children (age range 6 – 16 years). In addition, a prototype, piezo-resistive pressure platform (ISS), rigidly mounted to an AMTI force plate, was used to collect data from a different group of 15 healthy children (age range 6 – 16 years). Three footprints were collected from each foot. Both pressure platforms had a spatial resolution of 4 sensors per square cm, and collected data at 50Hz. A 12 camera, Vicon (Vicon, Oxford) system was used to collect synchronous data from markers placed on the feet, according to the Oxford Foot Model protocol (Stebbins, 2006). Co-ordinates from anatomical marker positions were projected onto the overall pressure map at a time corresponding to mid-stance, and used to automatically divide the footprint into five areas (medial heel, lateral heel, midfoot, medial forefoot, lateral forefoot) using a previously validated protocol (Giacomozzi, 2000). The peak force and peak pressure for each sub-area were compared between platforms, along with the intra-subject standard deviations.

RESULTS

Difference in the average peak force (normalized to body weight) obtained from each platform were minimal (Figure 1), (0.1-1.3 N/kg). Differences in peak pressure were slightly higher, with the least difference in the lateral forefoot (27kPa) and the most in the medial forefoot (157kPa). Differences in the intra-subject standard deviations were negligible in the peak force comparison (0.1-0.8N/kg). Differences in the peak pressure intra-subject standard deviations were much higher (16-98kPa). The co-efficient of variation for peak pressure was consistently lower for Emed. The lower accuracy of ISS was partially compensated for by instantaneously calibrating it with information from the force platform; this led to comparable peak force data, but little improvement was obtained for the accuracy of the pressure measurements. These results suggest that comparable data can be obtained from 2 different pressure platforms when the methodology is stanadardised, however accuracy of the pressure platform is critical.



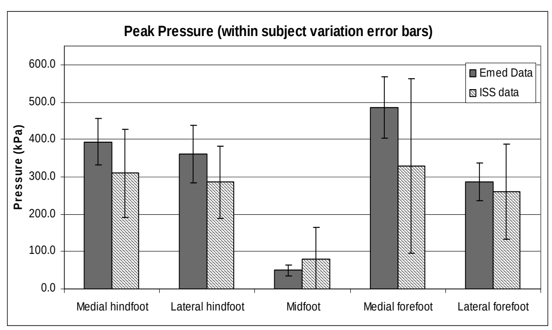


Figure 1: Comparison of average normalised force (above) and peak pressure (below)

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DO MOTION CONTROL RUNNING SHOES DECREASE SAGITTAL PLANE MOVEMENT OF THE MEDIAL LONGITUDINAL ARCH?

Thomas G. McPoil (1), Mark W. Cornwall (1), Melissa R. McNulty (1)

1. The Laboratory for Foot & Ankle Research, Physical Therapy Program, Northern Arizona University, Flagstaff, Arizona, USA

PURPOSE

While it has been proposed that motion control or stability running shoes are designed to control foot pronation, little research has been published to substantiate this claim. Cheung et al (2008) reported that motion control shoes, in comparison to neutral shoes, reduced midfoot forces during running but failed to consider contact area acting on the midfoot while walking. We hypothesized that the amount of midfoot contact area, indicative of medial and lateral longitudinal arch sagittal plane movement, would not increase between walking and running if a motion control shoe was effective in controlling foot mobility. The purpose of this study was to investigate the change in plantar surface contact area when walking and running in motion control shoes.

SUBJECTS

Ten females' subjects with a mean age of 25.5 years (range 22 to 34 years) with no history of congenital or traumatic deformity or foot problems volunteered to participate in study. All subjects ran at least 15 miles per week for the past 2 years and were fitted with new motion control shoes (Brooks Ravenna Stability) at the same local running footwear store.

METHODS

Foot measurements previously described were recorded in weight bearing so that the arch height ratio (AHR), foot mobility magnitude (FMM) and foot posture index (FPI) could be calculated for all subjects (McPoil et al, 2009). The FPI and AHR were used to determine foot posture with the FMM used to determine foot mobility. Each subject was asked to walk and run in their running shoes over a 42-meter indoor runway while in-shoe pressure data were collected using the PEDAR-X system. To determine the effect of the shoe contour on plantar surface area, each subject was asked to walk over the same distance while pressure data were collected while wearing a flat soled, non-contoured karate shoe. For all in-shoe pressure measurements, the sensor insoles were place over a flat piece of firm insole material (durometer = 58 Shore A) that replaced the original shoe sockliner. Ten consecutive steps from the middle 20 meters for the right foot only were selected for analysis. Novel percent mask and group mask software was used to determine contact area in the following plantar regions: medial heel, lateral heel, medial midfoot, lateral midfoot, medial forefoot, lateral forefoot, and hallux. A MANOVA was used to determine those plantar surface areas that were significantly difference. Based on those results, an ANOVA and Tukey's pairwise comparisons were used to further assess the medial and lateral midfoot regions.

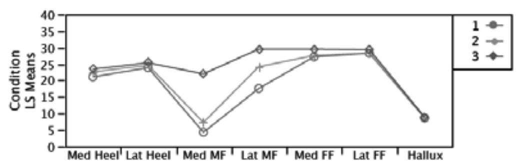

Figure 1: MANOVA results for plantar contact area (1 = karate shoe, 2 = walk in shoe; 3 = run in shoe).

RESULTS

The mean values for foot posture and mobility were: AHR = 0.335 ± 0.018; FMM = 1.58 ± 0.27; and FPI = 3.8 ± 2.35. Based on these measures only one subject was classified as excessively pronated.

Total plantar contact area was significantly increased in the medial midfoot ($p < .0001$) when running in the shoe in comparison to both walking conditions (shoe & karate). The mean surface area for the medial midfoot was 4.4 cm^2 for karate walk, 7.4 cm^2 for shoe walk, and 22.1 cm^2 for shoe run. Total plantar contact area was also significantly increased in the lateral midfoot ($p < .0001$) when running in the shoe in comparison to both walking conditions (shoe & karate). The mean surface area for the lateral midfoot was 17.7 cm^2 for karate walk, 24.3 cm^2 for shoe walk, and 29.6 cm^2 for shoe run.

CONCLUSION

The results indicate that motion control shoes, designed with a duo-density midsole and a firm heel counter, cannot prevent increased movement of the medial and lateral midfoot when running. This would suggest that sagittal plane movement of the midfoot is occurring within the motion control shoe while running.

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T31

LATERAL PLANTAR LOADING DURING CUTTING CORRELATES TO DECREASED FOOTWEAR RATING

Kevin R. Ford (1), Neil A. Manson (2), Gregory D. Myer (1), Richelle C. Gwin (3), Robert S. Heidt, Jr (3), Timothy E. Hewett (1)

1. Cincinnati Children's Hospital Medical Center, University of Cincinnati, Cincinnati OH, USA
2. Horizon Health, Saint John, NB and Dalhousie University, Halifax, NS, Canada
3. Wellington Orthopaedic and Sports Medicine, Cincinnati, OH, USA

INTRODUCTION

Plantar loading patterns have been shown to differ among cleat styles during complex sport maneuvers (Orendurff *et al*. 2008). The choice of footwear style, however, is often based on individual perceived ratings. The quantification of plantar loading patterns may help explain perceived footwear ratings during sports. The purpose of this study was to identify the relationship between foot loading patterns and perceived footwear rating.

METHODS

Seventeen male football players participated in the data collection (age: 16.9±0.9 years; height: 181.4±5.5 cm; mass: 79.8±10.9 kg). Each athlete was fitted with a new model of a noncleated turf American football training shoe (Nike Astrograbber). A flexible in-shoe pressure distribution measuring insole (Pedar, Novel) was inserted into the right shoe and sampled at 99Hz. Athletes ran a slalom course, modified from Eils *et al.* (2004), on a synthetic turf field with rubber infill. Nine side-cut steps were collected and analyzed with separate masks (medial and lateral heel, medial and lateral midfoot, medial, central and lateral forefoot, hallux, and the lesser toes). The force time integral in each individual region was divided by the force time integral for the total plantar foot surface in order to determine the relative load in each region.

Following the cutting trials, the athletes' perception of the footwear was assessed via visual analog scale (VAS) where 0mm = none/worst and 100 mm = complete/best. Categories included comfort, control, stability, traction, and safety. Athletes were also questioned whether they felt comfortable wearing and performing in the shoe at full intensity. One-way ANOVA was used to compare relative load and peak pressure between subjects that would wear (WW, n=11) and would not wear (NW, n=6) the noncleated shoe used during testing. Correlation coefficients were used to compare the relationship between pressure distribution and VAS.

RESULTS AND DISCUSSION

NW had increased relative loading in the lateral midfoot (p=0.002) and lateral forefoot (p=0.045) compared to WW. Peak pressure was significantly decreased in NW compared to WW at the medial forefoot (p=0.029). Increased lateral forefoot relative load significantly correlated to lower perceived ratings in control (r=-0.72), stability (r=-0.73), traction (r=-0.60) and safety (r=-0.66) (Figure 1). The perceived rating of comfort was not significantly correlated with relative load or peak pressure measures.

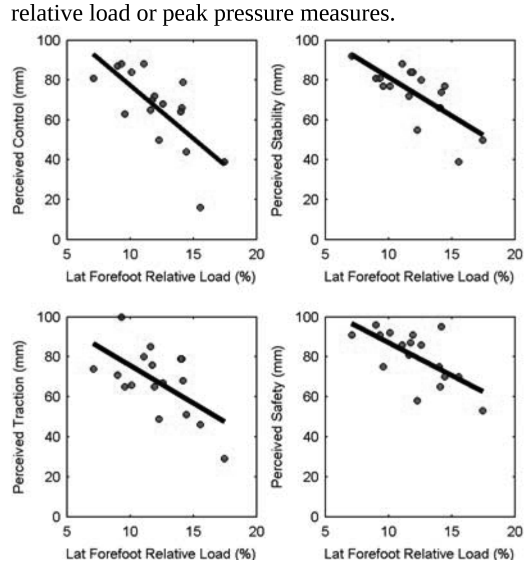

Figure 1: Relationship between visual analog scale and lateral forefoot relative load.

The differences in medial and lateral loading between groups likely relate to decreased stability and control during the cutting. The subset of athletes that did not feel comfortable performing in the noncleated footwear (NW) may have been accustomed to cleated footwear which led to greater lateral loading. Medial and lateral loading patterns have been found to differ with foot type (Queen *et al*. 2009). Additional investigation into the effects of foot type on perceived footwear rating and plantar loading are warranted.

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CROSSOVER SECOND TOE DEFORMITY: A ROBOTIC CADAVERIC STUDY OF INCREASED SECOND METATARSAL LENGTH AND PLANTAR PRESSURE

Eric C. Whittaker (1), Joel R. Weber (1), Patrick M Aubin (1,2), Bruce J. Sangeorzan (1,3), William R. Ledoux (1,3,4)

1. VA RR&D Center of Excellence, VA Puget Sound Health Care System, Seattle, WA
2. Department of Electrical Engineering, University of Washington, Seattle, WA
3. Orthopaedics & Sports Medicine, University of Washington, Seattle, WA
4. Mechanical Engineering, University of Washington, Seattle, WA

INTRODUCTION

The crossover toe deformity refers to an unstable second metatarsophalangeal joint (MTPJ) that leads to a progressive migration of the second toe in a dorsal and medial direction (Coughlin, 1993). The etiology is uncertain, but as summarized elsewhere (Kaz, 2007), it has been proposed that a long second metatarsal causes abnormal loading (i.e. increased plantar pressure) beneath the second MTPJ during stance phase. Over time, this causes MTPJ pain, agitation, medial collateral and lateral collateral ligament rupture, and toe migration. In the current study, we used a robotic gait simulator (RGS), cadaveric specimens and a surgical metatarsal lengthening process to further investigate this relationship. We hypothesized that lengthening of the second metatarsal would increase plantar pressure beneath the second MTPJ and reduce the peak pressure underneath the first metatarsal head.

METHODS

Dissection, including mid-tibia transection, exposure of the second metatarsal dorsally, and separation of nine extrinsic foot tendons, was performed for six specimens. Elongation of the second metatarsal was achieved through transection, a custom-made aluminum support bracket, and aluminum spacers of varying lengths (control, 2mm, 4mm, 6mm, 8mm). A robotic gait simulator (RGS) simulated physiologic tibial motion, tendon loading, and ground reaction forces (GRF) on the cadaveric feet. Stance phase was scaled to 10 seconds and the vertical GRF to half body weight. The GRF was measured with a force plate mounted to the RGS, and a Novel emed-sf pressure plate was mounted in series with the force plate. Custom pressure masks were created from weight bearing radiographs of each foot and used to determine the peak pressure (PP) and pressure-time integral (PTI) under the first and second MTPJs during gait. Statistical analysis was performed with a linear mixed effects model.

RESULTS

With increased spacer length, PP decreased under the first metatarsal head and increased under the second metatarsal head (Figure 1). Second metatarsal PP and PTI were positively correlated with an increase in second metatarsal length ($p = .0005$, $p < .0001$). First metatarsal PP and PTI were significantly negatively associated with second metatarsal length ($p = .029$, $p = .024$).

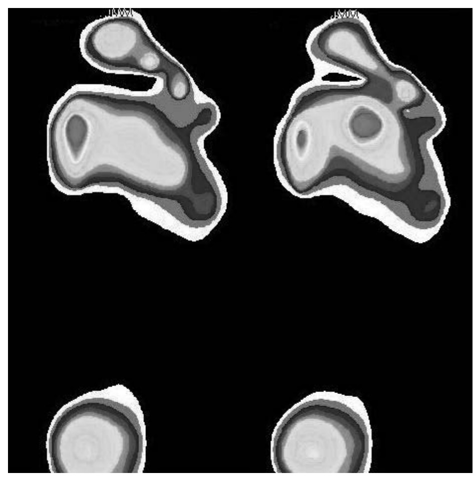

Figure 1: Peak plantar pressure for one foot in the RGS; left: control spacer, right: 8mm spacer.

DISCUSSION

Our results support the hypothesis that second metatarsal length is positively associated with plantar pressure underneath the second MTPJ. Differing from the first MTPJ, the second MTPJ capsule and soft tissue structures are not designed to support the large stance phase loads. It is thought that prolonged exposure to the plantar pressures measured in our study would lead to the observed joint subluxation and dislocation seen in crossover toe patients. Overall, these findings give a better understanding of the mechanisms responsible for the development of the deformity as well as clarify controversial associations of the deformity found in the literature.

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T33

HYPOTHENAR PRESSURE MAPPING PROVIDES INSIGHT FOR REDUCING ULNAR NERVE COMPRESSION IN CYCLISTS

Josh Slane (1), Mark Timmerman (2), Heidi Ploeg (1), Darryl Thelen (1)

1. Department of Biomedical Engineering, University of Wisconsin-Madison, Wisconsin, USA
2. School of Medicine and Public Health, University of Wisconsin-Madison, Wisconsin, USA

INTRODUCTION

Chronic ulnar nerve compression is believed to be a primary cause of sensory and motor impairments of the hand, which is common among cyclists. Prevention techniques include frequently changing hand position and wearing padded gloves. However to date, there has been no scientific examination into the effectiveness of these measures (Kennedy, 2008).

The purpose of this study was to evaluate the effects of hand position and glove padding on pressure over the ulnar nerve. Specifically, we considered three hand positions commonly used on road bicycles. We also compared gloves with either 3 mm or 5 mm gel or foam padding placed over the hypothenar eminence, thenar eminence and metacarpal heads.

METHODS

Thirty-six healthy adults (18 male, 18 female) were tested (38.9±13 yrs, 74.2±13.9 kg and 174±9cm). A stationary cycle was adjusted to match the dimensions of each subject's personal road bicycle. Subjects performed a series of trials in which hand position (Fig 1) and glove type were randomly varied while power and cadence levels were kept consistent. A pressure sensitive mat (229 sensors, Elastisens - FO44, Novel GmbH) was used to record pressure distribution over the hypothenar region of the hand. This region is where the ulnar nerve enters the hand through Guyon's Canal and is most susceptible to external compression (Black, 2007). Pressure from 12 consecutive pedal strokes were averaged to quantify both the pressure distribution and peak pressure over the hypothenar region.

Three-dimensional images of the subject's hand with and without the pressure mat were obtained using a laser scanner (Shape-Grabber Inc). These laser scans were registered with subject-specific pressure images to relate pressure profiles to the underlying anatomy (Fig 1). The compressive stiffness of the foam and gel padding was measured separately with a materials testing machine and a 15 mm diameter indenter.

RESULTS AND DISCUSSION

Padding in the hypothenar region of the glove reduced peak pressures found in the no glove condition from 21-28%, with the highest pressure reduction achieved using 3 mm of foam. The foam padding/glove system was found to be ~50% less stiff than the gel, suggesting that better pressure reduction was achieved using a more compliant interface. Absolute pressure magnitudes were greatest with the hands in the drops position (128-168 kPa), which were significantly greater ($p < 0.05$) than those found in the tops and hoods positions. Pressure magnitudes did not significantly vary between male and female cyclists.

To our knowledge, this is the first study that has measured high-resolution pressure distributions on cyclist's hands while riding. The data obtained is important for establishing a quantitative, scientific approach to design interventions aimed at reducing the prevalence of hand impairments in cyclists.

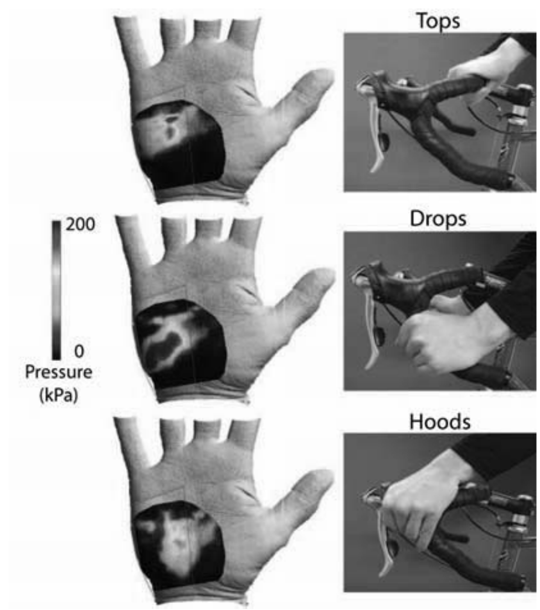

Figure 1: Pressure concentrations near Guyon's Canal were observed with the hands in the drops and hoods positions.

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USING CENTRE OF PRESSURE (COP) DATA TO IDENTIFY GOLF PUTTING TECHNIQUES

Pat McLaughlin and Russell Best

Institute for Sport, Exercise and Active Living, Victoria University, Melbourne, Australia

INTRODUCTION

Consider the scoring system in golf. On each hole, playing to "par" allows the golfer two putts per hole – a total of 36 putts over an 18-hole round, or about half the total number of shots allocated for par. Considering that players have 13 other clubs in their bag, and would tend to use most of them at some stage, the putter is the most often used piece of equipment that a player possesses. Whilst this simplistic logic may be an exaggeration for the elite golfer, data indicates that even at the elite level, around 38% of all strokes are putts (2009 PGA Tour data, www.pgatour.com)

Research into putting performance follows the same common theme - players are separated into groups based on handicap level. The question of interest seemingly "how do the elite/expert/low handicap golfers compare to the novice/high handicap golfers?" All authors assumed that putting performance was related to handicap level. On putting performance alone this was not definitively supported. Also, the authors often analysed only putts that were successful - that is putts that went in the hole (Paradisis and Rees, n.d.) or stopped over the target (Delay et al., 1997; Haltom, 1994; Zafiroglu, 1994). However, these types of studies do not investigate technique differences. In contrast, Pelz (2000) points out that there have been many different putting techniques used by professionals over the years.

The aim of the present study was to define if more than one putting technique truly exists in a sample of golfers of different handicap levels.

METHODOLOGY

A total of 38 players participated in the testing sessions. These players were members of a private suburban golf club in Melbourne and all were experienced golfers. The average age of the sample was 55.3 (±17.8) years and average handicap 15.3 (±6.9).

Simultaneous video (50Hx perpendicular to line of putt) and pressure data (16 x 16 pliance® pressure mat, 50Hz, novel, Germany) were collected for each putt. The data collection process required each player to putt whilst standing on the pressure mat which was positioned on the practice putting green. Each participant completed five putts at a hole cut into the green 4m (13ft) away.

RESULTS AND DISCUSSION

Combining the kinematic, temporal and COP data, 62 parameters were available for analysis. Players were not divided according to handicap, putting trials were analysed individually and all data entered into cluster analysis methods to define putting techniques. This allowed the creation of technique based on the movements exhibited by players during the putting stroke.

For cluster analysis, the most influential parameters in determining cluster membership were related to movement of the COP in the medio-lateral direction (COPx), namely: range of movement of COPx in the backswing; range of movement of COPx in the downswing; maximum velocity of COPx in the downswing and velocity of COPx at the time of ball contact (this was the most influential parameter).

Two distinct putting techniques were identified by cluster analysis:

1. Less movement (relative to cluster 2) of COPx in the backswing and downswing phases with velocity of COPx at ball contact closer to zero (on average). Low COPx velocity.
2. Larger movement (relative to cluster 1) of COPx in the backswing and downswing phases with velocity of COPx at ball contact non-zero. High COPx velocity.

CONCLUSIONS

Analysis of COP data is able to distinguish putting techniques in an amateur group of golfers. Cluster analysis methods allow performance techniques to be distinguished based on parameters related to execution of the skill, rather than handicap.

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EFFECT OF MATTRESS FOR PREVENTION OF DECUBITUS, BED POSITION AND SUBJECTS' BMI ON INTERFACE PRESSURE

Tae Soo Bae (1), Mu Seong Mun (1)

1. Biomechanics Team, Korea Orthopedics and Rehabilitation Engineering Cener

INTRODUCTION

Patients with poor blood circulation carry the risk of decubitus since it is generally unavoidable to use various bed positions in order to treat patients. Several mattresses have been introduced in order to prevent decubitus caused as such. However, since there are diverse kinds of them and many different methods to use them, patients' families and nurses experience many difficulties in selecting and using mattresses for prevention of decubitus. Therefore, in this study, the effects of diverse mattresses for prevention of decubitus, bed positions and subjects' body mass index (BMI) on contact pressures in the areas such as the scapula, the hip and the feet- heels was quantitatively measured, analyzed and compared with each other.

METHODS

Experiments were conducted on normal persons with under-, normal-, and over-weight (5males/ group) who were free of vascular system diseases. In this study, a general mattress and four types of mattresses for prevention of decubitus (latex, bubble, alternative, and air mattress) were largely used and five different bed positions were defined as experimental conditions using three-step electromotive beds. (Figure 1) To measure contact pressures, a order-made pressure measuring system (bedsize, Pliance FTX, v12.1.36, Novel Gmbh) was used. The mattress and the pressure sensor were stabilized for around 30 minutes and an additional process for stabilization for 30 minutes was included after the subject was laid on the mattress before the experiment in order to minimize the contact pressure measurement errors caused by air movements in the mattress. The same processes were repeated for five different bed positions. Maximum peak pressure (MPP) was separately measured using the predefined mask on supine body areas.

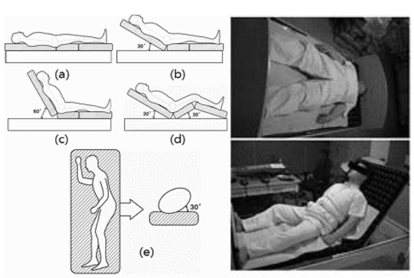

Figure 1: Various positions of Bed mattress and experiments (a)ground (b) Head 30 degress (c) Head 60 degree (d) Head 30 degree and foot 30 degree (e) Side

RESULTS

In the case of flat positions, the result indicated that only the bubble mattress was acceptable to the subject with normal body weight, latex or air mattress to the over-weighted subject and only air mattress to the low-weighted subject. In the case of the subject with normal body weight for 30^{o} position in the head area, MPP lower than the reference value were measured in all the mattresses except for latex and bubble mattress and in the case of the over-weighted subject, all mattresses were usable except for general mattresses, alternatives, and bubble mattresses but in the case of the under-weighted subject, all the mattresses showed MPP higher than the reference value. In the case of 60^{o} position in the head area, only the latex mattress was acceptable to the subject with normal body weight while all the mattresses showed MPP higher than the reference value in the case of both the over-weighted subject and the under-weighted subject. In the case of 30° positions in the head area in the leg area respectively, it was indicated that only general mattress and alternative mattresses were acceptable to the subject with normal body weight while only general mattresses, latex, and air mattresses were acceptable to the over-weighted subject. In the case of the low-weighted subject, latex, alternative, and air mattress showed MPP lower than the reference value. Lastly for side position, all the mattresses showed results exceeding the reference value in the case of the subject with normal body weight and the over-weighted subject while it was indicated that alternative and air mattresses were acceptable to the over-weighted subject.

Table 1: Experiment for measuring interface pressure. Each symbol(◎,◇, ◆, □) means acceptable position and mattress with respect to contact peak pressure of 32 mmHg

	normal					Over-weight					Under-weight				
	□	□	□	□	□	□	□	□	□	□	□	□	□	□	□
general		□							□						
latex			□			□	□		□					□	
bubble	□														
alternative		□								■				□	
air		□		□		□	□		□	■	□			□	

ground ② Head 30 deg ③ Head 60 deg ④Head-Foot 30 deg ⑤ side

CONCLUSION

Air mattresses showed generally even effects to prevent decubitus across all the bed positions followed by latex mattresses. It was indicated that decisions to use bubble or alternative mattresses for prevention of decubitus should be made more carefully.

A STUDY OF THE EFFECTS OF GEL LINER THICKNESS ON IN-SOCKET RESIDUAL LIMB PRESSURES IN TRANS-TIBIAL PROSTHESIS USERS

Rebecca Stine, MS (1-2), Andrew Hansen, PhD (2-3), Kerice Tucker, BS (2)
Erin Boutwell, MS (2), Mark Edwards, CP, MHPE (2,4), Steven Gard, PhD (1-2)

1. Jesse Brown VA Medical Center, Dept. of Veterans Affairs, Chicago, IL USA
2. Northwestern University Prosthetics Research Laboratory and Rehabilitation Engineering Research Program, Chicago, IL USA
3. Minneapolis VA Medical Center, Dept. of Veterans Affairs, Minneapolis, MN USA
4. Otto Bock Health Care, Minneapolis, MN USA

INTRODUCTION

Polliack et al. (2000) pointed out that measuring pressure between the prosthetic socket and residual limb could provide valuable information for the process of socket fabrication, modification, and fit. Previous studies have investigated how interface pressure distributions change as a result of prosthetic alignment (Seelen, 2003), adaptable ankle-foot components (Wolf, 2009), and walking surfaces such as ramps and stairs (Wolf, 2009 and Dou, 2006). Prosthetic liners may improve comfort by more evenly distributing pressure at the residual limb (Boutwell, 2009). It is also likely that changes in liner thickness will affect interface pressure. Therefore, the purpose of this study is to examine the effects of gel liner thickness (3mm and 9mm) on residual limb pressures.

METHODS

Eleven subjects with unilateral trans-tibial amputations agreed to participate in the study and signed IRB-approved consent forms. Mean age of the subjects was 56 ± 9 years with at least 6 months experience using a prosthesis and all were able to walk without undue fatigue. Two Alpha prosthetic gel (thermoplastic elastomer) liners (Ohio Willow Wood, Mt. Sterling, OH) were tested. Sockets for both gel liners were made for each subject using a CAM system called Squirt Shape (Rolock, 2000) that was developed in our laboratory. All other prosthetic components were the same for each subject. The same certified prosthetist fit and aligned the socket to each subject. Subjects were given an accommodation period of at least two weeks on each prosthesis prior to testing. A gait analysis was performed with the person walking at his normal self-selected speed for each liner condition. Mean walking speed across all subjects and all conditions was 1.14 ± 0.14 m/sec.

Prior to each gait evaluation, 6cm X 3cm pressure sensors (Pliance, Novel Electronics, Inc.) were placed over the patellar tendon (PT) region, the anterior distal tibia (DT), and the fibular head (FH). Pressure data were synchronized with the motion data to calculate gait cycles and recorded at 120 Hz.

RESULTS

Data from nine of the subjects were analyzed due to incomplete data sets from two subjects. Peak pressure values averaged over multiple gait cycles for each sensor matrix were calculated. Peak values during the first 40 percent of the gait cycle and the second 40 percent of the gait cycle were recorded and compared between the 3mm and the 9mm liners (Table 1). A nonparametric Wilcoxon signed ranks test with a Bonferroni correction was used to determine any significant median differences between the peak pressure values for each sensor matrix. No significant differences were found between liners.

Conditions		Median (IQR)		p-value
		3mm	9mm	
PT	Peak 1	210 (67)	167 (106)	0.575
	Peak 2	229 (78)	191 (116)	0.260
FH	Peak 1	258 (63)	203 (78)	0.110
	Peak 2	327 (61)	179 (80)	0.066
DT	Peak 1	217 (110)	263 (139)	0.678
	Peak 2	158 (87)	168 (129)	0.678

Table 1: Peak pressure values for each sensor matrix for each liner. (IQR = inter quartile range)

DISCUSSION AND CONCLUSION

Median peak pressure values were reduced for both peaks over the PT and the FH with the thicker liner, although the small sample size may have contributed to the lack of statistical differences between the two liners. The DT peak pressure increase may indicate that thicker gel liners provide a different pressure distribution pattern. Patient preference should be considered when determining liner thickness.

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FORCE AND IMPULSE OF KEYSTROKE DURING PIANO PLAYING. DIFFERENCES AMONG PROFESSIONAL PLAYERS IN CLASSICAL PASSAGES

H.Gaertner(1), R. Pozzo(2), H. Riehle (1)

1. University of Konstanz, Germany
2. University of Tor Vergata Rome, Italy

INTRODUCTION

An understanding of the force and impulse acting on the key by the finger associated to the sound generated is important because it makes the difference between interpretation and pure mechanical played notes. Some studies have been made in order to determine the force applied on the key and the relation with the level of loudness and the type of finger touch (Harding 1989, Kinoshita 2007, Parlitz 1998), providing important informations in relation to *piano* (*p*), *mezzo forte (mf)*, and *fortissimo (f f)* levels with *staccato* and *legato* touch. In the present study, using pressure sensors applied on the key surface, we made an analysis of the relationship between the force-time characteristics and the performance of pianists playing passages of classical music.

METHODS

Fifteen pianists (8 female and 6 male) with different level of expertise (from IP: winning international prices to TL: top level) played the first 2 measures of Schubert's Wanderer Phantasie (4 times in *ff* and 4 times in *pp*) on a grand piano (Bechstein). They were asked to play with theirs best interpretation and constancy; the finger sequences (fingering) were identical for all subjects. Five pressure sensors (S2013, 20x45 mm^2, <1,2 mm, 10-600 KPa, Pliance-Novel) were attached to the most relevant keys and collected at 300 Hz. Video records were taken via 3 Casio Cameras at 300 Hz for later 3-D analysis. Sound track was recorded via professional tools.

RESULTS

Peak force and impulse were calculated for each sensor and for each repetition. TL subjects showed 50% higher peak force in *ff* (9,6 N vs 6,4 N) but similar values in *pp* (2,7 N vs 2,6 N) compared to IP subjects. All pianists have a stable reproducibility of peak force as described by the coefficient of variation (CV 10%-20% for IP; CV 14%-27% for TL in *ff* and *pp* respectively).

Indeed, very high differences were noted in the total impulses. Overall average values for TL subjects were 73% (0,91 Ns vs 0,24 Ns in *ff*) and 69% (0,33 1 Ns vs 0,11 Ns in *pp*) higher then for IP. Significant difference were also found in the CV of impulse structure (11%-18% for IP vs 33%-35% for TL).

An efficiency index was calculated by computing the proportion of the impulse up to the peak force to the total impulse. This index reveals the capacity of the pianist to produce the desired loudness without exert unnecessary force. The relationship between this parameter and the number of touches played on each keys could reflect the stability of motor pattern and force sensibility of the pianist. Fig 1 shows exemplary the difference of impulse efficiency between IP and TL pianist (sbj. 4 and sbj. 12). It easy to recognize that for almost all the strikes subject 4 shows a significant better regulation of force production.

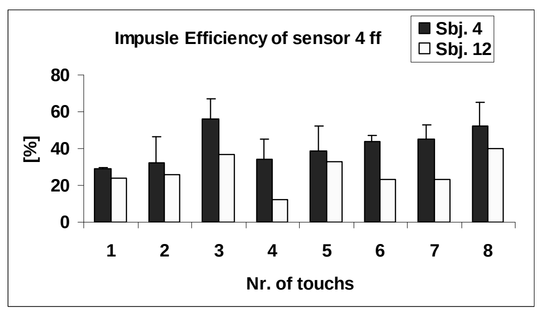

Figure 1. Mean and SD values of efficiency for two subjects with different level of expertise

Moreover an analysis of possible relationships between the force produced during playing and the force capacity of the fingers was carried out by testing maximal voluntary contraction force of each finger under isometric controlled conditions.

Sound characteristics were studied by acoustical analysis (Samplitude V7.0) and by a quality questionnaire administered to professional judges.

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THE EVOLUTION OF HUMAN RUNNING

Daniel E. Lieberman (1)

1. Department of Human Evolutionary Biology, Harvard University, Massachusetts

Ever since the human lineage diverged from the African apes, hominins have been bipeds of some sort. Comparative and fossil evidence suggest that the earliest hominins were capable, habitual bipedal walkers but were also adept at climbing trees. At some point, however, hominins lost the ability to climb trees very well, and became superlative long distance runners. Comparisons of human endurance running performance with other mammals show that we excel at speed, distance, and the climatic context in which we can run. Further, human distance running capabilities far exceed those of any other primate, and they match or even surpass the best mammalian runners in hot conditions over very long distances. These capabilities raise several questions, among them when humans became long distance runners, why these abilities evolved, and how the evolutionary history of endurance running may help avoid injury.

A review of the fossil evidence suggests that features which improve endurance running performance in humans first appear about 2 million years ago in the genus *Homo*, mostly in the species *H. erectus* (Bramble and Lieberman, 2004). These features include changes in the foot (e.g., shortened toes and the development of a full longitudinal arch), the hip (e.g., expansion of the cranial portion of the gluteus maximus), and the head (e.g., enlargement of the anterior and posterior semicircular canals). Many important adaptations for endurance running, however, are impossible to assess in the fossil record, and it is not known whether *H. erectus* was as capable a runner as modern humans.

A review of the ethnographic and archaeological records suggests that the primary advantage of long distance running was for persistence hunting. In this method of hunting, runners alternatively chase quadrupeds at a gallop speed, thus preventing them form panting, and track the animals (usually at a walk). Since quadrupeds cannot simultaneously pant and gallop, this form of hunting drives the animals into a state of hyperthermia making it easy to dispatch them without significant projectile technology, which was lacking until very recently in human evolution.

Finally, studies of habitually barefoot runners indicate that the human foot, like the rest of the body, is well designed for endurance running. People can run long distances at high speeds quite comfortably over very hard surfaces without using cushioned, high-heeled shoes. A barefoot "style" of running typically involves a forefoot or midfoot strike, both of which generate no discernable impact transient because of increased limb compliance and less effective mass at the moment the foot collides with the ground (Lieberman et al., 2010). This sort of light and gentle landing requires more calf and foot muscle strength, but generates less impact than heel-toe running in modern running shoes, and is presumably how humans ran for millions of years. Such insights raise the evolutionary medical hypothesis that more minimal shoes which foster a more natural, barefoot running style may have advantages over more cushioned shoes that facilitate heel striking.

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P1

THE EFFECT OF SPECIAL SHOE INSERT DESIGNED FOR DIABETIC PATIENTS ON PLANTAR FOOT PRESSURE DISTRIBUTION

Moshe Ayalon (1), Iftach Hetsroni (2), Meir Nyska (2)

1. The Zinman College of Physical Education, Wingate Institute, Israel
2. Meir Hospital, Sapir Medical Center, Kfar Saba, Israel

INTRODUCTION

Plantar callosities and ulcers are one of the main factors leading to serious complication in the diabetic foot. Increased plantar foot pressures are the main cause for plantar keratosis leading to foot ulcers in diabetic peripheral neuropathy. Common areas for ulcers are under the metatarsal heads and every effort should be done to reduce plantar foot pressures in trying to prevent callosities and plantar ulcers. Orthotic devices believed to be efficient in plantar foot pressure reduction.. In this study we present the effect of a novel orthotic insert that was designed to reduce the plantar pressure under the metatarsal heads. We hypothesized that with this special insert a reduction in planar pressures under the metatrsal heads will be seen compared to standard costum-made orthotics.

METHODS

Ten subjects were evaluated while walking (4.5 km/h) with: a) standard costum-made orthotics b) newly designed orthotic. The newly designed orthotic is based on integrating metatarsal bar and a pad (figure 1). In-shoe plantar pressure measurement was performed using Pedar-x system (Novel gmbh, Munich). The data of four steps for each leg were collected for each measuring condition. The average peak plantar pressure (N/cm^2) was calculated for 10 zones of interest of the foot. The analysis included comparison of standard-custom made orthotics to the newly designed orthotic.

Figure 1: The newly designed orthotic

RESULTS

The newly design orthotic reduced average peak pressures in the second and lateral metatarsals and increased average peak pressures in the heel, midfoot ,1st and 2nd toes (Figure 2).

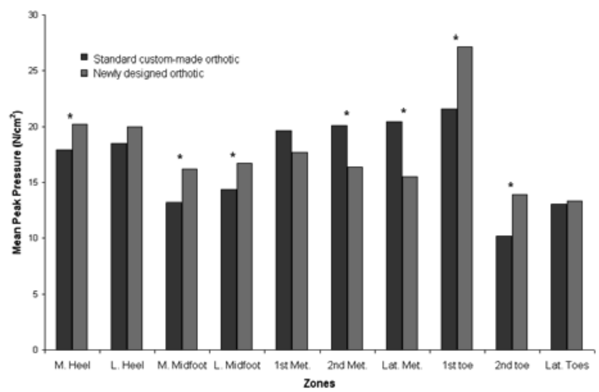

Figure 2: Comparasion of plantar pressure

DISCUSION

This study has shown that, on average, newly designed orthotic proved to reduce the plantar pressures in the 2nd metatarsal and lateral metatarsal by 22% and 31 % repectively. Being the metatrsals the most common area for callosities and plantar ulcers in a diabetic foot, the newly designed orthotic may be efficient in prevent ulceration and to improve wound healing. However, the tested orthotic increased the mean peak pressure in the mid foot and toes by 22% and 36 % repectively. This redistribution of load may put in risk especially the toes. Frequent monitoring may be required in order to make sure that not only callosities and ulcers improve in areas where plantar pressure is reduced by this device, but also that new callosities are not created in areas where pressure is increased with this special insert.

EFFECT OF GAIT SPEED CHANGES ON FOOT LOADING CHARACTERISTICS IN CHILDREN

Kerstin Bosch (1), Maike Westhues (2), Dieter Rosenbaum (1)

1. Movement Analysis Lab, Orthopaedic Department, University Hospital Muenster, Germany
2. University of Applied Science Muenster, Germany

INTRODUCTION

Plantar pressure measurements have been established as a method for the assessment of foot loading and functional restoration of foot-related problems in adults and children. For clinical problems, the effects of conservative or surgical treatment can be evaluated with repeated measurements that may help to demonstrate the changes induced by the chosen treatment option (Rosenbaum 1997). However, several factors have been identified as influential so that they should be taken into consideration.

For adults, gait speed changes may alter peak pressures and the regional load distribution but does not affect the whole foot uniformly (Rosenbaum 1994). For children in the early stages of gait acquisition, this effect has not been evaluated yet. Therefore, the present study investigated how changes in gait speed would affect foot loading characteristics in children.

MATERIALS AND METHODS

Twenty healthy children (11 male; 9 female) between 4 and 12 years of age participated in plantar pressure measurements at normal, slow and fast walking speeds. They were asked to walk barefoot across a capacitive pressure distribution platform (emed-ST) that was embedded flush in a 5.0 x 0.8 m walkway.

Gait speed was measured with infrared light gates (AF-Sport Timing System, Wesel; 10 ms resolution) that were adjusted to the child's shoulder height. After self-selected normal walking (**n**), the children were asked to walk slow (**s**) and fast (**f**) until 5 valid trials from both feet were stored at each gait speed.

Footprints were subdivided into 10 regions of interest: lateral and medial heel; lateral and medial midfoot; 1st, 2nd and 3rd-5th metatarsals; hallux; 2nd toe; toes 3-5.

As dynamic parameters peak pressure (kPa), contact time (ms) and maximum force (% body weight) were evaluated for the whole foot and the regions of interest.

The dynamic parameters were statistically tested by using the GLM (generalized linear model) for repeated measures. The level of significance was set at $p < 0.05$.

RESULTS

The average gait speed was 0.77±0.07 m/s for slow, 1.20±0.06 m/s for normal and 1.65±0.06 m/s for fast walking. The contact time decreased significantly from 808±118 ms (s) to 589±71 ms (n) and 443±44 ms (f, $p<0.001$). The maximum force increased from 105±11% body weight (s) to 115±11% (n) and 127±16% (f, $p<0.001$). The changes were most pronounced in the heel and hallux but showed even a decrease in the lateral midfoot and forefoot. The peak pressure of the whole foot increased significantly from 340±130 kPa (s) to 392±91 kPa (n) and 524±178 kPa (f, $p<0.001$), especially in the heel, 1st and 2nd metatarsals and toes but not in the lateral midfoot and metatarsals 3-5.

DISCUSSION AND CONCLUSION

The effects of systematically altering gait speed were demonstrated in children. Similar to the adults, the loading increase is not uniformly distributed across the whole plantar surface but is concentrated on the heel, 1st and 2nd metatarsals and the toes. The decreased loading in the metatarsals 3-5 that was seen in adults, could only partly be confirmed for the children.

In conclusion, the results underline the importance to ensure comparable gait speed for repeated measurements in a clinical follow-up of single patients also for children. If this cannot be achieved e.g. because of age-dependent developmental phases, comparisons should bear the potential changes in mind that were demonstrated in the present results.

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ACKNOWLEDGMENT

We wish to thank the children and their families for their voluntary participating.

EFFECTS OF INTENSE RUNNING TO EXHAUSTION ON THE IN-SHOE PLANTAR PRESSURE PATTERNS IN YOUNG MIDDLE-DISTANCE ATHLETES

François Fourchet (1), Luke Kelly (2), Cosmin Horobeanu (1), Grégoire P. Millet (3)

1. Health Centre, Aspire Academy for Sports Excellence, Doha, Qatar
2. Aspetar, Orthopaedic and Sports Medicine Hospital, Doha, Qatar
3. ISSUL Institute of Sport Sciences-Department of Physiology, Faculty of Biology and Medicine, University of Lausanne CH-1015, Lausanne, Switzerland.

INTRODUCTION

It has been reported that intense exhaustive running increased forefoot loading in experienced adult runners and may explain some foot/ankle overuse injuries (Weist 2004). However to our knowledge no studies investigated that topic in adolescent athletes. The aim of this study was to examine the effects of intense running to exhaustion (Tlim) on the in-shoe plantar pressure patterns in young elite middle-distance athletes

METHODS

Eleven male adolescent distance runners (age: 16.9±2.0, body mass: 54.6±8.6 kg, height: 170.6±10.9 cm, maximal aerobic speed: 18.7±1.5 km.h-1) performed a constant-pace Tlim at 95% of their maximal aerobic speed (17.8±1.4 km.h-1) on a treadmill (h/p/Cosmos, Nussdorf-Traunstein, Germany).

Plantar pressure measurements were performed with one sensor insole worn in the right running shoe (Novel, Munich, Germany).

Data were recorded first at 1 minute after the start (1 min) and 30 s prior exhaustion (max). A regional analysis was performed utilizing nine separate ‘‘masks’’ or areas of the foot (Groupmask Evaluation, Novel, Munich, Germany).

Mean area (mA) and contact time (CT) were determined for the nine selected regions. In addition, the relative load (RL%) was calculated as the force time integral in each individual region divided by the force time integral for the total plantar foot surface.

One Way ANOVA Repeated Measures was performed and Tukey post-hoc analyses were used. Statistical significance was accepted at $p < 0.05$.

RESULTS AND DISCUSSION

Running time to exhaustion was 8.8±3.4 min. Between 1 min and max, significant increase in CT ($P < 0.001$) was observed in all regions except Hallux. mA (Fig. 1) and RL% (Fig. 2) increased significantly under the medial midfoot area. The other measured parameters did not change.

The longer CT in fatiguing conditions may result from a reduced strength and stretch-shortening-cycle efficiency of plantar flexor muscles. In adults, the following over load of the medial arch of the foot is the first phase of a compensatory midfoot landing strategy (Willson 1999) and induced also medial and central heads of metatarsal harmful overload (Weist 2004). Conversely no forefoot overload effect was reported in the present results for adolescent athletes.

Then we may assume that the medial arch of the foot is a structure of primary interest to absorb the increased foot loadings in fatiguing conditions in young elite middle-distance runners. Therefore reinforcement of the muscles supporting the medial arch might be implemented in foot/ankle injuries prevention programs.

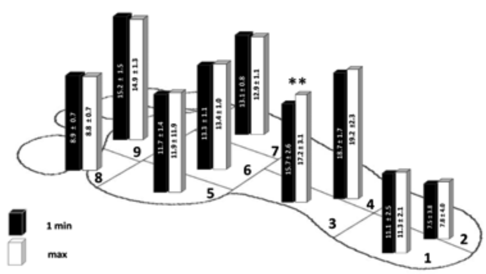

*Figure 1: Mean +/- SD mean area (mA) at 1 minute after the start (1 min) and prior exhaustion (max).**, P < 0.01*

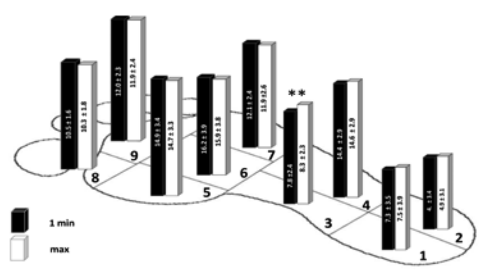

*Figure 2: Mean +/- SD relative load (RL%) at 1 minute after the start (1 min) and prior exhaustion (max).**, P < 0.01*

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CLINICAL APPLICATIONS OF A FOREFOOT CONIC CURVE MODEL

Howard Hillstrom (1,2), Andrew Kraszewski (1), Philip Demp (2), Betty Chow (1), Mark Lenhoff (1), Jinsup Song (2), Benjamin Heilman (2), Sonali Rajan (1), Andrea Woodley (1)

1. Leon Root MD, Motion Analysis Laboratory, Hospital For Special Surgery, New York, NY
2. Temple University School of Podiatric Medicine, Temple University, Philadelphia, PA

INTRODUCTION

Biomechanical differences between healthy feet (pes planus, rectus, and cavus) and those with pathology (diabetic hallux valgus) may be related to their fundamental forefoot geometries. The specific aim of this project is to quantify and compare conic curve parameters among healthy and diabetic feet.

METHODS

We hypothesized that asymptomatic healthy individuals with pes planus, rectus, and cavus feet and diabetics with hallux valgus will show differences in conic curve equation parameters. Sixty-one healthy test subjects were stratified according to resting calcaneal stance position and forefoot to rearfoot relationship into their foot types at HSS. Patients with Diabetes and hallux valgus were evaluated at TUSPM. The x-y-z metatarsal head (MTH) locations were acquired with 3D motion analysis. A unique conic curve[1] was fit to each set of points and its resulting equation normalized by the first term (see Figure 1). Parameters of interest were B, C, D, E, and curve eccentricity. Data were analyzed with a univariate mixed-effect analysis of variance model followed by Bonferroni post-hoc t tests. Foot type and trial were modeled as fixed and random effects.

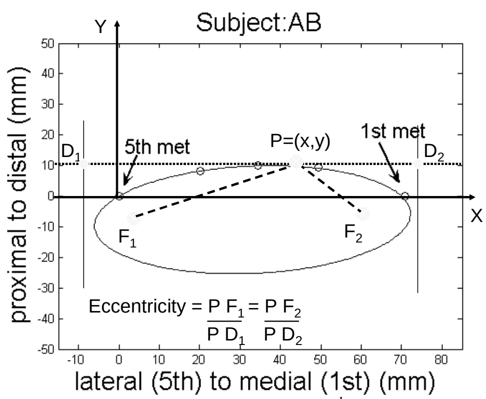

Figure 1: 2D Forefoot Model. The 5^{th} MTH was on the origin (0,0) and the 1^{st} MTH along the x-axis. Each conic (blue curve) is defined uniquely by the equation: $Ax^2+Bxy+Cy^2+Dx+Ey$, where A=1.

RESULTS

The conic model did not significantly distinguish among foot type with the exception of parameter D, the x-axis scaling factor in the conic curve equation; it distinguished rectus from diabetic feet with hallux valgus. As shown in Figure 2 several structural variables had significant Pearson correlations with model parameters; in particular parameter D was significantly correlated to most.

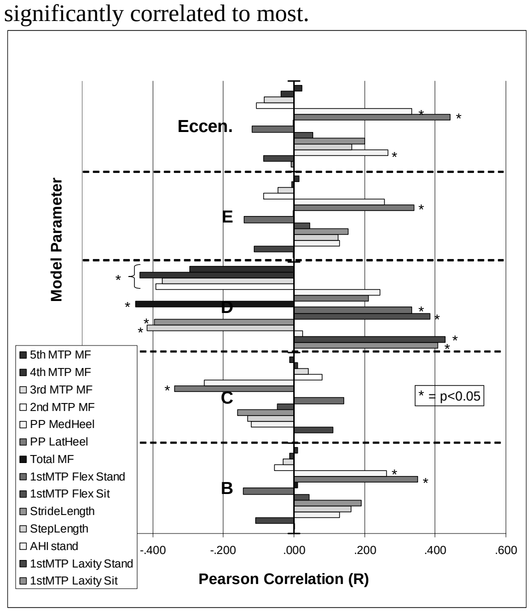

Figure 2: Pearson Correlations

DISCUSSION

The geometric forefoot model was significantly correlated with AHI, 1^{st} MPJ flexibility and laxity, total maximum plantar force and peak pressure beneath the heel. This suggests that simple models may be useful to differentiate normal vs. pathological feet.

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Variation of tactile cues reduce Nociceptive Capacity of Plantar Irritating Stimulus impact on walking gait

Marc JANIN PhD Pod. ; Podiatrist[12] and Philippe DUPUI PH-MCU [1]

1. Laboratory of de Physiology, Toulouse Medical College, 133 Route de Narbonne, 31062 Toulouse, France
2. Podiatrist office, 7 Rue de Treguel, 86000 Poitiers, France

INTRODUCTION: Nociceptive Capacity of a Plantar Irritating Stimulus (NCPIS) affects plantar cutaneous somaesthesia and influences kinetic posture by modifying control systems (1,9). This NCPIS doesn't express itself like reason of pain in podiatrist consultations but affects the postural control [1]. To understand better the repercussions of the NCPIS on gait analysis, we analysed the relationship between dynamic baropedograpphy through Latero-Medial Index (L/M , 2-4,7) of area: L/MA and force: L/MF ; Fig. n°1) of 15 subjects with NCPIS (pathological: P) and without NCPIS (non pathological: NP). Nociceptive sensory of NCIPS on walking track (H) was reduced by foam (F) during walking before and/or on baropedography force plate (8). 4 levels of sensory information were applied: start walking (S) and arrival (A) on the force plate were delivered the same way with the tactile information: hard (SHAH), foam (SFAF); start walking and arrival on the force plate crossed each other (SHAF ; SFAH). ANOVA and Schéffe post-hoc tests were used to test the possible difference between P and NP for L/MF and L/MA.

RESULTS: No significant differences were present among the L/MF: between P and NP ($F=0.080$; $p=0.779$), between H and F ($F =0.1622$; $p=0.190$), no inter action P/NP and H/F ($F=0.513$; $p=0.674$). Results were significantly different with regard to its L/MA: between P and NP ($F = 63.90$; $p<0.0001$), between H and F ($F= 299.37$; $p<0.0001$) and inter action P/NP and H/F (($F = 295.83$; $p<0.0001$). Post Hoc test showed no significantly difference into P and NP when start walking was on foam independently of the quality of arrival (H or F).

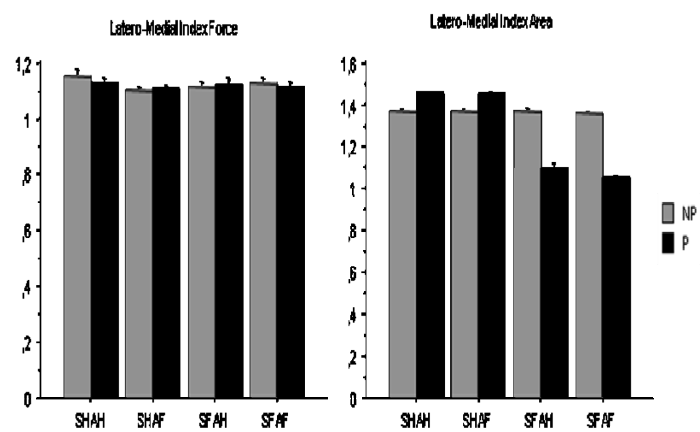
CONCLUSION: No modifications of L/MF are present because L/MF are modified when weight add up like the literature explain. For L/MA differences are observed. For P, arrival on foam reduces nociceptivce cue for NCPI (1,8). Their sensory information will come back as normal. Normalization of sensory feedback will be improving muscular responses (5,7). Their kinetic foot pattern will find back the pattern of subject to NP (no difference into SFAH, 5-6, 8).

Those results confirm the potential nociceptive capacity of the NCPI on postural control and posturo kinetic capacity.

FIGURES

Figure 1: Baropodometry to calculated L/M Index.

Figure 2: Latero-Medial Index Force and Aera. NP: non Pathological; P: pathological SHAH start and arrival on hard, SHAF start on hard and arrival on foam, SFAH start on foam and arrival on hard, SHAH start and arrival on foam.



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EFFECT OF INTERLOCKING PATTERN IN ELECTRICAL BED ON THE PREVENTION OF BEDSORE

Hyunho Choi, Jong Hyun Kim, Jae Soo Hong, Keyoung Jin Chun, #Dohyung Lim

Gerontechnology Center, Korea Institute of Industrial Technology, Chungnam, Korea
#Corresponding and Presenting Author: dli349@kitech.re.kr

INTRODUCTION

This study was motivated by a demand to prove scientifically the fact that applying an optimum interlocking pattern to electrical beds can be helpful to prevention of bedsore in bedridden patients (Evan and Loyal, 2007; Maki, 2009).

MATERIALS AND METHODS

Participants: Following Institutional Review Board approval, seven healthy female (65.6 ± 4.1 years, 151.2 ± 3.8cm, and 62.6 ± 5.6kg; mean±SD) were participated.

Electrical Bed Selection: Two electrical bed (Type A: KQ-86320, Paramount Bed, Japan and Type B: PZB-H3S, Platz Bed, Japan) were chosen by the analyses of market price, market share, sales volume, user approaching, brand awareness, test convenience, etc.

Electrical Bed Operation: Automatic interlocking patterns built-in to the electrical beds were used for the electrical bed operations. Here, automatic interlocking patterns were different from each other in characteristics.

Measurements and Data Acquisition: Contact area and pressure distribution were measured by pressure mapping system (Pliance FTX, Novel, Germany) to identify a possibility of bedsore occurrence and an effect of the automatic interlocking pattern change. They were also used in considering contact and boundary conditions in musculoskeletal modeling and to validate the musculoskeletal models described below. Three-dimensional (3D) motions were measured by 3D motion analysis system (VICON Motion System, VICON Ltd., England) to identify the characteristics of the automatic interlocking patterns of the electrical beds and the participant motions. Here, data of the 3D motions were used in musculoskeletal modeling described below.

Musculoskeletal Modeling and Analysis: To identify alteration of normal and shear forces on the body of the participant as changing the automatic interlocking pattern of the electrical bed, musculoskeletal models for the electrical beds and participants were developed by BRG. LifeMOD (LifeModeler, Inc., USA).

Statistical Analysis: A paired t-test and one-way ANOVA test with Tukey's-b post hoc multiple comparisons were used to identify existence and nonexistence of statistically significant differences. Here, the significance levels for all statistical tests were set at 0.05.

RESULTS AND DISCUSSIONS

Patterns and values of contact area and pressure distribution were significantly alerted as changing the automatic interlocking pattern of the electrical bed ($p<0.05$) (Figure. 1). Patterns and values of normal and shear forces were also altered by the automatic interlocking pattern changes (Figure 2). It was generally found that as using the automatic interlocking pattern of Type B bed, pressure values were repeatedly approached or exceed to the critical value of bedsore occurrence. Also, it was identified that normal and shear forces generated by using the automatic interlocking pattern of Type B bed were generally higher than those of Type A bed. These results indicated that using the automatic interlocking pattern of Type B bed could be more vulnerable to bedsore than using that of Type A bed. Therefore, from the current study, it is judged that considering an optimum interlocking pattern in manufacturing electrical bed may be one of important factors to prevention of bedsore.

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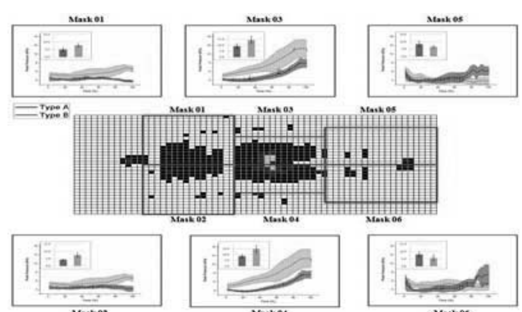

Figure 1: Peak pressure distribution. Here, the results of contact area were not shown because of limitation of page. However, a tendency of the results of contact area was similar to those of pressure distribution.

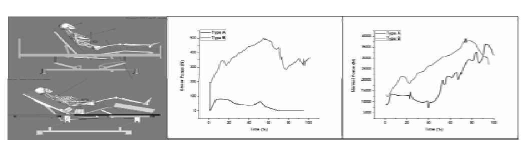

Figure 2: Normal and shear forces predicted from the musculoskeletal model analysis

BIOMECHANICAL FACTOR TO BE CONSIDERED DURING POWER-LIFT DESIGN TO REDUCE THE RISK OF MUSCULOSKELETAL DISORDERS

Hyunho Choi, Keyoung Jin Chun, #Dohyung Lim

Gerontechnology Center, Korea Institute of Industrial Technology
#Corresponding and Presenting Author: dli349@kitech.re.kr

INTRODUCTION

This study was aimed at 1) identification of the hypothesis that musculoskeletal disorders (MSDs) can be occurred to caregivers by repeated uses of power-lift (Sean, 2005) and 2) suggestion of biomechanical factors to be considered to reduce the risk of MSDs if the hypothesis is true.

MATERIAL AND METHODS

Participants: Following Institutional Review Board approval, one young healthy man (age: 26 years, height: 172cm, weight: 67 kg) was participated.

Power-Lift Selection: Two power-lifts (Type A: Bolero, ARJO, USA and Type B: MONA, Horcher, USA) were chosen by the analyses of market price, market share, sales volume, user approaching, brand awareness, test convenience, etc.

Power-Lift Operation: Operations to drive power-lift at strait and curve tracks were selected based on the analysis of power-lift workings performed frequently at care facilities. Then, the participant operated power-lift during ten minutes and took a rest for forty minutes between operations.

Measurements and Data Acquisition: Ten EMG sensors (Tringo Wireless EMG System, DELSYS, USA) were attached to the right muscles of the participant (Table 1). The EMG data were analyzed by using MDF (Median Frequency) technique. Contact area and pressure distribution were measured by pressure mapping system by using pedar and pliance system (Novel gmbh, Germany) to identify a risk of MSDs. They were also used in considering contact and boundary conditions in musculoskeletal modeling and to validate the musculoskeletal models described below. Three-dimensional (3D) motions were measured by VICON Motion System (VICON Ltd., England) to identify the characteristics of motions of the power-lift and the participant motions. Here, data of the 3D motions were used in musculoskeletal modeling described below.

Musculoskeletal Modeling and Analysis: To identify muscle forces and joint torques required for power-lift operations, musculoskeletal models for the power-lift and the participants were developed by BRG. LifeMOD (LifeModeler, Inc., USA). Additionally, the musculoskeletal models were used to identify how much effective in reduction of the risk of MSDs when the suggested biomechanical factors were considered in new power-lift design.

Statistical Analysis: A paired t-test was used to identify existence and nonexistence of statistically significant differences. Here, the significance level was set at 0.05.

RESULTS AND DISCUSSIONS

Possibility of muscle fatigue occurrence was generally increased during power-lift workings ($p<0.05$) (Table 1). This result indicates that our hypothesis is acceptable and further more power-lift design should be improved with consideration of biomechanical and ergonomic factors. The contact area and pressure generated as using Type A power-lift was generally lower and higher, respectively, than those as using Type B power-lift ($p<0.05$) (Figure 1). The results also showed that the contact area and pressure were depended on power-lift design, particular in handle and caster shapes (Figure 1). Therefore, from the results obtained the current study, it is supposed that improvement of power-lift design with consideration of biomechanical and ergonomic factors (here, handle and caster shapes) may be one of solutions for reduction of the risk of MSDs.

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Table 1: The rate of muscle fatigue occurrence

(Unit :%)			Straight		Curve	
	Function	Muscle	Type A	Type B	Type A	Type B
Upper Body	Protraction	Prapezius	2.5	6.5	10.5	33.1
	Adduction	Pectoralis Major.	—	13.7	—	14.5
	Flexion	Biceps Brachii	3.2	—	6.8	4.0
	Adduction	Triceps Brachii	—	7.6	—	0.6
	Extension	Erector spinae	—	5.5	—	5.2
Lower Body	Adduction	Gluteus Medius	—	0.5	8.2	7.0
	Extension	Rectus Femoris	1.9	7.5	5.8	—
	Extension	Biceps Femoris	—	3.0	0.7	—
	PlantarFlexion	Tibialis anterior	3.1	11.9	3.3	11.4
	PlantarFlexion	Gestrocnemius	1.5	12.1	10.7	11.2

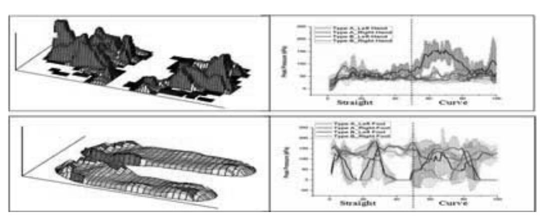

Figure 2: Peak pressure distribution of hands and feet. Here, the results of contact area were not shown because of limitation of page. However, a tendency of the results of contact area was similar to those of pressure distribution.

VARIATIONS IN PLANTAR LOADING PATTERNS IN INDIVIDUALS WITH SOFT TISSUE VERSUS BONEY REARFOOT TRAUMA: A PRELIMINARY STUDY

Stephanie Albin (1), Mark W. Cornwall (2), Thomas G. McPoil (2)

1. **Physical Therapy Foot & Ankle Clinic, The Orthopaedic Specialty Hospital, Salt Lake City Utah, USA**
2. **The Laboratory for Foot & Ankle Research, Physical Therapy Program, Northern Arizona University, Flagstaff, Arizona, USA**

PURPOSE

Little research has been conducted to determine the differences in plantar loading patterns following surgical repair of Achilles tendon ruptures versus displaced calcaneal fractures (Schepers et al, 2008; Hastings et al, 2000). While both types of traumatic injuries involve significant hindfoot structures, one could expect differences in plantar loading as a result of the characteristics of the involved tissues. The intent of this preliminary study was to assess the variations in plantar loading patterns in individuals who had undergone a surgical repair of a) Achilles tendon following rupture or 2) intra-articular calcaneal or tibial plafond fracture.

SUBJECTS

Ten male subjects participated in the study; five had Achilles tendon repairs (ATR) and five were post-Open Reduction Internal Fixation for hindfoot fractures including intra-articular calcaneal or tibial plafond fractures (HFFx). The mean age of the ATR patients was 51.2 years and 33.0 years for the HFFX. At the time of testing, approximately 3 to 4 months post-surgery, all subjects were able to ambulate without using an assistive device. All subjects had no history of congenital or traumatic deformity or foot pain on the non-involved side.

METHODS

Each subject was asked to walk over a 15-meter walkway while in-shoe pressure data were collected using the PEDAR-X system. For all in-shoe pressure measurements, the sensor insoles were positioned over a flat piece of firm insole material (durometer = 58 Shore A) that was placed in a flat rubber-soled, non-contoured karate shoe. Ten consecutive steps from the middle 8-meters of the walkway for both the involved and un-involved feet were selected for analysis. Novel percent mask and group mask software were used to determine maximum mean pressure (MMP) and normalized Maximum Force (MaxF) for each of the following plantar regions: medial heel (1), lateral heel (2), medial midfoot (3), lateral midfoot (4), medial forefoot (5), and lateral forefoot (6). A MANOVA was used to determine those plantar regions in which significant differences occurred between MMP and MaxF between the two groups. Based on those results, an ANOVA and Tukey's pairwise comparisons were used to further analyze those plantar regions that were statistically significant. For this abstract, only the plantar pressure results for the involved side will be presented.

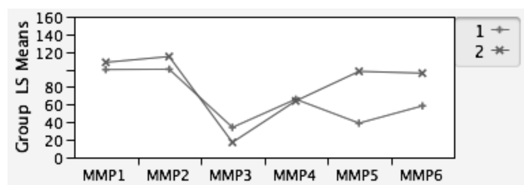

Figure 1: MANOVA results for Maximum Mean Pressure (MMP) (1 = HFFx; 2 = ATR).

RESULTS

MaxF was not significantly different between ATR and HFFx for any of the six plantar regions analyzed. MMP was significantly decreased ($p < .05$) in the medial midfoot and medial forefoot plantar regions for the HFFx group in comparison to the ATR group.

Stance phase durations were significantly decreased ($p < .05$) between the involved and non-involved limbs for the HFFx group but not for the ATR group. The mean stance duration for the involved and non-involved limbs was: 1.01 sec and 0.86 sec for HFFx and 0.78 sec and 0.77 for ATR.

CONCLUSIONS

Although a small cohort of subjects were evaluated in this preliminary study, it would appear that boney trauma of the hindfoot when compared to soft tissue injury, not only significantly slows walking speed, but also substantially reduces the amount of medial plantar loading of the midfoot and forefoot regions of the foot. These results have important implications for foot orthoses design.

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EFFECTS OF SUB-HALLUCIAL WEDGE AND MEDIAL ARCH SUPPORT ON DYNAMIC PLANTAR PRESSURE

Craig Nimick, Jaminelli Banks, Robert Marroquin, Christopher Hood, Dana Tango, Reagan Kane, James Furmato, Kendrick Whitney, Jinsup Song

Gait Study Center, Temple University School of Podiatric Medicine, Philadelphia, PA

INTRODUCTION

Functional hallux limitus (FHL) (Dananberg, 1986) has been described as a predecessor to structural and debilitating deformities such as hallux rigidus and lesser metatarsalgia. Compensatory changes include increased lesser metatarsal plantar pressures and decreased weight bearing under the hallux. Case reports have shown use of a sub-hallucial wedge to decrease lesser metatarsal pain and increase weight bearing pressures under the first metatarsal phalangeal joint (1st MTPJ). (Clough, 2005) Other studies have demonstrated that control of excessive STJ pronation with an orthotic arch support increases 1st MTPJ dorsiflexion in subjects with FHL. (Munuera, 2006) The purpose of this pilot study was to objectively evaluate the biomechanical effect of a sub hallux wedge and arch support in subjects with FHL.

MATERIALS AND METHODS

Ten asymptomatic healthy subjects with FHL were evaluated. All subjects had >50º of dorsiflexion of the 1st MTPJ in open kinetic chain and <30 º of dorsiflexion in weight bearing as measured with a goniometer. Dynamic plantar pressure was collected using a EMED-X (Novel LLC, St. Paul, MN) in three test conditions; barefoot (BF), barefoot with a sub hallucial wedge (CW), and barefoot with a sub hallucial wedge and arch support (CWA) during self-selected comfortable paced walking. Sub-hallucial wedge and medial arch support were placed onto each foot using a custom designed brace. Peak pressure time integrals (PTI) were analyzed under metatarsal heads 1-5 and the hallux. One way Analysis of Variance was performed at a significance level of 0.05.

RESULTS

Ten subjects (20 feet), with mean age of 26.8 years and BMI of 27.8 kg/m^2, participated in the study. There was no statistically significant difference in self-selected walking speed (mean 1.26m/s) across 3 conditions. As expected, PTI beneath 1st MTPJ was significantly lower than 2nd and 3rd MTPJs in barefoot walking. When subjects walked with CWA, significantly reduced PTI was noted under MTPJs 2-5, compared to both BF and CW conditions (Figure 1).

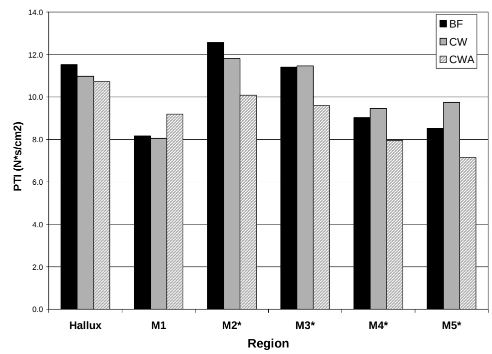

*Figure 1. Mean Peak Pressure Time Integrals (N*sec/cm^2) of the hallux and metatarsal heads 1-5 for each test condition. * denotes P<0.05*

CONCLUSION

A sub-hallucial wedge (CW) by itself did not alter dynamic plantar pressure when compared to barefoot comfortable paced walking. The combination of the sub hallucial wedge with a medial arch support significantly reduced PTI under all lesser metatarsal heads. The hallucial wedge and medial arch support appear to compliment one another, suggesting the importance of addressing both forefoot and rearfoot mechanics when treating patients with a functional limitation of the 1st MTPJ.

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THE EFFECTS OF SLIPPERS AND LOWER LIMB POSITIONING ON PLANTAR PRESSURES IN SELLECTED *BALLET* BALANCES

Érica Engrácia Valenti (1), Paula H. Lobo da Costa (1),

Kerstin Bosch (2), Dieter Rosenbaum (2)

1. Department of Physical Education, Federal University of Sao Carlos, Sao Carlos, Brazil.
2. Movement Analysis Laboratory, Orthopaedic Department, University Hospital, Münster, Germany.

INTRODUCTION

Classical *ballet* teaching and practice are strongly influenced by tradition, however, biomechanical investigations have recently contributed to ensure a more secure practice and an evidence-oriented teaching (IMURA *et al*., 2008). This study aims to describe the effects of slippers and limb positioning on plantar pressure variables produced by selected techniques of classical *ballet*.

METHODOLOGY

Fourteen non-professional female dancers aged between 15 and 25 years old, without any musculoskeletal impairment, who have practiced for at least seven years volunteered to this study. They performed three valid trials of standing for four seconds in the following balance postures: *attitude devant, derriére* and *attitude a la second*, with slippers and barefoot.

Plantar pressure variables were quantified with an EMED platform (Novel, Germany) at 50 Hz sampling rate. Peak pressures and contact areas where compared between slippers vs. barefoot conditions and for the three different lower limb positions with a univariate ANOVA-test. Significant level was determined for $p \leq 0.05$.

RESULTS AND DISCUSSION

Contact areas (Table 1) were statistically higher for the barefoot condition and no effect of the balance postures were found. The barefoot/slippers effect was not found for the contact area.

Contact area	*Attitude devant*	*Attitude derriére*	*Attitude a la second*
Barefoot	169.0* ±21.5	172.4* ±24.4	174.0* ±24.0
Slippers	155.2 ±12.7	157.6 ±14.2	158.4 ±14.3

*Table 1: Means (±SD) for contact areas (cm^2). N = 14. *significant difference for barefoot vs. slippers.*

Highest values of peak pressures were also found for the barefoot condition and again no differences were found among the three balance postures tested.

Peak pressure	*Attitude devant*	*Attitude derriére*	*Attitude a la second*
Barefoot	172.9* ±24.2	172.3* ±24.5	175.8* ±23.1
Slippers	155.7 ±13.5	157.7 ±14.2	170.3 ±44.5

*Table 2: Means (±SD) for peak pressures (kPa). N = 14. *significant difference for barefoot vs. slippers.*

These results are in accordance with the higher stability reported by the dancers during the barefoot trials, compared to the trials performed with slippers. The use of slippers may have slightly limited and compressed dancers' toes, a fact that may have reduced the contact area with the ground and affected the peak pressures.

In conclusion, our findings illustrate the foot loading characteristics during the performance of balance postures. Moreover, the foot training may be emphasized and have beneficial effects for the performance of these classical *ballet* techniques.

Figure: A ballet dancer during "Attitude devant" position on the EMED platform (with permission).

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ACKNOWLEDGEMENT

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DYNAMIC EFFECTS OF DIFFERENT ALTERNATING CYCLE TIME CONDITIONS ON SOFT TISSUE PERFUSION RECOVERY

Byeong Hee Won(1,2), Youn Jung Choi(2), Keyoung Jin Chun(2), Chang-seop Song(1)

1. Department of Mechanical Design and Mechatronics, Hanyang University, Republic of Korea
2. Gerontechnology Center, Korea Institute of Industrial Technology, Republic of Korea

INTRODUCTION

Pressure ulcers (PU) are a significant and costly problem in health care area. Alternating pressure air mattress (APAM) is one of the popular product for preventing PU. Usual operating parameters of APAM are alternating cycle time and air cell pressure level. The objective of current study is to quantify a tissue vitality with a variation of alternating cycle time through tissue perfusion and interface pressure measurement to understand proper operating conditions.

METHOD

Six male(age:68.6±5.1, BMI:24.5±3.2) and six female subjects(age:64.1±4.2, BMI:23.3±2.5) were selected to test the tissue perfusion and pressure distributions at the air mattress(3-cell type, average air cell pressure: 34~37mmHg, measuring time:1hour). For each 3 alternating cycle time conditions(3,5,10min). TCM4 Oximeters(Radiometer A/S, Denmark) were used to record transcutaneous partial pressures of oxygen($tcPO_2$) and carbon dioxide($tcPCO_2$) from the tissue. Electrodes were calibrated using known gases and were attached to the skin at chest, sacrum and heel. Pliance pressure mapping system (Novel, Germany) which consists of a thin mat containing grids of miniature pressure sensors, was placed between the subject and the mattress, one at a time, to electronically record information on pressure distributions. Force and peak, mean pressure data was collected with the Pliance.

RESULTS

The recovery & occlusion of tissue perfusion appeared according to the deflation & inflation of air cell. Perfusion recovery time(t_R) and recovered pressure(P_R) were defined as dynamic parameters of tissue perfusion from 5% to 95% of $tcPO_2$ values during tissue recovery period(Figure 1). Recovery time and recovered pressure for each alternating cycle time conditions are given at sacrum (Figure 2). From the figure, we can see the distinct results that shorter alternating cycle condition gives faster tissue recovery. And the recovered pressure decreased gradually with the increase of alternating cycle time. Thus these results means that shorter cycle time is more effective in the aspect of tissue vitality because faster and stronger recovery characteristics can be acquired. However peak pressure and force from interface pressure measurement was irregularly changed with alternating cycle time(Figure 3). This might be affected by time variant interface condition.

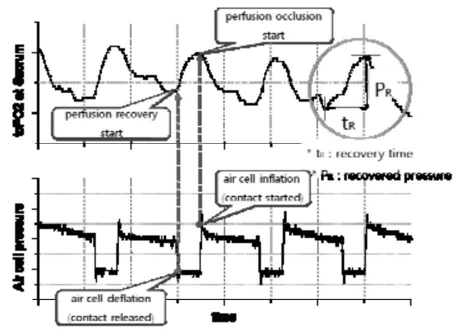

Figure 1: Results of tissue perfusion. Definition of recovery time(t_R), recovered pressure(P_R).

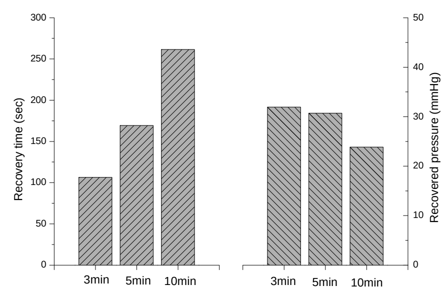

Figure 2: Recovery time and recovered pressure for each alternating cycle time.

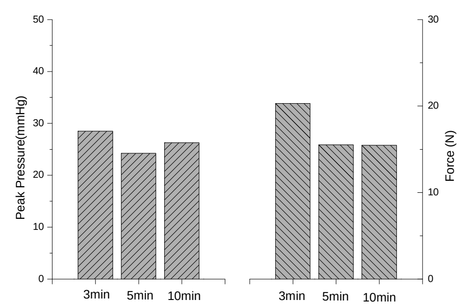

Figure 3: Peak pressure and force from interface pressure measurement for each alternating cycle time.

CONCLUSION

The current study may be valuable by quantifying tissue recovery characteristics, i.e., tissue perfusion recovery enhanced by shorter alternating cycle condition, of the APAM application for PU prevention. The fining may contribute to understanding dynamic effects of time-varying external contact to tissue

A NOVEL APPROACH USING A FE-FOOT MODEL FOR CLINICAL APPLICATIONS

Wyss, Christian[1,2]

1. Laboratory of Movement analysis, Childrens University Hospital Basel, Switzerland
2. Laboratory of Movement analysis, Cantonal Hospital Aarau, Switzerland

INTRODUCTION

Most FE models in the literatur [1, 2, 3] are based on boundary conditions without concerning muscular forces of the lower leg. The aim of our study was to get boundary conditions using data of a gait analysis in combination with ANYBODY muscle modelling.

METHODS

We examined 10 healthy subjects with a Vicon MX System (6 Cameras). Additionally we performed an AMTI force platform, a Novel SF pressure measurement System and a MegaWin EMG system, synchronized with the Vicon system. The muscle forces were determined using a slightly modified model from the repository AMMRV1.1 of ANYBODY TECHNOLOGY (Vaughan). Postprocessing with the EMG data was done. A modified finite element model from the Website ww.ulb.ac.be/project/vakhum/ which is based on CT Scan was used. Contacts between bones were considered with 3 different methods. 1. as bonded, 2. as joints with less friction and 3. as frictional contact with additional ligaments because of a underconstrained model. We were firstly interested in reaction forces/moments and not in stress and strain. Therefore we did a quasi static analysis computing the reaction forces/moments in every 10% of stance phase in each joint. The foot was fixed at the end of the tibia and the fibula. The position of the foot segments in the FE model were adapted due to the kinematic measurements. The reaction forces we measured with a force plate and an EMED SF as well as a PEDAR System. As a novel method to get additional boundary condition we used the model of Vaughan from the ANYBODY muscle modelling Repository.

RESULTS

Reaction forces and moments were with all 3 methods comparable, opposite to stress and strain which showed huge differences, especially in contact regions. Reaction forces and moments in the ankle joint computed with our FE model fit very well with results we get from the ANYBODY model. In other joints such as TMT and MTP joints we get similar results as for example Jacob[4] described.

DISCUSSSION & CONCLUSIONS

The above shown method to get the boundary conditions for an FE Model from gait analysis and ANYBODY modelling is probably a valuable method for clinicians in the future. The method seems to be satisfactory to get reaction forces and moments.

There are further investigations necessary, to simulate stress and strain. The main problem here seems to be an underconstrained model using frictionless contacts. Main reason for that are the unbalanced muscle forces from the ANYBODY modelling system in combination with the applicated geometry of the foot. Other possibilities to avoid a underconstrained model such as the introduction of ligaments or weak springs even though frictional contact is in our understanding not a satisfactory solution.

Figure 1: Mesh of the FE-Model

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RELIABILITY OF IN-SOLE PLANTAR PRESSURE USING SIMPLE AND DETAILED MASKS OF THE FOREFOOT, MIDFOOT, AND HINDFOOT

Jonathan S. Akins, Karen A. Keenan, Brian P. Dugan, Melanie Francis. Francis, John P. Abt, Timothy C. Sell, Scott M. Lephart

Neuromuscular Research Laboratory, University of Pittsburgh, PA

BACKGROUND

Plantar pressure measurements obtained with in-shoe pressure sensor insoles can be analyzed by dividing the foot into anatomical regions using masks. A simple mask could divide the foot into four regions: toes, forefoot, midfoot, and hindfoot. A detailed mask could further divide these regions to capture data from specific anatomical regions/structures. For example, dividing the forefoot into individual metatarsals may help clinicians and researchers indentify individuals who are at increased risk of stress-related injuries to these structures (e.g. stress fracture). Regardless of which mask is applied, these measures must be reliable. The purpose of this study was to obtain the reliability of in-sole plantar pressure measurements using simple and detailed masks and to determine which is more reliable.

METHODS

Ten healthy males (n=8) and females (n=2) participated in this study (age: 27.7 ± 4.1 years, mass: 77.6 ± 10.7 kg, height: 174.3 ± 7.0 cm). Subjects performed a gait task on two different days using the PEDAR-X® system (Novel GmbH, Munich, Germany) to measure in-sole plantar pressures (100Hz) using their own athletic footwear (without socks). After familiarization of the task, subjects performed three trials of 20 consecutive straight-line walking steps using usual gait across a level, laboratory floor.

Maximum force (MF), force-time integral (FTI), peak pressure (PP), pressure-time integral (PTI), and maximum mean pressure (MMP) were calculated for simple and detailed masks. Both masks divided the foot into three primary regions: the forefoot, midfoot, and hindfoot. The simple mask divided the forefoot into metatarsal 1, 2, and 3-5; and did not divide the midfoot and hindfoot. The detailed mask divided the forefoot into individual metatarsals and the midfoot and hindfoot into medial/lateral regions. Intraclass correlation coefficients (ICC) were calculated using a two-way random effects model (ICC [2,k]). ICCs were averaged for each variable to obtain a single ICC value for the simple and detailed masks. Left and right foot measurements were combined for the analyses. A t-test was used to identify significant differences between the simple and detailed masks (α=0.05).

RESULTS

No significant differences were found between the simple and detailed masks. Maximum force and force-time-integral resulted in excellent reliability (ICC>0.90) for all masks and regions. Peak pressure, pressure-time-integral, and maximum mean pressure resulted in good reliability (ICC>0.70) for all masks and regions.

Table 1: ICCs (mean ± SD) for simple and detailed masks of the forefoot, midfoot, and hindfoot

	Forefoot		
	Simple	Detailed	p-value
MF	0.962 ± 0.016	0.966 ± 0.014	0.653
FTI	0.932 ± 0.027	0.942 ± 0.024	0.456
PP	0.849 ± 0.164	0.891 ± 0.136	0.586
PTI	0.845 ± 0.078	0.891 ± 0.083	0.292
MMP	0.929 ± 0.038	0.943 ± 0.035	0.454
	Midfoot		
	Simple	Detailed	p-value
MF	0.975 ± 0.003	0.974 ± 0.010	0.916
FTI	0.955 ± 0.013	0.955 ± 0.019	0.972
PP	0.938 ± 0.044	0.944 ± 0.023	0.818
PTI	0.852 ± 0.108	0.887 ± 0.035	0.549
MMP	0.852 ± 0.055	0.751 ± 0.237	0.604
	Hindfoot		
	Simple	Detailed	p-value
MF	0.915 ± 0.018	0.911 ± 0.041	0.913
FTI	0.959 ± 0.023	0.952 ± 0.020	0.720
PP	0.870 ± 0.021	0.861 ± 0.047	0.799
PTI	0.908 ± 0.030	0.911 ± 0.024	0.912
MMP	0.890 ± 0.021	0.877 ± 0.047	0.731

DISCUSSION

In-sole plantar pressure measurements obtained demonstrated good reliability for all masks and regions indicating that the PEDAR-X® system is capable of collecting reliable data, regardless of which mask is utilized. These findings are similar those of previous studies (Boyd *et al*, 1997; Putti el al, 2007). The lowest ICC was for MMP using the detailed mask for the midfoot, which may be a function of the small contact area of the medial midfoot. Since the contact area is relatively small, it is possible that even small variation in the rollover pattern during gait may significantly affect measurements in this region. It is therefore recommended to not subdivide this region in order to maximize the reliability of the measurements obtained, both in the clinical and research settings.

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THE INFLUENCE OF FATIGUE, LIGAMENT LAXITY AND HORMONAL FLUCTUATION ON PLANTAR PRESSURE IN FEMALE ATHLETES

Roshna E. Wunderlich (1), Kristina T. Silke (1), Sarah F. Cebulski (2)

1. Department of Biology, James Madison University, Virginia
2. Department of Biology, Carleton University, Ottawa

INTRODUCTION

Overuse injuries of the lower extremity compose over 20% of injuries in collegiate athletes, up to 50% of injuries in pediatric athletics (Arendt, 2003, Bennell, *et al.*, 1996, Dalton, 1992) and a considerable portion of injuries in military recruits (Milgrom, *et al.*, 1985, Rauh, *et al.*, 2006). Like some traumatic injuries (acl tears) overuse/stress injuries are significantly more frequently observed in women (Taunton, *et al.*, 2002). The etiology of sex differences in overuse injuries has been associated with various factors including training, anatomical differences, hormonal differences, and joint laxity. Joint laxity has also been associated with fluctuating hormone levels. Specifically, estrogen has been associated with suppression of collagen synthesis and decreased ligament cross-sectional area (Abubaker, *et al.*, 1996, Booth, *et al.*, 1970, Liu, *et al.*, 1997, Samuel, *et al.*, 1996). As collagen serves to provide stiffness to biological tissues such as ligaments and tendons, its breakdown can play a key role in enhanced joint laxity. In this study we quantify hormone levels, ligament laxity, and plantar pressure distribution to investigate the extent to which hormone fluctuation and fatigue interact to influence plantar pressure distribution and lower limb load.

METHODS

Twenty-four female collegiate athletes were tested, 8 for a four-week protocol and 16 for a twelve-week protocol. Blood was drawn each visit and subsequently assayed for estradiol content using an ELISA kit. Ankle ligament laxity was measured using an ankle arthrometer, and subjects walked, ran and cut barefoot over an EMED-SF plantar pressure platform. Following a fatigue test (stair running followed by a modified hurdle beep test), lactate levels, ankle laxity and plantar pressure were re-measured. Repeated measures ANOVA was used to compare ligament laxity and plantar pressure in 7 foot regions across menstrual weeks and fatigue states.

RESULTS

Ankle ligament laxity increased following fatigue ($p<0.05$). There was also significant variation across subjects ($p<0.01$). Females had more ligament laxity than males ($p<0.01$), and there was an interaction between gender and fatigue such that females experienced a greater increase in ligament laxity post-fatigue than males. There was a significant interaction between menstrual week and fatigue such that the effect of fatigue was significantly greater in week 4 and lower in week 2 than any other menstrual week. While hormone levels may not themselves affect ligament laxity, the interaction of hormone level and fatigue may influence ligament laxity.

Peak plantar pressures exhibit variation across foot region, menstrual week and fatigue condition. Peak pressures in metatarsals 2-5 are significantly ($p<0.01$) lower post-fatigue than pre-fatigue. Peak pressures in the medial and lateral midfoot, however, vary based on menstrual week and fatigue condition. In weeks 1 and 2 post-fatigue midfoot peak pressure are higher post-fatigue than pre-fatigue ($p<0.05$), while post-fatigue pressures are slightly lower than pre-fatigue pressures in weeks 3 and 4. Week 2 is also the week of greatest ligament laxity, suggesting foot ligament laxity is allowing some midfoot collapse, especially post-fatigue when the plantarflexors are fatigued and providing less load resistance (Sharkey, *et al.*, 1999, Weist, *et al.*, 2004). This laxity may also reduce load transmission to the metatarsal heads.

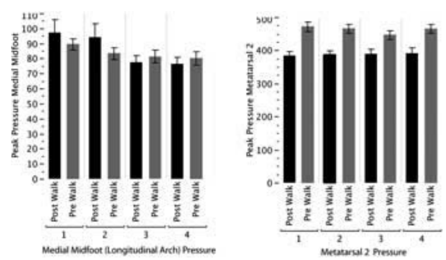

Figure 1. Peak pressures in medial midfoot and metatarsal 2 across weeks and fatigue state.

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COMPARISON OF PLANTAR PRESSURE MEASUREMENTS OBTAINED DURING BAREFOOT AND SHOD CONDITIONS

Jonathan S. Akins, Karen A. Keenan, Brian P. Dugan, Melanie Francis, John P. Abt, Timothy C. Sell, Scott M. Lephart

Neuromuscular Research Laboratory, University of Pittsburgh, PA

BACKGROUND

Pedobarographic platforms allow for assessment of plantar pressures at the foot-ground interface whereas pedobaragraphic insoles allow for assessment at the foot-shoe interface. It has been suggested that barefoot assessments may be more sensitive in detecting risk factors for the development of exercise related lower leg pain than shod assessments (Willems, 2007). Plantar pressures may be altered or attenuated by footwear, thereby making detection of small but significant variations in plantar pressure measurements more difficult in shod conditions. In addition, barefoot assessments require less subject preparation, may allow for data to be collected on more subjects in a shorter period of time, and require a smaller laboratory space, thereby potentially simplifying the data collection process. The purpose of this study was to compare plantar pressure measurements obtained while barefoot to those obtained while shod.

METHODS

Ten healthy males (n=8) and females (n=2) participated in this study (age: 27.7 ± 4.1 years, mass: 77.6 ± 10.7 kg, height: 174.3 ± 7.0 cm). Data were collected on two different days in both barefoot and shod conditions. Barefoot measurements were obtained with the EMED-X® (Novel GmbH, Munich, Germany), sampling at 100Hz. A two-step approach at a self-selected speed was utilized for all trials. After familiarization of the task, subjects performed 10 right-foot trials.

Shod measurements were obtained with the PEDAR-X® system (Novel GmbH, Munich, Germany), sampling at 100Hz. Subjects used their own athletic footwear, without socks, and were asked to practice straight-line walking at their "usual" walking speed. After familiarization, three trials of 20 consecutive straight-line steps across a level tiled floor were recorded.

For the barefoot trials, average maximum force (MF) and peak pressure (PP) were calculated for all trials. For the shod trials, the first and last two right steps were removed and MF and PP were calculated for the remaining right-foot steps. A similar mask was applied to both barefoot and shod trials with the following regions: medial heel, lateral heel, midfoot, each metatarsal (1-5), great toe, toe 2, and toes345. A t-test compared MF and PP obtained in the barefoot (EMED) and shod (PEDAR) conditions (α=0.05).

RESULTS

PEDAR and EMED measurements differed significantly in several regions (Table 1).

Region	PEDAR	EMED	
	Maximum Force		
GT	110.1 ± 33.2	126.1 ± 64.7	
T2	60.4 ± 18.3	29.4 ± 15.9	*
T345	75.6 ± 24.7	25.9 ± 20.0	*
MT1	145.6 ± 54.0	161.1 ± 62.8	
MT2	126.6 ± 37.7	186.4 ± 33.4	*
MT3	101.9 ± 29.0	181.5 ± 36.0	*
MT4	73.7 ± 23.7	101.9 ± 37.6	*
MT5	49.8 ± 18.1	38.0 ± 22.8	
Midfoot	184.6 ± 66.4	139.9 ± 87.0	
Lat Hindfoot	276.0 ± 60.5	232.5 ± 48.1	
Med Hindfoot	288.6 ± 67.9	282.0 ± 49.1	
	Peak Pressure		
GT	194.9 ± 46.8	334.4 ± 179.3	*
T2	165.6 ± 44.3	205.1 ± 103.3	
T345	111.5 ± 37.3	105.3 ± 56.7	
MT1	192.2 ± 41.3	266.6 ± 116.1	
MT2	193.3 ± 43.1	554.4 ± 278.2	*
MT3	185.8 ± 42.8	391.9 ± 143.1	*
MT4	159.9 ± 43.7	246.4 ± 109.4	*
MT5	117.8 ± 36.2	176.9 ± 137.3	
Midfoot	121.4 ± 32.3	118.2 ± 44.3	
Lat Hindfoot	185.9 ± 30.9	367.3 ± 124.0	*
Med Hindfoot	192.1 ± 33.2	367.0 ± 97.8	*

* $p < 0.05$

Table 1: Maximum force and peak pressure for each region (mean ± SD)

DISCUSSION

Statistically different plantar pressure measurements were obtained in barefoot and shod conditions. Peak pressure appears to be significantly attenuated in the hindfoot, MT2 through MT4, and the great toe in the shod condition. Maximum force was significantly attenuated in MT2 through MT4, T2 and T345. If the variable of interest is peak pressure, then barefoot assessments may be preferred. However, since most athletic activities are performed while shod and the results were inconsistent, it may be more appropriate to assess plantar pressure in the shod condition.

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FIRST RAY INSTABILITY IN HALLUX VALGUS DEFORMATY - A KINEMATIC RADIOGRAPHIC AND PEDOBAROGRAPHIC ANALYSIS

Dietze Albrecht[1], Ulf Bahlke[2], Heiner Martin[2], Mittlmeier Thomas[1]

Chirurgische Universitätsklinik Rostock, Abteilung für Unfall- und Wiederherstellungs-chirurgie[1], Rostock, Germany
Institut für Biomechanik Universität Rostock[2]

INTRODUCTION

Manifest hallux valgus deformity remains an abundant fore foot condition, whose entire or definite cause remains controversial. The extent of first ray instability and in particular of the first tarso-metatarsal (TMT-I) joint of the affected foot remains a key argument in the debate for optimal surgical treatment in hallux valgus conditions (Coughlin 2003). Numerous attempts have been undertaken to more precisely investigate and quantify first ray instability. Klaue (1994) designed an apparatus to evaluate static TMT-I instability and combined a clinical study with a cadaver radiograph analysis for validation of the device. Glasoe et al. (2000) modified the Rodgers apparatus in order to respect dynamic stabilization aspects of the foot. Additionally in the pedobarographic foot print of hallux valgus patients a deformity related reduced weight bearing of the first ray has been described (Lorei, 2006).

In our study, pedobarographic findings were correlated to the radiographically determined degree of first ray mobility of the foot.

PATIENTS AND METHOD

For our study 8 patients presenting a hallux valgus deformity on one foot or both feet, who were registered for corrective foot surgery out with our hospital, were enrolled. Seven were female one was male. Mean age was 44 (range 15-65). Clinical symptoms like foot pain, AAOF-Score and hallux valgus angle and intermetarsal angle were recorded. For simultaneous radiographic and pedobarographic measurement a mobile C-arm fluoroscope (ziehm, Nuremburg) and a plantar pedobarographic measurement platform (emed novel, Munich) were combined in an experimental setting to allow dynamic gait analysis in the entire stance phase of walking (including heel set down and fore foot take off). Pulsing radiograph recording (frequency 4 pulses/s) allowed digital analysis in selected frames and determination of motion among the joints of the first ray of the foot in particular the TMT-I joint (CAD program Solid Works, 2007). Simultaneously the pedobarographic footprint was recorded by the platform and peak pressure as well as maximum force were analyzed. Statistical analysis of hallux valgus radiological characteristics and pedographic findings of the first ray was by Pearson´s correlation.

RESULTS

All enrolled patients presented a manifest hallux valgus deformity and met requirements for operative treatment by an experienced foot surgeon blinded to the study set up. The mean intermetatarsal angle was 15°(range 8°-28°) and the mean hallux valgus angle was 45° (range 12°-80°). A large intermetatarsal angle correlated significantly with an increase of peak pressure under the first metatarsal (p=0,014). Further-more the radiological computer-assisted determined maximal dorsiflexion of the first ray (from talus till metatarsal head) significantly correlated with the extent of the intermetatarsal angle (p=0,012). Mobility of the first tarso-metatarsal (TMT-I) joint however showed a significant correlation to an increase of maximum force of the mid-forefoot region.

DISCUSSION

Mobility of the first metatarsal ray and in particular of the first tarso-metatarsal joint has been the subject of extensive research. Various devices had been developed to quantify first ray mobility of the foot. As in our study first ray deviation was mainly analyzed in the sagital plane. In our study we can support the notion of an enlarged intermetatarsal angle being associated with an increased dorsiflexion of the first ray of the foot during gait. Further more we found instability of the TMT-I –joint to have an increasing affect on mid-forefoot maximum force that may be relevant for pathological correlates.

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IS HALLUX VALGUS ASSOCIATED WITH DIFFERENT PEAK PLANTAR PRESSURE AND PRESSURE-TIME INTEGRALS?

Andrew M. Galica (1), Alyssa B. Dufour (1, 4), Howard J. Hillstrom (2), Mark W. Lenhoff (2), Jocelyn C. Frey (2), Virginia A. Casey (1), Marian T. Hannan (1, 3)

1. Institute for Aging Research, Hebrew SeniorLife, Boston, MA, USA
2. Hospital for Special Surgery, New York City, NY, USA
3. Harvard Medical School, Boston, MA, USA
4. School of Public Health, Boston University, Boston, MA USA

INTRODUCTION

While there are many clinical and case reported studies of hallux valgus (HV), the etiology and biomechanics of the pathology remain poorly understood. Although previous studies have noted differences in peak pressures in various regions of the foot between individuals with and without HV (Kernozek 2003, Yamamoto 1996, Yavuz 2009), results are inconsistent and have not been confirmed in larger studies. The purpose of this research is to describe peak plantar pressures and pressure-time integrals in an epidemiological population-based study and to investigate whether these measures differ between those with and without HV as defined by standardized foot examinations. We hypothesize that it is possible to distinguish individuals with and without HV based on differences in peak pressure and pressure-time integral measures.

METHODS

Data were obtained from a subset of participants enrolled in the Framingham Heart Study (N = 464; 57% female; mean age, 65 years; mean BMI, 28). Between 2002 and 2005, plantar pressure values were collected using a Tekscan Matscan system (model 3150, resolution of 1.4 sensels/cm^2) while participants walked at a comfortable pace barefoot across the mat. Data was imported into Novel software and masked into 12 segments (toes, submetatarsal heads 1-5, medial arch and heel, lateral arch and heel). Peak pressure and pressure time integral values were then calculated. Two-tailed Students T-tests assessed differences between those with and without HV.

TABLES

Table 1: Peak Pressure Differences

Measure	Peak Pressure, N/cm^2		
	No Hallux Valgus	Hallux Valgus	P Value
FEMALE	n = 172	n = 87	
Hallux	18.5 ± 6.1	18.4 ± 6.1	0.84
Submetatarsal head 1	17.7 ± 5.8	17.1 ± 5.8	0.46
Submetatarsal head 2	22.2 ± 4.8	21.9 ± 5.1	0.67
MALE	n = 155	n = 45	
Hallux	19.3 ± 7.0	18.0 ± 7.3	0.27
Submetatarsal head 1	18.9 ± 5.9	19.3 ± 7.6	0.80
Submetatarsal head 2	23.0 ± 3.9	22.8 ± 5.2	0.82

Table 2: Pressure Time Integral Differences

Measure	Pressure Time Integral, (N/cm^2)×sec		
	No Hallux Valgus	Hallux Valgus	P Value
FEMALE	n = 171	n = 87	
Hallux	7.9 ± 3.9	8.0 ± 4.6	0.90
Submetatarsal head 1	8.4 ± 3.7	8.0 ± 4.2	0.50
Submetatarsal head 2	10.7 ± 3.1	10.9 ± 5.1	0.72
MALE	n = 155	n = 45	
Hallux	7.7 ± 4.5	8.2 ± 5.8	0.61
Submetatarsal head 1	8.7 ± 3.7	10.5 ± 7.9	0.15
Submetatarsal head 2	10.9 ± 3.1	11.9 ± 7.0	0.38

RESULTS

Preliminary analysis revealed significant sex differences in plantar pressures. Therefore, subsequent analyses were stratified by gender. Since the right foot is considered more dominant, only these results are reported. Tables 1 and 2 present a subset of the 12 analyzed foot areas. Female subjects with HV exhibited greater peak pressure values under the lateral arch than those without HV ($p = 0.046$). No significant differences were found in pressure-time integrals for females with and without HV, regardless of foot area. Male subjects with HV exhibited greater peak pressure (p=0.007) and pressure-time integral (p=0.015) values than those without HV under groupings of the 3rd, 4th, and 5th toes.

DISCUSSION

Significant gender differences in peak pressures and pressure time integrals were noted. The data did not support our hypothesis as analyses did not distinguish between those with and without HV based on pressure-related measures. It is possible that the accuracy and resolution of Tekscan, in comparison to other foot mat systems, may have affected our results. It is also possible, since both groups contained concomitant pathologies, the plantar loading effects for HV washed out. Future work may investigate whether consideration of additional foot disorders or deformities may better use the plantar pressure measures to distinguish foot pathologies.

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COMPARISON OF GEOMETRY HALLUX ANGLES WITH RADIOGRAPHIC HALLUX ABDUCTUS ANGLES FOR PREDICTING HALLUX ABDUCTOVALGUS

James A. Furmato, Jinsup Song

Gait Study Center, Temple University School of Podiatric Medicine, Philadelphia, PA

INTRODUCTION

Several criteria determine the presence and severity of the bunion deformity (hallux abductovalgus; HAV), a common and disabling foot disorder. Radiographic relationships in the weight-bearing dorso-plantar view include the intermetatarsal angle, tibial sesamoid position, and hallux abductus angle (HAA). The HAA determines the amount of hallux deviation in HAV. While the radiographic technique is well accepted, it requires a small dose of radiation. Evaluating foot structure without ionizing radiation may be clinically useful.

The emed™ Geometry software (novel GmbH, Munich) calculates two hallux angles based on dynamic footprint: HA1 based on the relationship of the medial hallux to the medial foot border and HA2 relating the center of the Hallux to the estimated bisection of the angle between the first and second metatarsals, see Figure 1(left) and 1 (center). Alternatively, investigators wanted to examine the utility a graphical analog of HA1, constructed from a scan of the plantar weight bearing foot (MVIa), see Figure 1 (right).[1] However, there has been no comparative study between radiographic versus the previously mentioned alternates.

Investigators examined the utility of a using the angles form the Geometry software or a graphical analog of HA1 or HA2 based on a flatbed scan of the plantar weight bearing foot collected to determine the Malleolar Valgus Index (MVI)[1]. When HAV is present, the Hallux Abductus angle is greater than 15 degrees[2].

METHODS

This is a retrospective study. Twenty eight (28) feet with clinical diagnosis of HAV and 10 feet without were included. Clinical assessment of HAV, radiological measure of HAA, MVI foot scans and EMED-X dynamic footprints were available for each subject. Bivariate linear fit of HA1, HA2 and MVIa against HAA was evaluated using JMP 8 (SAS Institute, Cary, NC)

RESULTS

Correlation coefficients and root mean square errors of HA1, HA2, and MVIa to corresponding HAA measurements were calculated (0.707 and 4.17, 0.334 and 14.8, and 0.822 and 6.17, respectively (all p-values <0.01). Descriptive statistics are shown in Table 1. However, none of parameters was able to distinguish between those feet with HAV and those without when one way Analysis of Variance was performed .

Figure 1: Radiographic angle (left), emed Geometry angles (center), HA1 MVI analog (right).

Table 1: Mean and standard deviation (parentheses) of four measurements with (HAV+; n=28) and without (HAV-; n=10) clinical finding of hallux abductovalgus.

	HAA	HA1	HA2	MVIa
HAV+	25.8 (7.2)	16.1 (5.8)	28.6 (18.0)	19.2 (9.0)
HAV-	18.0 (8.9)	13.1 (5.7)	27.9 (15.4)	13.2 (9.2)

CONCLUSIONS

When a plantar weight-bearing scan of the foot is available, a reasonable estimate of the presence of bunion can be determined based on the relationship of the medial hallux to the medial foot border. However, none of these variable by itself was able discern feet with HAV when evaluated via univariate analysis. Additional work is needed.

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PLANTAR LOADING IN THE CAVUS FOOT

Andrew P. Kraszewski[1], Betty Chow[1], Sherry I. Backus[1], Jonathan T. Deland[1], Philip H. Demp[2], Jinsup Song[2], Benjamin Heilman[2], Sonali Rajan[1], Andrea Woodley[1], Howard J. Hillstrom[1,2]

1. Leon Root MD, Motion Analysis Laboratory, Hospital For Special Surgery, New York, NY
2. Temple University School of Podiatric Medicine, Temple University, Philadelphia, PA

INTRODUCTION

Data are available describing plantar loading planus and rectus foot types[1], but little is available describing the loading of the healthy cavus foot. The specific aim of this project was to compare plantar loading during gait for asymptomatic healthy individuals as a function of foot type (pes planus, rectus, and cavus).

METHODS

We hypothesized asymptomatic healthy individuals with pes planus, rectus, and pes cavus foot types will show differences in plantar loading. Sixty-one subjects were stratified according to the resting calcaneal stance position and the forefoot to rearfoot relationship into pes planus, rectus, and cavus groups. Plantar pressures were recorded with an Emed-X system (Novel, Munich, Germany) while subjects walked barefoot at their self-selected comfortable speed. Five steps per side were analyzed with Novel masking software. Peak pressure (PP), maximum force (MF), pressure-time integral (PTI), force-time integral (FTI), and contact area were calculated over ten masked regions. Data were analyzed with a univariate mixed-effect analysis of variance (ANOVA) model, followed by Bonferroni post-hoc t tests where significance was found. Foot type and replication were modeled as fixed and random effects, respectively. ANOVA significance was set at $p \leq 0.05$; Bonferroni post-hoc significance was set at $p \leq 0.017$.

RESULTS

Table 1 tabulates the results for PP and MF. Hallux PP and MF values were significantly higher for planus foot types. PP and MF beneath the 1st metatarsal head (MTH1) in planus feet were significantly lower than both rectus and cavus feet. The 5th MTH PP and MF values were significantly higher for rectus and cavus as compared to planus feet. PP and MF was lowest under the lateral arch for cavus feet. Cavus feet showed significantly higher PP at the lateral heel than rectus or planus. MF beneath the medial heel was significantly different between rectus and cavus feet. The planus medial arch had twice the contact area of rectus, which had twice the area of cavus.

Region	**Peak Pressure, N/cm²** Mean(SD)			
	Planus *N=27*	**Rectus** *N=22*	**Cavus** *N=12*	***P-hoc**
Hallux	43.9(6.9)	37.1(5.6)	32.5(5.8)	1,2
MTH1	28.1(8.0)	35.8(6.6)	37.3(6.8)	1,2
MTH2	51.1(5.8)	37.7(4.8)	38.8(4.9)	1,2
MTH3	40.4(4.3)	34.2(3.6)	34.8(3.6)	1,2
MTH4	27.8(3.6)	26(3.0)	25.6(3.1)	†
MTH5	20.1(5.8)	26.3(5.2)	25.3(5.3)	1,2
LatHeel	33.8(4.0)	33.6(3.3)	38.4(3.4)	1,3
MedHeel	37.2(4.8)	36.5(4.0)	38.8(4.1)	
LatArch	11.5(1.7)	10.9(1.4)	7.7(1.5)	1,3
MedArc	20.1(5.8)	26.3(5.2)	25.3(5.3)	1,3
	Maximum Force, N Mean(SD)			
Hallux	135(17.1)	108.8(14.1)	91.5(14.5)	1,2,3
MTH1	134.9(23.4)	148.2(19.2)	157.2(19.7)	1
MTH2	177.1(16.2)	151.1(13.3)	152.1(13.6)	1,2
MTH3	172.6(17.8)	150.7(14.6)	152.3(14.9)	1,2
MTH4	104.6(14.9)	98.9(12.3)	100.8(12.6)	
MTH5	41(10.7)	51.6(8.8)	58.4(9.0)	1,2
LatHeel	221.2(19.4)	215(15.9)	207.1(16.3)	
MedHeel	265.4(21.6)	255.6(17.7)	241.7(18.2)	1
LatArch	107(25.8)	79.7(20.0)	52.7(20.5)	1,2,3
MedArc	26.8(7.7)	13.7(6.3)	6.2(6.5)	1,2,3

*Table 1. Plantar loading measurements across foot types. *P-hoc: 1=Cavus vs. Planus; 2= Rectus vs. Planus; 3=Cavus vs. Rectus ; †blank cells indicate a non-significant ANOVA result*

DISCUSSION

In most plantar regions several measures of plantar loading were sensitive to foot type. The lower 1st MTH loads support the "hyper-mobile 1st-ray theory" where load is supported by the 2nd and 3rd MTH in many planus feet. PTI and FTI corroborate this observation. Higher 5th MTH cavus loads showed, as expected, differences to those of planus feet. These data will serve as a reference for future investigations of pedal pathology as related to foot types, and for designing and evaluating treatments.

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ACKNOWLEDGEMENTS

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FOOT STRUCTURE IS RELATED TO FOOT FUNCTION

Rajshree Mootanah (1), Jinsup Song (2), Sherry Backus (3), Jonathan Deland (3), Phillip Demp (2), Benjamin Heilman (3), Mark Lenhoff (3), Jocelyn Frey(3), Andrew Kraszewski (3), Howard J Hillstrom (3)

1. Anglia Ruskin University, Medical Engineering Research Group, Essex, UK.
2. Temple University, School of Podiatric medicine, PA, USA.
3. Hospital for Special Surgery, Rehabilitation Department, NY, USA.

INTRODUCTION

It has been suggested that foot structure may influence foot function (Song et al., 1996). The aim of this study is to investigate the relationship between foot structure and function as a predictive tool that could be used to plan more effective conservative and surgical treatments of pedal pathologies. We hypothesize that measures of foot structure (hindfoot alignment and arch height) are associated with biomechanical measures of foot function in asymptomatic healthy individuals.

METHODS

Foot structure was characterized by computing: (1) the maleolar valgus index (MVI) while standing, (2) arch height index (AHI) while sitting, and (3) AHI standing. MVI is a measure of static hindfoot alignment (Song et al, 1996). The subject's plantar foot is scanned while standing on a plexiglass platform over a flatbed scanner with a custom-made jig to register the lateral and medial malleolus. The deviation from the midpoint of the transmalleolar axis to the midpoint of the hindfoot, normalized to the foot width in this region, comprised the MVI. Note that AHI is the arch height at one-half of foot length normalized by the truncated foot length (Zifchock et al, 2006).

Foot function was characterized by calculating: (1) The center of pressure excursion index (CPEI - a measure of dynamic foot function), which is the lateral displacement of center of the pressure curve from the line constructed between the initial and the final center of pressure values, normalized by the foot width at the anterior one third of foot. (Song et al., 1996) The emed X system (Novel gmbh, Germany) and custom software, developed in C++, were employed to calculate the CPEI. Peak pressure (PP) and maximum force (MF) were calculated for the total plantar foot and each masked anatomical region using Novel software. Temporal-distance foot-fall parameters (e.g. step length, stride length, velocity, etc) were obtained with the GaitMatII (EQ systems, Glenside, PA). Each of 61 asymptomatic healthy adult test subjects walked at their comfortable self-selected speed across both the Emed-X and the GaitmatII systems to obtain the foot function data. Each subject also was structurally evaluated with MVI and AHI (sitting and standing).

RESULTS

Pearson correlation coefficients were calculated for each combination of structural and functional parameter for the entire cohort. The results are summarized in Table 1. MVI was significantly correlated with PP and MF at the hallux and negatively correlated with MF at the 1st MTPJ. AHI standing was correlated with PP and MF at the 1st MTPJ and negatively correlated with the PP at the 2nd MTPJ. AHI sitting was correlated with double support time and velocity. Step and stride lengths were negatively correlated with AHI sitting. AHI sitting was correlated with MF at the 5th MTPJ and negatively correlated with 2nd MTPJ PP and MF at the 1st MTPJ.

DISCUSSION

MVI was correlated with hallucial loading and negatively correlated with 1st MTPJ loading. This finding is consistent with the overpronation that accompanies valgus hindfeet and a hypermobile first ray. AHI was correlated with medial column loading as well as temporal-distance footfall parameters. Foot structure is correlated with foot function and one's basic gait pattern.

Table 1: correlation between foot structure & function

	AHI sit	AHI stand	MVI (%)
CPEI (%)			
Double support time	R=0.316, p=0.017		
Step Length	R=-0.354, p=0.007		
Stride Length	R=-0.340, p=0.010		
Velocity	R=0.339, p=0.010		
Peak Pressure-Hallux			R=0.380, p=0.004
Peak Pressure-1st MTPJ		R=0.320, p=0.015	
Peak Pressure-2nd MTPJ	R=-0.375, p=0.004	R=-0.316, p=0.017	
Maximum force Hallux			R=0.354, p=0.007
Maximum force 1st MTPJ	R=-0.344, p=0.009	R=0.265, p=0.046	R=-0.294, p=0.026
Maximum force 5th MTPJ	R=0.266, p=0.045		

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HUMAN WALKING: COMPARATIVE AND EVOLUTIONARY APPROACHES

Roshna E. Wunderlich

Department of Biology, James Madison University, Virginia

Human bipedal walking has a long and interesting evolutionary history. Fossilized feet and footprints have played a prominent role in the interpretation of the nature of bipedalism in early hominins. Recent discoveries of the fossil pedal remains of the Flores hominid (Figure 1), *Ardipithecus*, and ancient trackways have highlighted the diversity of early hominin bipeds and the mosaic of human and nonhuman skeletal features they possess. Proper interpretation of pedal function requires knowledge of form and function in living humans as well as our closest nonhuman primate relatives.

Figure 1: Late Pleistocene hominin foot from Flores, Indonesia (Jungers *et al*, 2009)

Human upright striding bipedal gait is unique among primates and is associated with a distinct foot roll-over pattern involving heel-strike, lateral midfoot pressure, lateral to medial pressure transfer across metatarsal heads, and a toe-off with high loads on metatarsals 1-2 and toe 1. Other primates exhibit higher medial midfoot pressures, higher lateral forefoot pressures, and lower toe pressure.

The human footfall pattern is associated with specialized anatomical features including broad calcanei and distinctly robust fifth metatarsals. The robust first metatarsal is broad dorsally, allowing a "close-packed" position during toe-off dorsiflexion rather than during grasping, and relatively short toes, decreasing mechanical work for toe flexors (Rolian *et al*, 2009).

Fossil hominids exhibit a unique combination of features. The Flores foot possesses a modern lateral column but a short hallux, long lateral toes and a weight bearing navicular (Figure 1). Our laboratory-based studies of functional anatomy in humans and other primates suggest that this hominin likely used a roll-over pattern resembling humans in early stance but similar to apes in later stance with medial midfoot weight bearing, a laterally-placed toe-off, and lack of full extension or high loads on the toes during toe-off (Figure 2). Analysis of fossil footprints provides additional insight into pedal function in early hominins when examined in the context of modern unshod human foot pressure. Taken together, paleontological and neontological studies of foot function provide deeper insights into modern human foot form, function, and pathology.

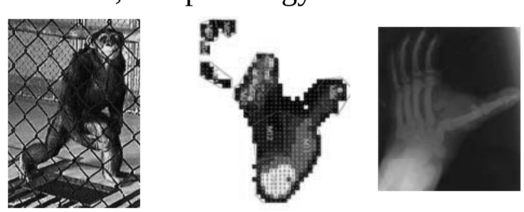

Figure 2: Chimpanzee walking bipedally Note the presence of pressure in the midfoot, the high pressures on metatarsals 2-3, and low toe pressures.

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EFFECT OF SHOE FLEXIBILITY ON PLANTAR LOADING IN CHILDREN WHO ARE LEARNING TO WALK

Howard J. Hillstrom (1,2), Melanie Buckland (1), Corinne McCarthy (1), David Scher (1), Leon Root (1), Sherry I. Backus (1), Andrew P. Kraszewski (1), Jocelyn Frey (1), Jinsup Song (2), Kendrick Whitney (2), Cherri Choate (2), Paul Scherer (3)

1. Leon Root MD, Motion Analysis Laboratory, Hospital For Special Surgery, New York, NY
2. Temple University School of Podiatric Medicine, Temple University, Philadelphia, PA
3. California College of Podiatric Medicine, Oakland, CA

INTRODUCTION

In a previous pilot study of ‘cruisers’ (non-independent ambulation), ‘early walkers’ (independent ambulation for 0 – 5 months), and ‘experienced walkers’ (independent ambulation for 6 – 12 months) developmental age significantly affected the children’s stability when walking and performing functional activities.[1] The purpose of this investigation is to examine how shoe structural characteristics affect plantar pressure distribution in early walkers.

METHODS

We hypothesized that torsional flexibility of children’s shoes affects plantar loading. Twenty-six children were evaluated in barefoot and each of four shoes that stratified a range of torsional flexibilities. The children were early walkers. Plantar pressures were recorded barefoot and shod with Emed-X and Pedar-X systems (Novel, Munich, Germany) respectively, at self-selected comfortable walking speeds. A minimum of five steps per side were analyzed using Novel masking software. Peak pressure (PP), maximum force (MF), pressure-time integral (PTI), force-time integral (FTI), and contact area were calculated over ten masked regions. Data were analyzed using a univariate mixed-effect ANOVA, followed by Bonferroni post-hoc t tests where significance was found. Footwear and replication were modeled as fixed and random effects, respectively. Significance was set at $p \leq 0.05$ for ANOVA and $p \leq 0.005$ for the post-hoc tests.

RESULTS

Peak pressure (Table 1, N/cm^2) for the right foot was significantly different across shoes for masked regions except the 5th MTPJ. In general, the stiffest shoe (ST) had the lowest PP while the most flexible shoe (UF) had the highest; ST and UF were also the most dissimilar and similar, respectively, to the barefoot pressures. Torsional flexibility (Figure 1, deg/Nm) showed a decreasing trend with increasing torsion angle across all footwear types. Highest flexibility was observed in the Ultraflex (pink line) footwear.

Peak Pressure (N/cm^2)	Barefoot (BF)	UltraFlex (UF)	MidFlex (MF)	LowFlex (LF)	Stiff (ST)	$p^{\dagger}$	*Post Hocs
Total	13.2(1.7)	13.3(2.2)	10.7(0.0)	11.8(1.9)	9.5(1.7)	**	2-5, 7,10
Hallux	11.2(1.5)	10.8(0.0)	10.8(0.0)	11.7(1.6)	8.3(0.0)	**	2,5, 8,10
1stMTPJ	8.1(1.1)	7.9(1.4)	7.5(0.0)	7.1(1.2)	6.2(0.0)	**	2-4, 7,9,10
2ndMTPJ	7.1(0.9)	8.1(1.1)	7.2(0.9)	7.4(1.0)	5.9(0.0)	**	1,4,5, 7,9,10
3rdMTPJ	6.0(0.6)	6.4(0.7)	5.3(0.0)	6.1(0.5)	5.5(0.0)	**	2,4,5, 7,9
4thMTPJ	4.9(0.8)	6.2(1.0)	5.8(0.0)	5.6(0.8)	5.6(0.0)	**	1,2,3, 4,7,9
5thMTPJ	3.9(0.8)	3.7(1.0)	3.5(0.9)	3.8(0.5)	3.9(1.0)	--	
LatHeel	8.2(1.2)	7.9(1.6)	6.0(0.0)	6.4(1.4)	5.2(0.0)	**	2-7, 9,10
MedHeel	8.9(1.4)	9.3(1.8)	6.5(0.0)	6.8(1.4)	5.3(0.0)	**	2-7, 9,10
LatArch	6.3(0.8)	7.2(0.9)	6.2(0.0)	6.7(0.8)	6.5(0.4)	**	1,5,8
MedArch	6.3(0.8)	7.1(1.0)	5.6(0.0)	5.8(0.8)	5.5(0.0)	**	1,2,4-6,7,9

*Table 1: Peak pressure values per plantar region across footwear type. *Post-hoc comparisons: 1=BFvsUF; 2=BFvsMF; 3=BFvsLF; 4=BFvsST; 5=UFvsMF; 6=UFvsLF; 7=UFvsST; 8=MFvsLF; 9= MFvsST; 10=LFvsST. †: --(p>0.05); **(p<0.001);*

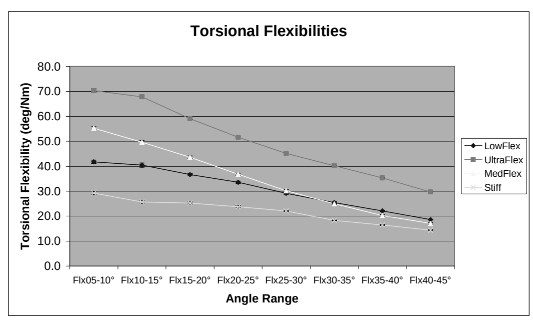

Figure 1: Plot of torsional flexibilities per shoe.

DISCUSSION

These data have implications for footwear design aiming to control plantar loading conditions that may also bear influence on a child’s proprioception when learning to walk.

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DYNAMIC PLANTAR PRESSURE CHANGES DURING LOADED GAIT

Stephen L. Goffar, Rett J. Reber, Bryan C. Christiansen, R. Benjamin Miller, Jacob A. Naylor, Brittany M. Rodriguez, Michael J. Walker, Deydre S. Teyhen

Doctoral Program in Physical Therapy, U.S. Army-Baylor University, Texas, USA

BACKGROUND

Lower extremity overuse injuries are reportedly the most common injuries in the military. Extreme values of static arch height (Williams, 2001) and the heavy loads commonly carried by military personnel in training and combat environments are associated with an increased risk of lower extremity overuse injuries (Bisiaux, 2008). While associations between static arch height and plantar pressure distributions have been demonstrated (Teyhen, 2009), limited knowledge exists regarding the impact of load carriage on plantar pressure distributions in the shod foot across arch types as delineated by AHI. The purpose of this study was to determine how load carriage affects dynamic plantar pressure distributions during gait in individuals with varying arch types.

METHODS

One hundred and fifteen healthy service members (97 males, 18 females, 31.3 ± 5.6 years, 177.1 ± 7.0 cm, 86.0 ± 11.0 kg) were enrolled in this study. They had no current medical condition which would preclude them from carrying up to a 40 kg load. Static measurements of heel to toe length (HTL) and arch height (AH) at 50% HTL were obtained with the Foot Assessment Platform System (FAPS) (McPoil, 2008). Arch Height Index (AHI) was calculated by dividing AH by HTL. Dynamic plantar pressure measurements were obtained using an in-shoe pressure measurement system (Pedar-x, Novel Electronics, Inc., St. Paul, MN, USA) while the subjects wore their own combat boots. Subjects walked for approximately 30 seconds at 3.0 mph on a treadmill under each of three levels of load: uniform without additional load (WB), 20 kg load including weapon, helmet, and body armor (20kg), 40 kg load adding weighted ruck sack (40kg). Load carriage sequence was counterbalanced.

DATA ANALYSIS

Participants were categorized by arch type based upon accepted cutoff values for AHI resulting in 28 high (AHI > .267), 61 normal, and 26 low (AHI < .229) arched right feet. An average of 9.8 ± .6 consecutive error-free steps were analyzed for each load condition. Maximum force (MaxF), force time integral (FTI), and pressure time integral (PTI), were calculated for regions of the plantar foot using a nine sector mask. Changes in each were analyzed with a 3×3 repeated measures ANOVA across the levels of load carriage and arch type.

RESULTS

There was a significant interaction between arch type and load for the MaxF (p=.001) and FTI (p≤.005) in the medial midfoot. Although MaxF and FTI increased in all regions of the foot with load (p<.001) regardless of foot type, the forces in the medial midfoot were greater in those with low arches.

There was a significant interaction between arch type and load for the MaxF (p=.004) in the medial forefoot. MaxF was greater in the high arched feet relative to normal and low arched feet (<.001) across all loads. The reverse was true at the great toe region, in which low and normal arched feet demonstrated greater MaxF (p ≤.004) compared to high arched feet.

The relative distribution of PTI in the nine regions of the plantar foot increased proportionately regardless of foot type under all load conditions.

DISCUSSION & CONCLUSIONS

Higher forces in the medial midfoot in low arched feet may be related to the increased surface area in this region or may represent increased pronation. However, the relative increases in medial midfoot forces in low arch feet did not increase disproportionately with increases in load compared to normal or high arched feet.

Force distributions in the 1st ray differed based on foot type. Those with high arched feet had greater forces in the medial forefoot region, while those with normal or low arched feet had greater forces in the great toe region, regardless of load. These differences in force distributions may demonstrate different strategies to generate a rigid lever during toe-off.

Regardless of foot type, increases in load did not alter the relative distribution of pressure over the plantar foot. These findings possibly indicate a negligible impact of loads ≤ 40kg on footwear and orthoses prescription. However, differences in dynamic plantar pressure during gait based on AHI were supported.

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'KIDFOOT MUENSTER' – NINE-YEAR RESULTS OF PLANTAR PRESSURE MEASUREMENTS IN CHILDRENS' FOOT DEVELOPMENT

Dieter Rosenbaum (1), Kerstin Bosch (1), Joachim Gerss (2)

(1) Movement Analysis Lab, Department of General Orthopaedics and Tumor Orthopaedic
(2) Dept. of Medical Informatics and Biomathematics, University Hospital Münster, Germany

INTRODUCTION

In 1999, the 'Kidfoot Münster' project was initiated. The main scope was to investigate the individual development of the child's growing foot by assessing foot loading patterns in a longitudinal project. Over 100 children were recruited for this purpose. In previous reports we described preliminary results (Bertsch 2004, Bosch 2007; 2009, Unger 2004). Now the first group of children has successfully completed their nine years of study involvement so that we can present the changes in foot loading characteristics of healthy children from the onset of independent walking to the end of their elementary school age.

MATERIAL & METHODS

As soon as the children were able to walk without support for several meters they were invited to the lab for participation in this long-term investigation. Between 1 and 10 years of age, they were asked to visit the lab on 17 occasions. Initial visits were every 3 months during the first year, twice a year until the age of 6 and final visits were once per year. By the end of 2009, complete data sets of 36 children were available.

Dynamic foot loading patterns were measured during free walking across a capacitive platform (emed ST or X, 4 sensors/cm^2). For each foot, 5 trials were recorded and stored in a database (Medical Professional 13.3.30, Novel GmbH Munich). Five regions of interest (H=Heel, MF=midfoot, FF=forefoot, HX=hallux, LT=lateral toes) were analyzed with standard pressure parameters (PP=peak pressure, MF=maximum force, CA=contact area, AI=arch index).

RESULTS

Average peak pressure values of the total foot increased over time from 141 kPa to 409 kPa. The highest values were initially located under the hallux but moved towards the heel and forefoot. The relative maximum force increased especially in the heel (71%) and forefoot (87%) but decreased in the midfoot by 63%. Relative contact area decreased markedly under the midfoot. The arch index gradually decreased by 44% up to an age of six to seven years and then leveled off. The range of values indicates pronounced inter-individual differences (Fig. 1).

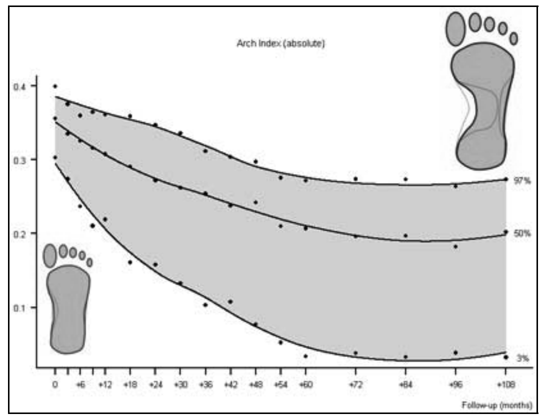

Fig. 1: Development of the arch index between the onset of walking and 9 years of walking experience (i.e. 1 to 10 years of age); 3rd, 50th & 97th percentile.

DISCUSSION

With the presented data we are able to describe – in more detail than before – the development of the child's healthy foot with respect to dynamic loading parameters. The results provide a range of normal values for the observed age range from one to ten years. These data may be used for clinical applications when potential pathologies in pediatric orthopedics shall be evaluated with comparable means.

While not overthrowing previous knowledge these data provide a more detailed insight into the individual foot development, the range of acceptable foot characteristics in certain ages, and finally the time course for the developmental changes.

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Contact: diro@uni-muenster.de, www.kidfoot.de

DYNAMIC FOOT LOADING PATTERNS IN CHILDREN WITH JUVENILE IDIOPATHIC ARTHRITIS (JIA)

Dieter Rosenbaum (1), Hartmut Michels (2), Matthias Hartmann (2)

(1) Movement Analysis Lab, Department of General Orthopaedics and Tumor Orthopaedics University Hospital Münster, Germany
(2) German Centre for Paediatric and Adolescent Rheumatology, Garmisch-

Introduction

Juvenile idiopathic arthritis (JIA is an auto-immune disease that may affect various joints leading to specific malpositions with compensatory movements. Foot involvement is frequently encountered (Spraul & Koenning). A pronounced pain stimulus causes the children to respond with a pain-relieving position thus causing muscular dysbalance. Children with oligoarthritis can partially compensate the joint problems in the neighbouring joints maintaining an asymmetric but usually fairly smooth gait pattern. In children with polyarticular arthritis the adjacent joints are also affected and compensation is hardly possible. Therefore, these patients may develop gait pattern changes. The present study evaluated characteristic gait patterns in these patients in order to evaluate the need for specific treatment.

Material & Methods

Forty-two children, 20 oligoarthritis patients (**OA**, 15 girls/5 boys; 11.0±3.5 years) and 22 polyarthritis patients (**PA**, 15 girls/7 boys; 14.2 ±3.6 years) were examined. Clinical and pedobarographic data were collected during inpatient stay. Every participant passed a clinical examination and the visual analogue scale was used to assess the current pain intensity. Plantar pressure measurements (emed ST 4, Novel; 50 Hz) were carried out with the instruction to walk normal at self-selected speed and recording a step in full gait. Five valid trials of each foot were stored for subsequent analyses. Dynamic foot loading parameters (**PP**=peak pressure, **FTI**=force time integral) were evaluated in ten plantar regions of interest. Forces were normalized to body mass. The averaged data of right and left foot were used for comparisons. The Mann-Whitney U-test was used for statistical analysis.

Results

While the maximum PP of the total foot did not show statistical differences, particularly higher PP were found under the hindfoot in PA children (Tab. 1). Furthermore, significantly higher PP in the first, second and third to fifth metatarsal head region were seen in PA children. The FTI showed higher values in the lateral and medial hindfoot and also in the third to fifth metatarsal in PA patients. The clinical findings indicated restricted dorsiflexion and plantarflexion in PA patients more frequently than in OA patients.

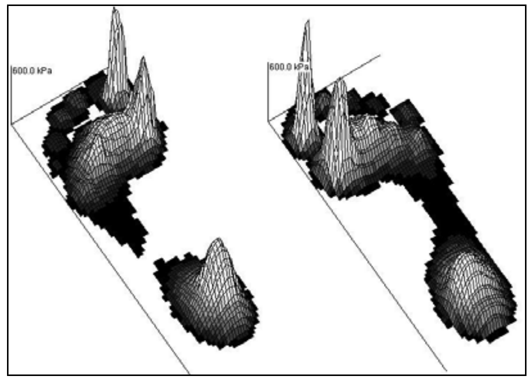

Fig. 1: Plantar pressure pattern of a selected PA patient #38 with elevated pressures under the first ray.

PP [kPa]	OA patients (n=20) Mean±SD	PA patients (n=22) Mean±SD	P-level
lat. heel	270±60	324±85	.016
med. heel	311±85	379±119	.037
MT-1	202±122	264±182	.049
MT-2	272±114	353±149	.022
MT-3-5	254±94	355±122	.008
FTI [Ns]			
lat. heel	29.9±14.0	41.4±18.4	.030
med. heel	36.8±15.2	50.8±22.6	.044
MT-3-5	49.7±20.9	68.3±33.2	.034

Tab. 1: Mean ± SD values of selected gait parameters of the two patient groups.

Discussion

The present clinical data in PA patients indicates changes in the loading response and terminal stance phase. This can be considered as a potential reason for higher PP and FTI values under the hindfoot and higher PP under the metatarsals in PA patients. Polyarticular arthritis may cause higher hindfoot and metatarsal loading as compared to oligoarthritis. This shows the need for preventive measures in these patients.

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acknowledgement

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Contact: diro@uni-muenster.de

PLANTAR FASCIITIS AND PAIN SYMPTOM ARE RELATED TO THE LONGITUDINAL ARCH SHAPE AND NOT TO THE PLANTAR PRESSURE DURING RUNNING

Ana Paula Ribeiro, Francis Trombini-Souza, Vitor Tessutti, Silvia M.A. João, Isabel C.N. Sacco

Physical Therapy, Speech and Occupational Therapy Dept., School of Medicine, University of São Paulo, Brazil.

INTRODUTION

Plantar fasciitis has been the third most common disease in runners, but its pathogenesis is still inconclusive. Changes in the longitudinal plantar arch and mechanical overload on the feet, have been described as risk factors for developing plantar fasciitis[1,2,3,4]. However, there are few studies investigating these factors during running in individuals with and without plantar fasciitis. Most of the literature investigated biomechanical parameters of the plantar fasciitis during gait and the results are still controversial, mainly because it is unclear the effect of pain associated with the disease over those parameters. Wearing *et al.* (2007)[3] discussed that symptomatic individuals make some adaptations during gait to reduce forces over the rearfoot and, consequently, increase loads over adjacent foot areas, such as over the mid and forefoot, as also observed by some authors[1,3]. According to biomechanical studies, it is possible that the plantar pressure distribution in individuals with plantar fasciitis would be different during the symptomatic and asymptomatic phases of the disease, and the pattern could be even more different during running. The purpose of this study was to investigate the association between plantar fasciitis and its pain symptom with the longitudinal plantar arch (LPA) shape and plantar pressure distribution during running.

METHODS

Ninety recreational runners were studied: 45 had plantar fasciitis (PF): symptomatic 30-SPF (45.4±8.1 yr, 69.6±14.0 kg, 1.68±9.2 m) and asymptomatic 15-APF (38.3±3.3 yr, 72.3±10.0 kg, 1.76±7.8 m) and 60 were healthy controls (CG) (35.0±9.0 yr, 66.8±12.0 kg, 1.71±9.0 m). Pain was assessed by a visual analogue scale (VAS). The LPA was evaluated by digital photogrammetry during weight bearing static posture[5]. The index between 0.22 – 0.25 was used to classified the LPA as normal; < 0.21 cavus; > 0.26 valgus[5]. The plantar pressure was evaluated by Pedar X system during running of 40 meters at a speed of 12±5% km/h. Runners used a common standardized sport footwear. Contact area, contact time and pressure peak were evaluated in 5 areas: rearfoot (medial and lateral), midfoot and forefoot (medial and lateral). Groups were compared using ANOVAs for repeated measures, followed by Tukey *post-hoc* tests ($p<0.05$).

RESULTS AND DISCUSSION

SPF group reported mean time since onset of pain of 7±2 months and pain levels of 5±2 cm. In APF and CG groups, the pain level was 0 cm. The LPA was more elevated in both groups with plantar fasciitis: symptomatic (0.17±0.08; cavus LPA; p=0.009) and asymptomatic (0.17±0.07; cavus LPA; p=0.008), compared to controls (0.22±0.05; normal LPA).

Table 1- Mean, standard deviation and comparison among groups with plantar fasciitis (symptomatic-SPF and asymptomatic-APF) and the control (CG) of plantar pressure variables during running.

Variable	*Group*	***Contact area*** (cm²)	***Contact Time*** (ms)	***Peak Pressure*** (kPa)
Rearfoot medial	SPF (1)	12.2±1.6	134.4±23.9	337±84.8
	APF (2)	11.4± 3.0	135.1±28.6	322±124.1
	CG (3)	12.5± 1.4	147.3±32.9	306.2±61
	p (tukey)	0.902 (1-2)	0.958 (1-2)	0.985 (1-2)
		0.998 (1-3)	0.998 (1-3)	0.588 (1-3)
		0.520 (2-3)	0.850 (2-3)	0.982 (2-3)
Rearfoot lateral	SPF (1)	10.4 ± 2.5	135.4± 36.2	346.1± 97.1
	APF (2)	9.8± 2.7	137.1± 46.0	291.5± 99.4
	CG (3)	10.8± 2.4	149.3± 38.8	331.1± 91.2
	p (tukey)	0.601 (1-2)	0.229 (1-2)	0.834 (1-2)
		0.809 (1-3)	0.085 (1-3)	0.892 (1-3)
		0.364 (2-3)	0.968 (2-3)	0.542 (2-3)
Midfoot	SPF (1)	39.5± 5.1	182.8± 37.1	129.0± 29.0
	APF (2)	38.3± 6.6	179.2± 38.2	106.7±20.9
	CG (3)	41.1± 5.4	198.0± 32.3	124.1± 30.6
	p (tukey)	0.109 (1-2)	0.822 (1-2)	0.998 (1-2)
		0.271 (1-3)	0.998 (1-3)	0.995 (1-3)
		0.202 (2-3)	0.916 (2-3)	0.991 (2-3)
Forefoot medial	SPF (1)	33.0± 2.6	207.6± 25.7	346.6±99.9
	APF (2)	32.1± 3.0	216.8± 28.1	312.4±99.2
	CG (3)	33.8± 2.6	217.6± 28.0	374.4± 96.4
	p (tukey)	0.883 (1-2)	0.461 (1-2)	0.818 (1-2)
		0.702 (1-3)	0.355 (1-3)	0.995 (1-3)
		0.883 (2-3)	0.493 (2-3)	0.884 (2-3)
Forefoot lateral	SPF (1)	37.0± 3.5	218.7± 25.5	284.3± 58.9
	APF (2)	36.5± 4.0	221.5± 27.9	242.4± 66.1
	CG (3)	37.8± 3.6	226.1± 26.4	266.5± 77.6
	p (tukey)	0.800 (1-2)	0.995 (1-2)	0.565 (1-2)
		0.734 (1-3)	0.619 (1-3)	0.924 (1-3)
		0.987 (2-3)	0.923 (2-3)	0.419 (2-3)

The cavus architecture of the LPA would lead to greater strain in the plantar fascia during static and, mostly, during dynamic activities, such as running. Chronically, these stresses will cause micro traumas in the plantar fascia and, probably, will lead to the progression of PF symptoms or even to the onset of PF. However, the plantar pressure distribution during running did not demonstrate association to the PF or its pain symptom.

CONCLUSION

The plantar fasciitis and the pain symptom are not associated to the plantar pressure distribution patterns during running. However, runners with plantar fasciitis with or without pain symptom present cavus LPA and more elevated arch compared to controls.

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FOOT PRESSURE DISTRIBUTION FOLLOWING OPERATIVE REDUCTION OF HIGH GRADE INTRA-ARTICULAR FRACTURES OF THE CALCANEUS

Moshe Ayalon (1), David Ben-Sira (1), Meir Nyska (2), Iftach Hetsroni (2)

1. The Zinman College of Physical Education, Wingate Institute, Israel
2. Meir Hospital, Sapir Medical Center, Kfar Saba, Israel

INTRODUCTION

Displaced intra-articular fractures of the calcaneus represent a complex injury typically affecting middle-aged active population. The ideal goal of open reduction with internal fixation in these fractures is to recreate the calcaneus width, height and length, the subtalar joint congruency, the soft tissue balancing, and consequently the foot and ankle kinematics. It is not known whether operative intervention for these injuries provide any advantage to non-operative management in terms of recreating the normal foot and ankle kinematics during the gait cycle. The current study was designed to assess the plantar pressure profiles of the foot following open reduction with internal fixation of high grade intra-articular fractures of the calcaneus.

METHODS

The study sample included 14 operated patients and 8 control healthy subjects. Plantar pressure distribution was collected during walking on a level floor at a natural preferred cadence with Pedar-x system (Novel gmbh, Munich). 10 zones of the plantar area of the foot were defined. The subjects walked using the same shoe model on a level floor at natural preferred cadence. Plantar pressure data were collected over 4 cycles. For each step the pressure time integral (PTI) [Ns/cm^2] was calculated for each zone. In the injured group, the sound limb was compared with the injured limb. In addition, the sound limb and the injured limb were separately compared with the control group (16 feet). Because some of the variables did not fulfill the assumptions of normality that underlie parametric statistics, a non-parametric approach was applied using the Mann-Whitney test for between samples comparisons. A p value of 0.05 was considered significant.

RESULTS

In eight of the ten masks PTI was significantly lower in the involved legs than in the healthy sample. PTI of the injured group was expressed as percentage of the healthy group. The relative means of the PTIIs of the involved legs were lesser in the medial aspect than the lateral aspect of the foot (MH-42% vs. LH-80%, MMF- 35% vs. LMF-103% ns, 1MTT-51% & 2MMT- 67% vs. LMTT-81% ns, 1TOE-23% & 2TOE-15% vs. LTOE-51%). Similar trends characterize the uninvolved leg. Thus, injured patients adopt a modified gait pattern typified by reduced plantar pressure, particularly on the medial aspect of the foot.

DISCUSION

Both limbs of operated patients demonstrated a similar PTI pattern of relative reduction, particularly in the medial aspect, compared to the healthy feet. The cause for this pattern remains unclear at this point. A potential explanation for the similar pattern of the injured and sound limbs is that over the years since the operative intervention and the rehabilitation process, patients may have implemented compensatory mechanisms in their non-operated limbs to prevent chronic over-loading of one side caused by gait asymmetry. The observed symmetry in gait adjustment may be related to the self-selected walking velocity chosen by each patient, and could indicate good functional recovery as related to walking. However, measuring at self selected gait velocities could potentially mask kinematical differences between operated and contra-lateral non-operated limbs that could emerge in higher locomotive velocities or during perturbations while running, walking on uneven terrains and side sloped surfaces.

PLANTAR PRESSURE DISTRIBUTION FOLLOWING OPERATIVE TREATMENT FOR PROXIMAL FEMUR FRACTURES

Brunk, M. ; Emmerich, J.; Mittlmeier; T

1. Dept. of Trauma and Reconstructive Surgery, University of Rostock, Germany

INTRODUCTION:

Hip fractures are regarded as the most common severe type of fall-related injury concerning elderly people and the most serious of the osteoporotic fractures because of their high morbidity, mortality and impairment in quality of life. (2-5).
Pertrochanteric and femoral neck fractures of the femur, with respect to healing, may be regarded as fracture-types with long rehabilitation time. (1)
Current rehabilitation protocols of patients following operative treatment recommend full weight-bearing on the operated extremity as early as possible. The present study seeks to highlight the reactions to early loading under the aspect of pain-control. In-shoe pressure redistribution to provide relief of postoperative mobilisation is based on assumed links between pressure and pain. However, little is known about pain-associated loading after operative treatment of hip fractures.

MATERIAL AND METHODS:

29 patients, who had an operative treatment of a fracture of the femoral neck or a pertrochanteric fracture, were allowed to full weight-bearing as tolerated on the injured limb. 12 man and 17 women, ranging in age from 45 to 96 years, took part in this study at our center. During postoperative mobilization elderly patients were allowed to practise full weight-bearing while loading of the injured limb was mostly limited due to pain in the early postoperative period.
Gait analysis was performed by using the novel PEDAR in-shoe plantar pressure measurement system (novel GmbH, Munich, Germany). Computerized gait-testing was performed at one, seven and twelve days postoperatively to quantify weight-bearing in association with pain.
Visual analog scale (VAS) score of pain were obtained from all subjects before and after testing with the PEDAR-system. Statistical analysis was used to analyze the relationship between the plantar pressure parameters and VAS scores in the period of hospitalisation. Pearson's correlation was applied to analyze the correlation between the changes in plantar pressure parameters and VAS scores. Statistical significance was set as $p < 0.05$.

RESULTS:

The average amount of weight these patients placed on the injured limb increased during the time of hospitalisation. During mobilisation maximum peak pressure (MPP), maximum pressure-time integral (PTI) increased at the seventh and twelfth day after surgery. In analogy, subjective pain scores decreased significantly during hospitalisation. 6 of 29 patients were measured at a fixed follow-up. For these 6 patients the increase in the PTI and MPP values were statistically correlated with the improvement in VAS scores ($r = 0.8$) in the course of follow up.
The average load supported by the injured limb was 70.4 (43.8-83.1) % of the uninjured limb after three days, and gradually increased to 91.1 (75.4-96.2) % at twelve days.

DISCUSSION:

This study investigated plantar pressure related to pain during gait in subjects at different time points after operative treatment of hip fractures. Plantar pressure came up to 40 to 90 % of the uninjured limb. Furthermore, a positive correlation of gait parameters with the improvement in VAS exists. Correspondingly, gait and balance disturbances caused by hip pain have an influence on dynamic plantar pressure distribution. Our subjects presented a worse load distribution pattern during gait at the beginning of hospitalisation overloading the midfoot and the rearfoot. These changes could be associated with pain pathogenesis that worsens the biomechanical condition of these patients and their clinical consequences

CONCLUSION:

Following operative treatment we recommend pain-associated full weight-bearing on the operated extremity as early as possible to avoid specific consequences on postural balance control; the reactions to early loading under the aspect of pain-control may provide a faster mobility after operative treatment.

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THE MĀORI FOOT: STATIC MORPHOLOGY AND DYNAMIC FUNCTION IN HEALTHY AND DIABETIC POPULATIONS

Jason Gurney (1), Celia Kuch (1), Dieter Rosenbaum (2), Uwe Kersting (3)

1. Department of Sport and Exercise Science, University of Auckland, New Zealand
2. Motion Analysis Lab, University Hospital Muenster, Germany
3. Center for Sensory-Motor Interaction, Aalborg University, Denmark

INTRODUCTION

Foot morphology and plantar loading may differ between ethnicities (Veves, 1995). Our previous work suggested altered loading strategies may exist between elite athletes of Māori and New Zealand Caucasian ('NZC') ethnicity (Gurney, 2009).

Māori suffer disproportionately from diabetes and particularly its complications, including foot ulceration (Joshy, 2006). If Māori exhibit altered foot morphology and function, perhaps different preventative strategies are required for reducing diabetic foot complications in Māori, such as footwear modifications.

Therefore the purpose of this study was to investigate the static morphology and dynamic function of the Māori foot in both healthy and diabetic populations compared to NZC controls. It was hypothesised that based on previous findings significant differences would be observed between ethnicities.

METHODS

A total of 40 participants were further divided into 10 Māori (8f; 42±11yrs; BMI 27±5) and 10 NZC (8f; 43±11yrs; BMI 26±4) who had no neuromusculoskeletal injury, plus 10 Māori (5f; 58±11yrs; BMI 32±5) and 10 NZC (5f; 60±10yrs; BMI 30±3) diagnosed with Type-2 diabetes.

Plantar pressure was evaluated using a Novel EMED-AT system (Novel GmBh, Munich, Germany) at a frequency of 50Hz during walking at a self-selected speed. Following familiarisation five trials were collected for each foot (Hughes et al., 1991).

Static morphology was measured using Harris mat techniques. Foot length, heel width, forefoot width, and arch index were determined. The latter was calculated as the width of the narrowest part of the midfoot divided by toeless foot length.

Plantar pressure data were analysed using Novel software, in which the foot was divided into 10 regions. Multiple pressure parameters were analysed.

Kolmogorov-Smirnov tests confirmed data normality. Independent T-tests were then used to test for significant differences between ethnic groups in both the healthy and diabetic participants (p = <0.05).

RESULTS

Arch index was found to be significantly greater in the healthy Māori compared to the healthy NZC, but this difference was not observed when comparing diabetic groups. No other static morphological differences were found between ethnicities.

Few significant differences were found between ethnicities, healthy or diabetic, in terms of dynamic plantar loading. Peak pressures under the central forefoot were found to be significantly greater in diabetic Māori compared to diabetic NZC (Fig. 1).

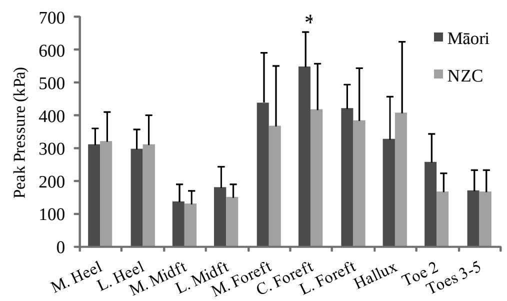

Figure 1: Peak pressure (kPa) data from Type-2 diabetic Māori and NZ Caucasians (=p<0.05).*

DISCUSSION AND CONCLUSIONS

The hypothesis that significant morphological and functional differences would be found between Māori and NZC feet was mostly proven incorrect. Since the compared groups were largely homogenous, we can conclude that Māori ethnicity alone may not affect foot morphology and function in healthy and diabetic populations. This may suggest that no special considerations are required when clinically treating or designing preventative care for diabetic Māori at risk of plantar ulceration. However further research is required on larger groups and non-urbanised Māori.

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ASYMMETRY IN PLANTAR LOADING DURING GAIT IN NATIVE AMERICANS WITH AND WITHOUT DIABETES AND WITH AND WITHOUT NEUROPATHY

Thomas W. Kernozek, Cassandra Heizler, John F. Greany

Health Professions Department, University of Wisconsin – La Crosse, WI USA

INTRODUCTION

The adjusted prevalence of diabetes among Native Americans has increased by 26.9%, from 6.7% to 8.5%, between 1996 and 2006 (Jernigan et al., 2010). On average, Native Americans are 2.3 times as likely to have diabetes as non-Hispanic whites of similar age (Burrows, 2000). Plantar ulceration is a common problem in individuals with diabetes. Among those with diabetes, 24% will require an amputation of the foot and/or leg (Lott et al., 2008).

Peak plantar pressure and pressure-time integral are two variables used to screen individuals with diabetes and neuropathy for the development of foot ulcers (Brown et al., 2004, Sauseng et al., 1999, Stess et al., 1997). The role asymmetry in plantar pressures has not investigated in Native Americans.

Our aim was to investigate plantar loading asymmetry during gait in Native Americans with no diabetes (ND), diabetes (D) and diabetes with neuropathy (D-NP).

METHODS

Ninety-eight Native American's volunteered to participate in the study (mean age 50.3 yrs; 19-86). Twenty three individuals had diabetes (D), 14 had diabetes with peripheral neuropathy (D-PN) and 61 did not have diabetes (ND). Neuropathy status was determined with a biothesiometer (> 25 V threshold). Twenty-four percent of participants were overweight (>25-29.9 BMI) and 59% were obese (>30 BMI).

Plantar pressure data were collected using an EMED-AT floor mounted capacitance based platform sampling at 50 Hz. The 2-step method was used to obtain plantar loading data from five trials of each foot. Each plantar loading trial was subdivided into ten specific plantar regions for analysis. Differences in the absolute value of peak pressure and peak pressure-time integral of feet were examined across metatarsal regions using an Analysis of Variance (ANOVA). Separate ANOVA's were run on peak pressure and pressure time integral for each of the metatarsal regions (alpha = 0.05).

RESULTS

Peak pressure asymmetry was different between feet across each the metatarsal regions 1-3 of the plantar surface during gait ($p < 0.05$). Table 1 depicts the asymmetry in pressure-time integrals between feet across metatarsal regions of the plantar surface ($p < 0.05$). Post hoc comparisons indicated that individuals without diabetes (ND) displayed less asymmetric loading described by peak pressure in metatarsal 1-3 regions and pressure-time integral across all metatarsal regions than those with diabetes (D) and diabetes with peripheral neuropathy (D-PN).

	MTH 1	MTH 2	MTH 3	MTH 4	MTH 5
ND	28.6 ± 31.4	27.1 ± 23.3	20.8 ± 17.4	15.4 ± 13.7	16.1 ± 13.5
D	49.5 ± 46.1	49.4 ± 50.6	33.7 ± 37.5	17.8 ± 11.9	21.2 ± 30.8
D-PN	58.8 ± 37.5	57.0 ± 60.2	41.9 ± 24.7	39.4 ± 40.1	37.3 ± 32.9

*Table 1: Mean (± Standard Deviation) Peak Pressure- time integral (kPA*s) for each group [Non-diabetic (ND), Diabetic (D) and Diabetic with Peripheral Neuropathy (D-PN)] for each Metatarsal Region (MTH1-MTH5).*

CONCLUSIONS

Native Americans with diabetes appeared to show greater asymmetry in plantar loading variables across the forefoot region compared to controls. Individuals with neuropathy and diabetes had the greatest amount of asymmetry with pressure time integral across these regions. Loading asymmetry may pose a role in the development of wounds in Native Americans with peripheral neuropathy and diabetes. Further research is warranted.

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T15

PLANTAR PRESSURE DISTRIBUTION PATTERNS IN MULTIPLE SCLEROSIS PATIENTS WITH DIFFERENT NEUROLOGICAL STATUS

Tatiana L. Tsvetkova (1), Igor D. Stolyarov (2), Olga L. Pichugina (2), Andrey M. Petrov (2), Alexander G. Ilves (2), Lidia N. Prakhova (2), Irina G. Nikiforova (2), Viatcheslav V. Lebedev (3), Fedor V. Yakushev (1), Vadim N. Tartakovskiy (1)

1. novel SPb, St.-Petersburg, Russia
2. Institute of Human Brain RAS, St.-Petersburg, Russia
3. novel, Munich, Germany

INTRODUCTION

Expanded Disability Status Scale (EDSS) is widely used to assess the disability of patients with multiple sclerosis (MS). The EDSS quantifies disability in eight functional systems (FS).

The aim of this study was to obtain the plantar pressure distribution images in MS patients with different neurological status (EDSS) and to estimate the plantar pressure distribution parameters changes.

METHODS

106 patients (33m/73f, age 39±10 years), diagnosed with relapsing-remitting MS (duration of MS 8±6 years) according to McDonald's criteria, were examined. Neurological status of examined patients was characterized with EDSS 3.4±1.2. Patients were divided into five groups in relation to estimated EDSS: EDSS [1,1.5] (no disability), EDSS [2,2.5] (minimal disability), EDSS [3,3.5] (disability in mild to moderate), EDSS [4,4.5] (severe disability), and EDSS [5,6.5] (increasing limitation in ability to walk, walking assistance is needed). All patients received permanent pathogenetic therapy.

Plantar pressure measurements were performed with emed-AT 25 system (novel, Munich, Germany). Five dynamic records of each foot were made with first step protocol. novel database medical was used to collect clinical and pressure measurement data. Peak pressure (PP), mean pressure (MP), maximum force (MF), pressure-time integrals (PTI), force-time integrals (FTI), contact time (CT) and arch index (AI) were calculated in novel-projects with novel automask for hindfoot, midfoot, five metatarsal heads (MH1-MH5), big toe, second toe and lateral toes. Parameters were calculated for each subject and for five groups. The difference in pressure distribution parameters under the foot areas was checked with ANOVA. Significance level was set as $p<0.001$.

RESULTS AND DISCUSSION

Each group had a specific plantar pressure distribution pattern that varied with a degree of disability status. The patients from Gr.2 (vs. Gr.1) were characterized with decreased MP under the hindfoot, increased MF and FTI under MH5 and FTI under MH4; increased CT under the foot and foot regions. Significantly decreased PP, MP, MF, FTI, and PTI under the hindfoot, MH4 and MH5, increased FTI, PTI, and contact time (% roll over process) under MH1, increased MF under the midfoot and AI were found in patients from Gr.3 (vs. Gr.2). Decreased PP, MF and FTI under MH2 were found in Gr.4 (vs. Gr.3). At the same time PP, MP, MF, PTI and FTI were significantly increased under MH4 and MH5 and decreased under MH1; MF and FTI were decreased under the midfoot together with AI and reached the values comparable with parameters from Gr.2. Significant decrease of PP, MP, MF under MH2, MH3, MH4 and midfoot, decreased AI, increased FTI and PTI under MH1 were found in Gr.5 (vs. Gr.4) in concert with significant increase of CT under the foot and all foot regions.

Medial shift of loading is comprehensible because the first ray is the most appropriate structure used in weight bearing to provide an additional security. Reduced parameters under the second and third metatarsal heads can be explained with the development of the transverse arch in the forefoot with an increase of spasticity. Lower loading of midfoot (with decreased arch index) is supposed to be because of an increase of longitudinal arch height. The hindfoot off-loading could be a sequence of the foot circumduction and inversion during the gait. First signs of significant changes of all parameters occurred in MS patients with moderate disability can be regarded as a result of disability in several function systems.

CONCLUSION

These results confirmed that patients with minimal disability have minimal gait impairments. Patients with mild and moderate disability reveal the changes in decreasing hindfoot and lateral forefoot loading with medial shift of loading. MS patients with severe disability showed decreased parameters in central forefoot. Patients with limited ability to walk have stable pathologically changed plantar pressure distribution pattern characterized with low values of all parameters and significantly decreased loading of hindfoot, midfoot, forefoot with medial shift of loading.

CAN GAIT INITIATION PROCESS BE EVALUATED WITH PRESSURE PLATFORMS?

Paula H. Lobo da Costa (1), Kerstin Bosch (2), Dieter Rosenbaum (2)

1. Department of Physical Education, Federal University of Sao Carlos, Sao Carlos, Brazil.
2. Movement Analysis Laboratory, Orthopaedic Department, University Hospital, Münster, Germany.

INTRODUCTION

The transition from standing posture to cyclic walking is a special challenge to be mastered by young children. It is the result of anticipatory postural adjustments before stepping. Studies on this matter with children have focused on force plate data and electromyographic responses (Stackhouse *et al.*, 2007; Wicart *et al.*, 2006) in clinical contexts. The aim of this study is to verify whether gait initiation can be measured with the EMED system in order to understand the relative contribution of foot structure in this process. Besides, as the medial longitudinal arch of young children is in progress, the role of the development of foot functions on this process could also be assessed.

METHODOLOGY

Antero-posterior and medio-lateral displacements of the center of pressure at the foot/floor interface and the Chippaux-Smirak Index (CSI) were measured during gait initiation in 20 Brazilian school children aged 3.7 (±0.7) years and with body mass of 17.3 (±2.3) kg. The dynamic data were obtained with the EMED-ST System (NOVEL, Germany), sampled at 50Hz and foot indexes were measured with foot prints in bipedal stance. Three to five valid trials of gait initiation were collected. The children stood still over the plate and after a "go" signal they performed the gait initiation and continued to walk for about 6 m. The walking speed was freely selected by each child and all were barefoot.

RESULTS & DISCUSSION

The CSI classified of 80% of the children as flat-footed (CSI ranged from 49% to 69% for right and left feet). The average displacements of the center of pressure in antero-posterior and medio-lateral directions relative to the plate system were analysed. The results of one child are presented (Fig.1 & Tab.1). At 70% of the stance phase the anticipatory behavior to the step begins. The range of COP_ML motion toward the stance foot is 9.2 (±1.44) cm, which represents 1.3 times the forefoot width, whereas the COP_AP range of motion is 12.6 (±0.,7) cm, equivalent to 0.80 times the foot length of this child. The maximum COP_ML and the minimum COP_AP motions represent the displacements toward the stepping foot and the backward direction respectively. It is interesting to note that these very small motions in the directions opposite to the step are insufficient to identify anticipatory behavior.

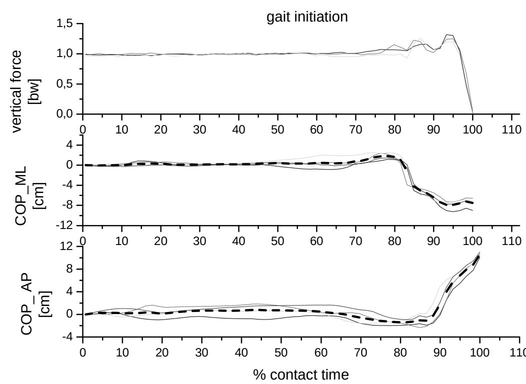

Fig. 1: Anterior-posterior (AP) and medio-lateral (ML) COP displacements and vertical force (body weight-BW) during gait initiation (one child, BR19, four trials).

GI	COP- AP [cm]			COP- ML [cm]		
	Min	max	range	min	max	range
Mean	-1.90	10.67	12.57	-7.70	1.50	9.2
SD	±0.62	±0.41	±0.72	±0.94	±1.01	±1,44

Tab. 1: Minimum, maximum and range of COP AP and ML displacements (one child, BR19, four trials).

Ledebt *et al.* (1998), on the other hand, found significant COP displacements for children of the same age. Since both feet were in contact with the platform, the system could not quantify the loading shift from one foot to the other as gait starts, but it was possible to measure the dynamic behavior of the COP before stepping, which may be of importance to follow individuals with gait and balance disorders.

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THE EFFECTS OF FOOTWEAR, LEARNING, AND FATIGUE ON CENTER OF PRESSURE EXCURSION DURING SINGLE LIMB BALANCE

Antonia M. Zaferiou, BE, Rebecca A. Zifchock, PhD, Allison M. Brown, PT, MA, Jocelyn C. Frey, BS, Howard J. Hillstrom, PhD
Hospital for Special Surgery, Leon Root, MD, Motion Analysis Laboratory, New York, NY

INTRODUCTION

Stability during single limb balance activities is important for functional tasks ranging from basketball lay-ups to putting on pants. During rehabilitation, single limb balance on an uneven surface is a common evaluation and neuromuscular treatment technique. Previous work has studied the relationship between postural sway during single limb posture and vision, varied support surfaces, and different athletic training backgrounds. However, little is known about the affect of footwear conditions on single limb stance stability.

Therefore, the purposes of this study were to investigate (1) how different footwear conditions affect stability, (2) how task learning and fatigue affects stability (by comparing the order of testing), and (3) if there is a coupling between AP and ML sway. No differences in stability were expected between footwear conditions. Anticipating a strong learning effect and minimal fatigue, stability was expected to increase with successive trials. Finally, it was predicted that the ratio of AP sway to ML sway will increase as an effect of learning.

METHODS

Five healthy male subjects (22±2.5yrs, 1.79±0.08m, 74.7±3.1kg) who participated in cutting sports were included in this study. Subjects balanced on their right legs on a 3” thick Airex® foam surface of a BioSway™ (Biodex Medical Systems, NY) for 30 seconds while center of pressure (COP) coordinates were recorded at 20Hz. Despite the challenge the compliant foam surface presented, participants maintained a posture with approximately 20° arm abduction and 45° left knee flexion. This study was part of a larger investigation comparing the following footwear conditions: barefoot (BARE), standard basketball sneaker (Nike Air Max Go™) (STD), the standard basketball sneaker with ankle taping (TAPE), and an experimental sneaker (EXP) designed to reduce ankle inversion. Participants ran and performed cutting maneuvers for about 30 minutes between footwear conditions, allowing for fatigue effects. The TAPE condition was used first for all subjects with the other conditions randomized.

The COP excursion was characterized by analyzing sway indices, defined as the standard deviation of the COP coordinates. COP sway indices were calculated independently in the AP and ML directions (denoted APSI and MLSI respectively) as well as in both directions combined into an overall sway index (OSI). One-way ANOVAs were used to test the effects of footwear condition and task order on OSI, APSI, and MLSI. Two-way ANOVAs were used to test the effects of footwear condition and task order separately as they relate to sway “direction” (APSI vs. MLSI). Due to the preliminary nature of the data, P≤0.10 was considered significant.

RESULTS AND DISCUSSION

Results are displayed in Table 1. No significant effect was found from footwear condition on OSI or APSI. However, there was a significant difference in MLSI detected between TAPE and BARE that may be caused by more sensory feedback in the BARE condition than in the TAPE condition. There was also a significant effect of sway direction throughout footwear conditions, showing greater sway in the AP direction than in the ML direction. This can be attributed to the greater lever arm of the foot in the AP direction. The comparison of the task order showed a significant effect of task order on OSI, suggesting task learning followed by fatigue. Also, a significant interaction between task order and direction of sway indicated that sway in the AP direction was consistently greater than ML direction sway. Further, the interaction also showed that AP sway may have been less affected by task learning than by fatigue, as compared to the ML sway (Fig 1).

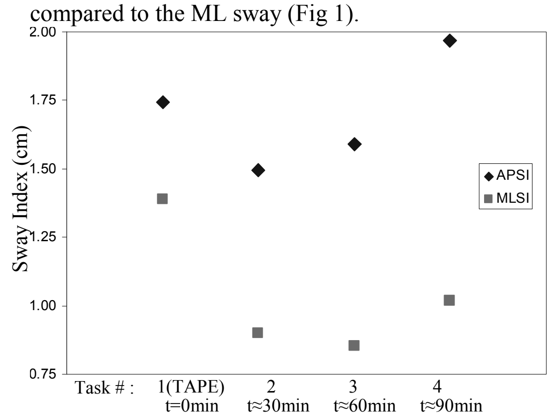

Figure 1: Average MLSI and APSI vs. task order.

While the study is underpowered and footwear condition comparison methods were not ideal, the results may indicate an interesting interplay between task learning and fatigue that overpowers footwear condition effects. This is important to consider during rehabilitation because once a patient has become fatigued, continuing to implement single limb balance tasks into a treatment session may increase injury risk.

Footwear	OSI	MLSI	APSI	Post-hoc relationship with	Task #	OSI	MLSI	APSI	Post-hoc relationship with
BARE	1.95	0.93	1.71	MLSI: TAPE	**1 (TAPE)**	2.25	1.39	1.74	OSI: 3 MLSI: 3
STD	1.96	0.89	1.73		**2**	1.75	0.90	1.50	OSI: 4 APSI: 4
TAPE	2.25	1.39	1.74	MLSI: BARE	**3**	1.81	0.85	1.59	OSI: 1, 4 MLSI: 1
EXP	1.89	0.95	1.62		**4**	2.22	1.02	1.97	OSI: 2, 3 APSI: 2
1 way ANOVA p-values	0.44	**0.05**	0.93			**0.04**	**0.04**	**0.06**	
2 way ANOVA p-values		**0.01**					**+ 0.10**		

Table 1: Average sway indices [cm] and ANOVA and Post-hoc results (+indicates statistical interaction).

RELATIONSHIP BETWEEN FOOT RANGE OF MOVEMENT AND PLANTAR PRESSURE DISTRIBUTION IN DIABETIC NEUROPATHIC PATIENTS

Cristina D. Sartor, Andreja P. Picon, Maria I. Roveri, Roberto C. Dinato, Isabel C. N. Sacco

Physical Therapy, Speech and Occupational Therapy Dept., School of Medicine, University of Sao Paulo, Brazil

INTRODUCTION

It is not widely known if foot static range of movement restriction influences abnormal plantar pressure distribution in patients with diabetic neuropathy, and thus be associated with higher chances of plantar ulcers occurrences. The purpose of this study was to investigate the static active range of motion (ROM) of the ankle (AJ) and the first metatarsophalangeal joints (1st MTF), plantar pressure distribution during gait, and the relationship between these pressure and ROM variables in diabetic neuropathic patients and non-diabetic individuals.

METHODS

Thirty six young adults participated in this study: control group (CG, n=18, 43±9yrs) and diabetic neuropathic group (NG, n=18, 57±4yrs), matched in body mass (p=.10), height (p=.98), but not in age (p<.01). Static active ROM of the 1st MTF and AJ joints were measured in sagittal plane using a manual goniometer and an electrogoniometer, respectively. Plantar pressure (Pedar X system, Novel) was analyzed during barefoot walking with antiskid socks, in 5 areas: rearfoot, midfoot, lateral and medial forefoot and hallux. General linear model for repeated measures ANOVAs were used to compare groups in the studied areas. ROM of both joints were compared between groups using t test. Pearson coefficients were calculated to test correlation between ROM and pressure variables in both groups.

RESULTS

NG presented smaller 1st MTF and AJ ROMs (table 1), smaller contact area at the rearfoot, higher pressure-time integral at the rearfoot, and smaller peak pressure at the hallux (table 2). There was a moderate and significant correlation between pressure-time integral at the hallux and 1st MTF ROM (r=.44, p=.06), and between pressure-time integral at the rearfoot and 1st MTF ROM (r=.48, p=.04) in the NG, but not in CG.

Table 1. Mean 1ST MTF and AJ ROM of CG and NG.

	1ST MTF ROM (º)	AJ ROM (º)
CG	80 ± 16	60 ± 10
NG	67 ± 18	46 ± 9
p	<.01	.029

DISCUSSION

The results demonstrated that the 1st MTP motion restriction play an important role in NG patients by changing the foot rollover mechanism during gait, and consequently the plantar pressure, particularly at the hallux. This may explain the lower pressures at the hallux and the increased pressure-time integral at rearfoot in diabetic subjects. These findings are in accordance with other authors that had also described the same trend and suggested that the neuropathy influences the loading and patterns of walking (Turner *et al*, 2007). Nurse and Nigg (2001) found that peak pressure and pressure-time integral were significantly higher in areas of normal sensitivity and lower at the insensate areas, and the center of pressure (COP) under the foot shifted away from areas of decreased sensitivity when sensory input is reduced from a portion of the foot. Sacco et al (2009a) found reduction of peak pressure at the hallux during gait in neuropathic patients wearing shoes. Giacomozzi and Caselli (2002) found that, in this population, the COP excursion was shorter longitudinally, so the metatarsal heads left the floor earlier in the stance phase. Sacco et al (2009b) studied this altered foot rollover and argued that it was associated with less dynamic mobility and altered plantar pressure distribution. Our results confirm that an altered ROM, especially at the hallux, can produce alterations in pressures, potentially increasing the risk of ulcer formation.

CONCLUSION

In clinical practice, we could use this non-sophisticated measuring tool (1st MTF ROM by manual goniometer) to predict a potential pressure alteration in neuropathic patients during walking. This ROM alteration would indicate a condition more susceptible to injuries and plantar ulcerations.

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Table 2. Mean values of Pressure-time Integrals, Contact Area and Peak Pressure statistically different between CG and NG

	Pressure-Time Integral (kPa.s)– heel	Contact Area rearfoot (cm^2)	Peak Pressure hallux (kPa)
CG	83.6 ± 10.3	30.8 ±2.2	263.6 ±37.2
NG	127.7 ± 102.7	29.2 ±4.9	227.3 ±91.3
p	.04	.01	.02

CAST-BOOT ANKLE-FOOT ORTHOSES YIELD GREATER FOREFOOT LOAD REDUCTION THAN TOTAL CONTACT CASTS

David J. Gutekunst, Kathryn L. Bohnert, Mary K. Hastings, and David R. Sinacore

Applied Kinesiology Laboratory, Washington University in St. Louis, MO USA

INTRODUCTION

Total contact casting (TCC) is a common off-loading strategy in individuals with plantar ulcers secondary to diabetes mellitus (DM) and peripheral neuropathy (PN). Cast-boot ankle-foot orthoses (AFO) may offer a lower-cost alternative to TCC.

The purpose of this analysis was to assess the off-loading capabilities of the cast-boot AFO by comparing the plantar loading patterns in TCC and AFO for subjects with DM, PN, and plantar ulcers. Specifically, we assessed the reduction in plantar loads in the hindfoot, midfoot, and forefoot in each off-loading device compared to unshod walking.

METHODS

Twenty-three subjects with DM, PN, and a plantar ulcer gave informed consent and were randomly assigned to off-loading treatment with either TCC (n=11) or AFO (n=12). Subjects first walked unshod across an EMED pressure platform (Novel Inc., St. Paul, MN). Next, subjects walked across a 6-meter walkway at preferred walking speed with a Novel Pedar pressure insole (Novel Inc., St. Paul, MN) placed in the off-loading device (TCC or AFO) on the ulcerated foot. The EMED trials effectively created an individualized “baseline” value for comparison with each subject’s TCC or AFO trial.

For both Pedar and EMED trials, the plantar pressure map was divided into 3 masks reflecting the hindfoot, midfoot, and forefoot. Multiple steps were averaged on each subject’s ulcerated side for contact area [cm^2], contact time [ms], maximum force [N], peak pressure [N/cm^2], force-time integral [N*s], and pressure-time integral [$N*s/cm^2$]. The effects of TCC and AFO were assessed using analysis of covariance, with walking speed and the EMED value of the outcome variable included as covariates. Mean values for TCC and AFO are reported as estimated marginal means in order to account for variability due to walking speed and EMED values.

RESULTS

No differences in estimated marginal means were found in the hindfoot mask, though there was a trend towards higher force-time integral in the AFO condition compared to TCC (p=0.071). Similarly, no differences were found in the midfoot mask, though there was a trend towards higher peak pressure in the TCC condition compared to AFO (p=0.052).

In the forefoot mask, the AFO showed greater off-loading capability for force and pressure-related outcomes. The TCC condition had significantly higher estimated marginal means than the AFO condition for maximum force (p=0.012) and force-time integral (p=0.010). Similarly, the TCC condition had significantly higher values compared to the AFO condition for peak pressure (p=0.011) and pressure-time integral (p=0.007).

Results are summarized in Table 1 below.

CONCLUSIONS

The cast-boot AFO yielded significantly greater reductions in force and pressure variables in the forefoot region of individuals with DM, PN, and plantar ulcers, suggesting that AFO may provide a viable, low-cost off-loading strategy for forefoot ulcers. Further research is needed to determine whether the load reductions seen in AFO correspond to improved healing rates.

ACKNOWLEDGMENTS

We acknowledge our funding sources: **R01 DK59224, NICHD T32 HD007434, R21 DK079457.**

TABLE 1: Comparison of forefoot force and pressure variables in EMED, TCC, and AFO conditions

	Estimated Marginal Means (Mean ± SE)			
Forefoot outcome variable	**EMED**	**TCC**	**AFO**	**TCC vs. AFO**
Maximum Force [N]	723.3 ± 27.9	182.5 ± 20.2	99.8 ± 19.3	p = 0.012
Force-Time Integral [N*s]	201.0 ± 21.2	66.1 ± 9.4	26.4 ± 9.0	p = 0.010
Peak Pressure [N/cm^2]	85.3 ± 32.9	13.4 ± 6.0	6.6 ± 5.0	p = 0.011
Pressure-Time Integral [$N*s/cm^2$]	35.9 ± 4.4	5.4 ± 0.8	2.0 ± 0.8	p = 0.007

PRESERVATION OF THE FIRST ROCKER IS RELATED TO INCREASES IN GAIT SPEED IN INDIVIDUALS WITH HEMIPLEGIA AND AFO

Karen J. Nolan, PhD (1,2), Mathew Yarossi, BS (1)

1. Kessler Foundation Research Center, Human Performance and Movement Analysis Laboratory, West Orange, NJ,USA, www.kesslerfoundation,org.
2. UMDNJ – New Jersey Medical School, Department of Physical Medicine and Rehabilitation, Newark, NJ, USA

BACKGROUND AND PURPOSE

Ankle foot orthotics are often prescribed to individuals with hemiplegia to assist with ambulation.[1] Gait speed has been used as a primary indicator of orthotic effectiveness, and improved functional ambulation.[2] Limited research has been conducted to understand the specific mechanisms leading to improved gait speed after orthotic intervention. Changes in impulse during the first rocker (braking force) and third rocker (propulsion force) may directly affect changes in gait speed after othotic intervention.[3] Therefore the purpose of this investigation was to objectively measure changes in impulse during double support and correlate those finding to changes in gait speed with and without AFO in individuals with hemiplegia.

METHODS

SUBJECTS: Fifteen individuals with hemiplegia (age 51 ± 12 y, height 1.72 ± 12.2 m, mass 85 ± 21 kg) greater than six months currently using an AFO during ambulation.

PROCEDURES: Subjects performed 10 walking trials at a self-selected pace in two conditions, with and without AFO. During all walking trials, foot pressure data was collected using the Pedar®-x Expert System (Novel Electronics Inc., St Paul, MN, USA). The main outcome measures were gait cycle time (sec), mean force (bodyweights), and impulse (mean force x time) in the, wholefoot, hindfoot (heel and arch) and forefoot (metatarsal heads, toes, and hallux), during initial double support (IDS) and terminal double support (TDS).

ANALYSIS: Wholefoot, hindfoot, and forefoot impulse were calculated in custom Matlab programs by integrating the local forces under specific anatomical regions during IDS and TDS. Independent sample t-tests were used to test for significant differences with and without AFO ($p \leq 0.05$).

RESULTS

Gait cycle timing significantly decreased with the AFO during the entire gait cycle (p=0.034), initial double support (IDS; p=0.034), single support (SS; p=.030), and terminal double support (TDS; p=0.048). During IDS, impulse on the affected limb significantly decreased in the wholefoot (with 0.052±0.016, without 0.086±0.57; p=0.016) and hindfoot (with 0.037±0.012, without 0.060±0.032; p=0.006), and remained relatively unchanged in the forefoot (with 0.016±0.007, without 0.025±0.027; p=0.14). Mean force during IDS on the affected limb also significantly decreased in the wholefoot (p=0.0029) and hindfoot (p=0.0069). During IDS, hindfoot impulse % change and velocity % change during a two-minute walk were significantly correlated (R^2=0.44, p=0.007, figure 1). During TDS, impulse on the affected limb was not significantly different in the wholefoot, hindfoot or forefoot (with 0.062±0.020, without 0.069±0.026; p=0.37). Mean force during TDS on the affected limb significantly increased in the wholefoot (p=0.0002) and remained relatively unchanged in the forefoot (with 0.30±0.11, without 0.27±0.10; p=0.28).

DISCUSSION AND CONCLUSION

Researchers have shown orthotics increase gait speed; this research suggests that the increase in speed is not due to increased propulsive forces at the end of TDS, but due to decreased breaking forces during IDS. The AFO provides increased dorsiflexion at footstrike creating a decreased impulse (breaking force) in the hindfoot thereby preserving the first ankle rocker and providing a more efficient weight acceptance and positively affected gait speed. Future research is required to fully understand the mechanisms underlying the increases in gait speed associated with orthotic intervention in adults with hemiplegia.

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EFFECT OF CAD DESIGNED PEDORTHOSIS WITH BUILD-IN WEDGE FOR CHILDREN WITH CLUBFOOT

X.C. Liu (1), R. Rizza (2), J. Thometz (1), R. Lyon (1), C. Tassone (1), S. Tarima (3)

1. Dept. of Orthopedic Surgery, Children's Hospital of WI, Medical College of WI, WI
2. Dept. of Mechanical Engineering, Milwaukee School of Engineering, WI
3. Division of Biostatistics, Medical College of WI, WI

INTRODUCTION:

A customized dynamic pedorthosis has been developed to prevent recurrence of the treated clubfoot. This new pedorthosis was developed using our OrthoticPro™, a software package that utilizes dynamic plantar pressure data, and computer aided engineering tools. The pedorthosis is constructed using rapid prototyping technologies (RP). The purpose of this research was to: 1) compare the pressure metrics between barefoot in regular shoes and the pedorthosis, 2) quantify the deviation of the center of pressure (COP) trajectory from the normal trajectory with and without the use of the pedorthosis, and 3) develop a FEA model to predict this effectiveness of wedges.

METHODS:

Five typically developing children (average age 7.2 years, 2 girls and 3 boys) and five clubfoot patients with (average age 6 years 1 girl and 4 boys) were recruited. The finite element model was generated using CT based geometry and plantar pressure (Emed, Novel Inc., MN). The pedorthosis was constructed from the CAD model and manufactured using the RP (Stereolitography). Eight-pedorthtoses were fitted to the children with clubfoot and the children were measured during walking with and without orthotics using the Pedar insole pressure system (Novel Inc., MN).

RESULTS:

There was significant reduction of the average COP deviation following the use of the pedorthosis. The maximum reduction of the average COP deviation occurred in the forefoot (7.87%) and then the midfoot (4.00%). There are no significant differences of any pressure measurements at the midfoot, medial forefoot, and entire toes. Significant reduction of maximal force, peak pressure, and loading at the heel and the lateral forefoot are identified following the use of the new pedorthosis ($P<0.05$) (Figure 1).

DISCUSSION:

This kinetic change may imply a reduced supination of the forefoot in children with residual clubfoot. Our short-term follow-up demonstrates that the pedorthosis improves the dynamic misalignments in the residual clubfoot. A customized wedge as predicted by FEA indicates a corrective magnitude of the wedge angle which varies along the forefoot and midfoot regions.

FIGURES AND TABLES

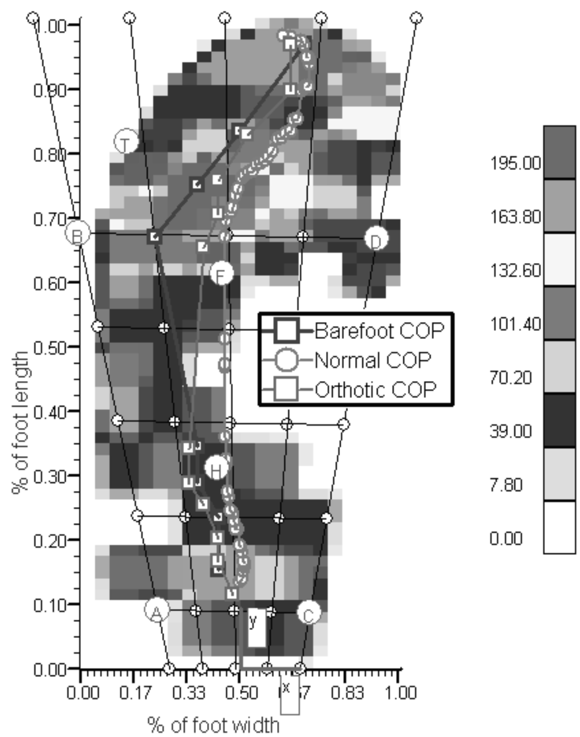


Figure 1: Average deviation of the COP trajectory from the normal trajectory in the hindfoot, midfoot and forefoot regions following the use of the new pedorthosis.

INTRAARTICULAR AND MUSCLE FORCE REACTIONS OF THE LEG USING DIFFERENT INSOLES

Wyss, Ch.[1,2], Roethlisberger, M[3], Gerber, H[3]

1. Laboratory of Movement analysis, Childrens University Hospital Basel, Switzerland
2. Laboratory of Movement analysis, Cantonal Hospital Aarau, Switzerland
3. Institute for Biomechanics. ETH Zurich. Switzerland

Introduction

In the literature the effect of insoles is discussed controversial [1, 2, 3]. The aim of our study was firstly to find a correlation between kinematic, kinetic and electromyographic changes during gait caused by the application of different insoles. Secondly we were interested in the change of the intraarticular force in the knee and the change of muscle forces in the leg.

Methods

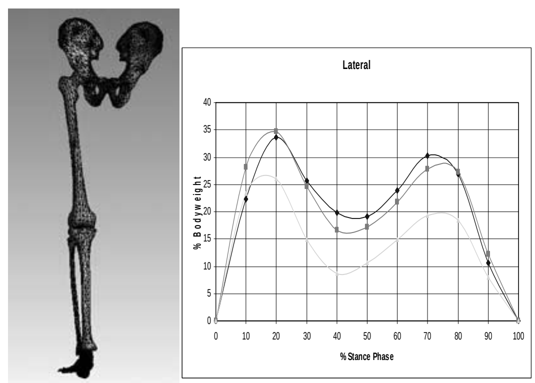
We examined 10 healthy subjects using 5 different insoles alternatively with a neutral insole. The kinematic measurements were performed with a Vicon MX System (6 Cameras). Additionally we used an AMTI force platform and a Novel SF pressure measurement System, even though a Pedar (Novel) insole pressure measurement system. The EMG was measured with a MegaWin system, synchronized with the Vicon system. The intraarticular reaction forces in the medial and the lateral compartment oft the knee were calculated with the help of a finite element model (ANSYS) of the leg which is considered as a rough estimate based on CT Scans. The muscle forces were determined using a slightly modified model from the repository AMMRV1.1 of ANYBODY TECHNOLOGY (Vaughan).

Results

Opposite to the literature we found that there is, for some kinematic parameters, a more or less unique change due to different insoles. The insoles with a medial wedge, for example, showed a significant decrease of eversion of the hindfoot and the insoles with a lateral wedge showed a significant increase of eversion. A more interesting result was the calculation of the intraarticular reaction force in the medial and lateral compartment of our knee model. In figure 1 the pink curve shows the reaction force in the lateral knee compartment in % Bodyweight with an insole with a medial wedge. Nearly the same curve (blue line) shows the measurement with a neutral insole. A significant decreased reaction force could only be observed when we simulated a stiff ankle joint that means inversion and eversion of the ankle joint was excluded in our simulation.

Using our muscle model we found no significant change of muscle forces in the whole leg. Opposite to that we found a significant increase of all thigh extensors comparing barefoot gait with gait in shoes with neutral insoles.

Fig. 1: Reaction force in the lateral knee compartment



DISCUSSSION & CONCLUSIONS

The findings concerning the kinematic parameters of the ankle joint confirm more or less the clinical assumptions. But the clinical conclusion, that the changing of these parameters influence the reaction forces in the knee, could not be confirmed without reservations. The results of our simulations shows clearly that the effect of an insole will be completely compensated by the ankle joint. There is no effect on muscle forces of the leg or joint reaction forces in the knee if the ankle is enough mobile.

Because the muscle forces are significantly increased, when gait analysis is performed with shoes, in the future it will be necessary to do the clinical gait analysis not only barefoot as hitherto.

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T23

RATIOS OF LATERAL TO MEDIAL PATELLOFEMORAL FORCES AND PRESSURES IN A SIMULATED OPERATIVE ENVIRONMENT OF TOTAL KNEE ARTHROPLASTY

Joseph J. King (1), Douglas L. Cerynik (1), Jeffrey I. Peretz (1), Aaron Black (2), Norman A. Johanson (1)

1. Department of Orthopaedic Surgery, Drexel University College of Medicine, Philadelphia, PA
2. Drexel University College of Medicine, Philadelphia, PA

INTRODUCTION

The differences of lateral and medial patellofemoral forces in different total knee arthroplasty (TKA) designs have not been extensively studied. The purpose of this study was to measure the distribution of patellofemoral forces in native and TKA cadaver knees utilizing sensor technology performed in a simulated operative environment. Currently the criteria for evaluating the success of soft tissue balancing in TKA involves only the surgeon's observation of intraoperative patellar tracking during a passive range of motion, but no good objective measure of patellar tracking has been described. We propose that the ratio of lateral to medial maximum force and peak pressure may be used as a surrogate marker for patellar tracking.

METHODS

The patellofemoral forces of 6 knees from 3 fresh-frozen half-body female cadavers (mean age 82 years) were evaluated with a capacitive sensor placed on the patellofemoral articulation in staged clinical scenarios. The half-body cadavers were placed supine with the pelvis stabilized such that all study aspects were similar to an actual operative procedure. The medial parapatellar approach was used and the patella was everted in the standard fashion. The sensor device, capacitive novel pliance system (Novel Electronics, St. Paul, MN), was sutured over the patellar surface and the joint capsule closed using 3 sutures.

Staged clinical scenarios tested were native knees (NKNP, 6 knees), total knee replacement without patellar resurfacing (RKNP, 6 knees), resurfaced knee and patella (RKRP, 6 knees), resurfaced knee and patella with lateral release (RKRP-LR, 3 knees), gender-specific knee with patella resurfacing (GKRP, 3 knees), and gender-specific knee with lateral release (GKRP-LR, 3 knees). Maximum force (N) and peak pressure (kPa) were simultaneously recorded over a set of 3-4 ranges of motion. Average values were compared for the medial and lateral patella compartments and for different clinical settings.

Component alignment, stability, and patellar tracking were clinically acceptable by direct visualization in all scenarios.

A student t-test with a pooled sample variance was used in comparisons, and the assumption of equal sample variances between lateral and medial sides was assessed using a F-test. P values of <0.05 were considered significant.

RESULTS

Significant differences in lateral and medial force and pressure differentials were seen in most scenarios despite clinically normal patellar tracking (Figures 1 and 2).

For the native knee (NKNP), a statistically significant difference was seen between the lateral and medial patellofemoral maximum force means ($p=0.04$) at a ratio between the lateral and medial sides of 1.63:1 with a trend ($p=0.11$) in the ratio of lateral to medial peak pressure (1.80:1). For RKNP scenario, an increase in the ratio of lateral to medial maximum force and peak pressure means was seen to 2.86:1 and 1.99:1 which was significant ($p<0.01$ and $p=0.04$).

For the resurfaced knee and resurfaced patella scenario (RKRP), statistically significant differences were seen in lateral to medial maximum force means and peak pressure means ($p<0.01$ and $p<0.01$) at the ratios of 2.75:1 and 2.57:1. The addition of a lateral release in this scenario (RKRP-LR) reduced the lateral to medial ratios of maximum force and peak pressure means to insignificant ratios (1.46:1 and 1.11:1, respectively).

For the gender-specific knee (GKRP), a statistically significant difference was seen in lateral to medial maximum force means ($p<0.01$) at the ratio of 1.96:1, but not in the lateral to medial peak pressure means (1.33:1). Addition of a lateral release to the gender-specific knees did not significantly alter the ratios of lateral to medial differentials.

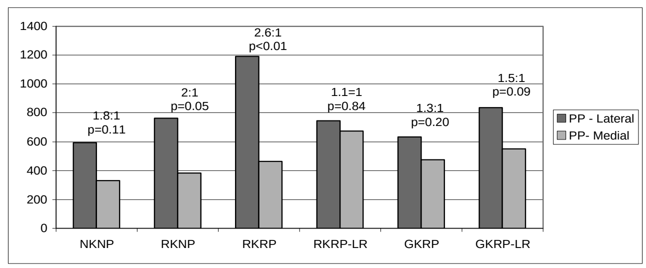

Figure 1: Peak pressure (PP) in the lateral and medial compartments including the ratio between the compartments

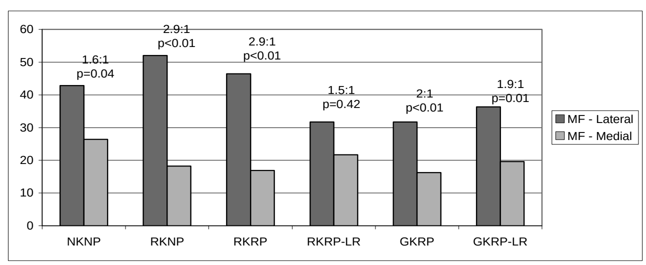

Figure 2: Maximum force (MF) in the lateral and medial compartments including the ratio between the compartments

DISCUSSION

Ratios of peak pressure and maximum force were increased in the lateral patellofemoral compartment on resurfaced knees which were not seen in gender knees. RKRP-LR and GKRP demonstrated lateral to medial patellofemoral force and pressure ratios most similar to the native patella (NKNP).

The addition of a lateral release in conventional total knee arthroplasty can equalize the medial-lateral force distributions.

The decreasing lateral to medial maximum force and peak pressure ratios in the lateral release group (RKRP-LR) after TKA and with gender-specific knees (GKRP) compared to the standard TKA (RKRP) demonstrates that these ratios are surrogate markers for patella tracking due to the known effects of lateral retinacular release on patella tracking and the design of gender-specific knees to improve tracking.

Intraoperative quantification of force and pressure differentials would enable the surgeon to assess proper patellar tracking, giving guidance for using lateral retinacular releases.

T24

CALCULATION OF PRESSURE TIME INTEGRAL, A DIFFERENT APPROACH.

Tom. Melai (1,2), Herman. IJzerman (2), Ton. LH. de Lange (1), Paul. B. Willems (2), Kenneth. Meijer (2), Nicolaas. C. Schaper (3), Hans. HCM. Savelberg (2)

1. Chair of Health Care & Technology for Quality of Life, Fontys University, The Netherlands
2. NUTRIM, Maastricht University Medical Centre, The Netherlands
3. CARIM, Maastricht University Medical Centre, The Netherlands.

INTRODUCTION

High plantar peak pressure values (PP) in people with diabetic polyneuropathy are correlated with ulceration (Boulton, 1983). An additional parameter that is commonly used to asses loading of the foot is the Pressure Time Integral (PTI) (Sauseng, 1999). However, a recent study (Waaijman, 2009) found high correlations between PP and PTI; and suggested that PTI would be of limited additional value.

Novel software calculates PTI as the product of contact time with PP in one mask (PTI_novel) instead of using pressure per sample. This is expected to give an overestimation of the true PTI. The Force Time Integral (FTI) is calculated using individual sensor pressures per sample. Therefore, a more accurate approach to obtain PTI per mask is dividing FTI by contact area (PTI_F).

We sought to test the differences between these two calculation methods in different populations with and without diabetes and diabetic polyneuropathy.

METHODS

Subjects were divided in a group of healthy elderly (HE), a group with diabetes type 2 (DM), and a group with diabetes and polyneuropathy (DPN). Subjects were asked to walk barefooted over a pressure platform at standardized walking speed (1.2±0.1m/s).

	N	Age	Length	Weight
HE	19	72.6±1.9	1.73±0.017	68.1±1.2
DM	33	76.5±3.4	1.69±0.017	74.7±2.7
DPN	76	78.0±2.0	1.74±0.009	82.4±2.7

Table 1: Subject Characteristics. (mean±std. Error).

Data was masked using the Novel 10 mask division (1=heel, 2=mid foot, 3-7=metatarsal region, 8= hallux, 9-10= toes).

RESULTS

Overall PTI values were lower for PTI_F compared to PTI_novel (PTI_novel= 9.1±5.4 Ns/cm^2, PTI_F= 2.6±1.1 Ns/cm^2). PP and PTI_novel showed a similarity of pattern, but PTI_F pattern differed from PTI_novel, especially in the region of the hallux. Figure 1 shows these differences for the DPN group. To operationalize the difference between both PTI calculations, PTI of the hallux was divided by PTI of the heel. These ratios differed significantly (p=0.000) between both calculations in each group (PTI_novel - PTI_F: HE 1.17 - 0.81; DM 1.34 - 0.79; DPN 1.19 - 0.80).

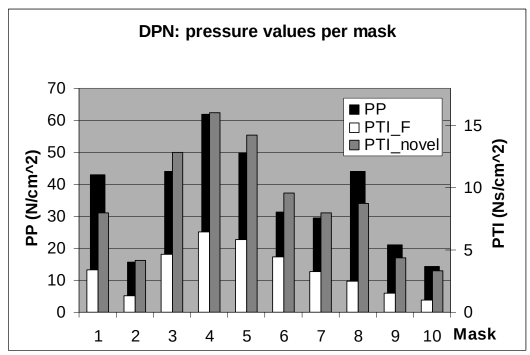

Figure 1: Pressure variables per mask in DPN group: PP (on the left axis), PTI_novel and PTI_F (on the right axis)

DISCUSSION

These results show that there is a difference in PTI calculation based on PP and FTI, especially in the hallux. Because the PTI_novel is based on PP, the additional value as described by Waaijman et al. (2009) may be limited.

In addition, when comparing groups, there is a similarity of differences between PP and PTI_novel (not shown here). However, the differences between PP and PTI_F show an inconsistency when comparing the different groups per mask.

Based on these results it is concluded that the more accurate calculation, as suggested here, may contribute to better understand foot sole loading.

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FOOTPRINT MASKS: REPRODUCIBILITY OF THE OXFORD ANATOMICALLY-BASED SELECTION BY MEANS OF NOVEL GEOMETRICAL MASKS.

Claudia. Giacomozzi (1), Julie. Stebbins (2,3), Axel. Kalpen (4)

1. Dept of Technology and Health, Istituto Superiore di Sanità, Rome, Italy
2. Oxford Gait Laboratory, Nuffield Orthopaedic Center, Oxford, UK
3. NDORMS, Oxford University, Oxford, UK
4. novel biomechanics lab, novel gmbh, Munich, Germany

INTRODUCTION

In the field of foot biomechanics, the clinical relevance of the integration of pressure, kinematics and force measurements has already been demonstrated (Giacomozzi, 2000). Previous work conducted by the authors in this field has been focused on two key areas: First, to produce a reliable combination of pressure instrumentation, a kinematic foot model (in this case the Oxford Foot Model (OFM) (Stebbins, 2006), procedures and software to provide automated masks for accurate pathologic footprint selection. Second to identify the most appropriate geometrical footprint selections which, in presence of regular footprints, compares reliably to a selection based on markers from the OFM. Since kinematic measurements are not always available, it is useful to know which geometric mask should be used to allow comparable results. The present study deals with this latter aim.

MATERIALS AND METHODS

Markers from the OFM were used to define the following 5 anatomically-based foot areas (Stebbins, 2006): 1) medial hindfoot (MH); 2) lateral hindfoot (LH); 3) midfoot (MI); 4) medial forefoot, toes included (MF); 5) lateral forefoot, toes included (LF). novel gmbh purposely modified the multimask software to integrate kinematic data from a VICON system acquired simultaneously with an emed pressure platform. As a result, the above areas were identified by vertically projecting the position of the relevant anatomical markers at midstance. Two automated masks (available in software) were found to be the most similar to the OFM based mask: i) BIS mask, based on the longitudinal bisecting line of the foot and on the two lines perpendicular to it placed at default distances from the bottom of the footprint (27% and 55% respectively); ii) HT2 mask: similar to the BIS mask, but based on the longitudinal line going from the middle of the hindfoot to 2nd toe (HT2 line). Since the OFM-based selection uses two different longitudinal axes for the hindfoot and the forefoot, two further geometrical masks were created based on a mixed use of the previous two masks: i) MIX: same hindfoot selection as BIS, same forefoot as HT2; ii) MIX1: same hindfoot as HT2, same forefoot as BIS. Thus, 5 masks were applied in all to 200 footprints acquired from a young healthy population of 19 girls and boys (38 feet). Up to now, 25 different, randomly selected footprints have been processed. %RMSEs were calculated along the whole loading process for vertical force (F), contact area (A), peak pressure (PP) and mean pressure (MP).

RESULTS

Very good matching was found in all the 4 geometric masks with the OFM-based mask – RMSE<5% for F, A and PP; RMSE≈6% for MP. BIS and MIX1 showed better results than HT2 and MIX, MIX1 giving the best results. MF was the most sensitive area as for F; MI was the most sensitive as for A, PP and MP. Table 1 reports the %RMSEs for MIX1 mask. Fig.1 shows the 5 masks on a regular footprint and the corresponding %RMSE averaged over F, A, PP and MP.

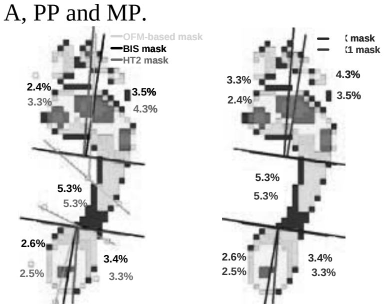

Figure 1: Masks on a regular footprint (left:OFM-based, BIS and HT2; right: MIX and MIX1). %RMSEs averaged over F, A, PP and MP.

Table 1. %RMSE (sd) of MIX1 with respect to OFM-based mask.

	F	A	PP	MP
MH	3.7(3.0)	2.0(1.5)	0.5(0.7)	3.8(2.7)
LH	3.6(3.4)	2.1(1.7)	2.8(3.9)	4.7(3.4)
MI	2.5(1.4)	4.1(1.8)	5.5(3.0)	9.0(4.6)
MF	3.5(2.5)	2.8(2.3)	1.2(2.7)	1.9(1.5)
LF	3.8(2.4)	3.7(2.2)	2.9(3.8)	3.8(2.1)
Total foot	3.9(2.0)	3.4(1.2)	3.8(2.1)	5.7(1.9)

DISCUSSION AND CONCLUSIONS

For regular footprints, the MIX1 geometric mask produced highly comparable results compared to the OFM-based anatomical mask, with %RMSE only slightly higher than intra-subject variability.

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A TECHNIQUE TO ASSESS PLANTAR SOFT TISSUE STIFFNESS DURING THE STANCE PHASE OF GAIT

Carolyn Roposh (1), Carole Tucker (2), Margo Orlin (1), Rami Seliktar (1)

1. Drexel University, Philadelphia, Pa
2. Temple University, Philadelphia, Pa

INTRODUCTION

The bones of the foot are largely supported by the soft tissue on the plantar surface, which acts as the interface with the ground (Tachdjian, 1985). Soft tissue stiffness could affect pressure distribution, shock absorption, or stability during gait. Assessment of soft tissue is important for foot structures that may be prone to increased stiffness such as in high arch structures that are commonly more rigid and less adaptable to the ground for impact cushioning (Van Boerum, 2003). Even though plantar stiffness has been evaluated using a variety of methods, there are no standardized methods to measure plantar stiffness that take into account the real-time, dynamic, weight bearing condition of stance phase of gait (Cavanaugh, 1999 & Rome, 2000). Therefore, we hypothesize that this technique will be able to describe the changes in plantar stiffness during stance based on differences in arch structures.

METHODS

Reflective markers were placed on the feet of 13 subjects (age range 18-21) with no foot pathologies. Trajectory data were collected with Vicon motion capture cameras during sitting trials and while walking over a Novel EMED-ST/E pedobarograph (Munich, Germany). To model the stiffness of the soft tissue, force and deformation had to be isolated. The deformation was the difference in distance between the markers on the foot and the ground during sitting unloaded trials and during walking. Deformation data were limited to the time each segment were in contact with the ground and decimated due differences in collection frequencies. Force data for each segment were matched to the corresponding deformation data, and all data were inputted into the equation for a simple spring to model stiffness.

RESULTS

When compared to the low arch subjects, the mean stiffness values for the high arched subjects were greatest in the heel and lowest in the hallux (Table 1). The real time stiffness plots show the changing stiffness during the weight bearing condition of stance phase (Figure 1).

FIGURES AND TABLES

	Low (N=4)	**Typical (N=3)**	**High (N=6)**
Heel	9.47 ± 3.6	16.19 ± 8.0	20.41 ± 6.3
Midfoot	2.01 ± 0.8	1.08 ± 0.3	0.90 ± 0.6
1st Met	1.51 ± 0.7	2.17 ± 1.8	5.78 ± 1.2
2nd Met	1.48 ± 0.6	2.50 ± 1.0	2.87 ± 0.9
5th Met	0.42 ± 0.1	2.35 ± 1.1	4.21 ± 1.0
Hallux	2.05 ± 0.9	3.69 ± 1.1	2.21 ± 0.9

Table 1: Mean Stiffness(N/mm) ± Standard Deviations

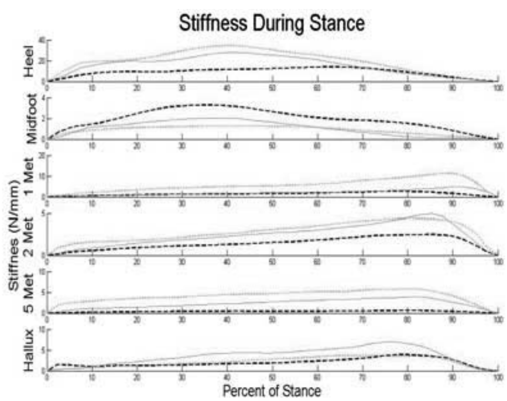

Figure 1: Changing stiffness during stance for high (dot), low (dashed), and typical (solid) arch structures

CONCLUSION

This work demonstrates the feasibility and utility of combining motion capture data with pedobarograph data to characterize plantar stiffness during stance for comparison between arch structures. Variations in plantar stiffness during loading and unloading could impact foot function and alter gait patterns, and therefore should be examined more thoroughly.

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RELIABILITY OF DYNAMIC FOOT GEOMETRY ASSESSMENT USING A PEDOBAROGRAPHIC PLATFORM AND A TWO-STEP APPROACH

Karen A. Keenan, Jonathan S. Akins, Brian P. Dugan, Melanie Francis, John P. Abt, Timothy C. Sell, Scott M. Lephart

Neuromuscular Research Laboratory, University of Pittsburgh, PA

BACKGROUND

Static measurements of foot structure have previously been used in both the clinical and research setting to describe and classify foot structure (Redmond, 2008). Attempts have been made to relate these different classifications to lower extremity injuries and to identify appropriate athletic footwear in an attempt to mitigate injury (Jenkins, 2007; Knapik, 2010). Such attempts, however, have yet to yield consistent results (Barnes, 2008; Burns, 2005). Since the foot is dynamically loaded during gait and sport activities, it may be more appropriate to classify foot structure based on dynamic geometric variables. Prior to this application, the reliability of these variables must be established. The purpose of this study was to determine the reliability of geometric variables obtained during gait at a self-selected speed.

METHODS

Ten healthy males (n=8) and females (n=2) participated in this study (age: 27.7 ± 4.1 years, mass: 77.6 ± 10.7 kg, height: 174.3 ± 7.0 cm). Data were collected on two different days using the EMED-X® system (Novel GmbH, Munich, Germany), with a sampling frequency of 100Hz. A two-step approach at a self-selected speed was utilized for all trials, which previously has been demonstrated to be reliable in gait analysis (McPoil, 1999). In order for a trial to be included the following criteria were met: only one foot contacted the platform, contact was made on the second step, subjects did not "target" the platform, and subjects appeared to walk with their normal gait and cadence. After familiarization of the task, subjects performed 5 right and 5 left trials.

Geometric variables were then obtained through the Novel software package. Intraclass correlation coefficients (ICC) were calculated using a two-way random effects model (ICC [2, k]) and means and standard errors were calculated for each foot for 21 geometric variables.

RESULTS

Means, standard errors of the measurement (SEMs), and ICCs for both the left and right feet are presented in Table 1. Excellent reliability (ICC>0.90) was demonstrated in 15 of the 21 geometric variables for the left foot and 16 of the 21 variables for the right foot. Good reliability (ICC>0.70) was demonstrated in 20 of the 21 variable for both the left and right feet.

	Mean ± SEM		ICC	
	Left	Right	Left	Right
Anterior plantar angle [°]	28.10 ± 0.79	28.46 ± 0.29	0.979	0.848
Posterior plantar angle [°]	28.03 ± 0.97	28.41 ± 0.32	0.975	0.775
Lateral tarsal angle [°]	154.02 ± 1.82	153.38 ± 1.62	0.708	0.629
Medial tarsal angle [°]	149.83 ± 1.49	149.76 ± 0.73	0.965	0.853
Lateral plantar angle [°]	7.32 ± 0.14	7.49 ± 0.08	0.991	0.971
Medial plantar angle [°]	7.32 ± 0.14	7.49 ± 0.08	0.991	0.971
Long plantar angle [°]	14.65 ± 0.28	14.98 ± 0.15	0.992	0.971
Transverse plantar angle [°]	16.61 ± 5.57	14.44 ± 5.00	0.785	0.733
Hallux angle [°]	4.42 ± 0.82	4.28 ± 0.75	0.986	0.983
Hallux angle (2) [°]	6.73 ± 1.46	5.80 ± 0.96	0.992	0.981
Forefoot angle [°]	113.94 ± 1.19	114.75 ± 2.17	0.736	0.921
Subarch angle [°]	114.73 ± 2.58	108.08 ± 2.73	0.972	0.975
Heel angle [°]	9.20 ± 1.15	10.06 ± 2.61	0.656	0.933
Foot progression angle [°]	7.46 ± 0.59	10.01 ± 0.54	0.990	0.988
Foot length [cm]	27.31 ± 0.15	27.43 ± 0.14	0.993	0.991
Forefoot width [cm]	9.75 ± 0.08	9.85 ± 0.13	0.970	0.989
Heel width [cm]	5.62 ± 0.05	5.63 ± 0.03	0.995	0.989
Coefficient of spreading	0.36 ± 0.00	0.36 ± 0.01	0.871	0.974
Arch index	0.24 ± 0.01	0.24 ± 0.01	0.965	0.985
Forefoot and heel coefficient	0.58 ± 0.01	0.57 ± 0.01	0.977	0.965
Forefoot coefficient	1.08 ± 0.01	1.09 ± 0.01	0.780	0.922

Table 1: Geometric variables: mean, SEM, and ICCs for left and right feet.

DISCUSSION

Reliable dynamic assessment of foot geometry can be obtained using the EMED-X pedobarography platform. These findings support the use of dynamic foot geometry assessment in future research to classify foot structure/type and to determine the relationship between foot geometry and lower extremity injuries.

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COMPARISON OF A SIX DEGREE-OF-FREEDOM FORCE SENSOR AND PRESSURE INSOLE MEASUREMENTS IN SELECTED SKIING MANOEUVRES

U.G. Kersting (1), N. Kurpiers (2), E. Hild (3), A. Kiefmann (4), V. Senner (4)

1. Center for Sensory-Motor Interaction, Aalborg University, Denmark
2. Dept. of Sport and Exercise Science, The University of Auckland, New Zealand
3. Novel GmbH, Munich, Germany
4. Dept. Sport Equipment and Materials, University of Technology Munich, Germany

INTRODUCTION

Alpine skiing has been characterized as particularly dangerous to the knee joint (Heir et al., 2005). Many of these may be avoidable with a greater understanding of biomechanical parameters related to skiing technique and equipment design. Attempts to measure the dynamic forces transferred to the skier have been described previously. Only few have successfully realized a full six degree-of-freedom measurement system due to the high technical challenges implied (Kiefmann et al., 2006). Alternatively, pressure insole systems have been used to collect data during skiing with different applications and studies (Schaff et al., 1989).

The purpose of this investigation was to compare a pressure insole in combination with a pressure sensor placed inside the shaft at the anterior aspect of the tibia to the data collected from a full six degree-of-freedom force sensor mounted between shoe and binding. It was hypothesized that vertical reaction forces and the moment about the ankle joint can be estimated from pressure measurements.

METHODS

Tests were carried out on two occasions on slopes of appr. 20° inclination. One subject performed various reference movements and relevant skiing motions including straight runs, short and long turns on a groomed slope. Secondly, six subjects performed runs on a mogul course. Skill levels ranged from advanced to elite. Two mobile 6 DoF force plates were mounted between ski boot and binding sampling to a biovision mini computer (500 Hz). A Pedar insole system (novel GmbH) consisting of an insole and a dorsal pad, sampling at 125 Hz was worn inside one of the boots. A synchronized video was recorded for all trials.

RESULTS

During both, the reference trials as well as skiing on the slope and moguls, anterior shaft contact forces of over 80% body weight were recorded. During single legged turns this was up to 100% body weight indicating a substantial load being transferred through the shaft of modern ski boots. For a relatively upright posture vertical forces of up to 80% of the reading of the force sensor were registered by the plantar insole.

DISCUSSION AND CONCLUSIONS

Results indicate a highly complex interaction between movement at the ankle joint, positioning of the body with respect to the ski and the transfer of loads through the boot-shaft system. It was thus not possible to get a reasonable representation of the vertical force by the pressure system. The combination of shaft and plantar pressure readings was used to estimate the anterior-posterior moment between leg and ski. In some relatively static riding situation the two systems were reasonably similar. It is thus indicated that the use of pressure inside a ski boot is not sufficient for characterizing the loads experienced by the body system during alpine skiing. However, for the assessment of boot fit, balancing tasks or biofeedback studies the use of pressure insoles will be valuable. Further, it might be possible to improve the pressure system setup by covering more area of the foot- and shank-boot interface. More studies are required to explore these relationships.

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DATA COLLECTION FROM TWO DIFFERENT PRESSURE PLATFORMS USING STANDARDISED METHODOLOGY PRODUCES COMPARABLE RESULTS.

Julie. Stebbins (1,2), Claudia. Giacomozzi (3)

1. Oxford Gait Laboratory, Nuffield Orthopaedic Center, Oxford, UK
2. NDORMS, Oxford University, Oxford, UK
3. Dept of Technology and Health, Istituto Superiore di Sanità, Rome, Italy

INTRODUCTION

It is widely believed that plantar pressure data is susceptible to a large degree of intra- and inter-subject variability. This reduces confidence in interpreting results, particularly following clinical intervention. There are now a range of pressure platforms available for collecting data, utilising a broad spectrum of technology. Along with this, there is currently a lack of standardization in methodology for collecting plantar pressure data, and reporting of results.

It is unclear how much of the variability reported in plantar pressure measurement is due to varying methods of data collection and instrumentation, and how much is attributable to real variation in the data.

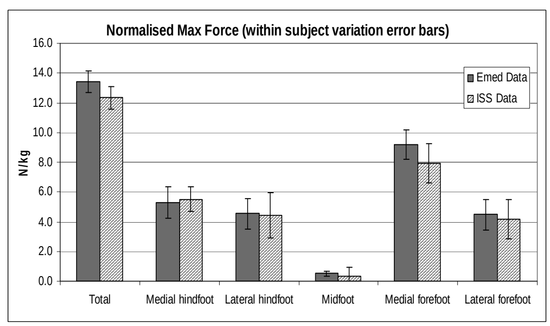
The aim of this study was to determine the amount of variability in plantar pressure assessment that is due to differences in the types of equipment used to collect the data. This was achieved by using a standardised protocol, and comparing the results and the intra- and inter-subject variability from two different pressure platforms.

METHODS

A Novel Emed-M capacitive pressure platform was used to collect data from 19 healthy children (age range 6 – 16 years). In addition, a prototype, piezo-resistive pressure platform (ISS), rigidly mounted to an AMTI force plate, was used to collect data from a different group of 15 healthy children (age range 6 – 16 years). Three footprints were collected from each foot. Both pressure platforms had a spatial resolution of 4 sensors per square cm, and collected data at 50Hz. A 12 camera, Vicon (Vicon, Oxford) system was used to collect synchronous data from markers placed on the feet, according to the Oxford Foot Model protocol (Stebbins, 2006). Co-ordinates from anatomical marker positions were projected onto the overall pressure map at a time corresponding to mid-stance, and used to automatically divide the footprint into five areas (medial heel, lateral heel, midfoot, medial forefoot, lateral forefoot) using a previously validated protocol (Giacomozzi, 2000). The peak force and peak pressure for each sub-area were compared between platforms, along with the intra-subject standard deviations.

RESULTS

Difference in the average peak force (normalized to body weight) obtained from each platform were minimal (Figure 1), (0.1-1.3 N/kg). Differences in peak pressure were slightly higher, with the least difference in the lateral forefoot (27kPa) and the most in the medial forefoot (157kPa). Differences in the intra-subject standard deviations were negligible in the peak force comparison (0.1-0.8N/kg). Differences in the peak pressure intra-subject standard deviations were much higher (16-98kPa). The co-efficient of variation for peak pressure was consistently lower for Emed. The lower accuracy of ISS was partially compensated for by instantaneously calibrating it with information from the force platform; this led to comparable peak force data, but little improvement was obtained for the accuracy of the pressure measurements. These results suggest that comparable data can be obtained from 2 different pressure platforms when the methodology is stanadardised, however accuracy of the pressure platform is critical.



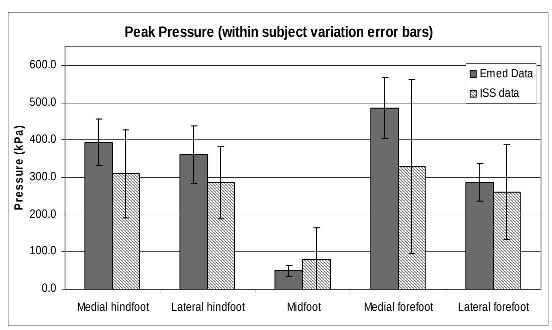


Figure 1: Comparison of average normalised force (above) and peak pressure (below)

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DO MOTION CONTROL RUNNING SHOES DECREASE SAGITTAL PLANE MOVEMENT OF THE MEDIAL LONGITUDINAL ARCH?

Thomas G. McPoil (1), Mark W. Cornwall (1), Melissa R. McNulty (1)

1. The Laboratory for Foot & Ankle Research, Physical Therapy Program, Northern Arizona University, Flagstaff, Arizona, USA

PURPOSE

While it has been proposed that motion control or stability running shoes are designed to control foot pronation, little research has been published to substantiate this claim. Cheung et al (2008) reported that motion control shoes, in comparison to neutral shoes, reduced midfoot forces during running but failed to consider contact area acting on the midfoot while walking. We hypothesized that the amount of midfoot contact area, indicative of medial and lateral longitudinal arch sagittal plane movement, would not increase between walking and running if a motion control shoe was effective in controlling foot mobility. The purpose of this study was to investigate the change in plantar surface contact area when walking and running in motion control shoes.

SUBJECTS

Ten females' subjects with a mean age of 25.5 years (range 22 to 34 years) with no history of congenital or traumatic deformity or foot problems volunteered to participate in study. All subjects ran at least 15 miles per week for the past 2 years and were fitted with new motion control shoes (Brooks Ravenna Stability) at the same local running footwear store.

METHODS

Foot measurements previously described were recorded in weight bearing so that the arch height ratio (AHR), foot mobility magnitude (FMM) and foot posture index (FPI) could be calculated for all subjects (McPoil et al, 2009). The FPI and AHR were used to determine foot posture with the FMM used to determine foot mobility. Each subject was asked to walk and run in their running shoes over a 42-meter indoor runway while in-shoe pressure data were collected using the PEDAR-X system. To determine the effect of the shoe contour on plantar surface area, each subject was asked to walk over the same distance while pressure data were collected while wearing a flat soled, non-contoured karate shoe. For all in-shoe pressure measurements, the sensor insoles were place over a flat piece of firm insole material (durometer = 58 Shore A) that replaced the original shoe sockliner. Ten consecutive steps from the middle 20 meters for the right foot only were selected for analysis. Novel percent mask and group mask software was used to determine contact area in the following plantar regions: medial heel, lateral heel, medial midfoot, lateral midfoot, medial forefoot, lateral forefoot, and hallux. A MANOVA was used to determine those plantar surface areas that were significantly difference. Based on those results, an ANOVA and Tukey's pairwise comparisons were used to further assess the medial and lateral midfoot regions.

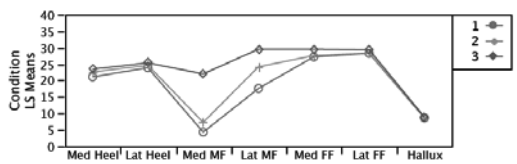

Figure 1: MANOVA results for plantar contact area (1 = karate shoe, 2 = walk in shoe; 3 = run in shoe).

RESULTS

The mean values for foot posture and mobility were: AHR = 0.335 ± 0.018; FMM = 1.58 ± 0.27; and FPI = 3.8 ± 2.35. Based on these measures only one subject was classified as excessively pronated.

Total plantar contact area was significantly increased in the medial midfoot ($p < .0001$) when running in the shoe in comparison to both walking conditions (shoe & karate). The mean surface area for the medial midfoot was 4.4 cm^2 for karate walk, 7.4 cm^2 for shoe walk, and 22.1 cm^2 for shoe run. Total plantar contact area was also significantly increased in the lateral midfoot ($p < .0001$) when running in the shoe in comparison to both walking conditions (shoe & karate). The mean surface area for the lateral midfoot was 17.7 cm^2 for karate walk, 24.3 cm^2 for shoe walk, and 29.6 cm^2 for shoe run.

CONCLUSION

The results indicate that motion control shoes, designed with a duo-density midsole and a firm heel counter, cannot prevent increased movement of the medial and lateral midfoot when running. This would suggest that sagittal plane movement of the midfoot is occurring within the motion control shoe while running.

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LATERAL PLANTAR LOADING DURING CUTTING CORRELATES TO DECREASED FOOTWEAR RATING

Kevin R. Ford (1), Neil A. Manson (2), Gregory D. Myer (1), Richelle C. Gwin (3), Robert S. Heidt, Jr (3), Timothy E. Hewett (1)

1. Cincinnati Children's Hospital Medical Center, University of Cincinnati, Cincinnati OH, USA
2. Horizon Health, Saint John, NB and Dalhousie University, Halifax, NS, Canada
3. Wellington Orthopaedic and Sports Medicine, Cincinnati, OH, USA

INTRODUCTION

Plantar loading patterns have been shown to differ among cleat styles during complex sport maneuvers (Orendurff *et al*. 2008). The choice of footwear style, however, is often based on individual perceived ratings. The quantification of plantar loading patterns may help explain perceived footwear ratings during sports. The purpose of this study was to identify the relationship between foot loading patterns and perceived footwear rating.

METHODS

Seventeen male football players participated in the data collection (age: 16.9±0.9 years; height: 181.4±5.5 cm; mass: 79.8±10.9 kg). Each athlete was fitted with a new model of a noncleated turf American football training shoe (Nike Astrograbber). A flexible in-shoe pressure distribution measuring insole (Pedar, Novel) was inserted into the right shoe and sampled at 99Hz. Athletes ran a slalom course, modified from Eils *et al.* (2004), on a synthetic turf field with rubber infill. Nine side-cut steps were collected and analyzed with separate masks (medial and lateral heel, medial and lateral midfoot, medial, central and lateral forefoot, hallux, and the lesser toes). The force time integral in each individual region was divided by the force time integral for the total plantar foot surface in order to determine the relative load in each region.

Following the cutting trials, the athletes' perception of the footwear was assessed via visual analog scale (VAS) where 0mm = none/worst and 100 mm = complete/best. Categories included comfort, control, stability, traction, and safety. Athletes were also questioned whether they felt comfortable wearing and performing in the shoe at full intensity. One-way ANOVA was used to compare relative load and peak pressure between subjects that would wear (WW, n=11) and would not wear (NW, n=6) the noncleated shoe used during testing. Correlation coefficients were used to compare the relationship between pressure distribution and VAS.

RESULTS AND DISCUSSION

NW had increased relative loading in the lateral midfoot (p=0.002) and lateral forefoot (p=0.045) compared to WW. Peak pressure was significantly decreased in NW compared to WW at the medial forefoot (p=0.029). Increased lateral forefoot relative load significantly correlated to lower perceived ratings in control (r=-0.72), stability (r=-0.73), traction (r=-0.60) and safety (r=-0.66) (Figure 1). The perceived rating of comfort was not significantly correlated with relative load or peak pressure measures.

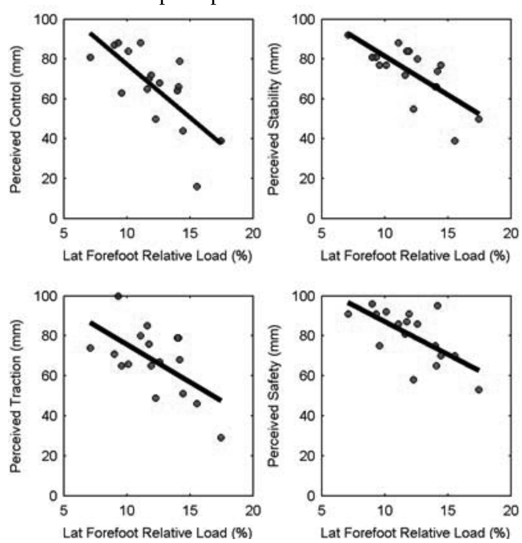

Figure 1: Relationship between visual analog scale and lateral forefoot relative load.

The differences in medial and lateral loading between groups likely relate to decreased stability and control during the cutting. The subset of athletes that did not feel comfortable performing in the noncleated footwear (NW) may have been accustomed to cleated footwear which led to greater lateral loading. Medial and lateral loading patterns have been found to differ with foot type (Queen *et al.* 2009). Additional investigation into the effects of foot type on perceived footwear rating and plantar loading are warranted.

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CROSSOVER SECOND TOE DEFORMITY: A ROBOTIC CADAVERIC STUDY OF INCREASED SECOND METATARSAL LENGTH AND PLANTAR PRESSURE

Eric C. Whittaker (1), Joel R. Weber (1), Patrick M Aubin (1,2), Bruce J. Sangeorzan (1,3), William R. Ledoux (1,3,4)

1. VA RR&D Center of Excellence, VA Puget Sound Health Care System, Seattle, WA
2. Department of Electrical Engineering, University of Washington, Seattle, WA
3. Orthopaedics & Sports Medicine, University of Washington, Seattle, WA
4. Mechanical Engineering, University of Washington, Seattle, WA

INTRODUCTION

The crossover toe deformity refers to an unstable second metatarsophalangeal joint (MTPJ) that leads to a progressive migration of the second toe in a dorsal and medial direction (Coughlin, 1993). The etiology is uncertain, but as summarized elsewhere (Kaz, 2007), it has been proposed that a long second metatarsal causes abnormal loading (i.e. increased plantar pressure) beneath the second MTPJ during stance phase. Over time, this causes MTPJ pain, agitation, medial collateral and lateral collateral ligament rupture, and toe migration. In the current study, we used a robotic gait simulator (RGS), cadaveric specimens and a surgical metatarsal lengthening process to further investigate this relationship. We hypothesized that lengthening of the second metatarsal would increase plantar pressure beneath the second MTPJ and reduce the peak pressure underneath the first metatarsal head.

METHODS

Dissection, including mid-tibia transection, exposure of the second metatarsal dorsally, and separation of nine extrinsic foot tendons, was performed for six specimens. Elongation of the second metatarsal was achieved through transection, a custom-made aluminum support bracket, and aluminum spacers of varying lengths (control, 2mm, 4mm, 6mm, 8mm). A robotic gait simulator (RGS) simulated physiologic tibial motion, tendon loading, and ground reaction forces (GRF) on the cadaveric feet. Stance phase was scaled to 10 seconds and the vertical GRF to half body weight. The GRF was measured with a force plate mounted to the RGS, and a Novel emed-sf pressure plate was mounted in series with the force plate. Custom pressure masks were created from weight bearing radiographs of each foot and used to determine the peak pressure (PP) and pressure-time integral (PTI) under the first and second MTPJs during gait. Statistical analysis was performed with a linear mixed effects model.

RESULTS

With increased spacer length, PP decreased under the first metatarsal head and increased under the second metatarsal head (Figure 1). Second metatarsal PP and PTI were positively correlated with an increase in second metatarsal length ($p = .0005$, $p < .0001$). First metatarsal PP and PTI were significantly negatively associated with second metatarsal length ($p = .029$, $p = .024$).

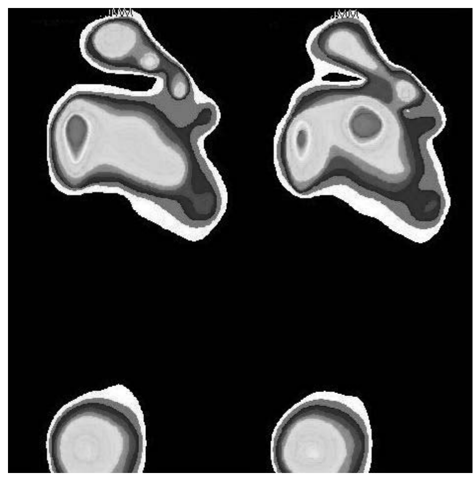

Figure 1: Peak plantar pressure for one foot in the RGS; left: control spacer, right: 8mm spacer.

DISCUSSION

Our results support the hypothesis that second metatarsal length is positively associated with plantar pressure underneath the second MTPJ. Differing from the first MTPJ, the second MTPJ capsule and soft tissue structures are not designed to support the large stance phase loads. It is thought that prolonged exposure to the plantar pressures measured in our study would lead to the observed joint subluxation and dislocation seen in crossover toe patients. Overall, these findings give a better understanding of the mechanisms responsible for the development of the deformity as well as clarify controversial associations of the deformity found in the literature.

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HYPOTHENAR PRESSURE MAPPING PROVIDES INSIGHT FOR REDUCING ULNAR NERVE COMPRESSION IN CYCLISTS

Josh Slane (1), Mark Timmerman (2), Heidi Ploeg (1), Darryl Thelen (1)

1. Department of Biomedical Engineering, University of Wisconsin-Madison, Wisconsin, USA
2. School of Medicine and Public Health, University of Wisconsin-Madison, Wisconsin, USA

INTRODUCTION

Chronic ulnar nerve compression is believed to be a primary cause of sensory and motor impairments of the hand, which is common among cyclists. Prevention techniques include frequently changing hand position and wearing padded gloves. However to date, there has been no scientific examination into the effectiveness of these measures (Kennedy, 2008).

The purpose of this study was to evaluate the effects of hand position and glove padding on pressure over the ulnar nerve. Specifically, we considered three hand positions commonly used on road bicycles. We also compared gloves with either 3 mm or 5 mm gel or foam padding placed over the hypothenar eminence, thenar eminence and metacarpal heads.

METHODS

Thirty-six healthy adults (18 male, 18 female) were tested (38.9±13 yrs, 74.2±13.9 kg and 174±9cm). A stationary cycle was adjusted to match the dimensions of each subject's personal road bicycle. Subjects performed a series of trials in which hand position (Fig 1) and glove type were randomly varied while power and cadence levels were kept consistent. A pressure sensitive mat (229 sensors, Elastisens - FO44, Novel GmbH) was used to record pressure distribution over the hypothenar region of the hand. This region is where the ulnar nerve enters the hand through Guyon's Canal and is most susceptible to external compression (Black, 2007). Pressure from 12 consecutive pedal strokes were averaged to quantify both the pressure distribution and peak pressure over the hypothenar region.

Three-dimensional images of the subject's hand with and without the pressure mat were obtained using a laser scanner (Shape-Grabber Inc). These laser scans were registered with subject-specific pressure images to relate pressure profiles to the underlying anatomy (Fig 1). The compressive stiffness of the foam and gel padding was measured separately with a materials testing machine and a 15 mm diameter indenter.

RESULTS AND DISCUSSION

Padding in the hypothenar region of the glove reduced peak pressures found in the no glove condition from 21-28%, with the highest pressure reduction achieved using 3 mm of foam. The foam padding/glove system was found to be ~50% less stiff than the gel, suggesting that better pressure reduction was achieved using a more compliant interface. Absolute pressure magnitudes were greatest with the hands in the drops position (128-168 kPa), which were significantly greater ($p < 0.05$) than those found in the tops and hoods positions. Pressure magnitudes did not significantly vary between male and female cyclists.

To our knowledge, this is the first study that has measured high-resolution pressure distributions on cyclist's hands while riding. The data obtained is important for establishing a quantitative, scientific approach to design interventions aimed at reducing the prevalence of hand impairments in cyclists.

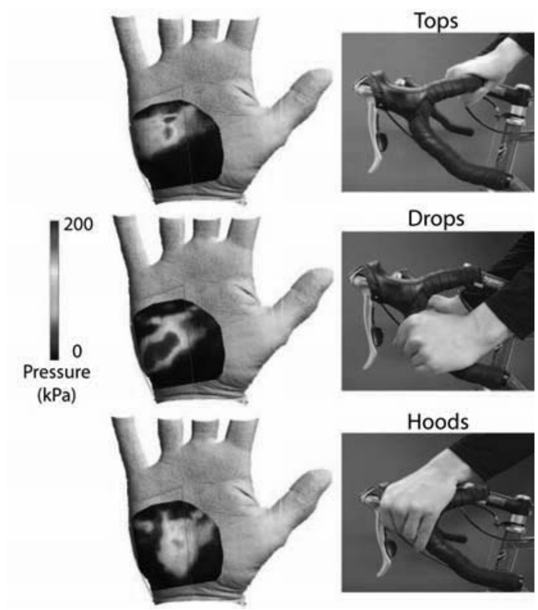

Figure 1: Pressure concentrations near Guyon's Canal were observed with the hands in the drops and hoods positions.

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USING CENTRE OF PRESSURE (COP) DATA TO IDENTIFY GOLF PUTTING TECHNIQUES

Pat McLaughlin and Russell Best

Institute for Sport, Exercise and Active Living, Victoria University, Melbourne, Australia

INTRODUCTION

Consider the scoring system in golf. On each hole, playing to "par" allows the golfer two putts per hole – a total of 36 putts over an 18-hole round, or about half the total number of shots allocated for par. Considering that players have 13 other clubs in their bag, and would tend to use most of them at some stage, the putter is the most often used piece of equipment that a player possesses. Whilst this simplistic logic may be an exaggeration for the elite golfer, data indicates that even at the elite level, around 38% of all strokes are putts (2009 PGA Tour data, www.pgatour.com)

Research into putting performance follows the same common theme - players are separated into groups based on handicap level. The question of interest seemingly "how do the elite/expert/low handicap golfers compare to the novice/high handicap golfers?" All authors assumed that putting performance was related to handicap level. On putting performance alone this was not definitively supported. Also, the authors often analysed only putts that were successful - that is putts that went in the hole (Paradisis and Rees, n.d.) or stopped over the target (Delay et al., 1997; Haltom, 1994; Zafiroglu, 1994). However, these types of studies do not investigate technique differences. In contrast, Pelz (2000) points out that there have been many different putting techniques used by professionals over the years.

The aim of the present study was to define if more than one putting technique truly exists in a sample of golfers of different handicap levels.

METHODOLOGY

A total of 38 players participated in the testing sessions. These players were members of a private suburban golf club in Melbourne and all were experienced golfers. The average age of the sample was 55.3 (±17.8) years and average handicap 15.3 (±6.9).

Simultaneous video (50Hx perpendicular to line of putt) and pressure data (16 x 16 pliance® pressure mat, 50Hz, novel, Germany) were collected for each putt. The data collection process required each player to putt whilst standing on the pressure mat which was positioned on the practice putting green. Each participant completed five putts at a hole cut into the green 4m (13ft) away.

RESULTS AND DISCUSSION

Combining the kinematic, temporal and COP data, 62 parameters were available for analysis. Players were not divided according to handicap, putting trials were analysed individually and all data entered into cluster analysis methods to define putting techniques. This allowed the creation of technique based on the movements exhibited by players during the putting stroke.

For cluster analysis, the most influential parameters in determining cluster membership were related to movement of the COP in the medio-lateral direction (COPx), namely: range of movement of COPx in the backswing; range of movement of COPx in the downswing; maximum velocity of COPx in the downswing and velocity of COPx at the time of ball contact (this was the most influential parameter).

Two distinct putting techniques were identified by cluster analysis:

1. Less movement (relative to cluster 2) of COPx in the backswing and downswing phases with velocity of COPx at ball contact closer to zero (on average). Low COPx velocity.
2. Larger movement (relative to cluster 1) of COPx in the backswing and downswing phases with velocity of COPx at ball contact non-zero. High COPx velocity.

CONCLUSIONS

Analysis of COP data is able to distinguish putting techniques in an amateur group of golfers. Cluster analysis methods allow performance techniques to be distinguished based on parameters related to execution of the skill, rather than handicap.

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EFFECT OF MATTRESS FOR PREVENTION OF DECUBITUS, BED POSITION AND SUBJECTS' BMI ON INTERFACE PRESSURE

Tae Soo Bae (1), Mu Seong Mun (1)

1. Biomechanics Team, Korea Orthopedics and Rehabilitation Engineering Cener

INTRODUCTION

Patients with poor blood circulation carry the risk of decubitus since it is generally unavoidable to use various bed positions in order to treat patients. Several mattresses have been introduced in order to prevent decubitus caused as such. However, since there are diverse kinds of them and many different methods to use them, patients' families and nurses experience many difficulties in selecting and using mattresses for prevention of decubitus. Therefore, in this study, the effects of diverse mattresses for prevention of decubitus, bed positions and subjects' body mass index (BMI) on contact pressures in the areas such as the scapula, the hip and the feet- heels was quantitatively measured, analyzed and compared with each other.

METHODS

Experiments were conducted on normal persons with under-, normal-, and over-weight (5males/ group) who were free of vascular system diseases. In this study, a general mattress and four types of mattresses for prevention of decubitus (latex, bubble, alternative, and air mattress) were largely used and five different bed positions were defined as experimental conditions using three-step electromotive beds. (Figure 1) To measure contact pressures, a order-made pressure measuring system (bedsize, Pliance FTX, v12.1.36, Novel Gmbh) was used. The mattress and the pressure sensor were stabilized for around 30 minutes and an additional process for stabilization for 30 minutes was included after the subject was laid on the mattress before the experiment in order to minimize the contact pressure measurement errors caused by air movements in the mattress. The same processes were repeated for five different bed positions. Maximum peak pressure (MPP) was separately measured using the predefined mask on supine body areas.

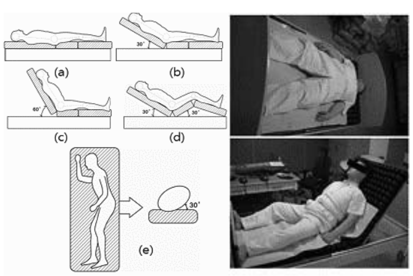

Figure 1: Various positions of Bed mattress and experiments (a)ground (b) Head 30 degress (c) Head 60 degree (d) Head 30 degree and foot 30 degree (e) Side

RESULTS

In the case of flat positions, the result indicated that only the bubble mattress was acceptable to the subject with normal body weight, latex or air mattress to the over-weighted subject and only air mattress to the low-weighted subject. In the case of the subject with normal body weight for 30^{o} position in the head area, MPP lower than the reference value were measured in all the mattresses except for latex and bubble mattress and in the case of the over-weighted subject, all mattresses were usable except for general mattresses, alternatives, and bubble mattresses but in the case of the under-weighted subject, all the mattresses showed MPP higher than the reference value. In the case of 60^{o} position in the head area, only the latex mattress was acceptable to the subject with normal body weight while all the mattresses showed MPP higher than the reference value in the case of both the over-weighted subject and the under-weighted subject. In the case of 30° positions in the head area in the leg area respectively, it was indicated that only general mattress and alternative mattresses were acceptable to the subject with normal body weight while only general mattresses, latex, and air mattresses were acceptable to the over-weighted subject. In the case of the low-weighted subject, latex, alternative, and air mattress showed MPP lower than the reference value. Lastly for side position, all the mattresses showed results exceeding the reference value in the case of the subject with normal body weight and the over-weighted subject while it was indicated that alternative and air mattresses were acceptable to the over-weighted subject.

Table 1: Experiment for measuring interface pressure. Each symbol(◎,◇, ◆, □) means acceptable position and mattress with respect to contact peak pressure of 32 mmHg

	normal					Over-weight					Under-weight				
	□	□	□	□	□	□	□	□	□	□	□	□	□	□	□
general		□							□						
latex			□			□	□		□					□	
bubble	□														
alternative		□								■				□	
air		□		□		□	□		□	■	□			□	

ground ② Head 30 deg ③ Head 60 deg ④Head-Foot 30 deg ⑤ side

CONCLUSION

Air mattresses showed generally even effects to prevent decubitus across all the bed positions followed by latex mattresses. It was indicated that decisions to use bubble or alternative mattresses for prevention of decubitus should be made more carefully.

A STUDY OF THE EFFECTS OF GEL LINER THICKNESS ON IN-SOCKET RESIDUAL LIMB PRESSURES IN TRANS-TIBIAL PROSTHESIS USERS

Rebecca Stine, MS (1-2), Andrew Hansen, PhD (2-3), Kerice Tucker, BS (2)
Erin Boutwell, MS (2), Mark Edwards, CP, MHPE (2,4), Steven Gard, PhD (1-2)

1. Jesse Brown VA Medical Center, Dept. of Veterans Affairs, Chicago, IL USA
2. Northwestern University Prosthetics Research Laboratory and Rehabilitation Engineering Research Program, Chicago, IL USA
3. Minneapolis VA Medical Center, Dept. of Veterans Affairs, Minneapolis, MN USA
4. Otto Bock Health Care, Minneapolis, MN USA

INTRODUCTION

Polliack et al. (2000) pointed out that measuring pressure between the prosthetic socket and residual limb could provide valuable information for the process of socket fabrication, modification, and fit. Previous studies have investigated how interface pressure distributions change as a result of prosthetic alignment (Seelen, 2003), adaptable ankle-foot components (Wolf, 2009), and walking surfaces such as ramps and stairs (Wolf, 2009 and Dou, 2006). Prosthetic liners may improve comfort by more evenly distributing pressure at the residual limb (Boutwell, 2009). It is also likely that changes in liner thickness will affect interface pressure. Therefore, the purpose of this study is to examine the effects of gel liner thickness (3mm and 9mm) on residual limb pressures.

METHODS

Eleven subjects with unilateral trans-tibial amputations agreed to participate in the study and signed IRB-approved consent forms. Mean age of the subjects was 56 ± 9 years with at least 6 months experience using a prosthesis and all were able to walk without undue fatigue. Two Alpha prosthetic gel (thermoplastic elastomer) liners (Ohio Willow Wood, Mt. Sterling, OH) were tested. Sockets for both gel liners were made for each subject using a CAM system called Squirt Shape (Rolock, 2000) that was developed in our laboratory. All other prosthetic components were the same for each subject. The same certified prosthetist fit and aligned the socket to each subject. Subjects were given an accommodation period of at least two weeks on each prosthesis prior to testing. A gait analysis was performed with the person walking at his normal self-selected speed for each liner condition. Mean walking speed across all subjects and all conditions was 1.14 ± 0.14 m/sec.

Prior to each gait evaluation, 6cm X 3cm pressure sensors (Pliance, Novel Electronics, Inc.) were placed over the patellar tendon (PT) region, the anterior distal tibia (DT), and the fibular head (FH). Pressure data were synchronized with the motion data to calculate gait cycles and recorded at 120 Hz.

RESULTS

Data from nine of the subjects were analyzed due to incomplete data sets from two subjects. Peak pressure values averaged over multiple gait cycles for each sensor matrix were calculated. Peak values during the first 40 percent of the gait cycle and the second 40 percent of the gait cycle were recorded and compared between the 3mm and the 9mm liners (Table 1). A nonparametric Wilcoxon signed ranks test with a Bonferroni correction was used to determine any significant median differences between the peak pressure values for each sensor matrix. No significant differences were found between liners.

Conditions		Median (IQR)		p-value
		3mm	9mm	
PT	Peak 1	210 (67)	167 (106)	0.575
	Peak 2	229 (78)	191 (116)	0.260
FH	Peak 1	258 (63)	203 (78)	0.110
	Peak 2	327 (61)	179 (80)	0.066
DT	Peak 1	217 (110)	263 (139)	0.678
	Peak 2	158 (87)	168 (129)	0.678

Table 1: Peak pressure values for each sensor matrix for each liner. (IQR = inter quartile range)

DISCUSSION AND CONCLUSION

Median peak pressure values were reduced for both peaks over the PT and the FH with the thicker liner, although the small sample size may have contributed to the lack of statistical differences between the two liners. The DT peak pressure increase may indicate that thicker gel liners provide a different pressure distribution pattern. Patient preference should be considered when determining liner thickness.

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FORCE AND IMPULSE OF KEYSTROKE DURING PIANO PLAYING. DIFFERENCES AMONG PROFESSIONAL PLAYERS IN CLASSICAL PASSAGES

H.Gaertner(1), R. Pozzo(2), H. Riehle (1)

1. University of Konstanz, Germany
2. University of Tor Vergata Rome, Italy

INTRODUCTION

An understanding of the force and impulse acting on the key by the finger associated to the sound generated is important because it makes the difference between interpretation and pure mechanical played notes. Some studies have been made in order to determine the force applied on the key and the relation with the level of loudness and the type of finger touch (Harding 1989, Kinoshita 2007, Parlitz 1998), providing important informations in relation to *piano* (*p*), *mezzo forte (mf)*, and *fortissimo (f f)* levels with *staccato* and *legato* touch. In the present study, using pressure sensors applied on the key surface, we made an analysis of the relationship between the force-time characteristics and the performance of pianists playing passages of classical music.

METHODS

Fifteen pianists (8 female and 6 male) with different level of expertise (from IP: winning international prices to TL: top level) played the first 2 measures of Schubert's Wanderer Phantasie (4 times in *ff* and 4 times in *pp*) on a grand piano (Bechstein). They were asked to play with theirs best interpretation and constancy; the finger sequences (fingering) were identical for all subjects. Five pressure sensors (S2013, 20x45 mm^2, <1,2 mm, 10-600 KPa, Pliance-Novel) were attached to the most relevant keys and collected at 300 Hz. Video records were taken via 3 Casio Cameras at 300 Hz for later 3-D analysis. Sound track was recorded via professional tools.

RESULTS

Peak force and impulse were calculated for each sensor and for each repetition. TL subjects showed 50% higher peak force in *ff* (9,6 N vs 6,4 N) but similar values in *pp* (2,7 N vs 2,6 N) compared to IP subjects. All pianists have a stable reproducibility of peak force as described by the coefficient of variation (CV 10%-20% for IP; CV 14%-27% for TL in *ff* and *pp* respectively).

Indeed, very high differences were noted in the total impulses. Overall average values for TL subjects were 73% (0,91 Ns vs 0,24 Ns in *ff*) and 69% (0,33 1 Ns vs 0,11 Ns in *pp*) higher then for IP. Significant difference were also found in the CV of impulse structure (11%-18% for IP vs 33%-35% for TL).

An efficiency index was calculated by computing the proportion of the impulse up to the peak force to the total impulse. This index reveals the capacity of the pianist to produce the desired loudness without exert unnecessary force. The relationship between this parameter and the number of touches played on each keys could reflect the stability of motor pattern and force sensibility of the pianist. Fig 1 shows exemplary the difference of impulse efficiency between IP and TL pianist (sbj. 4 and sbj. 12). It easy to recognize that for almost all the strikes subject 4 shows a significant better regulation of force production.

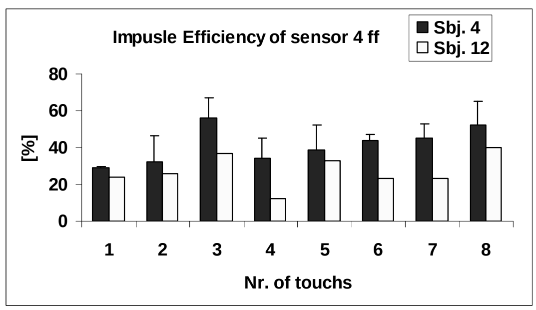

Figure 1. Mean and SD values of efficiency for two subjects with different level of expertise

Moreover an analysis of possible relationships between the force produced during playing and the force capacity of the fingers was carried out by testing maximal voluntary contraction force of each finger under isometric controlled conditions.

Sound characteristics were studied by acoustical analysis (Samplitude V7.0) and by a quality questionnaire administered to professional judges.

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THE EVOLUTION OF HUMAN RUNNING

Daniel E. Lieberman (1)

1. Department of Human Evolutionary Biology, Harvard University, Massachusetts

Ever since the human lineage diverged from the African apes, hominins have been bipeds of some sort. Comparative and fossil evidence suggest that the earliest hominins were capable, habitual bipedal walkers but were also adept at climbing trees. At some point, however, hominins lost the ability to climb trees very well, and became superlative long distance runners. Comparisons of human endurance running performance with other mammals show that we excel at speed, distance, and the climatic context in which we can run. Further, human distance running capabilities far exceed those of any other primate, and they match or even surpass the best mammalian runners in hot conditions over very long distances. These capabilities raise several questions, among them when humans became long distance runners, why these abilities evolved, and how the evolutionary history of endurance running may help avoid injury.

A review of the fossil evidence suggests that features which improve endurance running performance in humans first appear about 2 million years ago in the genus *Homo*, mostly in the species *H. erectus* (Bramble and Lieberman, 2004). These features include changes in the foot (e.g., shortened toes and the development of a full longitudinal arch), the hip (e.g., expansion of the cranial portion of the gluteus maximus), and the head (e.g., enlargement of the anterior and posterior semicircular canals). Many important adaptations for endurance running, however, are impossible to assess in the fossil record, and it is not known whether *H. erectus* was as capable a runner as modern humans.

A review of the ethnographic and archaeological records suggests that the primary advantage of long distance running was for persistence hunting. In this method of hunting, runners alternatively chase quadrupeds at a gallop speed, thus preventing them form panting, and track the animals (usually at a walk). Since quadrupeds cannot simultaneously pant and gallop, this form of hunting drives the animals into a state of hyperthermia making it easy to dispatch them without significant projectile technology, which was lacking until very recently in human evolution.

Finally, studies of habitually barefoot runners indicate that the human foot, like the rest of the body, is well designed for endurance running. People can run long distances at high speeds quite comfortably over very hard surfaces without using cushioned, high-heeled shoes. A barefoot "style" of running typically involves a forefoot or midfoot strike, both of which generate no discernable impact transient because of increased limb compliance and less effective mass at the moment the foot collides with the ground (Lieberman et al., 2010). This sort of light and gentle landing requires more calf and foot muscle strength, but generates less impact than heel-toe running in modern running shoes, and is presumably how humans ran for millions of years. Such insights raise the evolutionary medical hypothesis that more minimal shoes which foster a more natural, barefoot running style may have advantages over more cushioned shoes that facilitate heel striking.

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P1

THE EFFECT OF SPECIAL SHOE INSERT DESIGNED FOR DIABETIC PATIENTS ON PLANTAR FOOT PRESSURE DISTRIBUTION

Moshe Ayalon (1), Iftach Hetsroni (2), Meir Nyska (2)

1. The Zinman College of Physical Education, Wingate Institute, Israel
2. Meir Hospital, Sapir Medical Center, Kfar Saba, Israel

INTRODUCTION

Plantar callosities and ulcers are one of the main factors leading to serious complication in the diabetic foot. Increased plantar foot pressures are the main cause for plantar keratosis leading to foot ulcers in diabetic peripheral neuropathy. Common areas for ulcers are under the metatarsal heads and every effort should be done to reduce plantar foot pressures in trying to prevent callosities and plantar ulcers. Orthotic devices believed to be efficient in plantar foot pressure reduction.. In this study we present the effect of a novel orthotic insert that was designed to reduce the plantar pressure under the metatarsal heads. We hypothesized that with this special insert a reduction in planar pressures under the metatrsal heads will be seen compared to standard costum-made orthotics.

METHODS

Ten subjects were evaluated while walking (4.5 km/h) with: a) standard costum-made orthotics b) newly designed orthotic. The newly designed orthotic is based on integrating metatarsal bar and a pad (figure 1). In-shoe plantar pressure measurement was performed using Pedar-x system (Novel gmbh, Munich). The data of four steps for each leg were collected for each measuring condition. The average peak plantar pressure (N/cm^2) was calculated for 10 zones of interest of the foot. The analysis included comparison of standard-custom made orthotics to the newly designed orthotic.

Figure 1: The newly designed orthotic

RESULTS

The newly design orthotic reduced average peak pressures in the second and lateral metatarsals and increased average peak pressures in the heel, midfoot ,1st and 2nd toes (Figure 2).

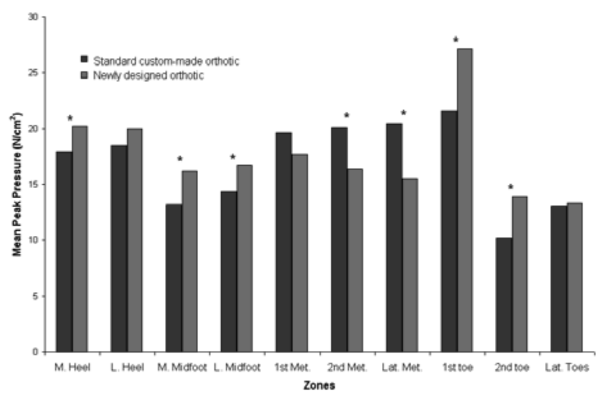

Figure 2: Comparasion of plantar pressure

DISCUSION

This study has shown that, on average, newly designed orthotic proved to reduce the plantar pressures in the 2nd metatarsal and lateral metatarsal by 22% and 31 % repectively. Being the metatrsals the most common area for callosities and plantar ulcers in a diabetic foot, the newly designed orthotic may be efficient in prevent ulceration and to improve wound healing. However, the tested orthotic increased the mean peak pressure in the mid foot and toes by 22% and 36 % repectively. This redistribution of load may put in risk especially the toes. Frequent monitoring may be required in order to make sure that not only callosities and ulcers improve in areas where plantar pressure is reduced by this device, but also that new callosities are not created in areas where pressure is increased with this special insert.

EFFECT OF GAIT SPEED CHANGES ON FOOT LOADING CHARACTERISTICS IN CHILDREN

Kerstin Bosch (1), Maike Westhues (2), Dieter Rosenbaum (1)

1. Movement Analysis Lab, Orthopaedic Department, University Hospital Muenster, Germany
2. University of Applied Science Muenster, Germany

INTRODUCTION

Plantar pressure measurements have been established as a method for the assessment of foot loading and functional restoration of foot-related problems in adults and children. For clinical problems, the effects of conservative or surgical treatment can be evaluated with repeated measurements that may help to demonstrate the changes induced by the chosen treatment option (Rosenbaum 1997). However, several factors have been identified as influential so that they should be taken into consideration.

For adults, gait speed changes may alter peak pressures and the regional load distribution but does not affect the whole foot uniformly (Rosenbaum 1994). For children in the early stages of gait acquisition, this effect has not been evaluated yet. Therefore, the present study investigated how changes in gait speed would affect foot loading characteristics in children.

MATERIALS AND METHODS

Twenty healthy children (11 male; 9 female) between 4 and 12 years of age participated in plantar pressure measurements at normal, slow and fast walking speeds. They were asked to walk barefoot across a capacitive pressure distribution platform (emed-ST) that was embedded flush in a 5.0 x 0.8 m walkway.

Gait speed was measured with infrared light gates (AF-Sport Timing System, Wesel; 10 ms resolution) that were adjusted to the child's shoulder height. After self-selected normal walking (**n**), the children were asked to walk slow (**s**) and fast (**f**) until 5 valid trials from both feet were stored at each gait speed.

Footprints were subdivided into 10 regions of interest: lateral and medial heel; lateral and medial midfoot; 1st, 2nd and 3rd-5th metatarsals; hallux; 2nd toe; toes 3-5.

As dynamic parameters peak pressure (kPa), contact time (ms) and maximum force (% body weight) were evaluated for the whole foot and the regions of interest.

The dynamic parameters were statistically tested by using the GLM (generalized linear model) for repeated measures. The level of significance was set at $p < 0.05$.

RESULTS

The average gait speed was 0.77±0.07 m/s for slow, 1.20±0.06 m/s for normal and 1.65±0.06 m/s for fast walking. The contact time decreased significantly from 808±118 ms (s) to 589±71 ms (n) and 443±44 ms (f, $p<0.001$). The maximum force increased from 105±11% body weight (s) to 115±11% (n) and 127±16% (f, $p<0.001$). The changes were most pronounced in the heel and hallux but showed even a decrease in the lateral midfoot and forefoot. The peak pressure of the whole foot increased significantly from 340±130 kPa (s) to 392±91 kPa (n) and 524±178 kPa (f, $p<0.001$), especially in the heel, 1st and 2nd metatarsals and toes but not in the lateral midfoot and metatarsals 3-5.

DISCUSSION AND CONCLUSION

The effects of systematically altering gait speed were demonstrated in children. Similar to the adults, the loading increase is not uniformly distributed across the whole plantar surface but is concentrated on the heel, 1st and 2nd metatarsals and the toes. The decreased loading in the metatarsals 3-5 that was seen in adults, could only partly be confirmed for the children.

In conclusion, the results underline the importance to ensure comparable gait speed for repeated measurements in a clinical follow-up of single patients also for children. If this cannot be achieved e.g. because of age-dependent developmental phases, comparisons should bear the potential changes in mind that were demonstrated in the present results.

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ACKNOWLEDGMENT

We wish to thank the children and their families for their voluntary participating.

EFFECTS OF INTENSE RUNNING TO EXHAUSTION ON THE IN-SHOE PLANTAR PRESSURE PATTERNS IN YOUNG MIDDLE-DISTANCE ATHLETES

François Fourchet (1), Luke Kelly (2), Cosmin Horobeanu (1), Grégoire P. Millet (3)

1. Health Centre, Aspire Academy for Sports Excellence, Doha, Qatar
2. Aspetar, Orthopaedic and Sports Medicine Hospital, Doha, Qatar
3. ISSUL Institute of Sport Sciences-Department of Physiology, Faculty of Biology and Medicine, University of Lausanne CH-1015, Lausanne, Switzerland.

INTRODUCTION

It has been reported that intense exhaustive running increased forefoot loading in experienced adult runners and may explain some foot/ankle overuse injuries (Weist 2004). However to our knowledge no studies investigated that topic in adolescent athletes. The aim of this study was to examine the effects of intense running to exhaustion (Tlim) on the in-shoe plantar pressure patterns in young elite middle-distance athletes

METHODS

Eleven male adolescent distance runners (age: 16.9±2.0, body mass: 54.6±8.6 kg, height: 170.6±10.9 cm, maximal aerobic speed: 18.7±1.5 km.h-1) performed a constant-pace Tlim at 95% of their maximal aerobic speed (17.8±1.4 km.h-1) on a treadmill (h/p/Cosmos, Nussdorf-Traunstein, Germany).

Plantar pressure measurements were performed with one sensor insole worn in the right running shoe (Novel, Munich, Germany).

Data were recorded first at 1 minute after the start (1 min) and 30 s prior exhaustion (max). A regional analysis was performed utilizing nine separate ‘‘masks’’ or areas of the foot (Groupmask Evaluation, Novel, Munich, Germany).

Mean area (mA) and contact time (CT) were determined for the nine selected regions. In addition, the relative load (RL%) was calculated as the force time integral in each individual region divided by the force time integral for the total plantar foot surface.

One Way ANOVA Repeated Measures was performed and Tukey post-hoc analyses were used. Statistical significance was accepted at $p < 0.05$.

RESULTS AND DISCUSSION

Running time to exhaustion was 8.8±3.4 min. Between 1 min and max, significant increase in CT ($P < 0.001$) was observed in all regions except Hallux. mA (Fig. 1) and RL% (Fig. 2) increased significantly under the medial midfoot area. The other measured parameters did not change.

The longer CT in fatiguing conditions may result from a reduced strength and stretch-shortening-cycle efficiency of plantar flexor muscles. In adults, the following over load of the medial arch of the foot is the first phase of a compensatory midfoot landing strategy (Willson 1999) and induced also medial and central heads of metatarsal harmful overload (Weist 2004). Conversely no forefoot overload effect was reported in the present results for adolescent athletes.

Then we may assume that the medial arch of the foot is a structure of primary interest to absorb the increased foot loadings in fatiguing conditions in young elite middle-distance runners. Therefore reinforcement of the muscles supporting the medial arch might be implemented in foot/ankle injuries prevention programs.

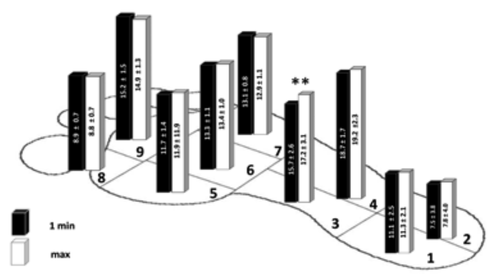

*Figure 1: Mean +/- SD mean area (mA) at 1 minute after the start (1 min) and prior exhaustion (max).**, P < 0.01*

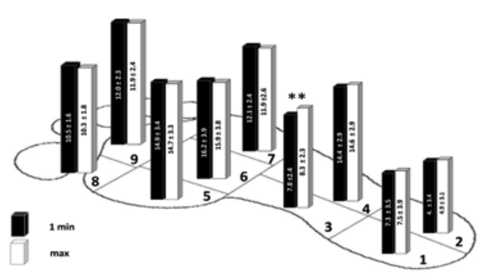

*Figure 2: Mean +/- SD relative load (RL%) at 1 minute after the start (1 min) and prior exhaustion (max).**, P < 0.01*

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CLINICAL APPLICATIONS OF A FOREFOOT CONIC CURVE MODEL

Howard Hillstrom (1,2), Andrew Kraszewski (1), Philip Demp (2), Betty Chow (1), Mark Lenhoff (1), Jinsup Song (2), Benjamin Heilman (2), Sonali Rajan (1), Andrea Woodley (1)

1. Leon Root MD, Motion Analysis Laboratory, Hospital For Special Surgery, New York, NY
2. Temple University School of Podiatric Medicine, Temple University, Philadelphia, PA

INTRODUCTION

Biomechanical differences between healthy feet (pes planus, rectus, and cavus) and those with pathology (diabetic hallux valgus) may be related to their fundamental forefoot geometries. The specific aim of this project is to quantify and compare conic curve parameters among healthy and diabetic feet.

METHODS

We hypothesized that asymptomatic healthy individuals with pes planus, rectus, and cavus feet and diabetics with hallux valgus will show differences in conic curve equation parameters. Sixty-one healthy test subjects were stratified according to resting calcaneal stance position and forefoot to rearfoot relationship into their foot types at HSS. Patients with Diabetes and hallux valgus were evaluated at TUSPM. The x-y-z metatarsal head (MTH) locations were acquired with 3D motion analysis. A unique conic curve[1] was fit to each set of points and its resulting equation normalized by the first term (see Figure 1). Parameters of interest were B, C, D, E, and curve eccentricity. Data were analyzed with a univariate mixed-effect analysis of variance model followed by Bonferroni post-hoc t tests. Foot type and trial were modeled as fixed and random effects.

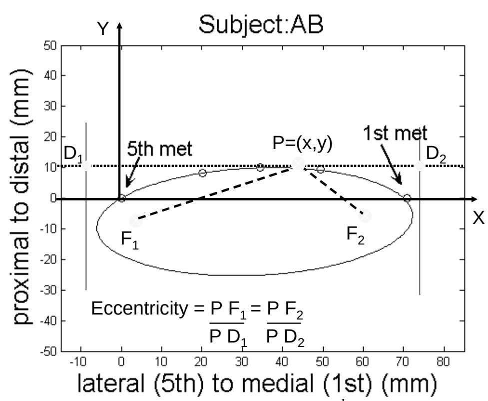

Figure 1: 2D Forefoot Model. The 5^{th} MTH was on the origin (0,0) and the 1^{st} MTH along the x-axis. Each conic (blue curve) is defined uniquely by the equation: $Ax^2+Bxy+Cy^2+Dx+Ey$, where A=1.

RESULTS

The conic model did not significantly distinguish among foot type with the exception of parameter D, the x-axis scaling factor in the conic curve equation; it distinguished rectus from diabetic feet with hallux valgus. As shown in Figure 2 several structural variables had significant Pearson correlations with model parameters; in particular parameter D was significantly correlated to most.

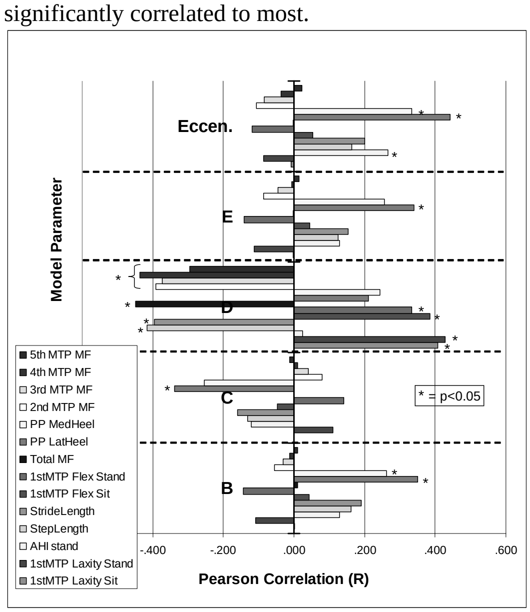

Figure 2: Pearson Correlations

DISCUSSION

The geometric forefoot model was significantly correlated with AHI, 1^{st} MPJ flexibility and laxity, total maximum plantar force and peak pressure beneath the heel. This suggests that simple models may be useful to differentiate normal vs. pathological feet.

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Variation of tactile cues reduce Nociceptive Capacity of Plantar Irritating Stimulus impact on walking gait

Marc JANIN PhD Pod. ; Podiatrist[12] and Philippe DUPUI PH-MCU [1]

1. Laboratory of de Physiology, Toulouse Medical College, 133 Route de Narbonne, 31062 Toulouse, France
2. Podiatrist office, 7 Rue de Treguel, 86000 Poitiers, France

INTRODUCTION: Nociceptive Capacity of a Plantar Irritating Stimulus (NCPIS) affects plantar cutaneous somaesthesia and influences kinetic posture by modifying control systems (1,9). This NCPIS doesn't express itself like reason of pain in podiatrist consultations but affects the postural control [1]. To understand better the repercussions of the NCPIS on gait analysis, we analysed the relationship between dynamic baropedograpphy through Latero-Medial Index (L/M , 2-4,7) of area: L/MA and force: L/MF ; Fig. n°1) of 15 subjects with NCPIS (pathological: P) and without NCPIS (non pathological: NP). Nociceptive sensory of NCIPS on walking track (H) was reduced by foam (F) during walking before and/or on baropedography force plate (8). 4 levels of sensory information were applied: start walking (S) and arrival (A) on the force plate were delivered the same way with the tactile information: hard (SHAH), foam (SFAF); start walking and arrival on the force plate crossed each other (SHAF ; SFAH). ANOVA and Schéffe post-hoc tests were used to test the possible difference between P and NP for L/MF and L/MA.

RESULTS: No significant differences were present among the L/MF: between P and NP ($F=0.080$; $p=0.779$), between H and F ($F=0.1622$; $p=0.190$), no inter action P/NP and H/F ($F=0.513$; $p=0.674$). Results were significantly different with regard to its L/MA: between P and NP ($F = 63.90$; $p<0.0001$), between H and F ($F= 299.37$; $p<0.0001$) and inter action P/NP and H/F (($F = 295.83$; $p<0.0001$). Post Hoc test showed no significantly difference into P and NP when start walking was on foam independently of the quality of arrival (H or F).

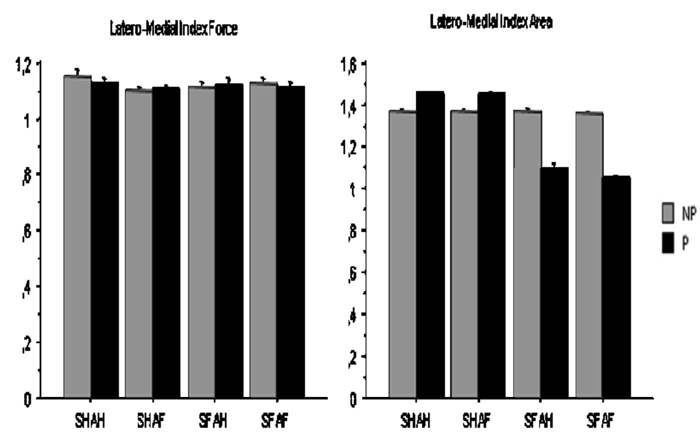
CONCLUSION: No modifications of L/MF are present because L/MF are modified when weight add up like the literature explain. For L/MA differences are observed. For P, arrival on foam reduces nociceptivce cue for NCPI (1,8). Their sensory information will come back as normal. Normalization of sensory feedback will be improving muscular responses (5,7). Their kinetic foot pattern will find back the pattern of subject to NP (no difference into SFAH, 5-6, 8).

Those results confirm the potential nociceptive capacity of the NCPI on postural control and posturo kinetic capacity.

FIGURES

Figure 1: Baropodometry to calculated L/M Index.

Figure 2: Latero-Medial Index Force and Aera. NP: non Pathological; P: pathological SHAH start and arrival on hard, SHAF start on hard and arrival on foam, SFAH start on foam and arrival on hard, SHAH start and arrival on foam.



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EFFECT OF INTERLOCKING PATTERN IN ELECTRICAL BED ON THE PREVENTION OF BEDSORE

Hyunho Choi, Jong Hyun Kim, Jae Soo Hong, Keyoung Jin Chun, #Dohyung Lim

Gerontechnology Center, Korea Institute of Industrial Technology, Chungnam, Korea
#Corresponding and Presenting Author: dli349@kitech.re.kr

INTRODUCTION

This study was motivated by a demand to prove scientifically the fact that applying an optimum interlocking pattern to electrical beds can be helpful to prevention of bedsore in bedridden patients (Evan and Loyal, 2007; Maki, 2009).

MATERIALS AND METHODS

Participants: Following Institutional Review Board approval, seven healthy female (65.6 ± 4.1 years, 151.2 ± 3.8cm, and 62.6 ± 5.6kg; mean±SD) were participated.

Electrical Bed Selection: Two electrical bed (Type A: KQ-86320, Paramount Bed, Japan and Type B: PZB-H3S, Platz Bed, Japan) were chosen by the analyses of market price, market share, sales volume, user approaching, brand awareness, test convenience, etc.

Electrical Bed Operation: Automatic interlocking patterns built-in to the electrical beds were used for the electrical bed operations. Here, automatic interlocking patterns were different from each other in characteristics.

Measurements and Data Acquisition: Contact area and pressure distribution were measured by pressure mapping system (Pliance FTX, Novel, Germany) to identify a possibility of bedsore occurrence and an effect of the automatic interlocking pattern change. They were also used in considering contact and boundary conditions in musculoskeletal modeling and to validate the musculoskeletal models described below. Three-dimensional (3D) motions were measured by 3D motion analysis system (VICON Motion System, VICON Ltd., England) to identify the characteristics of the automatic interlocking patterns of the electrical beds and the participant motions. Here, data of the 3D motions were used in musculoskeletal modeling described below.

Musculoskeletal Modeling and Analysis: To identify alteration of normal and shear forces on the body of the participant as changing the automatic interlocking pattern of the electrical bed, musculoskeletal models for the electrical beds and participants were developed by BRG. LifeMOD (LifeModeler, Inc., USA).

Statistical Analysis: A paired t-test and one-way ANOVA test with Tukey's-b post hoc multiple comparisons were used to identify existence and nonexistence of statistically significant differences. Here, the significance levels for all statistical tests were set at 0.05.

RESULTS AND DISCUSSIONS

Patterns and values of contact area and pressure distribution were significantly alerted as changing the automatic interlocking pattern of the electrical bed ($p<0.05$) (Figure. 1). Patterns and values of normal and shear forces were also altered by the automatic interlocking pattern changes (Figure 2). It was generally found that as using the automatic interlocking pattern of Type B bed, pressure values were repeatedly approached or exceed to the critical value of bedsore occurrence. Also, it was identified that normal and shear forces generated by using the automatic interlocking pattern of Type B bed were generally higher than those of Type A bed. These results indicated that using the automatic interlocking pattern of Type B bed could be more vulnerable to bedsore than using that of Type A bed. Therefore, from the current study, it is judged that considering an optimum interlocking pattern in manufacturing electrical bed may be one of important factors to prevention of bedsore.

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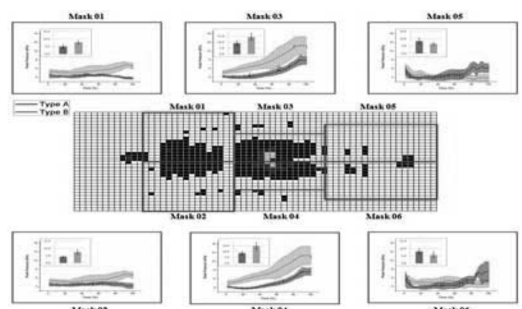

Figure 1: Peak pressure distribution. Here, the results of contact area were not shown because of limitation of page. However, a tendency of the results of contact area was similar to those of pressure distribution.

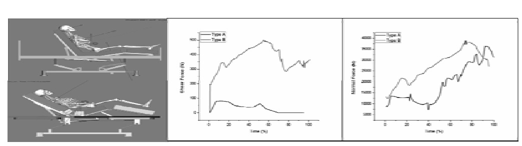

Figure 2: Normal and shear forces predicted from the musculoskeletal model analysis

BIOMECHANICAL FACTOR TO BE CONSIDERED DURING POWER-LIFT DESIGN TO REDUCE THE RISK OF MUSCULOSKELETAL DISORDERS

Hyunho Choi, Keyoung Jin Chun, #Dohyung Lim

Gerontechnology Center, Korea Institute of Industrial Technology
#Corresponding and Presenting Author: dli349@kitech.re.kr

INTRODUCTION

This study was aimed at 1) identification of the hypothesis that musculoskeletal disorders (MSDs) can be occurred to caregivers by repeated uses of power-lift (Sean, 2005) and 2) suggestion of biomechanical factors to be considered to reduce the risk of MSDs if the hypothesis is true.

MATERIAL AND METHODS

Participants: Following Institutional Review Board approval, one young healthy man (age: 26 years, height: 172cm, weight: 67 kg) was participated.

Power-Lift Selection: Two power-lifts (Type A: Bolero, ARJO, USA and Type B: MONA, Horcher, USA) were chosen by the analyses of market price, market share, sales volume, user approaching, brand awareness, test convenience, etc.

Power-Lift Operation: Operations to drive power-lift at strait and curve tracks were selected based on the analysis of power-lift workings performed frequently at care facilities. Then, the participant operated power-lift during ten minutes and took a rest for forty minutes between operations.

Measurements and Data Acquisition: Ten EMG sensors (Tringo Wireless EMG System, DELSYS, USA) were attached to the right muscles of the participant (Table 1). The EMG data were analyzed by using MDF (Median Frequency) technique. Contact area and pressure distribution were measured by pressure mapping system by using pedar and pliance system (Novel gmbh, Germany) to identify a risk of MSDs. They were also used in considering contact and boundary conditions in musculoskeletal modeling and to validate the musculoskeletal models described below. Three-dimensional (3D) motions were measured by VICON Motion System (VICON Ltd., England) to identify the characteristics of motions of the power-lift and the participant motions. Here, data of the 3D motions were used in musculoskeletal modeling described below.

Musculoskeletal Modeling and Analysis: To identify muscle forces and joint torques required for power-lift operations, musculoskeletal models for the power-lift and the participants were developed by BRG. LifeMOD (LifeModeler, Inc., USA). Additionally, the musculoskeletal models were used to identify how much effective in reduction of the risk of MSDs when the suggested biomechanical factors were considered in new power-lift design.

Statistical Analysis: A paired t-test was used to identify existence and nonexistence of statistically significant differences. Here, the significance level was set at 0.05.

RESULTS AND DISCUSSIONS

Possibility of muscle fatigue occurrence was generally increased during power-lift workings ($p<0.05$) (Table 1). This result indicates that our hypothesis is acceptable and further more power-lift design should be improved with consideration of biomechanical and ergonomic factors. The contact area and pressure generated as using Type A power-lift was generally lower and higher, respectively, than those as using Type B power-lift ($p<0.05$) (Figure 1). The results also showed that the contact area and pressure were depended on power-lift design, particular in handle and caster shapes (Figure 1). Therefore, from the results obtained the current study, it is supposed that improvement of power-lift design with consideration of biomechanical and ergonomic factors (here, handle and caster shapes) may be one of solutions for reduction of the risk of MSDs.

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Table 1: The rate of muscle fatigue occurrence

(Unit :%)			Straight		Curve	
	Function	Muscle	Type A	Type B	Type A	Type B
Upper Body	Protraction	Prapezius	2.5	6.5	10.5	33.1
	Adduction	Pectoralis Major.	—	13.7	—	14.5
	Flexion	Biceps Brachii	3.2	—	6.8	4.0
	Adduction	Triceps Brachii	—	7.6	—	0.6
	Extension	Erector spinae	—	5.5	—	5.2
Lower Body	Adduction	Gluteus Medius	—	0.5	8.2	7.0
	Extension	Rectus Femoris	1.9	7.5	5.8	—
	Extension	Biceps Femoris	—	3.0	0.7	—
	PlantarFlexion	Tibialis anterior	3.1	11.9	3.3	11.4
	PlantarFlexion	Gestrocnemius	1.5	12.1	10.7	11.2

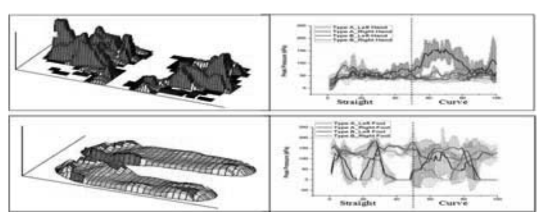

Figure 2: Peak pressure distribution of hands and feet. Here, the results of contact area were not shown because of limitation of page. However, a tendency of the results of contact area was similar to those of pressure distribution.

VARIATIONS IN PLANTAR LOADING PATTERNS IN INDIVIDUALS WITH SOFT TISSUE VERSUS BONEY REARFOOT TRAUMA: A PRELIMINARY STUDY

Stephanie Albin (1), Mark W. Cornwall (2), Thomas G. McPoil (2)

1. **Physical Therapy Foot & Ankle Clinic, The Orthopaedic Specialty Hospital, Salt Lake City Utah, USA**
2. **The Laboratory for Foot & Ankle Research, Physical Therapy Program, Northern Arizona University, Flagstaff, Arizona, USA**

PURPOSE

Little research has been conducted to determine the differences in plantar loading patterns following surgical repair of Achilles tendon ruptures versus displaced calcaneal fractures (Schepers et al, 2008; Hastings et al, 2000). While both types of traumatic injuries involve significant hindfoot structures, one could expect differences in plantar loading as a result of the characteristics of the involved tissues. The intent of this preliminary study was to assess the variations in plantar loading patterns in individuals who had undergone a surgical repair of a) Achilles tendon following rupture or 2) intra-articular calcaneal or tibial plafond fracture.

SUBJECTS

Ten male subjects participated in the study; five had Achilles tendon repairs (ATR) and five were post-Open Reduction Internal Fixation for hindfoot fractures including intra-articular calcaneal or tibial plafond fractures (HFFx). The mean age of the ATR patients was 51.2 years and 33.0 years for the HFFX. At the time of testing, approximately 3 to 4 months post-surgery, all subjects were able to ambulate without using an assistive device. All subjects had no history of congenital or traumatic deformity or foot pain on the non-involved side.

METHODS

Each subject was asked to walk over a 15-meter walkway while in-shoe pressure data were collected using the PEDAR-X system. For all in-shoe pressure measurements, the sensor insoles were positioned over a flat piece of firm insole material (durometer = 58 Shore A) that was placed in a flat rubber-soled, non-contoured karate shoe. Ten consecutive steps from the middle 8-meters of the walkway for both the involved and un-involved feet were selected for analysis. Novel percent mask and group mask software were used to determine maximum mean pressure (MMP) and normalized Maximum Force (MaxF) for each of the following plantar regions: medial heel (1), lateral heel (2), medial midfoot (3), lateral midfoot (4), medial forefoot (5), and lateral forefoot (6). A MANOVA was used to determine those plantar regions in which significant differences occurred between MMP and MaxF between the two groups. Based on those results, an ANOVA and Tukey's pairwise comparisons were used to further analyze those plantar regions that were statistically significant. For this abstract, only the plantar pressure results for the involved side will be presented.

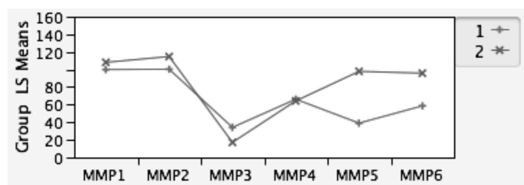

Figure 1: MANOVA results for Maximum Mean Pressure (MMP) (1 = HFFx; 2 = ATR).

RESULTS

MaxF was not significantly different between ATR and HFFx for any of the six plantar regions analyzed. MMP was significantly decreased ($p < .05$) in the medial midfoot and medial forefoot plantar regions for the HFFx group in comparison to the ATR group.

Stance phase durations were significantly decreased ($p < .05$) between the involved and non-involved limbs for the HFFx group but not for the ATR group. The mean stance duration for the involved and non-involved limbs was: 1.01 sec and 0.86 sec for HFFx and 0.78 sec and 0.77 for ATR.

CONCLUSIONS

Although a small cohort of subjects were evaluated in this preliminary study, it would appear that boney trauma of the hindfoot when compared to soft tissue injury, not only significantly slows walking speed, but also substantially reduces the amount of medial plantar loading of the midfoot and forefoot regions of the foot. These results have important implications for foot orthoses design.

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EFFECTS OF SUB-HALLUCIAL WEDGE AND MEDIAL ARCH SUPPORT ON DYNAMIC PLANTAR PRESSURE

Craig Nimick, Jaminelli Banks, Robert Marroquin, Christopher Hood, Dana Tango, Reagan Kane, James Furmato, Kendrick Whitney, Jinsup Song

Gait Study Center, Temple University School of Podiatric Medicine, Philadelphia, PA

INTRODUCTION

Functional hallux limitus (FHL) (Dananberg, 1986) has been described as a predecessor to structural and debilitating deformities such as hallux rigidus and lesser metatarsalgia. Compensatory changes include increased lesser metatarsal plantar pressures and decreased weight bearing under the hallux. Case reports have shown use of a sub-hallucial wedge to decrease lesser metatarsal pain and increase weight bearing pressures under the first metatarsal phalangeal joint (1st MTPJ). (Clough, 2005) Other studies have demonstrated that control of excessive STJ pronation with an orthotic arch support increases 1st MTPJ dorsiflexion in subjects with FHL. (Munuera, 2006) The purpose of this pilot study was to objectively evaluate the biomechanical effect of a sub hallux wedge and arch support in subjects with FHL.

MATERIALS AND METHODS

Ten asymptomatic healthy subjects with FHL were evaluated. All subjects had >50º of dorsiflexion of the 1st MTPJ in open kinetic chain and <30 º of dorsiflexion in weight bearing as measured with a goniometer. Dynamic plantar pressure was collected using a EMED-X (Novel LLC, St. Paul, MN) in three test conditions; barefoot (BF), barefoot with a sub hallucial wedge (CW), and barefoot with a sub hallucial wedge and arch support (CWA) during self-selected comfortable paced walking. Sub-hallucial wedge and medial arch support were placed onto each foot using a custom designed brace. Peak pressure time integrals (PTI) were analyzed under metatarsal heads 1-5 and the hallux. One way Analysis of Variance was performed at a significance level of 0.05.

RESULTS

Ten subjects (20 feet), with mean age of 26.8 years and BMI of 27.8 kg/m^2, participated in the study. There was no statistically significant difference in self-selected walking speed (mean 1.26m/s) across 3 conditions. As expected, PTI beneath 1st MTPJ was significantly lower than 2nd and 3rd MTPJs in barefoot walking. When subjects walked with CWA, significantly reduced PTI was noted under MTPJs 2-5, compared to both BF and CW conditions (Figure 1).

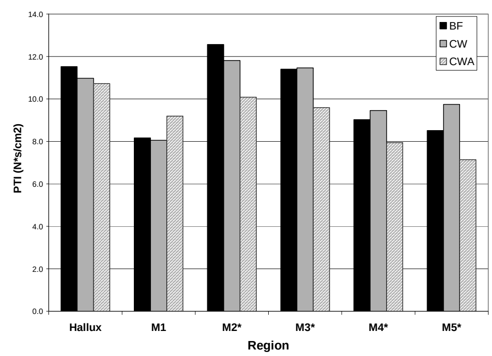

*Figure 1. Mean Peak Pressure Time Integrals (N*sec/cm^2) of the hallux and metatarsal heads 1-5 for each test condition. * denotes P<0.05*

CONCLUSION

A sub-hallucial wedge (CW) by itself did not alter dynamic plantar pressure when compared to barefoot comfortable paced walking. The combination of the sub hallucial wedge with a medial arch support significantly reduced PTI under all lesser metatarsal heads. The hallucial wedge and medial arch support appear to compliment one another, suggesting the importance of addressing both forefoot and rearfoot mechanics when treating patients with a functional limitation of the 1st MTPJ.

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This study was partially funded by a grant from APSMA.

P10

THE EFFECTS OF SLIPPERS AND LOWER LIMB POSITIONING ON PLANTAR PRESSURES IN SELLECTED *BALLET* BALANCES

Érica Engrácia Valenti (1), Paula H. Lobo da Costa (1),

Kerstin Bosch (2), Dieter Rosenbaum (2)

1. Department of Physical Education, Federal University of Sao Carlos, Sao Carlos, Brazil.
2. Movement Analysis Laboratory, Orthopaedic Department, University Hospital, Münster, Germany.

INTRODUCTION

Classical *ballet* teaching and practice are strongly influenced by tradition, however, biomechanical investigations have recently contributed to ensure a more secure practice and an evidence-oriented teaching (IMURA *et al*., 2008). This study aims to describe the effects of slippers and limb positioning on plantar pressure variables produced by selected techniques of classical *ballet*.

METHODOLOGY

Fourteen non-professional female dancers aged between 15 and 25 years old, without any musculoskeletal impairment, who have practiced for at least seven years volunteered to this study. They performed three valid trials of standing for four seconds in the following balance postures: *attitude devant, derriére* and *attitude a la second*, with slippers and barefoot.

Plantar pressure variables were quantified with an EMED platform (Novel, Germany) at 50 Hz sampling rate. Peak pressures and contact areas where compared between slippers vs. barefoot conditions and for the three different lower limb positions with a univariate ANOVA-test. Significant level was determined for $p \leq 0.05$.

RESULTS AND DISCUSSION

Contact areas (Table 1) were statistically higher for the barefoot condition and no effect of the balance postures were found. The barefoot/slippers effect was not found for the contact area.

Contact area	*Attitude devant*	*Attitude derriére*	*Attitude a la second*
Barefoot	169.0* ±21.5	172.4* ±24.4	174.0* ±24.0
Slippers	155.2 ±12.7	157.6 ±14.2	158.4 ±14.3

*Table 1: Means (±SD) for contact areas (cm^2). N = 14. *significant difference for barefoot vs. slippers.*

Highest values of peak pressures were also found for the barefoot condition and again no differences were found among the three balance postures tested.

Peak pressure	*Attitude devant*	*Attitude derriére*	*Attitude a la second*
Barefoot	172.9* ±24.2	172.3* ±24.5	175.8* ±23.1
Slippers	155.7 ±13.5	157.7 ±14.2	170.3 ±44.5

*Table 2: Means (±SD) for peak pressures (kPa). N = 14. *significant difference for barefoot vs. slippers.*

These results are in accordance with the higher stability reported by the dancers during the barefoot trials, compared to the trials performed with slippers. The use of slippers may have slightly limited and compressed dancers' toes, a fact that may have reduced the contact area with the ground and affected the peak pressures.

In conclusion, our findings illustrate the foot loading characteristics during the performance of balance postures. Moreover, the foot training may be emphasized and have beneficial effects for the performance of these classical *ballet* techniques.

Figure: A ballet dancer during "Attitude devant" position on the EMED platform (with permission).

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ACKNOWLEDGEMENT

Partial financial support by the Deutsche Forschungsgemeinschaft is gratefully acknowledged.

P11

DYNAMIC EFFECTS OF DIFFERENT ALTERNATING CYCLE TIME CONDITIONS ON SOFT TISSUE PERFUSION RECOVERY

Byeong Hee Won(1,2), Youn Jung Choi(2), Keyoung Jin Chun(2), Chang-seop Song(1)

1. Department of Mechanical Design and Mechatronics, Hanyang University, Republic of Korea
2. Gerontechnology Center, Korea Institute of Industrial Technology, Republic of Korea

INTRODUCTION

Pressure ulcers (PU) are a significant and costly problem in health care area. Alternating pressure air mattress (APAM) is one of the popular product for preventing PU. Usual operating parameters of APAM are alternating cycle time and air cell pressure level. The objective of current study is to quantify a tissue vitality with a variation of alternating cycle time through tissue perfusion and interface pressure measurement to understand proper operating conditions.

METHOD

Six male(age:68.6±5.1, BMI:24.5±3.2) and six female subjects(age:64.1±4.2, BMI:23.3±2.5) were selected to test the tissue perfusion and pressure distributions at the air mattress(3-cell type, average air cell pressure: 34~37mmHg, measuring time:1hour). For each 3 alternating cycle time conditions(3,5,10min). TCM4 Oximeters(Radiometer A/S, Denmark) were used to record transcutaneous partial pressures of oxygen($tcPO_2$) and carbon dioxide($tcPCO_2$) from the tissue. Electrodes were calibrated using known gases and were attached to the skin at chest, sacrum and heel. Pliance pressure mapping system (Novel, Germany) which consists of a thin mat containing grids of miniature pressure sensors, was placed between the subject and the mattress, one at a time, to electronically record information on pressure distributions. Force and peak, mean pressure data was collected with the Pliance.

RESULTS

The recovery & occlusion of tissue perfusion appeared according to the deflation & inflation of air cell. Perfusion recovery time(t_R) and recovered pressure(P_R) were defined as dynamic parameters of tissue perfusion from 5% to 95% of $tcPO_2$ values during tissue recovery period(Figure 1). Recovery time and recovered pressure for each alternating cycle time conditions are given at sacrum (Figure 2). From the figure, we can see the distinct results that shorter alternating cycle condition gives faster tissue recovery. And the recovered pressure decreased gradually with the increase of alternating cycle time. Thus these results means that shorter cycle time is more effective in the aspect of tissue vitality because faster and stronger recovery characteristics can be acquired. However peak pressure and force from interface pressure measurement was irregularly changed with alternating cycle time(Figure 3). This might be affected by time variant interface condition.

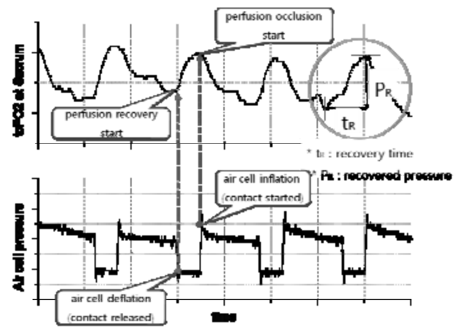

Figure 1: Results of tissue perfusion. Definition of recovery time(t_R), recovered pressure(P_R).

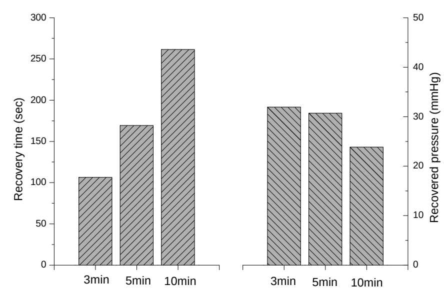

Figure 2: Recovery time and recovered pressure for each alternating cycle time.

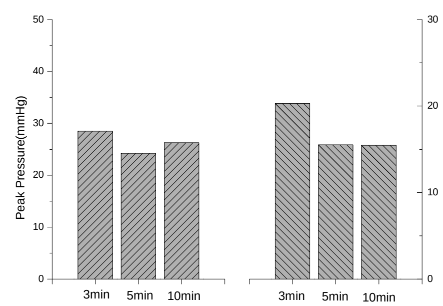

Figure 3: Peak pressure and force from interface pressure measurement for each alternating cycle time.

CONCLUSION

The current study may be valuable by quantifying tissue recovery characteristics, i.e., tissue perfusion recovery enhanced by shorter alternating cycle condition, of the APAM application for PU prevention. The fining may contribute to understanding dynamic effects of time-varying external contact to tissue

A NOVEL APPROACH USING A FE-FOOT MODEL FOR CLINICAL APPLICATIONS

Wyss, Christian[1,2]

1. Laboratory of Movement analysis, Childrens University Hospital Basel, Switzerland
2. Laboratory of Movement analysis, Cantonal Hospital Aarau, Switzerland

INTRODUCTION

Most FE models in the literatur [1, 2, 3] are based on boundary conditions without concerning muscular forces of the lower leg. The aim of our study was to get boundary conditions using data of a gait analysis in combination with ANYBODY muscle modelling.

METHODS

We examined 10 healthy subjects with a Vicon MX System (6 Cameras). Additionally we performed an AMTI force platform, a Novel SF pressure measurement System and a MegaWin EMG system, synchronized with the Vicon system. The muscle forces were determined using a slightly modified model from the repository AMMRV1.1 of ANYBODY TECHNOLOGY (Vaughan). Postprocessing with the EMG data was done. A modified finite element model from the Website ww.ulb.ac.be/project/vakhum/ which is based on CT Scan was used. Contacts between bones were considered with 3 different methods. 1. as bonded, 2. as joints with less friction and 3. as frictional contact with additional ligaments because of a underconstrained model. We were firstly interested in reaction forces/moments and not in stress and strain. Therefore we did a quasi static analysis computing the reaction forces/moments in every 10% of stance phase in each joint. The foot was fixed at the end of the tibia and the fibula. The position of the foot segments in the FE model were adapted due to the kinematic measurements. The reaction forces we measured with a force plate and an EMED SF as well as a PEDAR System. As a novel method to get additional boundary condition we used the model of Vaughan from the ANYBODY muscle modelling Repository.

RESULTS

Reaction forces and moments were with all 3 methods comparable, opposite to stress and strain which showed huge differences, especially in contact regions. Reaction forces and moments in the ankle joint computed with our FE model fit very well with results we get from the ANYBODY model. In other joints such as TMT and MTP joints we get similar results as for example Jacob[4] described.

DISCUSSSION & CONCLUSIONS

The above shown method to get the boundary conditions for an FE Model from gait analysis and ANYBODY modelling is probably a valuable method for clinicians in the future. The method seems to be satisfactory to get reaction forces and moments.

There are further investigations necessary, to simulate stress and strain. The main problem here seems to be an underconstrained model using frictionless contacts. Main reason for that are the unbalanced muscle forces from the ANYBODY modelling system in combination with the applicated geometry of the foot. Other possibilities to avoid a underconstrained model such as the introduction of ligaments or weak springs even though frictional contact is in our understanding not a satisfactory solution.

Figure 1: Mesh of the FE-Model

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RELIABILITY OF IN-SOLE PLANTAR PRESSURE USING SIMPLE AND DETAILED MASKS OF THE FOREFOOT, MIDFOOT, AND HINDFOOT

Jonathan S. Akins, Karen A. Keenan, Brian P. Dugan, Melanie Francis. Francis, John P. Abt, Timothy C. Sell, Scott M. Lephart

Neuromuscular Research Laboratory, University of Pittsburgh, PA

BACKGROUND

Plantar pressure measurements obtained with in-shoe pressure sensor insoles can be analyzed by dividing the foot into anatomical regions using masks. A simple mask could divide the foot into four regions: toes, forefoot, midfoot, and hindfoot. A detailed mask could further divide these regions to capture data from specific anatomical regions/structures. For example, dividing the forefoot into individual metatarsals may help clinicians and researchers indentify individuals who are at increased risk of stress-related injuries to these structures (e.g. stress fracture). Regardless of which mask is applied, these measures must be reliable. The purpose of this study was to obtain the reliability of in-sole plantar pressure measurements using simple and detailed masks and to determine which is more reliable.

METHODS

Ten healthy males (n=8) and females (n=2) participated in this study (age: 27.7 ± 4.1 years, mass: 77.6 ± 10.7 kg, height: 174.3 ± 7.0 cm). Subjects performed a gait task on two different days using the PEDAR-X® system (Novel GmbH, Munich, Germany) to measure in-sole plantar pressures (100Hz) using their own athletic footwear (without socks). After familiarization of the task, subjects performed three trials of 20 consecutive straight-line walking steps using usual gait across a level, laboratory floor.

Maximum force (MF), force-time integral (FTI), peak pressure (PP), pressure-time integral (PTI), and maximum mean pressure (MMP) were calculated for simple and detailed masks. Both masks divided the foot into three primary regions: the forefoot, midfoot, and hindfoot. The simple mask divided the forefoot into metatarsal 1, 2, and 3-5; and did not divide the midfoot and hindfoot. The detailed mask divided the forefoot into individual metatarsals and the midfoot and hindfoot into medial/lateral regions. Intraclass correlation coefficients (ICC) were calculated using a two-way random effects model (ICC [2,k]). ICCs were averaged for each variable to obtain a single ICC value for the simple and detailed masks. Left and right foot measurements were combined for the analyses. A t-test was used to identify significant differences between the simple and detailed masks (α=0.05).

RESULTS

No significant differences were found between the simple and detailed masks. Maximum force and force-time-integral resulted in excellent reliability (ICC>0.90) for all masks and regions. Peak pressure, pressure-time-integral, and maximum mean pressure resulted in good reliability (ICC>0.70) for all masks and regions.

Table 1: ICCs (mean ± SD) for simple and detailed masks of the forefoot, midfoot, and hindfoot

	Forefoot		
	Simple	Detailed	p-value
MF	0.962 ± 0.016	0.966 ± 0.014	0.653
FTI	0.932 ± 0.027	0.942 ± 0.024	0.456
PP	0.849 ± 0.164	0.891 ± 0.136	0.586
PTI	0.845 ± 0.078	0.891 ± 0.083	0.292
MMP	0.929 ± 0.038	0.943 ± 0.035	0.454
	Midfoot		
	Simple	Detailed	p-value
MF	0.975 ± 0.003	0.974 ± 0.010	0.916
FTI	0.955 ± 0.013	0.955 ± 0.019	0.972
PP	0.938 ± 0.044	0.944 ± 0.023	0.818
PTI	0.852 ± 0.108	0.887 ± 0.035	0.549
MMP	0.852 ± 0.055	0.751 ± 0.237	0.604
	Hindfoot		
	Simple	Detailed	p-value
MF	0.915 ± 0.018	0.911 ± 0.041	0.913
FTI	0.959 ± 0.023	0.952 ± 0.020	0.720
PP	0.870 ± 0.021	0.861 ± 0.047	0.799
PTI	0.908 ± 0.030	0.911 ± 0.024	0.912
MMP	0.890 ± 0.021	0.877 ± 0.047	0.731

DISCUSSION

In-sole plantar pressure measurements obtained demonstrated good reliability for all masks and regions indicating that the PEDAR-X® system is capable of collecting reliable data, regardless of which mask is utilized. These findings are similar those of previous studies (Boyd *et al*, 1997; Putti el al, 2007). The lowest ICC was for MMP using the detailed mask for the midfoot, which may be a function of the small contact area of the medial midfoot. Since the contact area is relatively small, it is possible that even small variation in the rollover pattern during gait may significantly affect measurements in this region. It is therefore recommended to not subdivide this region in order to maximize the reliability of the measurements obtained, both in the clinical and research settings.

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THE INFLUENCE OF FATIGUE, LIGAMENT LAXITY AND HORMONAL FLUCTUATION ON PLANTAR PRESSURE IN FEMALE ATHLETES

Roshna E. Wunderlich (1), Kristina T. Silke (1), Sarah F. Cebulski (2)

1. Department of Biology, James Madison University, Virginia
2. Department of Biology, Carleton University, Ottawa

INTRODUCTION

Overuse injuries of the lower extremity compose over 20% of injuries in collegiate athletes, up to 50% of injuries in pediatric athletics (Arendt, 2003, Bennell, *et al.*, 1996, Dalton, 1992) and a considerable portion of injuries in military recruits (Milgrom, *et al.*, 1985, Rauh, *et al.*, 2006). Like some traumatic injuries (acl tears) overuse/stress injuries are significantly more frequently observed in women (Taunton, *et al.*, 2002). The etiology of sex differences in overuse injuries has been associated with various factors including training, anatomical differences, hormonal differences, and joint laxity. Joint laxity has also been associated with fluctuating hormone levels. Specifically, estrogen has been associated with suppression of collagen synthesis and decreased ligament cross-sectional area (Abubaker, *et al.*, 1996, Booth, *et al.*, 1970, Liu, *et al.*, 1997, Samuel, *et al.*, 1996). As collagen serves to provide stiffness to biological tissues such as ligaments and tendons, its breakdown can play a key role in enhanced joint laxity. In this study we quantify hormone levels, ligament laxity, and plantar pressure distribution to investigate the extent to which hormone fluctuation and fatigue interact to influence plantar pressure distribution and lower limb load.

METHODS

Twenty-four female collegiate athletes were tested, 8 for a four-week protocol and 16 for a twelve-week protocol. Blood was drawn each visit and subsequently assayed for estradiol content using an ELISA kit. Ankle ligament laxity was measured using an ankle arthrometer, and subjects walked, ran and cut barefoot over an EMED-SF plantar pressure platform. Following a fatigue test (stair running followed by a modified hurdle beep test), lactate levels, ankle laxity and plantar pressure were re-measured. Repeated measures ANOVA was used to compare ligament laxity and plantar pressure in 7 foot regions across menstrual weeks and fatigue states.

RESULTS

Ankle ligament laxity increased following fatigue ($p<0.05$). There was also significant variation across subjects ($p<0.01$). Females had more ligament laxity than males ($p<0.01$), and there was an interaction between gender and fatigue such that females experienced a greater increase in ligament laxity post-fatigue than males. There was a significant interaction between menstrual week and fatigue such that the effect of fatigue was significantly greater in week 4 and lower in week 2 than any other menstrual week. While hormone levels may not themselves affect ligament laxity, the interaction of hormone level and fatigue may influence ligament laxity.

Peak plantar pressures exhibit variation across foot region, menstrual week and fatigue condition. Peak pressures in metatarsals 2-5 are significantly ($p<0.01$) lower post-fatigue than pre-fatigue. Peak pressures in the medial and lateral midfoot, however, vary based on menstrual week and fatigue condition. In weeks 1 and 2 post-fatigue midfoot peak pressure are higher post-fatigue than pre-fatigue ($p<0.05$), while post-fatigue pressures are slightly lower than pre-fatigue pressures in weeks 3 and 4. Week 2 is also the week of greatest ligament laxity, suggesting foot ligament laxity is allowing some midfoot collapse, especially post-fatigue when the plantarflexors are fatigued and providing less load resistance (Sharkey, *et al.*, 1999, Weist, *et al.*, 2004). This laxity may also reduce load transmission to the metatarsal heads.

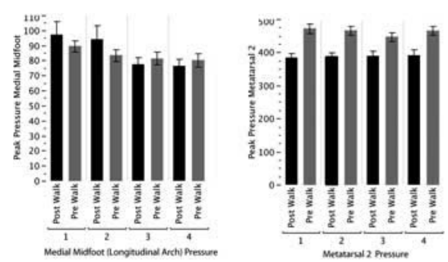

Figure 1. Peak pressures in medial midfoot and metatarsal 2 across weeks and fatigue state.

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COMPARISON OF PLANTAR PRESSURE MEASUREMENTS OBTAINED DURING BAREFOOT AND SHOD CONDITIONS

Jonathan S. Akins, Karen A. Keenan, Brian P. Dugan, Melanie Francis, John P. Abt, Timothy C. Sell, Scott M. Lephart

Neuromuscular Research Laboratory, University of Pittsburgh, PA

BACKGROUND

Pedobarographic platforms allow for assessment of plantar pressures at the foot-ground interface whereas pedobaragraphic insoles allow for assessment at the foot-shoe interface. It has been suggested that barefoot assessments may be more sensitive in detecting risk factors for the development of exercise related lower leg pain than shod assessments (Willems, 2007). Plantar pressures may be altered or attenuated by footwear, thereby making detection of small but significant variations in plantar pressure measurements more difficult in shod conditions. In addition, barefoot assessments require less subject preparation, may allow for data to be collected on more subjects in a shorter period of time, and require a smaller laboratory space, thereby potentially simplifying the data collection process. The purpose of this study was to compare plantar pressure measurements obtained while barefoot to those obtained while shod.

METHODS

Ten healthy males (n=8) and females (n=2) participated in this study (age: 27.7 ± 4.1 years, mass: 77.6 ± 10.7 kg, height: 174.3 ± 7.0 cm). Data were collected on two different days in both barefoot and shod conditions. Barefoot measurements were obtained with the EMED-X® (Novel GmbH, Munich, Germany), sampling at 100Hz. A two-step approach at a self-selected speed was utilized for all trials. After familiarization of the task, subjects performed 10 right-foot trials.

Shod measurements were obtained with the PEDAR-X® system (Novel GmbH, Munich, Germany), sampling at 100Hz. Subjects used their own athletic footwear, without socks, and were asked to practice straight-line walking at their "usual" walking speed. After familiarization, three trials of 20 consecutive straight-line steps across a level tiled floor were recorded.

For the barefoot trials, average maximum force (MF) and peak pressure (PP) were calculated for all trials. For the shod trials, the first and last two right steps were removed and MF and PP were calculated for the remaining right-foot steps. A similar mask was applied to both barefoot and shod trials with the following regions: medial heel, lateral heel, midfoot, each metatarsal (1-5), great toe, toe 2, and toes345. A t-test compared MF and PP obtained in the barefoot (EMED) and shod (PEDAR) conditions (α=0.05).

RESULTS

PEDAR and EMED measurements differed significantly in several regions (Table 1).

	PEDAR	EMED	
Region	Maximum Force		
GT	110.1 ± 33.2	126.1 ± 64.7	
T2	60.4 ± 18.3	29.4 ± 15.9	*
T345	75.6 ± 24.7	25.9 ± 20.0	*
MT1	145.6 ± 54.0	161.1 ± 62.8	
MT2	126.6 ± 37.7	186.4 ± 33.4	*
MT3	101.9 ± 29.0	181.5 ± 36.0	*
MT4	73.7 ± 23.7	101.9 ± 37.6	*
MT5	49.8 ± 18.1	38.0 ± 22.8	
Midfoot	184.6 ± 66.4	139.9 ± 87.0	
Lat Hindfoot	276.0 ± 60.5	232.5 ± 48.1	
Med Hindfoot	288.6 ± 67.9	282.0 ± 49.1	
	Peak Pressure		
GT	194.9 ± 46.8	334.4 ± 179.3	*
T2	165.6 ± 44.3	205.1 ± 103.3	
T345	111.5 ± 37.3	105.3 ± 56.7	
MT1	192.2 ± 41.3	266.6 ± 116.1	
MT2	193.3 ± 43.1	554.4 ± 278.2	*
MT3	185.8 ± 42.8	391.9 ± 143.1	*
MT4	159.9 ± 43.7	246.4 ± 109.4	*
MT5	117.8 ± 36.2	176.9 ± 137.3	
Midfoot	121.4 ± 32.3	118.2 ± 44.3	
Lat Hindfoot	185.9 ± 30.9	367.3 ± 124.0	*
Med Hindfoot	192.1 ± 33.2	367.0 ± 97.8	*

* $p < 0.05$

Table 1: Maximum force and peak pressure for each region (mean ± SD)

DISCUSSION

Statistically different plantar pressure measurements were obtained in barefoot and shod conditions. Peak pressure appears to be significantly attenuated in the hindfoot, MT2 through MT4, and the great toe in the shod condition. Maximum force was significantly attenuated in MT2 through MT4, T2 and T345. If the variable of interest is peak pressure, then barefoot assessments may be preferred. However, since most athletic activities are performed while shod and the results were inconsistent, it may be more appropriate to assess plantar pressure in the shod condition.

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ESM 2010 Program at-a-glance

Time		Saturday, August 14th	Sunday, August 15th	Monday, August 16th	Tuesday, August 17th	Time	
06:00				**Breakfast Pick-up** *Lobby* **6:00**		06:00	
06:30				**Buses depart 6:30**		06:30	
07:30	08:15		**Breakfast, Registration** *Lobby*	**Block Island Activity Day** see also page 10		07:30	08:15
08:15	08:30		**Welcome** *Ballroom*		**Breakfast (from 8:00)** *Lobby*	08:15	08:30
08:30	09:00		**Session 1** Foot Deformities *Ballroom*			08:30	09:00
09:00	09:30				**Session 4** Gait and Orthoses *Ballroom*	09:00	09:30
09:30	10:00					09:30	10:00
10:00	10:30		**Break**			10:00	10:30
10:30	11:00	**Registration** for workshop attendees *Hotel Providence Lobby*	**Session 2** The Evolution of Gait Keynote lecture 1 *Ballroom*		**Break**	10:30	11:00
11:00	11:30				**Session 5** Analysis Methods *Ballroom*	11:00	11:30
11:30	12:00	**Bus from Hotel to Workshops 11:30**				11:30	12:00
12:00	12:30	**Workshop I (12:00 – 13:00)** *Center for Restoration & Regenerative Medicine*	**Lunch**		**Lunch**	12:00	12:30
12:30	13:00					12:30	13:00
13:00	13:30	**Workshop II (13:10 – 14:10)** *Center for Restoration & Regenerative Medicine*	**Session 3** Pressure Distribution Measurement as a Diagnostic Tool *Ballroom*		**Session 6** Data Collection Methods and Gait *Ballroom*	13:00	13:30
13:30	14:00					13:30	14:00
14:00	14:30	**Workshop III (14:20 –16:00)** *Center for Restoration & Regenerative Medicine*			**Break**	14:00	14:30
14:30	15:00				**Session 7** Dynamic Load Distribution Keynote lecture 2 *Ballroom*	14:30	15:00
15:00	15:30	**Tour of the Lab (15:00 –15:30)**	**Poster & Break** *Johnson & Wales and RISD Gallery rooms*			15:00	15:30
15:30	16:00	**Workshop IV (16:00 –17:00)** *Center for Restoration & Regenerative Medicine*				15:30	16:00
16:00	16:30		**Podium Discussion I** *Ballroom*			16:00	16:30
16:30	17:00				**Podium Discussion II** *Ballroom*	16:30	17:00
17:00	17:30	**Bus from Workshops to Hotel 17:00**			**Closing remarks**	17:00	17:30
17:30	18:00	**Registration** *Hotel Providence Lobby* **Opening Reception** *Hotel Providence Blackstone Terrace*				17:30	18:00
18:00	18:30		**Evening Event** *RISD Museum of Art*			18:00	18:30
18:30	19:00					18:30	19:00
19:00	19:30			**Walk back to Old Harbour 19:00**	**Banquet Dinner** *Aspire Restaurant*	19:00	19:30
19:30	...	**Waterfire** *Providence Memorial Boulevard*		**Ferry depart 19:30**		19:30	...