i-FAB 2010 Program and Abstracts

DAY 1: THURSDAY SEPTEMBER 16, 2010

SESSION 1: OPENING CEREMONY AND KEYNOTE
1:00pm-2:15pm Thursday September 16, 2010

Keynote Speaker (Introduced by Dieter Rosenbaum)
1.1 1:15pm-2:15pm: **M Richter**. Coburg Clinical Center, Germany
What has Science Done for the Foot and Ankle Surgeon?

SESSION 2: FOOT STRUCTURE AND FUNCTION
2:25pm-3:50pm Thursday September 16, 2010

Invited Speaker (Introduced by Harold Kitaoka)
2.1 2:25pm-2:45pm: **BJ Sangeorzan**. University of Washington
Foot Structure and Function: Six Blind Men and the Elephant Revisited

Free Communications (Session Chairs: Howard Hillstrom and Dieter Rosenbaum)
2.2 2:45pm-2:57pm
Metatarsal Loading Pattern Differences with Age and Sex in Adolescent Athletes
KR Ford, GD Myer, KD Barber Foss, TE Hewett. Cincinnati Children's Hospital.

2.3 2:57pm-3:09pm
Spatiotemporal Volumetric Analysis of Dynamic Pedobarographic Data
TC Pataky, C Maiwald. Shinshu University.

2.4 3:09pm-3:21pm
Deep Plantarflexor Strength Increase Changes Rearfoot Motion in Shod Running
M Hagen, S Lescher, A Gerhardt, S Spichalla, EM Hennig, S Felber. University of Duisburg-Essen.

2.5 3:21pm-3:33pm
Measuring Dorsal Arch Height in Different Stances with Plantar Foot Muscles Passive and Active: A Reliable Determinant of MLA Articulation in Older People
PJ Latey, D Cobbin, C Zaslawski, NF Smith. University of Technology Sydney.

2.6 3:33pm-3:45pm
The Relevance of Subtalar-joint-anatomy for Chronic Overuse Injuries of the Lower Limbs
CA Reule, WW Alt. University of Stuttgart.

SESSION 3: POSTERS
3:50pm-4:10pm Thursday September 16, 2010

3.1 Foot Structure and Function
Click here for abstracts

3.2 Clinical Foot and Ankle
Click here for abstracts

3.3 Modeling
Click here for abstracts

3.4 Cadaver Simulation
Click here for abstracts

3.5 Diabetic Foot
Click here for abstracts

3.6 Footwear
Click here for abstracts

3.7 Non-human Primate
Click here for abstracts

3.8 Foot and Ankle Imaging
Click here for abstracts

3.9 Methods
Click here for abstracts

3.10 Foot Kinematics
Click here for abstracts

SESSION 4: CLINICAL FOOT AND ANKLE I

4:10pm-5:30pm Thursday September 16, 2010

Invited Speaker (Introduced by Bruce Sangeorzan)

4.1 4:10pm-4:40pm: **ST Hansen**. University of Washington.
A 40-Year Perspective on Foot and Ankle Surgery

Free Communications (Session Chairs: Wilfried Alt and Harold Kitaoka)

4.2 4:40pm-4:52pm
Functional Gait Analysis of Ankle Arthrodesis and Arthroplasty
ME Hahn, ES Wright, A Segal, M Orendurff, WR Ledoux, BJ Sangeorzan. VA Puget Sound, Seattle.

4.3 4:52pm-5:04pm
Reliability and Clinical Utility of Active Foot Range of Motion in Individuals with Midfoot Arthritis and Matched Controls
S Rao, JF Baumhauer, DA Nawoczenski. New York University.

4.4 5:04pm-5:16pm
In-vivo Kinematics of the Three Components of an Innovative Ligament-compatible Total Ankle Replacement: A Fluoroscopic Study
F Cenni, A Leardini, C Belvedere, F Catani, K Cremonini, S Giannini. Istituto Ortopedico Rizzoli.

4.5 5:16pm-5:28pm
Center of Rotation Position in Prosthetic Feet
AB Sawers, ME Hahn. VA Puget Sound, Seattle.

SESSION 5: POSTER RECEPTION
5:30pm-7pm Thursday September 16, 2010
See Abstracts in Session 3

SESSION 6: PEDOBAROGRAPHY INTEREST GROUP
7pm-8pm Thursday September 16, 2010

DAY 2: FRIDAY SEPTEMBER 17, 2010

SESSION 7: COMPUTATIONAL MODELING
8am-10:10am Friday September 17, 2010

Keynote Speaker (Introduced by Christopher Nester)
7.1 8:00am-9:00am: **T van den Bogert**, Orchard Kinetics.
Brain, Body, and Foot: A Multidomain Approach for Predictive Computational Modeling

Invited Speaker: (Introduced by Ahmet Erdemir)
7.2 9:00am-9:20am: **NM Grosland,** VA Magnotta, KH Shivanna. University of Iowa.
Patient-Specific Finite Element Modeling Techniques

Free communications
7.3 9:20am-9:32am (Session Chairs: Jason Cheung and Don Anderson)
Equivalence of Elastic Contact and Finite Element Models of Patient-Specific Contact Stress Exposure in the Human Ankle
AM Kern, DD Anderson, TD Brown. The University of Iowa.

7.4 9:32am-9:44am

Direct Determination of Toe Flexor Muscle Forces Based on Sub-metatarsal/Toe Pad Load Sharing by Using Finite Element Method

W-M Chen, PV Lee, VPW Shim, SB Park, SJ Lee, T Lee. National University of Singapore.

7.5 9:44am-9:56am

Prediction of Plantar Pressure Distribution in Flat-foot Children

I Pauk, V Ezerskiy, M Ihnatouski, B Najafi. Bialystok Technical University.

7.6 9:56am-10:08am

Prediction of the Effect of a Subject-specific AFO on the Gait of a Healthy Test Subject

V Creylman, L Muraru, H Vertommen, I Jonkers, J Vander Sloten, L Peeraer. University College Kempen.

SESSION 8: POSTERS

10:10am-10:30am Friday September 17, 2010

See Abstracts in Session 3

SESSION 9: CLINICAL FOOT AND ANKLE II

10:30am-11:45am Friday September 17, 2010

Free communications (Session Chairs: Vassilios Vardaxis and Michael Orendurff)

9.1 10:30-10:42am

Mechanical and Functional Measures Reveal a Coherent Structure to Ankle Instability: A Principal Component Analysis

TW Croy, LE Chinn, J Hertel. University of Virginia.

9.2 10:42-10:54am

Muscle Activation Patterns for Functional Ankle Instability and Normal Subjects During Anticipated and Unanticipated Jump-landings

MJ Coglianese, JT Hopkins, TR Dunn, MK Seeley. Brigham Young University.

9.3 10:54-11:06am

Preparatory Muscle Activation during a Lateral Hop Before and After Fatigue in Those With and Without Chronic Ankle Instability

KA Webster, BG Pietrosimone, PA Gribble. University of Toledo.

9.4 11:06-11:18am

Plantar Loading in the Cavus Foot

A Kraszewski, B Chow, M Lenhoff, S Backus, J Deland, P Demp, J Song, B Heilman, S Rajan, A Woodley, H Hillstrom. Hospital for Special Surgery.

9.5 11:18-11:30am

Does Foot Type Affect Foot Contact Dynamics?

R Mootanah, J Frey, RA Zifchock, SB Chow, AP Kraszewski, MW Lenhoff, SI Backus, J Deland, P Demp, J Song, HJ Hillstrom. Anglia Ruskin University.

9.6 11:30-11:42am

Effect of Modified Low-Dye Taping on First Ray Mobility in Individuals with Pronated Foot

M Tai, W L Hsi, SF Wang, MH Jan, H Chai. National Taiwan University.

SESSION 10: FOOT AND ANKLE WORLDWIDE

12:00pm-12:45pm Friday September 17, 2010

SESSION 11: CADAVERIC SIMULATION

1:00pm-2:20pm Friday September 17, 2010

Invited Speaker (Introduced by Jay Hertel)

11.1 1:00pm-1:20pm: **NA Sharkey**, Penn State University

What We Have Learned From Dynamic Gait Simulation

Free communications (Session Chairs: Alberto Leardini and Christopher Nester)

11.2 1:20pm-1:32pm

Effect of Subtalar Arthroereisis on Tibiotalar Contact Characteristics in a Cadaveric Flat Foot Model

N Martinelli, D Rosenbaum, M Schulze, A Marinozzi, V Denaro. University Campus Bio-medico, Rome.

11.3 1:32pm-1:44pm

The Sensitivity of Plantar Pressure and Talonavicular Alignment to Lateral Column Lengthening

I Oh, D Choi, B Williams C Imhauser, S Ellis, J Deland. Hospital for Special Surgery.

11.4 1:44pm-1:56pm

A Cadaveric Robotic Gait Simulator with Fuzzy Logic Vertical Ground Reaction Force Control

PM Aubin, E Whittaker, WR Ledoux. VA Puget Sound, Seattle.

11.5 1:56pm-2:08pm

In Vitro Description of Foot Bony Motion Using a Cadaveric Robotic Gait Simulator

E Whittaker, PM Aubin, WR Ledoux. VA Puget Sound, Seattle.

11.6 2:08pm-2:20pm

Comparison of Joint Pressure Changes with Adult Acquired Flatfoot and Talonavicular Joint Fusion for Correction of the Adult Acquired Flatfoot

ED Ward, WB Edwards, JR Cocheba, TR Derrick. Central Iowa foot Clinic, Perry, Iowa.

SESSION 12: POSTERS

2:20pm-2:45pm Friday September 17, 2010

See Abstracts in Session 3

SESSION 13: THE DIABETIC FOOT

2:45pm-4:25pm Friday September 17, 2010

Invited Speakers and Panel Discussion (Introduced by Peter Cavanagh)

13.1 2:45pm-3:00pm **EJ Boyko,** University of Washington
The Diabetic Foot: Unresolved and Future Issues

13.2 3:00pm-3:15pm **MJ Mueller**, Washington University of St Louis
The Diabetic Foot: Unresolved and Future Issues

13.3 3:15pm-3:30pm **SA Bus**. University of Amsterdam.
The Diabetic Foot: Unresolved and Future Issues

3:30pm-3:45pm Discussion

Free communications (Session Chairs: Joshua Burns and Sicco Bus)

13.4 3:45pm-3:57pm
Feasibility of Using Plantar Temperature to Assess the Biomechanics of the Plantar Foot
J Gerhard, R Semma, DA Wood, I Nwokolo, BL Davis, J Patel, M Matassini, VJ Hetherington, M Yavuz. Ohio College of Podiatric Medicine.

13.5 3:57pm-4:09pm
The Compressive Mechanical Properties of Diabetic Plantar Soft Tissue
S Pai, WR Ledoux. VA Puget Sound, Seattle.

13.6 4:09pm-4:21pm
A Longitudinal Investigation into Functional and Physical Durability of Insoles Used for the Preventative Management of Neuropathic Diabetic Feet
JS Paton, E Stenhouse, G Bruce, RB Jones. University of Plymouth.

DAY 3: SATURDAY SEPTEMBER 18, 2010

SESSION 14: FOOTWEAR

8:00am-9:50am Saturday September 18, 2010

Alex Stacoff Memorial Lecture: Keynote Speaker (Introduced by Alberto Leardini)

14.1 8:00am-8:55am: **BM Nigg**. University of Calgary.
Orthotics, Inserts and Shoes - Aligning the Skeleton

8:55am

14.2 *Announcement of the 10th Footwear Biomechanics Symposium in Tübingen, Germany. June 29-Jul 1, 2011*
S. Grau. Universitätsklinik Tübingen.

Free communications (Session Chairs: Jill McNitt-Gray and Smita Rao)

14.3 9:00am-9:12am

Relevance Ranking of Features Involved in Modelling Dorsal Pressures on the Foot Surface

J D Martín, M J Rupérez, C Monserrat, C Nester, M Alcañiz. University of Valencia.

14.4 9:12am-9:24am

The Impact of a Health Flip Flop on Asymptomatic Gait

C Price, R K Jones, P Graham-Smith. University of Salford.

14.5 9:24am-9:36am

Five-Toed Socks with Grippers on the Foot Sole Improve Static Postural Control Among Healthy Young Adults as Measured with Time-to-Boundary Analysis

J Shinohara, PA Gribble. The University of Toledo.

14.6 9:36am-9:48am

The Effect of Flexible Ski Boots on Knee Joint Loading and Muscle Activation

UG Kersting, P McAlpine, N Kurpiers, M de Zee, Aalborg University.

SESSION 15: POSTERS

9:50am-10:10am Saturday September 18, 2010

See Abstracts in Session 3

SESSION 16: NON-HUMAN PRIMATE FEET

10:10am-10:50am Saturday September 18, 2010

Invited Speaker (Introduced by Jill McNitt-Gray)

16.1 10:10am-10:30am: **R. Wunderlich**. James Madison University.

Evolution of the Human Foot and Bipedalism

Invited Speaker (Introduced by Christopher Nester)

16.2 10:30am-10:50am **TC Pataky**, R Savage, WI Sellers, RH Crompton. Shinshu University.

Footprint-based Gait Reconstruction of the 3.75 Ma Laetoli Hominin

SESSION 17: CLINICAL FOOT AND ANKLE III

10:50am-11:45am Saturday September 18, 2010

Free Communications: (Session Chairs: Amy Zavatsky and Bijan Najafi)

17.1 10:50am-11:02am

Postural Control and Tone of Gastrocnemius Muscle in Male Soccer Players and Endurance Trained Athletes

H Gapeyeva, H Aibast, T Kums, J Ereline, A Vain, K Jansen, H Lemberg, M Pääsuke. University of Tartu.

17.2 11:02am-11:14am

Real-World Locomotor Behavior Following Clubfoot Treatment: 10-Year Outcomes

MS Orendurff, KL Tulchin, VK Do, K Jeans, D Tabakin, LA Karol. Orthocare Innovations.

17.3 11:14am-11:26am

Strength Training of the Foot and Ankle in Paediatric Neuromuscular Disease

J Burns, JR Raymond, RA Ouvrier. University of Sydney.

17.4 11:26am-11:38am

The Effects of Foot and Ankle Strengthening with the AFX (Ankle Foot MaXimizer) on Athletic Performance in Male Varsity Basketball Players

SL Mann, RS Hall, NA Nembhard, TE Milner, JE Taunton. Progressive Health Innovations.

SESSION 18: TERMINOLOGY FOR THE FOOT AND ANKLE

12:00pm-12:45pm Saturday September 18, 2010

TM Greiner. University of Wisconsin.

SESSION 19: FOOT KINEMATICS I

1:00pm-3:00pm Saturday September 18, 2010

Keynote Speaker (Introduced by Joshua Burns)

19.1 1:00pm-2:00pm **A Leardini**, S Giannini. Istituto Ortopedico Rizzoli.

Kinematics of the Foot and Ankle: Review of Techniques and Findings

Free communications (Session Chairs: Stacie Ringleb and Tung-Wu Lu)

19.2 2:00pm-2:12pm

Medial Longitudinal Arch Deformation during Walking and Running

ER Hageman, ED Ward, TR Derrick. Iowa State University.

19.3 2:12pm-2:24pm

The Relationship Between Static Arch Rigidity and Foot Kinematics During Gait

MB Pohl, M Rabbito, R Ferber. University of Calgary.

19.4 2:24pm-2:36pm

Foot Bone Motion in Cavus, Neutral, and Planus Feet Using an In Vivo Kinematic Foot Model

E Whittaker, ME Hahn, WE Ledoux. VA Puget Sound, Seattle.

19.5 2:36pm-2:48pm

Should Linear Regression Be Used to Assess the Relationship between Multi-segment Foot Kinematics and Plantar Pressures in the Pediatric Patients with Foot Pathology?

KL Tulchin, MS Orendurff, LA Karol. Texas Scottish Rite Hospital.

19.6 2:48pm-3:00pm

Peri-talar Kinematics and Kinetics in Psoriatic Arthritis Patients with Achilles Tendon Enthesitis

E Hyslop, J Woodburn, IB McInnes, DE Turner. Glasgow Caledonian University.

SESSION 20: POSTERS

3:00pm-3:25pm Saturday September 18, 2010

See Abstracts in Session 3

SESSION 21: FOOT KINEMATICS II

3:25pm-4:25pm Saturday September 18, 2010

Free communications (Chairs: Debbie Nawoczenski and Dirk De Clercq)

21.1 3:25pm-3:37pm

Multi-segment Foot Kinematics During Barefoot Treadmill Running

JR Leitch, J Stebbins, AB Zavatsky. University of Oxford.

21.2 3:37pm-3:49pm

Evaluation of the Contact Surface at the Ankle during Walking and Slow Running

P Wolf, G Pron, R Jones, A Liu, C Nester, P Lundgren, A Lundberg, A Arndt. ETH Zurich.

21.3 3:49pm-4:01pm

Invasive in vivo Description of the Effect of Foot Orthoses on Foot Kinematics

Liu A, Nester CJ, Jones RK, Arndt T, Wolf P, Lundgren P, Lundberg A. University of Salford.

21.4 4:01pm-4:13pm

An In-shoe Comparison of Foot Kinematics in Normals Versus Mechanical Foot Pain

J Halstead, D McGonagle, AM Keenan, PG Conaghan, AC Redmond. Leeds Institute of Molecular Medicine.

21.5 4:13pm-4:25pm

Reliability of the Oxford Foot Model during Gait in Healthy Adults

CJ Wright, TG Coffey, BL Arnold. Virginia Commonwealth University.

SESSION 22: FOOT AND ANKLE IMAGING

4:30pm-5:30pm Saturday September 18, 2010

Free communications (Chairs: Michael Mueller and Tim Derrick)

22.1 4:30pm-4:42pm

in vivo Kinematics of Two-component Total Ankle Arthroplasty during Gait

S Yamaguchi, Y Takakura, Y Tanaka, S Kosugi, T Sasho, K Takahashi, S Banks, Orthop Surg. Chiba University.

22.2 4:42pm-4:54pm

Soft Tissue Thickness under the Metatarsals: Is it Reduced in Those with Toe Deformities?

K Mickle, B Munro, S Lord, H Menz, J Steele. University of Wollongong.

22.3 4:54pm -5:06pm

Distribution of Intrinsic Foot Muscles in Healthy and Plantar Fasciitis

R Chang, JA Kent-Braun, REA Van Emmerik, J Hamill. Kintec Footlabs Inc.

22.4 5:06pm-5:18pm

Skeletal Demands on the Ankles of Female Ballet Dancers Evident from Orthopaedic Imaging

JA Russell, RM Shave. Univ. of California-Irvine.

22.5 5:18pm-5:30pm

Structural Polymorphisms in Midtarsal Bone Alignment Lead to Focal Midfoot Pressures

DJ Gutekunst, L Liu, T Ju, PK Commean, KE Smith, MK Hastings, DR Sinacore. Washington University.

SESSION 23: AWARDS, I-FAB BUSINESS MEETING, CLOSING CEREMONY

POSTER PRESENTATIONS:

3.1 Foot Structure and Function Posters

3.1.1 *Anticipatory Effects on the Ankle Joint Loading During Cutting in Female Soccer Players*
IS Shin, JH Lee, JH Shon, SH Seok, EJ Park, JJ Ryue, YJ Yu, KK Lee. Seoul National University.

3.1.2 *Effect of Fatigue on GRF and Ankle Joint Loading during Drop Landing*
KK Lee, JH Lee, JH Shon, SJ Kong, EJ Park, JJ Ryue, YJ Yu. Kookmin University.

3.1.3 *Static Load Effect on Sagittal Deformation of the Foot*
VG Vardaxis, S Wells, G Vardaxis. Des Moines University.

3.1.4 *Foot Architecture Affects First Metatarsophalangeal Joint Function*
S Rao, J Song, P Demp, S Backus, B Chow, J Frey , S Ellis, M Lenhoff, J Deland, A Kraszewski, H Hillstrom. New York University.

3.1.5 *The Arch Index as a Potential Measure of Dynamic Foot Function in Children*
CM Kerr, J Stebbins, T Theologis, AB Zavatsky. University of Oxford.

3.1.6 *Impact of Chronic Gout on Foot Function: Case-Control Study*
K Rome, S Survepalli, A Sanders, M Lobo, FM McQueen, PJ McNair, N Dalbeth. AUT University.

3.1.7 *Using Plantar Pressure Regression Factor Scores to Discriminate Foot Posture*
B Najafi, RT Crews, JM. Fascione, JS. Wrobel. Rosalind Franklin University.

3.1.8 *Comparison of In-shoe Plantar Loading during Walking, Ascending and Descending Stairs*
O-Y Lo, L Iannuzzi, K Mroczek, S Rao. New York University.

3.1.9 *Reliability Characteristics of Tibialis Posterior EMG*
R Semple, D Rafferty, D Turner, J Woodburn. Glasgow Caledonian University.

3.1.10 *Non-rigid Work in Human Walking: Are Hard Collisions in Fact Soft?*
KE Zelik, AD Kuo. University of Michigan.

3.1.11 *Lower Extremity Movement Strategies in Individuals with Achilles Tendinopathy*
Y-J Chang, S Arya, RJ Gregor, K Kulig. University of Southern California.

3.1.12 *Geometric Forefoot Model*
H Hillstrom, A Kraszewski, P Demp, B Chow, M Lenhoff, S Backus, J Deland, J Song, B Heilman, S Rajan, A Woodley. Hospital for Special Surgery.

3.1.13 *Intra-tester Reliability of a Hand-held Device to Measure Heel Pad Stiffness*
M Frecklington, K Rome, P Webb, PJ McNair. AUT University.

3.2 Clinical Foot and Ankle Posters

3.2.1 *Characteristics of Participants with Recurrent Sprains*
EJ Nightingale, CE Hiller, C-WC Lin, E Delahunt, GF Coughlan, B Caulfield. University of Sydney.

3.2.2 *The Effects of Ankle Bracing on Mechanical Ankle Restraint in Individuals With and Without Chronic Ankle Instability*
PA Gribble, S. Cattoni. The University of Toledo

3.2.3 *Attentional Demands of Dynamic Postural Stability Control in Healthy Subjects and Patients with Functional Ankle Instability*
L Rahnama, B Akhbari, M Salavati, A Kazemnezhad. Shahid Beheshti Medical University.

3.2.4 *Altered Posture-dependent Hoffmann Reflex Modulation with Chronic Ankle Instability*
KM Kim, CD Ingersoll, J Hertel. University of Virginia.

3.2.5 *Leg Muscle Activation and Foot Pressure Related to Functional Ankle Instability*
JT Hopkins, M Coglianese, T Dunn, S Reese, MK Seeley. Brigham Young University.

3.2.6 *Effect of Chronic Ankle Instability on Knee Joint Position Sense*
F Pourkazemi, N Naseri, H Bagheri, Z Fakhari. Sydney University.

3.2.7 *Exploring Ankle Ligament Laxity Differences Between Male and Female Collegiate Athletes*
K Liu, G Gustavsen, T Kaminski. University of Delaware.

3.2.8 *Greater Inversion Laxity is Associated with More Inverted Rearfoot Positioning during Gait in Subjects with Chronic Ankle Instability*
J Hertel, S.Y. Lee, P.O. McKeon. University of Virginia.

3.2.9 *Relationships Between Measures of Posterior Talar Glide and Ankle Dorsiflexion in Healthy Individuals*
NL Cosby, J Hertel. University of Virginia.

3.2.10 *Kinematics, Kinetics, and Muscle Activity during an Ankle Injury: A Case Study*
EM Davis, L Stirling, S Landry, BM Nigg. University of Calgary.

3.2.11 *Invertor vs. Evertor Activation and Foot COP During Walking in Ankle Instability*
T Dunn, M Coglianese, M Seeley, S Reese, JT Hopkins.Brigham Young University.

3.2.12 *Intraoperative Pedography Improves Clinical Outcome at Follow-up of More than 20 Months*
M Richter, S Zech. Coburg Clinical Center.

3.2.13 *Three-dimensional Analysis of the Ludloff Osteotomy*
G Wu, W Mao, J-Z Zhang. Beijing Tongren HospitaL.

3.2.14 *Improvement in Dynamic Foot Pressure in Patients after Minimally Invasive Percutaneous Distal Metatarsal Osteotomy for Hallux Valgus*
T-W Lu, C-F Chang, K-S Shih, Y-T Chien. National Taiwan University.

3.2.15 *Significant Reduction of Abnormal Forces and Impulses of the Lesser Toes in Patients after Chevron Osteotomy for Hallux Valgus*
C-C Kuo, H-C Hsu, T-W Lu. National Taiwan University.

3.2.16 *Gait Changes in Patients after Minimally Invasive Percutaneous Distal Metatarsal Osteotomy for Hallux Valgus*
K-S Shih, C-F Chang, Y-P Chen, T-W Lu. Far Eastern Memorial Hospital.

3.2.17 *Detection of Forefoot Pain Based on Plantar Pressure Parameters*
NLW Keijsers, NMStolwijk, JWK Louwerens, J Duysens. Sint Maartenskliniek.

3.2.18 *The Impact of an Actuated Ankle-foot Orthosis on the Walking Performance in Healthy Subjects and Spinal Cord Injured Patients: A Systematic Review*
S Duerinck, E. Swinnen, P. Beyl, P. Van Roy, P. Vaes. Vrije Universiteit Brussel.

3.2.19 *Characterizing Multisegment Foot Kinematics, Kinetics and Plantar Pressure during Gait of Severely Deformed Feet in Rheumatoid Arthritis: A Case Study.*
S Del Din, Z Sawacha, A Guiotto, E Carraro, A Gravina, S Masiero, C Cobelli. University of Padova.

3.2.20 *Foot and Ankle Joint Kinematics in Rheumatoid Arthritis Cannot Only Be Explained by Alteration in Walking Speed*
R Dubbeldam, A Nene, J Buurke, H Baan, H Hermens. Roessingh Research and Development.

3.2.21 *Effects of Kinesio-Taping of the Gastrocnemius on Muscle Activity Patterns of the Lower Leg*
D Rosenbaum, H Klingelmann. University Hospital Münster.

3.2.22 *Effect of Elastic Taping on Motions of Ankle Syndesmosis Joint*
H Chai, YH Liu , JJ Lin. National Taiwan University.

3.2.23 *Time of Day Influences the Diameter of the Plantar Fascia*
ST Skou, MS Rathleff, CM Moelgaard, JL Olesen. Aalborg University.

3.2.24 *Running Has No Effect on the Diameter of the Plantar Fascia*
ST Skou, MS Rathleff, JL Olesen. Aalborg University.

3.2.25 *Tibia Internal Rotation Measured with a Kinematic Sensor –A Reliability Study*
BV Nilsen, R Moe-Nilssen. University of Bergen.

3.2.26 *Is a Neutral Foot Posture Optimal for Runners: A Comparative Study of Different Foot Postures on Injury Survival*
RG Nielsen, UG Kersting, S Rasmussen. Aarhus University Hospital.

3.2.27 *The Effect of Invertor/Evertor and Plantar-/Dorsiflexor Fatigue on Plantar Pressure Distribution*
PE Olivier. Nelson Mandela Metropolitan University.

3.2.28 *Soleus T-reflex Amplitude Modulation when Standing Humans Adopt a Challenging Stance*
GR Chalmers. Western Washington University.

3.2.29 *Investigation of Spread of Foot Deformity by Using Digital Podoscope Device*
SS Azad, E Karimi, FT Ghomshe. Azad University.

3.2.30 *Ankle Injuries and Their Impact in the Community*
CE Hiller, EJ Nightingale, SL Kilbreath, JR Raymond, J Burns, KM Refshauge. University of Sydney.

3.3 Modeling Posters

3.3.1 *An Advanced Kinematic Model for Enhanced Calculation of Foot Bone Kinematics.*
K Peeters, F Burg, I Jonkers, G Dereymaeker, J Vander Sloten. KU Leuven

3.3.2 *A Closed-chain 3D Finite Element Foot-ankle Model - Biomechanical Perspective on Forefoot Plantar Stress Redistribution Following Tendo-Achilles Lengthening*
W-M Chen, VPW Shim, PV Lee, JW Lee, SJ Lee, T Lee. National University of Singapore.

3.3.3 *A Novel Approach Using a FE-Foot Model for Clinical Applications*
Ch Wyss. Children's University Basel.

3.3.4 *A Comparison of the Performance of Hexahedral and Tetrahedral Elements in Finite Element Models of the Foot*
SC Tadepalli, A Erdemir, S Sett, PR Cavanagh. University of Washington.

3.4 Cadaver Simulation Posters

3.4.1 *Medial Ankle Instability Assessment Using Heel Varus in an External Rotation Stress Test*
T Vaseenon, N Muhlenbruch, DD Anderson, Y Tochigi, JE Femino. The University of Iowa.

3.4.2 *Biomechanical Study of Combination of Fixing the Posterior Malleolus and Mending Anteroinferior Tibiofibular Ligament to Reconstruct Syndesmotic Stability*
Z-Y Zhang, W Wang, L Zhang, X-Z Zhang. The Fifth Affiliated Hospital of Liaoning Medical University.

3.4.3 *Kinematics of Adult Acquired Flatfoot and Correction of Adult Acquired Flatfoot with Talonavicular Joint Fusion*
TR Derrick, ED Ward, JR Cocheba, WD Edwards, ER Hageman. Iowa State University.

3.5 Diabetic Foot Posters

3.5.1 *RF Score as a Reliable Measurement to Assess Improved Dynamic Plantar Loading Improving Post Foot Reconstruction Surgery*
B Najafi, T Crews, DG Armstrong, J Wrobel. Rosalind Franklin University.

3.5.2 *Different Foot Kinematics, Kinetics and Plantar Pressure Patterns within the Gait of Diabetic Subjects: Cluster Analysis*
A Guiotto, Z Sawacha, G Guarneri, A Avogaro, C Cobelli. University of Padova.

3.5.3 *The Biomechanical Foot Differences between Subjects with Diabetes and Hallux Valgus Compared to Asymptomatic Non-diabetic Individuals*
J Song, J Furmato, B Heilman, D Tango, E Zoltick, A Kraszewski, B Chow, M Lenhoff, S Backus, J Deland, P Demp, S Rajan, A Woodley, H Hilstrom. Temple University.

3.5.4 *Association of Diabetic Foot Ulcer Healing and Offloading Compliance: Initial Findings*
RT Crews, B-J Shen, AJM Boulton, L Vileikyte. Rosalind Franklin University.

3.5.5 *Offloading the Foot by Varying Fiber Density within Diabetic Hosiery*
JT Weiland, RT Crews. Rosalind Franklin University.

3.5.6 *Histomorphological Evaluation of Diabetic and Non-diabetic Plantar Soft Tissue*
YN Wang, K Lee, WR Ledoux. VA Puget Sound, Seattle.

3.5.7 *Target Strain Errors in Plantar Tissue Testing*
S Pai, WR Ledoux. VA Puget Sound, Seattle.

3.6 Footwear Posters

3.6.1 *Relationships between Foot Strike Patterns and Foot Pronation in Running*
M Sakaguchi, N Shimizu, H Kanehisa, T Yanai, Y Kawakami. Waseda University.

3.6.2 *An Investigation into Lower Limb Muscle Activation during the Golf Swing*
P Worsfold. University of Chester.

3.6.3 *The Influence of Sprint Spike Stiffness on Sprinting Performance and Metatarsophalangeal Joint Kinematics*
G Smith, M Lake, T Sterzing, P Worsfold. University of Chester.

3.6.4 *Effect of Footwear on the Gait of Children: A Systematic Review*
C Wegener, AE Hunt, B Vanwanseele, J Burns, RM Smith. The University of Sydney.

3.6.5 *In-vivo Investigation of the Existing Shear Reducing Diabetic Insoles*
J Patel, M Matassini, DA Wood, T Ky, F Luckine, J Gerhard, I Nwokolo, A Craig, VJ Heterington, M Yavuz. Ohio College of Podiatric Medicine.

3.6.6 *Foot Orthoses Contribute to Better Gait Initiation*
B Najafi, D Miller, BD Jarrett, JS Wrobel. Rosalind Franklin University.

3.6.7 *Assessment of the Body's Adaptation to Prescribed Foot Orthoses Using Dual Task Paradigm and Gait Initiation*
B Najafi, D Miller, BD Jarrett, JS Wrobel. Rosalind Franklin University.

3.6.8 *Biomimetic Ankle-foot Prosthesis with Compliant Joints and Segmented Foot*
J Zhu, Q Wang, L Wang. Peking University.

3.6.9 *Effect of Heel Height on the Distributions of Plantar Triaxial Stresses*
Y Cong, M Zhang. The Hong Kong Polytechnic University.

3.6.10 *Hard Sole or Soft Sole? A Study into Shoe-specific Ground Reaction Model Parameters*
R Naemi, N Chockalingam. Staffordshire University.

3.6.11 *Effects of Temperature on the Performance of Footwear Insole Foams Subjected to Quasi- Static Compression Loading*
MR Shariatmadari, R English, G Rothwell. Liverpool John Moores University.

3.6.12 *Effect of Negative Heel Profile Shoes on Triceps Surae: A Pilot Study*
Helen Branthwaite, Nachiappan Chockalingam, Anand Pandyan. Staffordshire University.

3.6.13 *The Effect of Memory Elastic Alloy Reinforced Shoes on Come-off Distance*
TW Kim, KK Lee, JH Lee, JH Shon, SJ Kong, SHSeok, EJ Park, JJ Ryue, YJ Yu. Kookmin University.

3.7 Non-human Primate Posters

3.7.1 *Kinematic Study of Load Response in the Human and Non-Human Primate Foot: Clues for the Evolutionary Development of Human Bipedalism*
TM Greiner, KA Ball. University of Wisconsin - La Crosse.

3.8 Foot and Ankle Imaging Posters

3.8.1 *Quantification of Foot Bone Motion Using Biplane Fluoroscopy*
WR Ledoux, R Tsai, M Fassbind, BJ Sangeorzan, DR Haynor. VA Puget Sound, Seattle.

3.8.2 *Design of a 1st Metatarsophalangeal Hemi-Arthroplasty Implant Based on Morphological Data*
A Kumar, B Donley, PR Cavanagh. University of Washington.

3.8.3 *Real-time Statistical Analysis of Plantar Pressure Data: A Preliminary Feasibility Study*
TC Pataky. Shinshu University.

3.8.4 *Talocrural Joint Contact Modelling with Thin-plate Splines*
N. Ying, W Kim, A Veloso. Nanyang Technological University.

3.9 Methods Posters

3.9.1 *Mechanical Testing of the Lapidus Plate for First Metatarsocuneiform Joint (MCJ-1) Arthrodesis*
D Rosenbaum, L Martinelli. University Hospital Münster.

3.9.2 *Low-cost Bi-axial Shear Stress Transducer: A Step Closer to Shear Distribution Measurement*
HC Lau, S Wearing, B Stansfield, S Solomonidis, W Spence. University of Strathclyde.

3.10 Foot Kinematics Posters

3.10.1 *A Systematic Review of Kinematic Models Used in Foot and Ankle Biomechanics*
C Bishop, D Thewlis, G Paul. University of South Australia.

3.10.2 *A Comparison of Two Multi-segment Foot Models in High-arched and Low-arched Athletes*
D Powell, D.S. Williams, S Zhang. University of Texas of the Permian Basin.

3.10.3 *A Reliability Study of Biomechanical Foot Function in Psoriatic Arthritis*
E Hyslop, J Woodburn. IB McInnes, R Semple, L Newcombe, G Hendry, D Rafferty, S De Mits, DE Turner. Glasgow Caledonian University.

3.10.4 *A Clinically Applicable Six-segmented Foot Model*
S De Mits, V Segers, J Woodburn, D Elewaut, D De Clercq, P Roosen. Ghent University.

3.10.5 *Repeatability of the Salford Multi-segment Foot Model*
H Jarvis, CJ Nester, P Bowden, RK Jones, S Forghaney. University of Salford.

3.10.6 *Classification of Mid-foot Break Using a Multi-segment Foot Model*
JD Maurer, AH Black, CM Alvarez, V Ward, KR Davies, RD Beauchamp. Sunny Hill Health Centre for Children.

3.10.7 *Use of Multiple Calibration in Multisegment 3D Foot Kinematics*
ZS Sawacha, A Guiotto, C Fassina, S Fantozzi, R Stagni, L Tersi, C Cobelli. University of Padova.

3.10.8 *A Novel Technique for Improving Marker Placement Accuracy*
S Telfer, G Morlan, E Hyslop, R Semple, D Rafferty, J Woodburn. Glasgow Caledonian University.

3.10.9 *Two Foot Marker Placement Methods Reveal Different Coordination Patterns during Running*
J Becker, S James, L Osternig, L-S Chou. University of Oregon.

3.10.24 *Personalised Orthotics for Rheumatoid Arthritis (po4ra): Development of 3D Kinematic and Plantar Pressure Based Design Rules for Commercial CAD Software*
K Gibson, S Telfer, KW Dalgarno, J Pallari, J Woodburn. Glasgow Caledonian University.

3.10.25 *Gait Analysis after Bilateral Total Ankle Replacement – A 12-month Follow-up Case Study*
H Wang, E Slaven, J Frame. Ball State University.

3.10.26 *Patients with Medial Tibial Stress Syndrome Are Characterized by Low Foot Kinematics Complexity and High Muscle Activation Complexity*
MS Rathleff, CG Olesen, A Samani, UG Kersting, P Madeleine. Aalborg Hospital.

3.10.27 *Kinematic and Electromygraphic Asymmetries of Ankle Joint during Drop Landing*
WX Niu, Y Wang, Y He, YB Fan, QP Zhao. Beihang University.

3.10.28 *Center of Pressure Trajectories during Landings with a Secondary Redirection Task*
L Held, JL McNitt-Gray, H Flashner. University of Southern California.

3.10.29 *Foot Loading after Several Days of Long Distance Walking*
NM Stolwijk, J Duysens, JWK Louwerens, NLW Keijsers. Sint Maartenskliniek.

3.10.30 *Validation and Relevance of Methodologies for Anatomical and Geometrical Identification of Plantar Sub-regions in Baropodometry*
J Stebbins, T Tsvetkova, C Giacomozzi. Nuffield Orthopaedic Centre.

What has Science done for the Foot and Ankle surgeon?

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INTRODUCTION

Foot and ankle surgery is a fast growing field within orthopaedic and trauma surgery. As in many other fields Evidence Based Medicine (EBM) is used as a solid and safe basis for decision making and treatment. Guidelines, Evidence Level and Grade of Recommendation are three different "tools" of EBM. The basis for high level of evidence is outcome research.

Basic science including experimental and other labratory research is a basis for altered treatment and outcome. The foot and ankle surgeon does of course profit from science and some examples of novel methods are given such as Intraoperative 3D-Imaging, Computer Assisted Surgery and Pedography including Intraoperative Pedography.

INTRAOPERATIVE 3D-IMAGING

Intraoperative 3D-imaging with the first available device (ISO-C-3D, Siemens, Germany) has shown potential benefit in foot and ankle surgery. The aim of a study was to assess the clinical use of the second generation device (ARCADIS-3D, Siemens, Germany) in comparison with earlier experience with the first generation device.

In a matched pair study, the ISO-C-3D/ARCADIS-3D was used for intraoperative visualization after reduction/correction and internal fixation. On average, the operation was interrupted for 440/320 seconds (ISO-C-3D/ARCADIS-3D); 120/60 seconds, on average, for the scan and 210/180 seconds, on average, for evaluation of the images by the surgeon. In 39%/34% of the cases (24/21 of 62), the reduction and/or implant position was corrected during the same procedure after the ISO-C-3D/ARCADIS-3D scan.

COMPUTER ASSISTED SURGERY (CAS)

CAS has shown the potential to increase the accuracy of surgical procedures in different fields of orthopedic surgery. The clinical experiences of 100 cases with CAS guided correction arthrodeses were evaluated. The time needed for preparation was 356 seconds (5 minutes, 56 seconds) (4 – 30 minutes), the correction process took 28 seconds (12-140 seconds). The CAS system encountered malfunctions in 3 procedures (3%). In the remaining cases, all the achieved angles or translations were within a maximum deviation of 2° or 2mm when compared to the planned correction ($p<.05$).

INTRAOPERATIVE PEDOGRAPHY (IP)

The purpose of a randomized prospective study was to assess the clinical use, and to analyze the potential clinical benefit of intraoperative pedography (IP) in a sufficient number of cases in comparison with cases treated without IP.

One hundred cases were included until April 11, 2008 In 24 of the 52 patients (46%), the correction was modified after IP during the same operation. All patients completed follow-up after 6 to 24 months (12 months on average). The follow-up scores were AOFAS, 82.5±19.7; VAS FA, 80.2±13.4; SF36, 84.3±18.8. The scores from the IP group were significantly higher than the scores from the group without IP (t-test, $p<.05$).

CONCLUSION

Intraoperative three-dimensional visualization with the ISO-C-3D/ARCADIS-3D can provide useful information that cannot be obtained from plain films or conventional C-arms. The second generation (ARCADIS-3D) provides faster scan and evaluation that reduces time spent. No other benefits were seen.

With CAS guidance for the correction of deformities of the ankle, hindfoot and midfoot, a surgeon is provided with a high accuracy and the ability for a fast correction process to take place. The significance of the introduced method is high in those cases, because the high accuracy may lead to an optimized clinical outcome.

In 46% of the cases a modification of the surgical correction were made after IP in the same surgical procedure. The follow-up scores were higher in the group with IP than in the group without IP. Modifications after IP improve the biomechanical function of the foot which improve the clinical outcome.

The shown results of science help the surgeon.

Foot Structure and Function:
Six Blind Men and the Elephant Revisited

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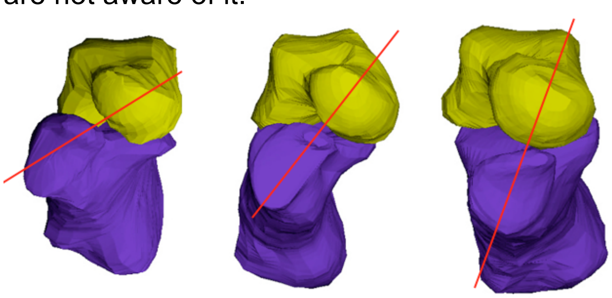
INTRODUCTION

A clinician and a biomechanist may look at the same foot and see very different things. A clinician usually doesn't see the foot at all unless it hurts, is deformed, infected, stiff or can't fit in a shoe. The clinician view focuses on the history, the exam and the images. The biomechanics view is usually associated with increased performance or reduction of pressure. It is focused on shape, motion and tissue properties in the present.

There are some fairly universal truisms about the foot. First, the foot follows a spectrum of shapes from very low to very high arch, the former accompanied by valgus and abduction and the latter accompanied by varus and adduction. All shapes are 'normal.' Flat feet are typically more flexible, more likely to progress and more easily corrected when correction is needed. High arched feet tend to be more rigid and are not easily restored to neutral shape. Many inactive people have cavus shape and are not aware of it.



The diagram shows a flatfoot on left and a cavus foot on the right with a neutral foot in between. The red line represents the axis of flexion of Chopart's joint.

Secondly, the hindfoot bones determine the midfoot function. When the hindfoot is in valgus, the calcaneus tends to be posterior and lateral to the talar plumb line. As it migrates in that direction the midfoot becomes more horizontal. The cuboid moves more lateral to the navicular, leaving the axis of Chopart's joint more parallel to the ground. Conversely when the hindfoot is in varus, the calcaneus is more medial and anterior under the talus. The cuboid is pushed forward to reside plantar to the navicular. The Chopart axis rotates more horizontal making the joint resistant to motion in the sagittal plane.

The final mechanical truism is that the midfoot acts as a universal joint connecting the dynamic hindfoot to the more static forefoot that acts as a launch platform for forward propulsion. The forefoot just has to meet the floor evenly in flatfoot position and in heel rise. Since the hindfoot changes from flexible valgus to rigid varus during this process, the midfoot must act as a universal joint.

METHODS

We will review clinical images/studies and mechanical studies to find the common ground.

DISCUSSION

Clinicians and scientists working together creates the best opportunity to solve the structural problems.

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Metatarsal Loading Pattern Differences with Age and Sex in Adolescent Athletes

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INTRODUCTION

Metatarsal stress fractures are a common sports injury [1]. While sex differences in the incidence rate of metatarsal stress fractures are inconclusive, the current data indicates a potential increased risk of fifth metatarsal fracture in males relative to females [2, 3]. The purpose of this study was to examine the effects of age and sex on dynamic metatarsal loading patterns in adolescent athletes.

METHODS

Subjects included middle and high school age soccer athletes between 12 and 17 years old (male n=64; female n=77). Dynamic barefoot plantar pressure distribution was obtained as each subject walked on a 6 m walkway. An emed-x system (Novel) platform that consists of a 48x32cm matrix of capacitive sensors (4 sensors/cm^2) collected at 100Hz was mounted in the center of the walkway and level to the walking surface. From each walking trial the foot was subdivided into 10 anatomical regions as described by Cavanagh et al [4]. The analyses were focused on the medial forefoot (metatarsal 1), central forefoot (metatarsal 2) and lateral forefoot (metatarsals 3-5). Peak pressure and force time integral were calculated in each foot region. A 2X6 ANOVA was used to investigate the effects of sex and age on the evaluated dependent variables.

RESULTS

Force time integral increased with age under each forefoot region ($p<0.05$, Figure 1) in both sexes. Peak pressure was increased with age (Table 1) in the lateral forefoot ($p<0.05$) but not the medial and central regions ($p>0.05$). More mature males demonstrated greater lateral forefoot force time integral compared to females ($p<0.05$). No sex differences in force time integral were found in the medial and central forefoot ($p>0.05$).

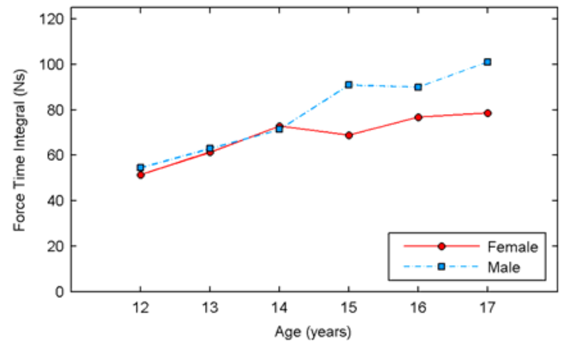

Figure 1: Force time integral under the lateral forefoot region for male and female athletes.

DISCUSSION

The increased force time integral in the lateral forefoot, as males mature, may underlie the increased risk of lateral forefoot stress fractures in male athletes. Future longitudinal investigations into the effects of forefoot load distribution on injury incidence are warranted to aid interventions designed to reduce lateral forefoot stress fractures.

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Table 1. Peak pressure under the medial, central and lateral forefoot regions

Age	Medial Forefoot (kPa)		Central Forefoot (kPa)		Lateral Forefoot (kPa)*	
	Female	Male	Female	Male	Female	Male
12	240.4 ± 97.5	204.4 ± 90.3	351.6 ± 95.3	354.7 ± 106.4	287.8 ± 45.7	315.0 ± 86.5
13	257.8 ± 87.5	293.9 ± 127.8	418.9 ± 154.1	452.6 ± 223.7	359.7 ± 109.0	326.4 ± 121.5
14	234.0 ± 37.5	277.7 ± 143.6	482.8 ± 221.1	356.3 ± 95.7	445.8 ± 149.6	305.6 ± 69.3
15	339.9 ± 214.2	273.8 ± 92.7	421.8 ± 190.4	435.8 ± 219.6	406.3 ± 141.5	398.8 ± 116.6
16	282.8 ± 103.7	240.3 ± 59.9	343.7 ± 75.1	384.0 ± 69.8	417.2 ± 150.0	368.4 ± 109.6
17	272.4 ± 127.5	328.1 ± 97.4	468.3 ± 194.3	429.3 ± 74.2	430.2 ± 119.5	452.9 ± 207.4

** indicates significant main effect of age*

Spatiotemporal Volumetric Analysis of Dynamic Pedobarographic Data

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INTRODUCTION
Recent studies have demonstrated visual and theoretical benefits of continuous topological statistical analysis [1] of 2D pedobarographic data [2]. The current study extends these techniques to the (3D) spatiotemporal domain by examining the pressure correlates of walking speed over the entire stance phase.

METHODS
Data from a single subject [2] were re-analysed. Twenty trials of each of slow, normal, and fast walking were performed in a random order. Pedobarographic data were collected at 500 Hz (0.5m Footscan3D, RSscan, Olen, Belgium) and walking speed data were collected at 100 Hz (ProReflex, Qualisys, Gothenburg, Sweden). Average walking speeds were 0.8, 1.2, and 1.9 m/s. The 3D data (x, y, t) were registered using an optimal spatial rigid body transformation and linear interpolation over stance phase. Recorded walking speed was then regressed against the voxel pressure data to compute a 'SPM{t}' image [1]. Statistical significance was conducted using random field theory [1].

RESULTS
Mean pressures broadly increased with walking speed and faster speeds were associated with faster forefoot transfer (Figure 1). Statistical volumes (Figure 2) emphasize that these behaviours were highly significant topologically ($p<0.001$) and highlight the relative timing differences of the walking speeds.

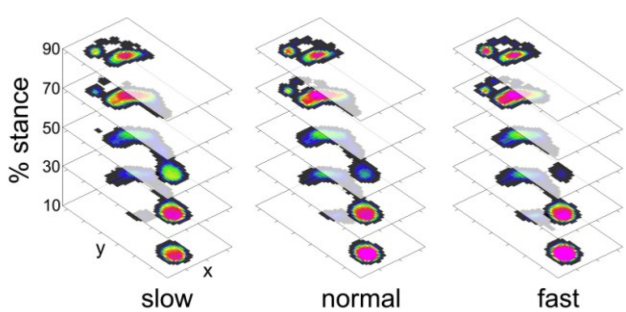

Figure 1: Mean pressure distributions.

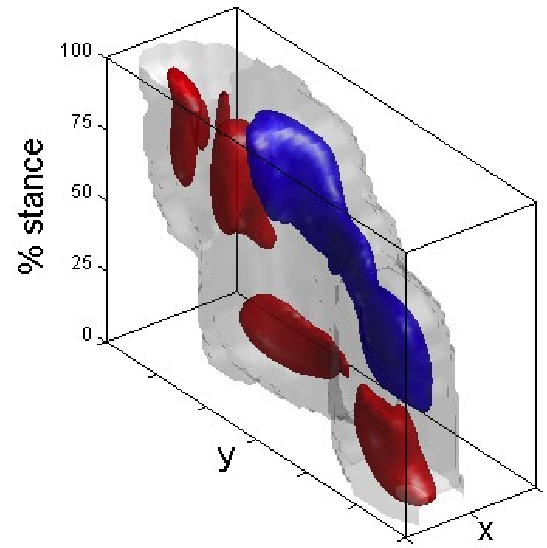

Figure 2: Volumetric statistical image (thresholded at SPM{t}>3.0) depicting areas of significant positive (red) and negative (blue) correlation with walking speed. Light grey areas indicate the search volume (thresholded at pressure > 0.5 Ncm^{-2}). All clusters were significant ($p<0.001$).

DISCUSSION
The current results are consistent with the notion that the windlass mechanism increases foot rigidity [3] for conceivable propulsive benefit via active pre-loading of the plantar aponeurosis during pre- or early-stance [2]. Multi-subject analyses are currently underway to test for the current effects at the population level. The proposed topological technique permits analysis of the entire spatiotemporal pedobarographic volume and, as such, has the potential to afford unique insights into dynamic foot function.

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Deep plantarflexor strength increase changes rearfoot motion in shod running

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INTRODUCTION

Several running injuries have been linked to excessive rearfoot motion [3]. Although bio-positive effects after strength training interventions have been observed [1,2], the role of shank muscle strength is not well understood. Purpose of this study was to identify the morphological and biomechanical effects of functional pronator and supinator strength training. For an efficient strengthening of the pronators and supinators a new training machine (FSTM) was developed with its rotational axis aligned to the subtalar joint axis as identified by Inman [4]. It was hypothesized that the specific training of the deep plantarflexors (Tibialis Posterior, Flexor Hallucis Longus, Flexor Digitorum Longus) and the peroneal muscles will stiffen the ankle joint complex and reduce the subtalar range of motion during ground contact in running.

METHODS

For the intervention study 29 healthy male rearfoot runners were randomly assigned into an experimental group (n=22) and a control group (n=7). Over a period of ten weeks the subjects went for 3 training sessions per week with single-set strength training until task failure within eight to ten repetitions. The subjects of the experimental group (FST - Functional Strength Training) performed supinations and pronations at the FSTM with their right leg. The left leg was trained with plantar and dorsiflexions at traditional training machines and served as intraindividual control leg (TT - Traditional Training). The control group (CG) performed plantar and dorsiflexions with both legs.

In a pre-post-test design pronator and supinator MVC as well as and rearfoot motion (electrogoniometer) during shod running across a Kistler force plate in 3.3 m/s were measured. The subjects ran in two test conditions: A=regular running shoe; B=same shoe with inserted valgus wedges (height difference: 6mm). Additionally, muscle volume was quantified by magnetic resonance images from both lower leg muscles of 9 randomly chosen test persons of the experimental group. The results were analyzed with a two-way repeated measures ANOVA.

RESULTS

Compared to TT, FST resulted in significantly higher pronator (14% vs. 8%, $p<0.01$) and supinator MVC (25% vs. 12%, $p<0.01$). Both, FST and TT increased the supination angle at touchdown in running for both shoes A, B ($p<0.01$). In shoe B pronation velocity was reduced about 16% in FST (FST vs TT: $p<0.01$) and delayed (FST: +23%; TT: +3%; $p<0.05$). The median power frequency of the vertical ground reaction force signal increased in A ($p<0.05$) and B ($p=0.05$). MRI recordings showed training specific increase in muscle volume of TP (+10%; FST vs. TT: $p=0.2$), FHL (+6%; FST vs. TT: p=0,09) after FST. No leg dominant effects in any outcome measures were observed in CG.

DISCUSSION

Both, FST and TT induced strength gains that stiffened the ankle joint complex and increased supination angle at touchdown in running. Compared to TT, the muscular control of rearfoot motion is enhanced after FST. It can be concluded that specific strength increase of the deep plantarflexors, especially FHL and TP, leads to reduced and delayed pronation velocity (shoe B). This study shows the beneficial effects of functional pronator and supinator strength training for runners in controlling rearfoot motion.

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Measuring dorsal arch height in different stances with plantar foot muscles passive and active: A reliable determinant of MLA articulation in older people

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INTRODUCTION

At present there are no measures of medial longitudinal arch (MLA) height while activating the plantar muscles. This study measured dorsal MLA height with plantar foot muscles passive, then active, in three stances (seated, standing, standing knee bend) to determine intra rater reliability and to examine MLA range of motion in older people.

METHODS

Study participants comprised 31 adults (mean age 60; sd 7.3), reporting minor foot and/or lower limb problems and low levels of physical activity. At both measurement collections, which were four weeks apart, dorsal MLA height was recorded during two tasks (intrinsic plantar muscles passive or active; in arch elevation) for all three stances. The dorsal arch height (DAH) measured with a vernier height gauge was divided by the truncated foot length to determine the dorsal arch height ratio (DAHR).

RESULTS

Within rater reliability levels (from intraclass correlation coefficients) were near perfect for DAH for both passive (0.97 to 0.99) and elevation tasks (0.96 to 0.98) across stances. The only statistically significant differences between stances for DAH were that seated was higher, than either standing or knee bend (ANOVA). This applied to both tasks.

There were statistically significant changes from passive to elevated tasks for all three stances for both DAH and DAHR methods of measurement (paired t tests).

Measures of passive and elevated DAH and DAHR in all stances showed almost perfect significant correlations ($r>0.95$). There was minimal correlation between the passive to difference and elevated to difference DAH(R) values (all $r^2<0.01$). While correlations for both passive and elevated DAH(R) in all three positions were only moderate ($r=0.44$ to 0.57, those between DAH(R) for task difference for each position were near perfect ($r=0.97$).

DISCUSSION

The paper grip test [1] and custom pulley system [2] are the sole measures of plantar muscle activation reported. Since arch height is in part regulated by the intrinsic foot muscles [2,3] measures of arch elevation may be a useful clinical test of intrinsic plantar muscle activation and MLA range of movement.

The new test, elevated DAH(R) in different stances, has similar high reliability as passive DAH(R) measures [4,5] and provides information about MLA articulation. The measure of difference is specific and discrete and might clarify MLA dynamic function. Further research is warranted as practising arch elevation may augment intrinsic foot muscle activation.

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The relevance of subtalar-joint-anatomy for chronic overuse injuries of the lower limbs.

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INTRODUCTION

Chronic overuse injuries (COI) especially at the Achilles tendon are of major importance to athletes, especially high performance athletes. Little is known of the predisposing extrinsic and intrinsic risk factors. It has been speculated that individual anatomy, described by calculated joint axes or arch index of the foot, could be such risk factors. Therefore the aim of this study was to identify the relationship between individual spatial orientation of the subtalar-joint-axis (STA), Arch index and COI in runners.

METHODS

An ultrasonic pulse-echo based measurement system (Zebris®) determines the spatial orientation of STA in-vivo and in real-time. Inclination angle (projection of the axis to the sagittal plane) and deviation (projection to the transverse plane) have been calculated. Arch-index and the angle of gait was determined by treadmill analysis with plantar pressure measurement (Zebris®). Information about previous injuries and running performance was collected with a questionnaire. 495 subjects mainly including long distance runners with a running performance of at least 25 km per week and a running history of 3 years were measured. 307 subjects were included.

RESULTS

There was a significant (p=0.002) mean difference between the deviation angle in people with Achilles tendon problems (ATP) and people without and also between men and women (Table 1). 69% out of 307 subjects have been injured before. We counted a total of 664 injuries. 22% of these injuries were located at the ankle, 21% at the knee and 14 % at the Achilles tendon. By means of the Arch-index we found out, that 42% of people with ATP had a high arch. In people without ATP only 34% had a high arch. The average of the inclination angle of 614 STA was 42° ± 16° and the deviation angle 11° ± 23° (Table 1).

DISCUSSION

The deviation angle of subjects with ATP was higher (18°) than in people without ATPs. Based on these results long distance runners with higher deviation angle are at a higher risk to suffer from Achilles tendon problems.

The injury-rate in this study was higher (69%) than stated by Mayer [1] (30%). The Results of Arch-index confirmed the proposition from Lohrer [2] that high arch is a common reason for ATP. The data of the mean inclination angle of the STA approximates the results of Isman [3] (Table 1). In contrast, the mean deviation angle measured in vivo is smaller and the axis closer to the foot bisection than measured by Isman.

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Table 1: Results of this study compared to other studies from Literature.

STA	Results of this Study					Literature	
	Total (N=614)	**ATP (N=95)**	**No ATP (519)**	**Female (N=178)**	**Male (N=436)**	**Isman [3] (N=46)**	**Lewis [4] (N=24)**
Inclination angle [°]	42 ± 11	44 ± 16	42 ± 17	45 ± 17	41 ± 16	41 ± 9	33 ± 11
Deviation angle [°]	11 ± 23	18 ± 23	10 ± 23	6 ± 22	13 ± 23	23 ± 11	18 ± 10

Anticipatory effects on the ankle joint loading during cutting in female soccer players

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INTRODUCTION

Female athletes showed higher anterior cruciate ligament (ACL) injury rates than male athletes [1]. Most non-contact lower extremity injuries in soccer occur during a cutting movement [2]. The ankle joint relies on the dynamic restraints of the lower extremity in order to prevent injury. The purpose of this study was to compare the ankle joint loading during preplanned and unanticipated running and cutting maneuvers.

METHODS

Thirteen female college soccer players participated in the study (age: 20.2 ± 0.37yrs, mass: 55.2 ± 4.44kg, and height 162.5 ± 5.17cm). The tasks for this experiment included a side cut to 45° and a crossover cut to 45°.

Subjects were asked to run with preferred speed then jumped at 1.5 meter from the force plate and cut to preplanned or unanticipated direction which was controlled by random visual cue. Kinematic, ground reaction force (GRF), and kinetic data were collected using a six-camera, 200-Hz VICON motion analysis system and a AMTI force plate at 2000 Hz.

The three-dimensional ankle joint moments were determined by inverse dynamics. Paired t-test was conducted to determine significant differences between conditions with $p < 0.05$.

RESULT

There were no significant differences of ground reaction force and ankle moment in the preplanned and the unanticipated conditions. Eversion moment, time to peak anterior and contact time were significantly different in the preplanned and the unanticipated conditions.

DISCUSSION

We anticipated that it would be loaded in ankle during unanticpated situation with cutting task, but the results were different. Thor studied that Cutting maneuvers performed without adequate planning may increase the risk of noncontact knee ligament injury due to the increased external varus/valgus and internal/external rotation moments applied to the knee[3]. We also form a hypothesis that unanticipated condition would be increased ankle moment. However the results differ from our hypothesis. The results are probably due to increase contact time during landing motion. We need to study about the relation between motion and eletromygraphic(on-off muscle).

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Table 1: Mean(±standard deviation) of kinematic and kinetic variables during the cutting motion

Variables	Dorsiflexion moment(Nm)	Plantarflexion moment(Nm)	Eversion moment(Nm)	Inversion ankle moment(Nm)	range_foot_ progression_angle(°)
Preplanned	2945.02 ±635.76	211.98±161.24	**367.4 ± 197.73**	105.65 ± 84.01	15.23 ± 8.19
Un-anticipated	2721.08±413.76	169.93±139.37	**223.03 ± 35.57**	119.73 ± 93.03	13.39 ± 13.75
	Contact _time(sec)	time to peak_Medial(sec)	time to peak_Ant(sec)	time to peak_Posterior (sec)	time to peak_Vertical (sec)
Preplanned	**0.27 ± 0.04**	0.026 ± 0.009	**0.23 ± 0.04**	0.033 ± 0.012	0.02 ± 0.01
Un-anticipated	**0.31 ± 0.05**	0.03 ± 0.007	**0.26 ± 0.04**	0.027 ± 0.008	0.03 ± 0.03

Bolded number indicates significant differences at $\alpha = 0.05$

Effect of fatigue on GRF and ankle joint loading during drop landing

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INTRODUCTION

The ankle sprain is the one of the most frequent sport injuries and 90% of sprains occurred during landing after the blocking movements [1]. Although many researchers were interested in the functional instability of ankle joint, they did not provide the exact causes of this phenomenon following acute ankle sprain [2]. Ankle injuries are frequently occurred when athletes are in a fatigued state [3]. The aim of this study was to investigate the fatigue effect on stability of landing and ankle joint loading.

METHODS

Thirteen female college soccer players were participated. Their age was 20.2±0.37 yrs, weight and height were 55.2±4.44 kg and 162±5.17 cm, respectively. Subjects were excluded who had a history of previous ankle injury, surgery and pain in lower extremity.

Subjects performed landing from a chair (height: 50cm) onto the force platform with a single one leg (right). Landings were measured before and after the fatigue protocol. To give the fatigue, each subject was asked to do squat exercise with lifting 30kg of barbells on the shoulder (15times/1set, 5 reps).

In order to collect the kinetic and kinematic data, a AMTI force platform at 2000Hz and six Vicon cameras at 200Hz were used and synchronized. To identify landing pattern, vertical, medio-lateral, and anterior-posterior ground reaction forces, and area of center of pressure were analyzed, The three-dimensional ankle joint moments were determined by inverse dynamics. Paired sample t-test was used to test the difference of the ankle joint stability after the fatigue protocol. Significance level was set to p<.05.

Figure 1: Single leg drop landing(right)

RESULTS

There was no significant difference in the ankle moment, COP area and GRF(frontal plane) after the fatigue protocol during drop landings. Anterior GRF showed significant difference between pre and post execution of the fatigue protocol(p= .016).

DISCUSSION

The purpose of this study was to investigate the stability of the ankle joint during drop landing motion after performing the execution of the fatigue protocol. The results of this study didn't show significant changes in the characteristics of ankle & GRF(frontal plane). It could be so that participants did not feel fatigue. If these problems consider, Further analyzes will be done to get possible statistically significant correlation.

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This research project was supported by the Sports Promotion Fund of Seoul Olympic Sports Promotion Foundation from Ministry of Culture, Sports and Tourism

Table 1: Mean(±standard deviation) of kinetic variables during the landing

Variables	Posterior GRF(%BW)	Anterior GRF(%BW)	Lateral GRF(%BW)	Medial GRF(%BW)	Vertical GRF(%BW)
Pre-fatigue	0.6456 ± 0.1431	**0.2113 ± 0.1629**	0.1034 ± 0.0788	0.1752 ± 0.0629	5.242± 0.725
Fatigue	0.6747 ± 0.1130	**0.0830 ± 0.0798**	0.0916 ± 0.0784	0.2150 ± 0.0684	5.101± 0.806

	COP Area(mm^2)	Inversion Moment(Nm)	Eversion Momen(Nm)	Dorsiflexion Moment(Nm)
Pre-fatigue	9843.96 ± 277.26	502.80 ± 220.87	99.10 ± 67.80	2308.64 ± 564.55
Fatigue	8355.89 ± 4191.20	509.89 ± 179.17	58.46 ± 40.89	2348.99 ± 463.35

Bolded number indicates significant differences at α=0.05

Foot Architecture Affects 1^{st} Metatarsophalangeal Joint Function

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INTRODUCTION

Foot architecture, characterized by the medial longitudinal arch, has been identified as a key contributor to function of the 1^{st} metatarsophalangeal (MTP) joint. In particular, low arch foot structure has been postulated to contribute to decreased 1^{st} MTP joint motion. [1] However, limited objective data exist elucidating 1^{st} MTP joint function across a range of medial longitudinal arch heights. The purpose of this study was to examine 1^{st} MTP joint motion and flexibility in individuals with high, normal and low arch foot structure.

METHODS

All procedures were approved by the Institutional Review Board at Hospital for Special Surgery. Asymptomatic individuals (n=61), with high, normal and low arches, based on relaxed calcaneal stance position (RCSP,°) and forefoot to rearfoot position (FF-RF,°) [2], participated in this study. Additional measures of foot architecture, including the malleolar valgus index (MVI), arch height (AH) and arch height index (AHI) were measured in bilateral stance.

1^{st} MTP joint dorsiflexion and stiffness were measured using a specially constructed jig. This device involves the application of a moment about the sagittal axis of the first MTP joint, with the subject in bilateral stance. The resulting angular excursion was measured by a potentiometer, while a torque transducer measured the applied moment. The slope of the angle versus moment curves in the first 25% of the joint's range of motion was termed early flexibility (°/N.cm), and during the last 25%, late flexibility (°/N.cm). Maximum dorsiflexion attained was recorded as peak dorsiflexion (DF,°).

A one-way ANOVA with Bonferroni adjusted post-hoc comparisons was used to assess between-group differences in arch structure, 1^{st} MTP joint flexibility and peak DF. Significant post hoc differences were denoted by * (Normal vs. Low), # (Normal vs. High), and + (Low vs. High).

RESULTS

Between-group differences were found in RCSP, FF-RF, and MVI. 1^{st} MTP late flexibility was significantly higher in individuals with low arch compared to normal arch structure. (Table 1) RCSP was linearly related to FF-RF (r=-0.79, p<0.01) and MVI (r=-0.58, p<0.01). 1^{st} MTP joint Late Flexibility was linearly related to RCSP (r=-0.29, p=0.01) and AHI (r=-0.25, p=0.03).

	Low Arch	Normal Arch	High Arch
	n=22	*n=27*	*n=12*
RCSP	- 6 $(2)^{*+}$	-1 $(1)^{*}$	0 $(1)^{+}$
FF-RF	7 $(5)^{+,*}$	2 $(1)^{*,\#}$	-2 $(1)^{+,\#}$
MVI	$13.7(5.1)^{+}$	$7.5(4.0)^{*}$	6.3 $(3.3)^{+}$
AH	6.4 (1.5)	6.4 (0.6)	6.9 (0.5)
AHI	0.34 (0.084)	0.36 (0.047)	0.38 (0.031)
Peak DF	73.5 (12.4)	73.1 (10.2)	77.3 (9.1)
Early Flexibility	17.4 (13.7)	23.1 (14.5)	17.7 (10.7)
Late Flexibility	27.8 (22.4)*	16.1 (8.6)*	19.0 (8.5)

Table 1. Mean (SD) summary of arch structure and 1^{st} MTP joint function. –ve values indicate valgus.

DISCUSSION

The findings of this study provide objective evidence quantifying differences in 1^{st} MTP joint function in asymptomatic individuals with varying arch height. Contrary to our hypothesis, individuals with low arches demonstrated increased 1^{st} MTP joint late flexibility compared to individuals with normal arch structure. Additional studies are indicated to examine effects of foot structure and 1^{st} MTP joint function on plantar loading in the presence of pathology.

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Static Load Effect on Sagittal deformation of the Foot

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INTRODUCTION

Longitudinal foot arch characteristics and mobility are used extensively to evaluate and assess predisposition to lower extremity injuries [1,2,3]. The change in these arch measures (deformation), under various foot static load conditions, is considered clinically efficacious procedure for arch mobility assessment. However, relationship, amongst these commonly used measures, and their systematic change with load has not yet been demonstrated. The radiographic changes of the medial / lateral longitudinal arch characteristics of the foot were evaluated under 3 percent body weight (%BW) load conditions replicating clinically feasible methodology. The association amongst arch characteristic and mobility measures was also quantified.

METHODS

Forty adults without foot deformities and lower extremity pathologies participated in the study. The inclination of the calcaneus (CI), metatarsal one (M1I) and metatarsal five (M5I); the medial (MLA), lateral (LLA) longitudinal arch angles; and the navicular height (NH), truncated foot length (TFL), and arch height index (AHI) parameters were digitally evaluated from radiographic images using MATLAB. All measurements were taken unilaterally (right foot only) under 3 randomized load conditions. In all three conditions subjects were instructed to maintain a relaxed upright stance position with the right tibia vertical (monitored by 2 orthogonal inclinometers, anchored on the tibia) while supporting their BW: (1) on the left leg, (2) bilaterally, and (3) on the right leg. These 3 load conditions represented <20%, 50%, and >80% of body weight (BW) on the right foot. While in unilateral stance, subjects were instructed to lightly support themselves using toe touching with the contralateral foot for balance purposes. One-way ANOVA repeated measures design and stepwise regression analysis was used for data analysis (SPSS, 17).

RESULTS & DISCUSSION

The changes in the foot parameters studied with load are presented in Table 1. Most of the measured parameters reflect statistically significant progressive foot deformation with load, other than M1I, M5I and LLA, which characterize the forefoot and the lateral foot structures, with M5I showing an increasing trend with load. Significant high to moderate associations were found as expected amongst the structural parameters of the foot, and were maintained or strengthened under the higher load conditions. Interestingly when the mobility parameters were assessed only: MLA and M1I, LLA and M5I, and AHI and NH were highly associated to each other, while LLA and CI showed a moderate association, indicating that changes in the foot structure with load is predominantly localized in the forefoot and the lateral foot structures.

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Table 1: Change in foot parameter values between static load conditions

	Percent (%) Static Load					One-way
Parameter	**<20%**		**50%**		**>80%**	**p value**
CI (º)	26.12 ± 3.92	**	24.49 ± 3.48	**	23.83 ± 3.29	.000
M1I (º)	19.64 ± 3.35	*	18.85 ± 2.74	NS	18.99 ± 2.70	NS
M5I (º)	6.50 ± 2.52	NS	6.99 ± 2.58	**	7.63 ± 2.76	.001
MLA (º)	134.23 ± 5.34	**	136.66 ± 5.18	**	137.18 ± 4.78	.000
LLA (º)	147.37 ± 5.12	**	148.52 ± 4.61	NS	148.53 ± 4.58	.014
NH (% change)		**	8.79 ± 7.07	**	3.07 ± 5.61	.000
TFL (% change)		**	-0.83 ± 0.98	**	-0.22 ± 0.86	.000
AHI (ratio)	0.168 ± 0.026	**	0.152 ± 0.026	**	0.146 ± 0.024	.000

* change, @ p<.05; ** change, @ p<.001; NS = Not significant; (-) % change indicates elongation

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The Arch Index as a Potential Measure of Dynamic Foot Function in Children

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INTRODUCTION

The term 'flexible flatfoot' refers to a foot which has a normal arch during non-weight-bearing and a flattening of the arch during stance [1]. Treatment of feet which are clearly flat because they are deformed or injured is not controversial, but there is uncertainty about whether a flexible flatfoot showing no clinical signs should be treated. In order to make an informed decision about this, dynamic foot function in children must be well quantified. The Arch Index (AI) of Cavanagh and Rogers [2] is one method used to assess foot arch structure. Normally it is evaluated during static stance or at the point of mid-stance in gait, but knowing exactly how the AI changes with time during stance could be just as important. The aim of this study was to investigate the variation in AI during walking and to see if this measure has the potential to identify subtle functional differences between children with neutral feet and those with flatfoot.

METHODS

Two children (age 12.0 and 13.4 yrs) participated in this preliminary study. One child was classified as having a neutral foot posture bilaterally (foot posture index [3], FPI = 0), whilst the other had flexible flatfoot bilaterally (FPI = 7). Spherical reflective markers were located at known anatomical landmarks, including the foot [4]. Plantar pressure data for quiet standing and for level walking were collected at 50 Hz using a Novel Emed-m pressure plate (Novel Gmbh, Munich, Germany). Kinematic data were collected at 100 Hz using a 12-camera Vicon MX system (Vicon Motion Systems, Oxford, UK) and processed using Vicon Nexus software. Pressure maps and marker trajectories were exported to Matlab (The MathWorks, Inc., Natick, MA, USA), in which calculations to find the AI versus time were implemented.

RESULTS

As expected, the flat foot had a larger static AI than the neutral foot (Fig. 1). During walking, however, there was less difference in AI values between the two subjects, although the flatfoot had AI values which were highest at the beginning of stance and decreased thereafter (Fig. 1, bottom). The AI patterns for the neutral foot were much less consistent (Fig. 1, top).

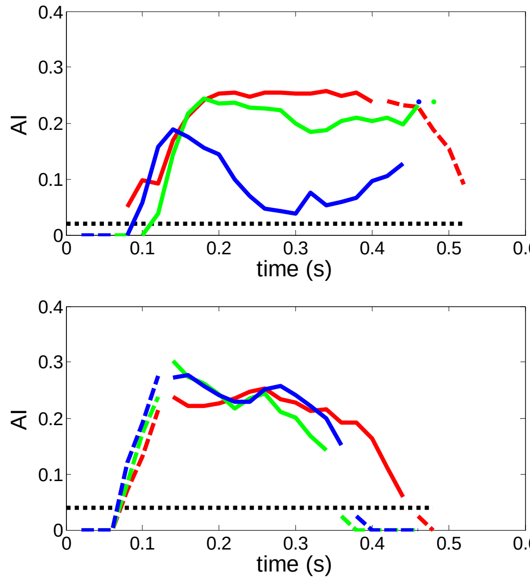

Figure 1: AI versus time (stance phase). Dotted line, static. Walking (3 trials): solid line, mid stance with whole foot in floor contact; dashed line, initial and terminal stance. Top, neutral foot. Bottom, flatfoot.

DISCUSSION

This study shows that it is possible to calculate changes in AI over the stance phase of gait and that there may be subtle differences in the AI patterns between neutral and flat feet. More subjects are currently being tested to see if the AI patterns for each group are consistent.

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Impact of Chronic Gout on Foot Function: Case-Control Study

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INTRODUCTION

Gout is a common form of arthritis caused by the inflammatory response to monosodium urate crystals within the joint [1]. The disease initially presents as self-limiting attacks of severe joint inflammation, and in the presence of persistent hyperuricaemia, chronic tophaceous disease may also develop. Tophi typically occur in both subcutaneous tissues and within affected joints, and may cause pain, mechanical obstruction of joint movement, disability and joint destruction [2]. Evaluating the function of the foot in patients with chronic gout may be helpful in understanding the pathways leading from underlying disease processes, to localised impairment and subsequently to loss of function. The aim of this study was to describe the functional and biomechanical characteristics of foot disease in chronic gout.

METHODS

Cases with gout (n=25) were recruited from rheumatology outpatient clinics. All cases had a history of acute gout according to ACR diagnostic criteria (median disease duration 21 years, flare frequency 2.92/year, 44% with tophi).

Cases were excluded if they were experiencing an acute gout flare at the time of assessment, lower limb amputation or diabetes mellitus. Age, sex and BMI-matched control participants (n=25) without arthritis, lower limb amputation or diabetes mellitus were also analysed. Plantar pressures were recorded using an in-shoe system to determine peak pressure and pressure-time integrals under ten regions of the foot.

An instrumented walkway was used to capture spatial and temporal gait parameters. Disease impact was measured using the Leeds Foot Impact Scale. To preserve data independence, data from the right foot of each participant were analysed.

RESULTS

Significant differences in all foot measures were observed between cases and controls. In particular, significant differences were present in the pressure-time integrals across all foot regions except under the hallux and lesser toes, with higher pressures over time in the gout group. Gait parameters that included walking speed, cadence and double-support were also impaired in cases with gout. Patient reported scores of disease impact were significantly higher in the cases.

DISCUSSION

Chronic gout is associated with important changes in load-bearing function across the entire foot, which may contribute to the development of pain and disability in this disease.

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Using Plantar Pressure Regression Factor Scores to Discriminate Foot Posture

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Introduction

Abnormal foot loading places the lower extremities at risk for injuries. Recently a plantar pressure regression factor (RF) score was developed to objectively discriminate the altered foot loading associated with Charcot neuropathy[1]. This score represents the similarity of the actual pressure distribution with a normal distribution and is independent of gait speed. The purpose of this investigation was to see if RF scores can differentiate abnormal foot postures.

Methods

Thirty subjects were recruited for this study. A single investigator determined the Foot Posture Index (FPI) for each subject (1.33 ± 1.605) [2]. Plantar pressure for a single step was collected for each subject. Subject's walked a 10m walkway in the middle of which was placed a EMed XR pressure platform. All trials were performed barefoot and walking velocity was assessed via a Physilog data logger and gyroscopes attached to the lower limbs [3]. Using the pressure profile of each full step, RF scores were calculated for each subject.

Results

Results suggest that RF score was independent of gait speed (r=-0.3, p>0.05) and BMI (r=0.06, p>0.05, see figure 1). Interestingly, RF score has a significant correlation with square value of PFI (r=-0.45, p=0.01, 95%CI=[-0.7,-0.1], see figure 2). This correlation is negative suggesting that when FPI is close to zero, the RF score is higher than when the FPI is either highly negative or highly positive.

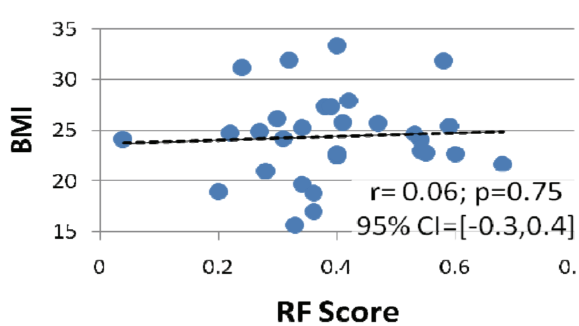

Figure 1: Correlation between RF score and Body Mass Index (BMI).

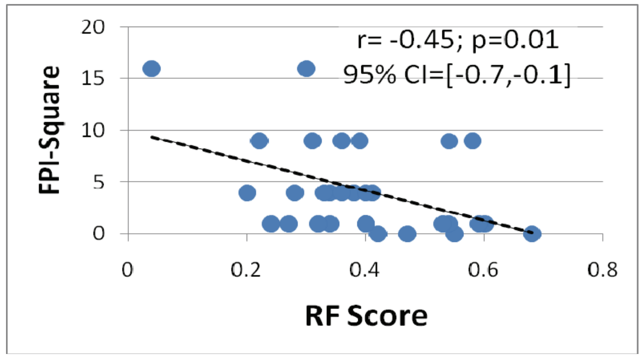

Figure 2: Correlation between RF score and square value of Foot Posture Index (FPI)

Discussion

These initial results indicate the ability of RF scores to differentiate the loading associated with different foot postures. As the posture deviated from neutral towards either pronation or supination, the RF scores declined. The key advantage of this score is that contrary to peak of plantar pressure, it is independent of gait speed and BMI. Additional work should seek to determine whether RF scores may be used to evaluate foot posture intervention strategies such as motion control shoes.

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Comparison of In-shoe Plantar Loading During Walking, Ascending and Descending Stairs

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INTRODUCTION

Studies indicate that non-gait activities of daily living (ADL) affect regional plantar pressure, however, the magnitude and location of changes in plantar pressure accompanying non-gait activities are contentious.[1, 2] The purpose of this study is to assess regional peak plantar pressure and pressure-time integral during walking, ascending and descending stairs. The objective findings from asymptomatic subjects in this study may help provide valuable baseline data, and could potentially be used for designing rehabilitation programs and footwear.

METHODS

7 active adults (4 males: 27.5±5.07years, BMI: 23.5±1.54kg/m^2; 3 females: 26.67±5.07years, BMI: 23.5±1.54kg/m^2) were recruited. They performed walking (W), ascending stairs (A), and descending stairs (D), at their self-selected speeds. In-shoe plantar pressures were acquired with the Pedar-X system (Novel, Munich, Germany) at 50 Hz. A minimum of 10 steps per activity were analyzed with NovelWin software to obtain Peak pressure (PP, kPa) and pressure-time integral (PTI, kPa-s) in the following six masks: heel, midfoot, medial forefoot, central forefoot, lateral forefoot, and hallux.

A two way repeated measures ANOVA was used to assess the effect of activity and mask on PP and PTI (SPSS v16, Chicago, IL). If interaction (activity x region) was significant ($\alpha < 0.05$), simple effects of activity on each mask were assessed. Post-hoc testing was performed using Bonferroni adjusted comparisons.

RESULTS

A significant activity-by-foot-region interaction was found for PP (P = 0.05) and PTI (P = 0.01). Subsequently, significant simple effects were noted at the heel (P = 0.007), midfoot (P = 0.034), medial forefoot (P = 0.032), central forefoot (P < 0.001), and lateral forefoot (P = 0.008) and not at the hallux (p=0.166). For PTI, significant simple effects were noted at the midfoot (P = 0.020), medial forefoot (P = 0.002), central forefoot (P = 0.009), and hallux (P = 0.003), but not at the heel (P=0.101) or lateral forefoot (P=0.066). The results of pair-wise comparisons of activity on PP and PTI at each foot area are summarized in table 1.

DISCUSSION

Our results indicate that PP sustained during stair ascent and descent is significantly lower than that sustained during walking. The reduction was most dramatic at the heel (50% decrease) and forefoot (20% decrease). In contrast, PTI sustained during stair ascent and descent was higher (20% increase) than that sustained walking. Taken together, these findings underscore changes in plantar load distribution that occur with stair climbing.

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ACKNOWLEDGEMENTS

This study was partly supported by the New York Physical Therapy Association.

Table 1: Summary of PP and PTI in six masks. * indicates significantly different from walking.

	Peak Pressure (kPa)			Pressure time Integral (kPa/cm^2.s)		
	Walking	Ascent	Descent	Walking	Ascent	Descent
Heel	208.00±32.74	101.97±9.99*	104.49±7.73*	52.43±19.54	54.89±16.58	40.19±14.40
Mid	94.96±7.67	79.49±4.91*	89.01±6.12	31.11±8.77*	45.13±13.28	41.48±10.79
M.Fore	171.29±12.25	132.53±15.31*	152.85±21.35	42.26±8.44	52.18±17.30	65.62±20.40*
C.Fore	176.75±13.23	128.73±12.23	121.72±11.56*	46.87±6.02	52.65±11.83	59.00±12.10*
L.Fore	137.71±8.60	108.21±11.68	101.34±7.51*	43.59±7.8	49.01±9.84	51.55±10.22
Hallux	197.25±23.01	154.38±22.33	175.63±24.41	42.67±18.47	51.57±17.03	73.02±17.51*

Reliability Characteristics of Tibialis Posterior EMG

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INTRODUCTION

Tibialis posterior (TP) tendinopathy is a common and disabling condition and is cited as the most common cause of adult acquired flatfoot [1]. Preliminary evidence in a variety of flatfoot cohorts shows increased TP activity in an attempt to support the collapsing MLA [2-3]. The aim of this study was to establish the reliability characteristics of TP EMG, barefoot and shod, to inform an intervention study.

METHODS

Six healthy adult subjects undertook gait analysis, barefoot and shod, on two occasions seven days apart. 3-D motion analysis was used to define key events of the gait cycle. Intramuscular bi-polar stainless steel fine wire electrodes (Motion Lab Systems Inc) were inserted under ultrasound guidance. Data were normalised to stride and maximum voluntary contractions (MVC). Reliability was analysed using the coefficient of multiple correlation (CMC), intra-class correlation coefficients (ICC) and the standard error of measurement (SEM).

RESULTS

TP activation displayed two clear bursts of activity during gait (Figure 1). Excellent within day reliability was found for TP EMG with the CMCs ranging from 0.828 to 0.854. Between day reliability was moderate to good with CMCs of 0.651 for shod and 0.709 for barefoot trials (Table 1). Barefoot trials showed greater reliability than shod trials for discrete variables with ICCs of 0.503 to 0.806 for barefoot and -0.121 to 0.860 (Table 2) for shod trials. The standard error of measurement for these variables ranged from 0.65% to 13.51% of MVC for barefoot trials and 1.20% to 20.13% of MVC for shod trials.

Table 1: Coefficient of multiple correlation BF: barefoot, SH: shod, (st deviation).

Variable	CMC within day Day 1	CMC within day Day 2	CMC between day
TP EMG BF	0.828 (0.132)	0.838 (0.120)	0.709 (0.136)
TP EMG SH	0.854 (0.073)	0.835 (0.139)	0.651 (0.059)

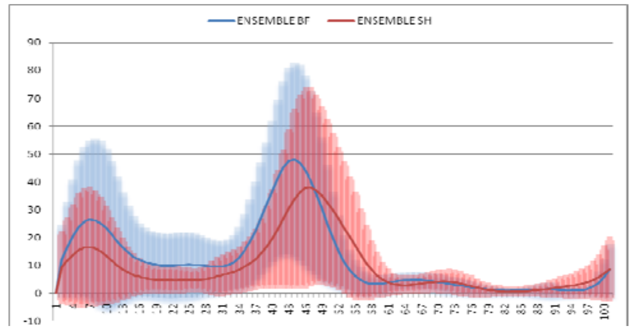

Figure 1: Ensemble average of TP activation barefoot (blue) and shod (red) for 6 subjects normalised to stride and MVCs.

DISCUSSION

In the barefoot and shod groups of this cohort both the shape of the curve and the temporal variables were more reliable than the magnitude characteristics of the signal; consistent with published literature [4]. The findings of this study suggest EMG variables must be selected with care as an outcome for intervention studies. Possible limitations of these findings are; the lack of acclimatisation time for the footwear, MVC normalisation method and the small sample size.

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ACKNOWLEDGEMENTS

RS and DT are funded by Arthritis Research UK

Table 2: Intra class correlation coefficient and standard error of measurement (%MVC) for discrete variables (MS:midstance, ***: ICC failed)

Variable	ICC Barefoot (95% CI)	ICC Shod (95% CI)	SEM Barefoot (95% CI)	SEM Shod (95% CI)
Peak EMG contact phase	0.602 (-2.6 to 0.9)	0.121 (79.4 to 0.8)	13.51 (-14.9 to 26.5)	7.558 (-19.4 to 19.4)
Time to peak contact phase	0.604 (-6.3 to 0.7)	0.860 (-0.2 to 0.9)	0.654 (-2.5 to 0.8)	1.536 (-4.7 to 3.1)
Peak EMG MS/ propulsion	0.806 (-0.4 to 0.9)	***	12.123 (-23.6 to 38.6)	20.183 (-50.7 to 53.0)
Time to peak MS/propulsion	0.503 (-2.8 to 0.9)	0.729 (0.4 to 0.9)	2.489 (4.4 to 8.4)	1.204 (-1.5 to 4.5)

Non-rigid work in human walking: are hard collisions in fact soft?

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INTRODUCTION

Muscles and tendons perform work about the lower-limb joints to power gait, but significant work may also be done by non-rigid deformations elsewhere in the body. Steady level-ground walking requires zero net mechanical work per stride, yet rigid-body estimates yield less negative joint work than positive [1]. This suggests that joint work measures fail to capture all the work performed by the body. We propose that work is done by soft tissue deformations, which could influence walking economy or risk of tissue injury. We hypothesize that these non-rigid bodies perform significant negative work during the collision of the leg with the ground after heelstrike. We investigated soft tissue work in human walking, specifically the role of the ankle-foot-shoe.

METHODS

We use two measures to indicate work not captured by standard inverse dynamics. The first, inter-segmental work, estimates work done by the ankle, foot and shoe on the rest of the body. This includes non-rigid deformations (e.g., compression of shoe or heel pad) in addition to rotational ankle work. The second measure, center-of-mass (COM) work, quantifies work done by the entire leg on the body, regardless of whether or not it is from joint rotations. We propose that the differences between these two measures and conventional, rigid-body joint work can serve as indicators of non-rigid deformations.

We measured healthy subjects walking on an instrumented treadmill at various speeds (N=10, 0.7-2.0 m/s). Rigid-body joint powers for the ankle, knee and hip were calculated from conventional inverse dynamics using Visual3D software. We computed COM work rate for each limb, the 3D dot product of each limb's ground reaction force with COM velocity. Inter-segmental work rate from ankle-foot-shoe is defined as summed translational and rotational work performed on the distal shank [2]. Work values were integrated from powers during each phase of gait – Collision, Rebound, Preload, Push-off – as defined by alternating regions of positive/negative COM work (Fig. 1).

RESULTS & DISCUSSION

We found strong evidence of non-rigid work performed during the Collision phase of gait (Fig. 1). At 1.25 m/s, summed ankle-knee-hip work failed to capture about 7.5 J, or 60% of the negative COM Collision work [3]. And rigid-body ankle joint work failed to account for about 2 J of negative inter-segmental work, so ankle-foot-shoe deformation may perform about 15% of the total negative Collision work. By both measures, non-rigid Collision work increased with speed. The overall results suggest that soft deformations may in fact dominate walking collisions.

We also observed evidence of a non-rigid elastic response after Collision. Based on differences between COM and summed joint work, about 4 J of the total non-rigid Collision work may be stored and then returned elastically during the Rebound phase. On average, less than half of the ankle-foot-shoe work (<1 J) contributed to elastic energy return. In summary, we believe non-rigid deformations play a significant energetic role in both positive and negative work during walking.

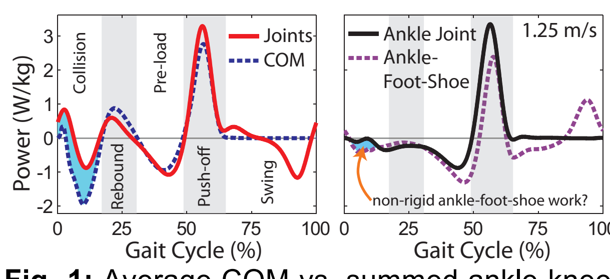

Fig. 1: Average COM vs. summed ankle-knee-hip joint power (left), inter-segmental work rate from ankle-foot-shoe vs. rigid-body ankle joint power (right). Fluctuating regions of positive and negative COM work define phases of gait.

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Lower Extremity Movement Strategies in Individuals with Achilles Tendinopathy

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INTRODUCTION

The tendinopathic Achilles tendon exhibits lower stiffness. [1] This alteration in tissue material property may impact physical performance. The aim of this study is to investigate the lower extremity movement strategies employed by individuals with and without Achilles tendinopathy during novel unipedal ballistic activity.

METHODS

Twelve male subjects (five with Achilles tendinopathy) participated (aged 45.5±8.2y/o). Lower extremity kinematics (Vicon 612 motion analysis system, Oxford, UK; 120Hz), and kinetics (AMTI force plate, Watertown, MA; 1,560Hz) were obtained during single-legged hopping at 2.2Hz. Sagittal plane Net Joint Moments (NJM) and Net Joint Moment Powers (NJP) were obtained using a standard inverse dynamics approach. Support moment and sum of powers were calculated as the sum of NJM and NJP for hip, knee and ankle joints, respectively. The eccentric phase was defined by all three joints exhibiting a negative NJP and the concentric phase was defined by positive NJP. The period of time, when the NJP was not unidirectional in all joints, was not included in these analyses.

RESULTS

Support moment and sum of power during concentric and eccentric phases were not different between groups. (Table 1) The tendinopathic ankle contributed less and the corresponding hip contributed more to the support moment during both phases. The same analysis of NJP showed similar, though less prominent contribution shifts. Tendinopathic subjects used less [(HC-AT)/HC] sagittal ankle angular displacement (4.8%Ecc, 9.6%Conc), and exhibited lower averaged angular velocity (22.8%Conc), averaged NJM (13.7%Ecc, 12.3%Conc) and averaged NJP (10.1%Ecc, 12.5%Conc).

DISCUSSION

The well known energy storage-release mechanism at the calf might be altered in the presence of a much more compliant Achilles tendon. Lower joint powers, observed in tendinopathic subjects, suggests that the musculotendinous unit is less capable of contributing to power output at a higher rate. Furthermore, because joint power is the product of joint moment and angular velocity; in this cohort both lower joint moment and lower angular velocity contributed to a lower joint power. The tendinopathic ankle joint moved slower with a smaller angular displacement. This might be attributed to either the subjects' inability to move rapidly or that the pathology altered the motor control strategy limiting the angular displacement. Further studies will address (1) validation of these findings by increasing the sample size; and (2) examination of the muscle activation pattern in order to better understand the motor control strategy in the presence of Achilles tendinopathy.

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Table 1. Averaged NJM and NJP during concentric and eccentric phases of single-legged hopping. Joint contributions to sum of NJM and NJP are in parentheses.

		Healthy Control		Achilles tendinopathy	
		Concentric (% of Sum)	Eccentric (% of Sum)	Concentric (% of Sum)	Eccentric (% of Sum)
Power (W/kg)					
	Ankle	4.73±1.56 (55.6)	-4.64±1.73 (46.4)	4.14±1.56 (47.6)	-4.17±1.97 (44.4)
	Knee	2.71±1.14 (36.5)	-3.19±1.3 (38.4)	3.44±0.95 (40.8)	-3.95±1.06 (37.1)
	Hip	0.67±0.37 (8.3)	-1.64±0.6 (15.4)	0.99±0.45 (12.0)	-2.14±0.69 (19.0)
	Sum of Power	8.95±1.79	-9.97±1.37	8.55±2.6	-10.26±3.06
Moment (Nm/kg)					
	Ankle	-2.12±0.46 (60.4)	-2.02±0.57 (53.2)	-1.86±0.36 (51.9)	-1.74±0.48 (46.6)
	Knee	-1.25±0.09 (29.2)	-1.31±0.28 (31.1)	-1.4±0.28 (32.2)	-1.36±0.32 (31.7)
	Hip	-0.4±0.27 (10.4)	-0.59±0.11 (15.8)	-0.6±0.24 (15.9)	-0.81±0.11 (21.7)
	Support Moment	-3.76±0.61	-3.91±0.61	-3.66±0.4	-4.12±1.33

GEOMETRIC FOREFOOT MODEL

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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.

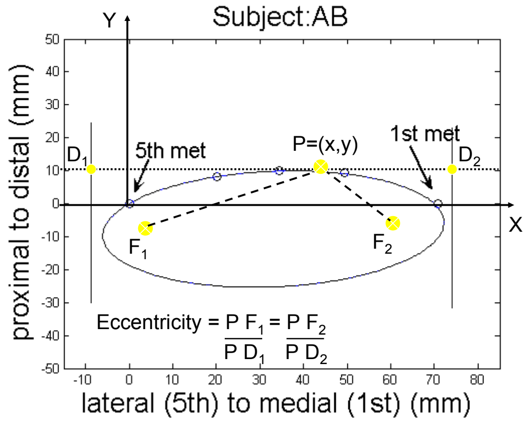

Figure 1. 2D Forefoot Model. The metatarsal head points were aligned such that the 5[th] was on the origin (0,0) and the 1[st] along the x-axis. Each conic (blue curve) is defined uniquely by the equation: $Ax^2+Bxy+Cy^2+Dx+Ey$, where A=1.

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. Type and side were modeled as fixed and random effects.

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 (Table1). It should be noted that Malleolar valgus Index (MVI) and Center of Pressure Excursion Index[2] (CPEI) did distinguish foot types.

DISCUSSION

The D parameter represents the linear distance from the 1[st] to the 5[th] metatarsal joint, which we would expect to be different given that all the diabetic subjects had hallux valgus. This suggests that simple models may be useful to differentiate normal vs. pathological feet.

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ACKNOWLEDGEMENTS

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Table 1	Mean (SD)				p-value	($p \le 0.0083$)
	Planus	**Rectus**	**Cavus**	**Diabetic**	**ANOVA**	**Post-Hoc***
MVI (%)	13.3(4.2)	7.4(3.6)	6.7(5.2)	15.3(6.2)	0.000	1,2,4,6
AvgCPEI (%)	18.4(4.7)	22.0(8.4)	23.7(6.2)	16.0(4.2)	0.000	1,2,4,6
B	1.2(10.4)	-0.03(8.4)	5.3(18.4)	0.9(13.0)	0.412	
C	-2.4(519.6)	-7.7(296.4)	-100.0(158)	5.7(223.6)	0.292	
D	-75.0(6.2)	-70.7(5.1)	-74.4(5.5)	-78.7(8.0)	0.001	4
E	64.2(1380.1)	202.0(1098.5)	887.5(1217.3)	67.0(1721.8)	0.262	
Eccentricity	2.0(2.3)	2.1(2.0)	2.7(2.1)	1.6(2.4)	0.691	

1=Cavus vs Planus; 2=Rectus vs Planus ; 3=Cavus vs Rectus ; 4=Diabetic vs Rectus ; 5=Diabetic vs Planus; 6=Diabetic vs Cavus

Intra-tester reliability of a hand-held device to measure heel pad stiffness

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INTRODUCTION

Subjective palpation remains the most common modality of assessing the heel pad within the clinical setting. Quantifying the heel pad has been limited largely to the laboratory setting. Determining the reliability of evaluating the heel pad mechanical properties from one occasion to the next have been reported in the literature but limited to short time frames [1,2]. Evaluating reliability that mirrors clinical practice has yet to be determined. Therefore, the aim of the current study was to investigate the within and between intra-tester reliability of a hand-held indenter to measure heel pad stiffness.

METHODS

Thirty asymptomatic subjects, mean age of 30.5 (SD: 12.1) years old with no history of heel pain or previous foot surgery, were recruited from AUT University. A heel pad indenter was used to measure the heel pad stiffness (N/mm^{-1}). A 30mm cylindrical probe was attached to a load cell, which passed through an aluminum front plate. Displacement of the plate was measured by a linear voltage displacement transducer. A specifically designed software package generated force-displacement curves that calculated the full and final 10% stiffness. Subjects were tested on two separate sessions, a mean of 32.2 (SD: 12.1) days. The procedure was standardised for subject position and indenter placement. Data was analysed using ICC's (95%CI) and SEM.

RESULTS

Table 1 demonstrates the intra-tester within and between session reliability.

DISCUSSION

Reasons for the poor intra-tester reliability maybe linked to the speed and force applied to the soft tissue by the operator. The development of a step motor may help to control these variables so that a clinical device could be used in everyday practice.

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Table 1 demonstrates the intra-tester within and between session reliability.

Curve Parameters	Within-Session ICC [95% CI]	Within-Session SEM	Between-Session ICC [95% CI]	Between-Session SEM
Full curve (N/mm^{-1})	0.62 [0.2-0.77]	12.6	0.24 [-1.6-0.4]	7.1
Final 10% (N/mm^{-1})	0.67 [0.4-0.8]	15.8	0.17 [-0.7-0.6]	11.7

Characteristics of Participants with Recurrent Sprains

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INTRODUCTION

The most common ankle injury is sprain with approximately half of these injuries making a full recovery [1]. Common residual problems include recurrent injury, giving way and decreased functional abilities. These problems are generally described by the umbrella term of chronic ankle instability (CAI). It has been proposed that CAI consists of three major groups: recurrent sprain, mechanical instability and functional instability [2]. This review explored the characteristics of participants with recurrent sprains.

METHODS

This was a systematic review of the literature using the AMED, CINAHL, Medline, PEDro, Pubmed and SportDiscus databases. Included articles had to compare a recurrent sprain group (≥ 2 ankle sprains) and a healthy control group. Outcomes were grouped into physical tests, strength, postural stability, proprioception, response to a perturbation, biomechanics and functional tests.

Search results and data extraction were conducted by two independent reviewers with a third person adjudicating if a consensus was not reached.

RESULTS

A total of 55 papers were found to fit the criterion. Papers were skewed towards the younger male population with 52 of the papers having an average age below 30 and 26 papers using a solely or majority male population.

A wide variety of outcomes were explored in the literature using a broad spectrum of methodologies and measures. As a result the outcome groups were small and papers could not be directly compared and a meta-analysis was not attempted.

Differentiating features between participants with recurrent sprain and healthy controls appear to be: the anatomy of the talus, position of the foot during gait, postural stability during more strenuous challenges (i.e. with the eyes closed or on an unstable platform), and stabilization time after a jump or hop.

There were no differences between groups in more commonly researched and clinically used indicators such as range of motion, strength (measured by peak torque and concentric in/eversion strength) and peroneal latency following perturbation or a jump.

DISCUSSION

Characterising participants with recurrent ankle sprain is still open for debate. This review suggests several directions which need to be pursued and others which could be abandoned. For example, strength measures such as peak torque and concentric in/eversion could be abandoned but power and endurance could be pursued. These measures may more accurately reflect the functional incapacity in this group.

Furthermore, clinically many of these tests are not appropriate requiring laboratory equipment to gain the sensitivity required to demonstrate differences. Characterising and predicting which ankle sprains will have ongoing difficulties remains a problem for the clinician.

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The Effects of Ankle Bracing on Mechanical Ankle Restraint in Individuals With and Without Chronic Ankle Instability

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INTRODUCTION

Ankle braces are implemented commonly to provide ankle stability during physical activity. Mechanical restraint is provided in laboratory settings in healthy ankles [1], but there is limited information to know if braces provide adequate support for those with and without chronic ankle instability (CAI) [2]. The purpose of this study was to quantify ankle mechanical stability with and without an ankle brace between subjects with and without CAI.

METHODS

Thirteen subjects with unilateral CAI (age: 21.50±1.38yrs; height: 166.37±9.19cm, mass: 68.42±10.96kg) and thirteen Control subjects (age: 21.83±2.71yrs; height: 169.34±6.93cm, mass: 63.50±8.84kg) volunteered to participate. CAI subjects had a history of at least 1 significant lateral ankle sprain, at least 2 episodes of the ankle giving way in the previous 3 months, and had scores of <90% on the Foot and Ankle Disability Index (FADI) and <80% on the FADI-Sport Scale.

Ankle mechanical stability was assessed with and without a lace-up style ankle brace using an ankle arthrometer. To assess ankle stability, anterior/posterior (A/P) loading was performed first followed by inversion/eversion (I/E) loading. Stability was measured on the affected ankle of the CAI group. A matched limb was assessed in the Control subjects. Three trials were completed in each direction with and without the ankle brace applied.

The dependant variables were the average total A/P (mm) and total I/E (degrees) excursions.

For each dependant variable, a Condition (Brace, No-Brace) by Group (CAI, Control) repeated measures ANOVA was performed. Statistical significance was set *a priori* at P<.05. Cohen's *d* values are reported to represent effect sizes.

RESULTS

For A/P stability, a significant main effect for Brace existed ($F_{1,24}$=26.623; p<.001). The brace limited A/P motion (11.74±5.37) compared to bare foot (14.97±4.3) (*d*=0.66). For I/E stability, a significant main effect for Brace existed ($F_{1,24}$=291.91; p<.001). The brace limited I/E motion (20.85±6.80) compared to bare foot (46.31±11.47) (*d*=2.70). (Table 1)

DISCUSSION

The ankle brace provided significant restraint to movement, especially with I/E. While only a moderate effect size was observed for A/P restraint, a very large effect size supports the brace's I/E restraint capabilities. There were no significant differences related to the presence of CAI, with both groups experiencing similar reductions in movement with the brace. CAI subjects were categorized with more of the inclusion criteria founded in functional deficits (ie FADI scores). Future research should include subjects with residual mechanical instability to determine what benefits the brace may provide.

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Table 1: Brace by Group Data for A/P and I/E stability measures

		CAI	Healthy	Brace Main Effect
A/P (mm)	No Brace	15.27 ± 5.56 mm	14.67 ± 2.73 mm	p<.001 *d* = 0.66
	Brace	12.78 ± 6.21 mm	10.69 ± 4.38 mm	
I/E (degrees)	No Brace	47.78 ± 10.07 mm	44.83 ± 12.96 mm	p<.001 *d* = 2.70
	Brace	21.15 ± 5.11 mm	20.56 ± 8.40 mm	

Attentional demands of dynamic postural stability control in healthy subjects and patients with functional ankle instability

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INTRODUCTION

Reduction of attentional resources is readily apparent in cases of reduced or conflicting sensory information[1]. The purpose of this study is to compare attentional demands of postural control in athletes with functional ankle instability (FAI) and healthy matched individuals.

METHODS

15 functional unstable ankles college athletes and 15 healthy matched Athletes were selected conveniently. Overall (OSI), Medial-lateral(MLSI), and Anterior-posterior stability index (APSI) were measured using Biodex Balance System (BBS). Subjects experiencing 2 postural task difficulties (stability levels of 5 and 7 of BBS) and 2 cognitiveloadings (no load and counting backward) while assessed postural stability.

RESULTS

2×2×2 (2 groups, 2 postural task difficulties, and 2 cognitive loadings) Mixed ANOVA showed significant group × condition interaction both for OSI and MLSI. Subjects with FAI had significantly more increase in OSI ($P<0.01$) and MLSI ($P=0.02$) when confronting cognitive task than healthy subjects. Besides at stability level 5,they showed a significantly greater degree of OSI($P<0.01$) and MLSI ($P<0.01$) demonstrating poor postural control ability in subjects with FAI.

DISCUSSION

Results show poorer balance ability in more difficult postural conditions following ankle injury Higher reliance of body on proprioceptive inputs to maintain balance with increased instability of support surface may explain the discrimination of postural control deficit under more challenging postural conditions in patients with FAI[2].Rozzi et al demonstrated the similar findings in patients with FAI during standing on a less stable platform (stability level 2)[3]. Regarding dual tasking effects on postural control, maintenance of balance during single leg stance is more attention demanding for subjects with FAI than for healthy individuals. The association of sensorimotor deficit and automaticity in postural control supported by the above results has been previously investigated in the literature[4].Finally,it seems that FAI is associated with increased attentional demands to dynamic postural control. Cognitive loading may be considered in any exercise program for subjects with FAI as an effective strategy to improve balance abilities. However, it needs further investigation.

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Altered Posture-Dependent Hoffmann Reflex Modulation with Chronic Ankle Instability

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INTRODUCTION

Hoffmann (H) reflex modulation between different body positions has been linked to postural stability [1]. It is hypothesized that the altered modulation between increasingly complex postures may be a potential mechanism of postural instability [2], such as that seen with chronic ankle instability (CAI). Our purpose was to assess H-reflex modulation of the peroneals and soleus in two postures (prone and bipedal stance) in subjects with and without CAI.

METHODS

Sixteen subjects with unilateral CAI (10 males, 6 females; age=21±6.9 yrs; height=173.9±7.4 cm; mass=72.6±11.9 kg) and 15 matched controls without any history of ankle sprains (9 males, 6 females; age=19.33±4.3 yrs; height=175.8±9.7 cm; mass=71.3±17.8 kg) participated. The independent variables were group (CAI, control) and limb (involved, uninvolved). Limbs of the controls were side matched to the involved limbs of the CAI subjects. Maximum H-reflexes and motor (M) waves were recorded bilaterally from the soleus and peroneals while subjects lied prone and then stood in quiet bipedal stance. The maximum H-reflexes were normalized to the maximum motor (M) waves to obtain H_{max}:M_{max} ratios for the two postures. The dependent variable was the measure of reflex modulation. It was quantified in a percent change score with the formula below;

$$\frac{\text{Prone} - \text{Standing Hmax: Mmax Ratios}}{\text{Prone Hmax: Mmax Ratio}} * 100$$

Two-way ANOVAs with repeated measures on limb were performed to compare H-reflex modulation between groups and limbs for both muscles. Tukey's HSD tests were conducted for post-hoc comparisons. The alpha level was set at <0.05.

RESULTS

Means and SD for the measure of reflex modulation for soleus and peroneals are shown in the Table 1. There was significant group by limb interaction for reflex modulation of the soleus (P=.002). In the CAI group the H-reflex modulation in the involved limb was significantly lower than the contralateral uninvolved limb and the both limbs in the control group. There were no significant side-to-side differences in the control group for the soleus. For the peroneals, neither the group by limb interaction (P=.184) nor the group main effect (P=.656) were statistically significant.

DISCUSSION

Decreased H-reflex modulation in the soleus, as assessed by the percent change scores in H_{max}:M_{max} ratios between lying and bipedal standing, was present in the CAI involved limbs compared to the CAI uninvolved limbs and both limbs of the control group. Similar results were not found in the peroneals. The reduced ability of sensorimotor system to down regulate H-reflex in more demanding postures suggests altered postural mechanisms related to CAI. This may be a potential mechanism of postural control deficits associated with CAI.

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TABLE 1. Means (±SD) for reflex modulation during a posture transition between prone and bipedal stance (percent change scores)

Soleus

Group \ Limb	Involved	Uninvolved
C A I	14.06(±9.1)*§	26.44(±11.8)
Control	26.37(±14.2)	27.59(±16.6)

Peroneals

C A I	14.27(±11.2)	20.58(±12.8)
Control	18.50(±7.8)	19.20(±9.7)

*significant difference between limbs in CAI subjects
§significant difference of involved limbs between groups

Leg Muscle Activation and Foot Pressure Related To Functional Ankle Instability

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INTRODUCTION

Chronic ankle instability could affect more than 40% of individuals who suffer a lateral ankle sprain and is often associated with frequent ankle inversion injury [1]. Some evidence suggests that chronic ankle instability patients may have a laterally deviated foot center of pressure (COP) during the stance phase of functional movement [2]. Evertor dysfunction is well documented in patients with ankle instability. However, no direct evidence supports the idea that invertor excitation could couple with evertor inhibition, and result in a laterally deviated COP and increased ankle injury susceptibility. The purpose of this study was to examine evertor/invertor activation and foot COP during a demanding functional movement. We hypothesized that functional anskle instability (FAI) subjects would exhibit invertor/evertor imbalances and laterally deviated COP trajectories during the stance phase of a forward lunge.

METHODS

12 FAI subjects (5 males, 7 females; age = 23 ± 4 yr; height = 1.74 ± 0.14 m; mass = 71.6 ± 17.6 kg) and 12 controls (5 males, 7 females; age = 23 ± 4 yr; height = 1.76 ± 0.17 m, mass = 71.4 ± 16.3 kg) participated in this study. Tibialis anterior (TA) and peroneus longus (PL) EMG were recorded during a forward lunge. COP trajectory was measured with pressure insoles. EMG and COP of the involved leg were recorded during three trials. Functional ANOVA determined differences between FAI and control subjects for normalized EMG and medial-lateral COP position during the lunge (α = 0.05). The functional ANOVA compared EMG and COP across the entire lunge task, rather than only at discrete points.

RESULTS

The analysis revealed no differences between groups during the stance phase of the lunge for PL EMG or COP. The FAI group did exhibit significantly less TA activation at the beginning and end of stance. TA activation was increased for the FAI group during the loading portion of the lunge stance phase (Figure 1).

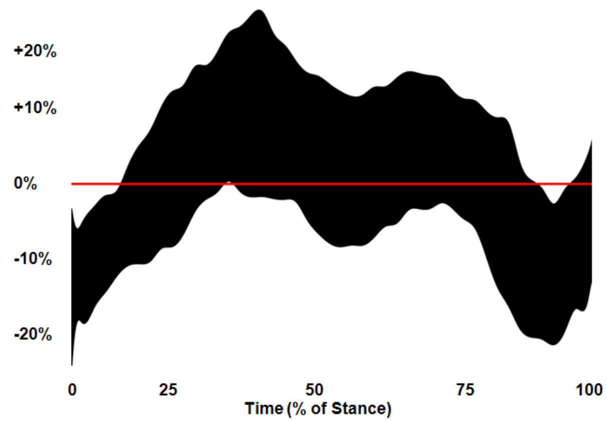

Figure 1: Results from the functional ANOVA for TA EMG. Percent change between the FAI and control groups is indicated on the vertical axis. The shaded area represents the mean and 95% confidence intervals for percent change between the FAI and control groups across stance.

DISCUSSION

These results do not support the idea that FAI subjects exhibit lateral COP relative to matched controls. This is in contrast to lateral COP reported during the stance phase of gait [2]. Also, we did not detect lowered PL activation as was anticipated during this movement considering previously reported findings of peroneal dysfunction [3]. However, we did observe increased TA activation in the FAI group relative to the control group during the loading portion of stance. Increased TA activation during loading without a concurrent increase in PL activation could provide a potentially dangerous strategy. The observed reduction in the FAI group during initial and terminal stance may indicate that patients with FAI poorly control dynamic foot position in high loading environments.

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Effect of Chronic Ankle Instability on Knee Joint Position Sense

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INTRODUCTION

Lateral ankle sprain (LAS) is an extremely common athletic injury. Despite extensive clinical and basic science research, the recurrence rate remains high. Chronic ankle instability (CAI) following LAS is hypothesized to predispose individuals to re-injury because of neuromuscular deficits which result following injury. The likely influence of a localized damage in a distal joint on the function of proximal joints and muscles is an important consideration in assessment and treatment of musculoskeletal problems. However, little experimental evidence in humans exists in this area. The relevance of the knee to the ankle is supported by the close biomechanical relationship between the two joints and the knowledge that knee joint pathologies can affect ankle motor control strategies [1]. However, no investigation has been carried out on the existence of joint position (JPS) deficits in the knee joint of patients with CAI. We performed this study to monitor the knee joint position sense in this type of patients.

METHODS

Ten female patients with CAI and ten healthy control subjects participated in this study. JPS was evaluated by reproduction of the angles in two standing and sitting positions, and in each position two target angles (20° and 60°) were tested. The knee joints in both lower limbs of patients and the dominant knee-limb of healthy subjects were evaluated. The knee angles were measured by using a system comprised of skin markers, digital photography, and Auto CAD software. Absolute error was considered as a dependant variable.

RESULTS

The analysis of median (non-parametric test) showed that in both standing and sitting positions and also two target angles, there were no significant differences between two knee joints (right and left) in patients ($p > 0.05$). There were also no significant differences between knee JPS in affected limb of the patients and matched knee joints of the control group.

Moreover, no significant difference between the knee JPS of dominant leg in healthy subjects and both knee joints of patients was observed (p>0/05).

DISCUSSION

The result of this study suggests that subjects with CAI do not have deficit in knee JPS comparing their injured and intact limb. This finding is in consistent with the studies carried out on patients with ACL reconstruction [2, 3]. They suggested that the afferent information from the intra-articular and peri-articular receptors in one limb could affect the muscle spindles signalling in the contra-lateral limb, therefore no difference can be observed. In addition, our results showed no evidence of impaired JPS in both positions and target angles in subjects with CAI compared to control group. In different research studies of lower limb function and motor-control changes in patients with CAI, controversial results were observed [4, 5, 6]. These different outcomes can be the result of different measuring systems, testing modes, measured variables, control of confounding variables, appropriate matching of the groups and subjects' characteristics.

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Exploring Ankle Ligament Laxity Differences Between Male and Female Collegiate Athletes

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INTRODUCTION
The ankle is the most commonly injured joint in sport [1]. With the rising trend of female participation in sport, there has been an increase of injuries, specifically at the lower extremity [2]. Females have a higher amount of knee laxity when compared to males [3], but there are few studies that have examined gender differences in ankle laxity. The increase in laxity at the knee has been shown to result in more ACL injuries in females, but we do not know if the same holds true in the ankle. The purpose of this study is to determine if differences in ankle ligament laxity exist between male and female collegiate athletes.

METHODS
A total of 65 Division I collegiate athletes were recruited for this study. Ankle ligament laxity measurements were taken from 33 females (age=19±1yrs, height=173.4±9.7cm, mass= 69.0±13.2kg), for a total of 66 female ankle measurements and 32 males (age=19±1yrs, height=183.9± 8.2cm, mass=84.8±10.3kg), for a total of 64 male ankle measurements.

Measurements for inversion and eversion rotation, along with anterior displacement of the ankle were obtained using an instrumented ankle arthrometer (Blue Bay Research, Inc., Milton, FL) with a custom written software program using Labview. Subjects were positioned supine on a table with the foot extended over the edge with a strap placed above the malleoli, securing the lower leg. Then the subject's foot was placed in the arthrometer footplate, secured with a heel and dorsal clamp. The tibial pad was secured onto the lower leg. Once the arthrometer was secured, the ankle was placed in a neutral position for all tests. To measure anterior displacement, an anterior load of 125N was placed on the ankle. For inversion-eversion rotation, a 4Nm torque was placed on the ankle.

Data were analyzed using an independent t-test comparing the arthrometry measurements between genders. Statistical significance was set at $p \leq 0.05$.

RESULTS
There were significant differences in inversion ($p=0.03$) and eversion ($p=0.02$) rotation laxity between male and females athletes (Table 1). In both measurements females had more laxity than their male counterparts. There were no significant differences between groups for the anterior displacement measurements.

DISCUSSION
Though the increase in laxity at the knee has been shown to result in more ACL injuries in females, we do not know if the same trend occurs at the ankle. Although we did find differences in ankle laxity between genders, further investigations should attempt to understand physiological differences in ligament laxity of the lower extremity between genders. Perhaps the greater body height and mass of our male subjects accounted for the lower laxity values compared to the females in this study. In addition, we did not account for the point in the menstrual cycle our female subjects were at during the test session; and acknowledge the fact that there may be hormonal influences on ligament laxity that were not accounted for in this study. Ongoing research can better understand the direct correlation of ankle ligament laxity and ankle injuries.

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Table 1: Ankle ligament laxity values of male and female collegiate athletes.

	Inversion Rotation (°)	Eversion Rotation (°)	Anterior Displacement (mm)
Male	32.9 ± 8.7 *	20.9 ± 6.4 *	8.3 ± 3.2
Female	36.7 ± 11.1 *	23.6 ± 6.7 *	7.7 ± 3.3

* indicates significance at $p \leq 0.05$.

Greater Inversion Laxity Is Associated With More Inverted Rearfoot Positioning During Gait In Subjects With Chronic Ankle Instability

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INTRODUCTION

Individuals with chronic ankle instability (CAI) have been shown to have increased inversion talar tilt laxity[1] and to exhibit more rearfoot inversion motion during gait[2], however these two measures have never been assessed together in the same study. Our purpose was to examine if two groups of CAI subjects, one with high inversion laxity and one with low inversion laxity, exhibited different frontal plane rearfoot kinematics during walking.

METHODS

Eighteen volunteers (7 males, 11 females) with self-reported CAI participated. An ankle arthrometer was used to perform an inversion talar tilt test to 4000 N-mm. Laxity was quantified as the amount of maximal inversion displacement. Subjects were divided into two groups based on their amount of inversion laxity. Subjects at or above the 50th percentile of inversion laxity were placed in the High Laxity Group (3 males, 6 females, inversion laxity=33.0º±7.4º), while those below the 50th percentile were placed in the Low Laxity Group (4 males, 5 females, inversion laxity=20.0º±4.3º).

Subjects then walked (1.32m/s) barefoot on a treadmill with embedded force plates while a 10 camera motion analysis system collected kinematic data of the lower extremities. Five 15-second trials were collected as subjects walked continuously. For each trial, frontal plane ankle kinematic data from all gait cycles within each trial were averaged and resampled to 100 data points, representing 100% of a stride cycle for each limb. Angular displacement data were averaged across trials and 95% confidence intervals (CI) were calculated for each of the 100 data points for each group. Time periods where the CI bands for the groups did not cross each other were identified and the mean differences between groups were then calculated at these intervals.

RESULTS

The High Laxity Group exhibited significantly more inversion positioning than the Low Laxity Group beginning just after midstance, extending through the entire swing phase, until soon after initial contact. (see figure 1) The mean(±SD) differences between groups across this entire interval was 3.8°±0.8°.

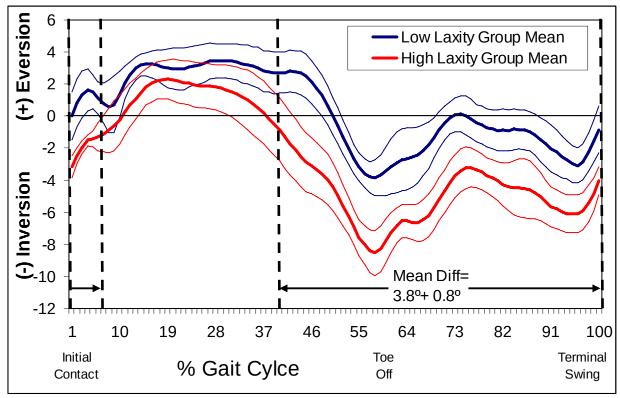

Figure 1. Frontal plane angular displacement (mean±95% CI) for the High and Low Laxity Groups. Vertical dashed lines represent time intervals where the confidence intervals for the two groups do not overlap, thus indicating group differences.

DISCUSSION

These results indicate that the amount of inversion talar tilt laxity is associated with the amount of inversion positioning during gait in subjects with CAI. It has been previously speculated that the greater inversion positioning seen in CAI patients during gait was primarily due to neuromuscular alterations ("functional instability") about the ankle.[2] Our current findings, however, suggest that the amount pathological laxity ("mechanical instability") in the ankle also plays an important role in the greater inversion positioning seen during gait. Clinicians should be cognizant of both mechanical and functional instability deficits when designing interventions for CAI patients.

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Relationships between measures of posterior talar glide and ankle dorsiflexion in healthy individuals

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INTRODUCTION

Arthrokinematics of the talocrural joint during the open kinetic chain (OKC) involve the talus gliding posteriorly on the tibia. This accessory motion allows dorisflexion range of motion (DF ROM) to occur and, normally, increased posterior talar glide should be associated with increased DF ROM. Recent research indicates that limited DF ROM has been associated with a lack of posterior talar glide on the tibia in subjects suffering from ankle sprain.[1,2]

Therefore, the purpose of this study was to examine the relationships between four different positioning measurements of DF ROM and the amount of talar glide as assessed manually with the posterior talar glide test and with an instrumented ankle arthrometer

METHODS

Forty-seven healthy participants participated. Variables measured included: posterior talar glide assessed manually, posterior talar displacement assessed with an arthrometer, and DF ROM in seated straight knee, prone bent knee, standing straight knee, and standing bent knee positions.

Pearson product moment correlation coefficients were calculated to determine the strength of relationships between the four DF ROM measurements, talar glide (manual) and displacement (arthrometer) measures. Data for the right and left limbs was analyzed separately.

RESULTS

Descriptive statistics are shown in Table 1 and the correlation results are presented in Tables 2 and 3.

Table 1. Means and Standard Deviations

Measure	Right (95% CI)	Left (95% CI)
SSK	9.6 ± 6.3 (9.0 – 12.6)	10.8 ± 6.3 (7.8 – 11.3)
PBK	14.0 ± 6.5 (13.0 – 16.7)	14.8 ± 6.4 (12.1 – 15.8)
STBK	34.8 ± 6.7 (33.7 – 37.5)	35.6 ± 6.7 (36.7 – 32.9)
STSK	32.5 ± 6.7 (35.4 – 31.6)	33.5 ± 6.7 (30.6 – 34.4)
Glide	26.0 ± 7.0 (26.4 – 30.35)	25.7 ± 7.0 (24.0 – 28.0)
Displacement	4.73 ± 1.61 (4.25 – 5.2)	5.17 ± 2.82 (4.35 – 6.00)

Table 2. Correlation results for the right limb data. SSK = sitting straight knee, PBK = prone bent knee, STSK = standing straight knee, STBK = standing bent knee (*$p \le .05$), †($p < .01$)

	PBK	SSK	STSK	STBK	Talar Glide	Displacement
PBK						
Pearson *r*	1					
Sig. (2-tailed)						
SSK						
Pearson *r*	.33*	1				
Sig. (2-tailed)	.02					
STSK						
Pearson *r*	.36*	.26	1			
Sig. (2-tailed)	.01	.08				
STBK						
Pearson *r*	.30*	.27	.69†	1		
Sig. (2-tailed)	.04	.07	.001			
Talar Glide						
Pearson *r*	.12	.27	.43†	.25	1	
Sig. (2-tailed)	.43	.07	.002	.09		
Displacement						
Pearson *r*	.25	.29	.07	.31*	.13	1
Sig. (2-tailed)	.09	.051	.64	.04	.39	

Table 3. Correlation results for the left limb data. SSK = sitting straight knee, PBK = prone bent knee, STSK = standing straight knee, STBK = standing bent knee (*$p \le .05$), †($p < .01$)

	PBK	SSK	STSK	STBK	Talar Glide	Displacement
PBK						
Pearson *r*	1					
Sig. (2-tailed)						
SSK						
Pearson *r*	.45*	1				
Sig. (2-tailed)	.002					
STSK						
Pearson *r*	.20	.18	1			
Sig. (2-tailed)	.18	.24				
STBK						
Pearson *r*	.33*	.36*	.60†	1		
Sig. (2-tailed)	.03	.01	.001			
Talar Glide						
Pearson *r*	.12	.22	.50†	.30*	1	
Sig. (2-tailed)	.41	.14	.001	.04		
Displacement						
Pearson *r*	.24	.28	.04	.27	-.04	1
Sig. (2-tailed)	.11	.055	.80	.07	.79	

DISCUSSION

We observed several statistically significant, albeit weak to moderate strength, correlations between the DF ROM measures and posterior talar glide as measured both manually and with the arthrometer. These findings support the hypotheses that a relationship exists between measures of posterior talar glide and DF ROM. Additionally, the weakest relationships among the different DF ROM measures were between the two OKC measurements. These results suggest that the two OKC measurements were not capturing the same osteokinematic motion occurring at the talocrural joint.

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Kinematics, kinetics, and muscle activity during an ankle injury: a case study

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INTRODUCTION

Participation in sport is coupled with the risk of injury, and in basketball the majority of these injuries involve the foot and ankle.[1] This case study provides real biomechanical data of the timing of various events including joint rotations, joint loading and muscle activation during an ankle injury.

METHODS

Kinetic data were collected with a 0.6 x 0.9 m force platform (model Z4852C, Kistler AG, Winterthur, Switzerland) sampling at 2000 Hz. Kinematic data were collected simultaneously with the kinetic data using an eight-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA, USA) operating at 200 Hz. Twenty-five retro reflective markers were placed on each subjects' right foot, leg and hips.

Data of the injured trial, the 4 preceding trials of that day, as well as the data from 7 other subjects were analyzed. For each repetition of the drill, the step taken on the forceplate was identified and analyzed. The timing of heel strike and toe-off were found in relation to the pre-drill synchronizing impact. This time was then used to extract the corresponding step in the EMG data.

RESULTS

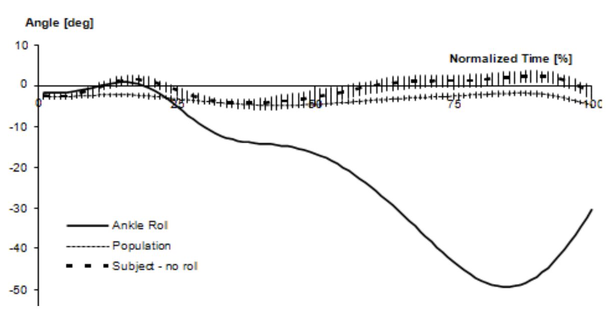

Figure 1: Ankle Inversion-eversion angle normalized to stance phase

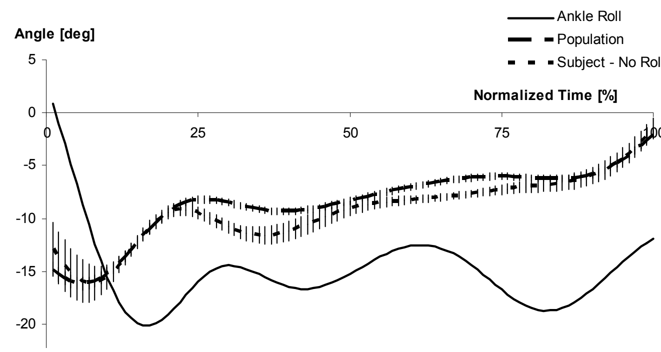

Figure 2: Knee internal-external rotation angle normalized to stance phase

DISCUSSION

The first compelling event differentiating the ankle-roll trial from the remaining data is the knee external rotation angle at touch down (Figure 2). The subject is approximately 15 degrees more externally rotated at touch down compared to previous trials, which as a result corresponds to a substantial increase in the knee internal rotation velocity. It has been suggested that lateral ligamentous injuries result from supination and inversion of the foot with external rotation of the tibia. [2] The results of this study suggest that an external rotation of the tibia at touch down and the subsequent rapid internal rotation to realign the knee joint initiated the series of events culminating in the injury. Interpretation of the same figure suggests that the subject over-compensated resulting in a joint position 5-10 degrees more internally rotated in the non-injured trials. The rapid ankle inversion that causes a joint position that can potentially injure tissue does not start until approximately 25% of stance (Figure 1).

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Invertor vs. Evertor Activation and Foot COP During Walking in Ankle Instability

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INTRODUCTION

Some evidence suggests that ankle instability patients have a laterally deviated foot center of pressure (COP) during gait [1]. While evertor dysfunction is well documented in subjects with ankle instability, there is no direct evidence to support the idea that invertor excitation could couple with evertor inhibition to result in a laterally deviated COP. The purpose of this study was to examine evertor/invertor activation and foot COP during walking.

METHODS

Subjects (n=12, 5 males and 7 females; age = 23 ± 4 yr; Height = 1.74 +/- .14 m, Mass = 71.6 kg +/- 17.6 kg) were identified with functional ankle instability (FAI) via the Functional Ankle Ability Measure (FAAM), the Modified Ankle Instability Index (MAII) and a physical examination. Controls (n=12, 5 males and 7 females; age = 23 ± 4 yr; 1.76m +/- .165m, Mass = 71.4 kg +/- 16.28 kg) had normal scores on the FAAM and MAII. EMG of the tibialis anterior (TA) and peroneus longus (PL) muscles was recorded during walking at a speed normalized to leg length. COP was measured with pressure insoles. Data from three consecutive stance phases were used for analysis. A functional analysis of variance was used to determine differences between FAI and control subjects with respect to normalized EMG and COP trajectory (α = 0.05). This analysis compares variables as polynomial functions rather than discrete values.

RESULTS

The functional analysis revealed a lateral COP trajectory in FAI subjects (Figure 1a). TA activation was increased in the FAI group during midstance, while the PL trended lower (Figure 1b & 1c respectively). The PL also displayed increased activation at heel strike and toe off in the FAI group (Figure 1c).

DISCUSSION

Our findings suggest COP trajectory is lateral in FAI subjects relative to controls, which is consistent with previously reported data [1]. Altered evertor and invertor activation coupled with lateral COP during midstance potentially places the foot in a dangerous position. The increased PL activation observed in the FAI group at heel strike and toe off may be a splinting mechanism designed to protect the unstable ankle during high impact forces (heel strike) or in vulnerable positions (plantarflexion during toe off).

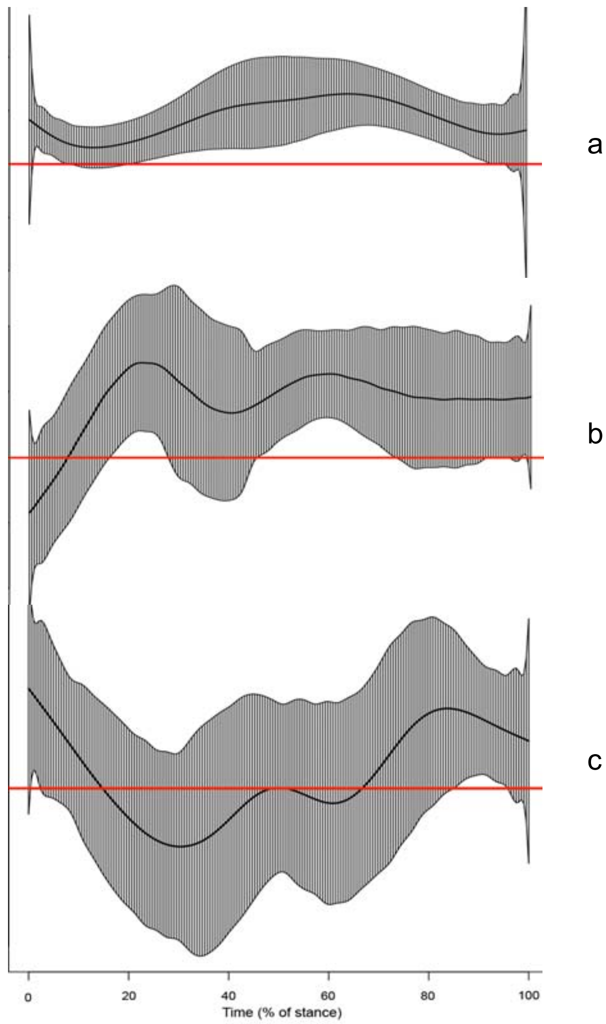

Figure 1: Functional analysis of (a) COP, (b) TA EMG, (c) PL EMG. Shaded areas = 95% confidence intervals. Red line divides groups. Confidence interval completely above red line indicates significantly greater FAI values. Confidence interval completely below red line indicates significantly greater control values.

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Intraoperative Pedography Improves Clinical Outcome at Follow-up of more than 20 months

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INTRODUCTION

The purpose of this study was to analyze the potential clinical benefit of intraoperative pedography (IP) in a sufficient number of cases in comparison with cases treated without IP. We hypothesized that the intraoperative changes after use of IP might improve clinical outcome scores in comparison with patients treated without IP and consequently no intraoperative changes for intended biomechanical improvements made.

METHODS

Patients (age 18 years and older) which sustained an arthrodesis and/or correction arthrodesis of the foot and ankle were included, starting September 1, 2006.

All subjects receive preoperative clinical and radiographic assessment and standard dynamic pedography. The subjects are randomized into two groups, a) use of IP, versus b) no use of IP.

The IP of the foot that was operated on was used *after* the surgeon considered the correction process and the internal fixation to be optimal based on the surgeons´ experience including the evaluation of the clinical appearance of the foot, and c-arm images. In the IP group, the contralateral foot and the involved foot before correction were measured in the preparation area after the beginning of the anaesthesia.

The following scores are used: American Orthopaedic Foot and Ankle Society (AOFAS), Visual-Analogue-Scale Foot and Ankle (VAS FA), Short-Form 36 (SF36, standardized to 100-point-maximum).

Intraoperative consequences after the use of IP and any adverse effects were recorded. Follow-up including clinical examination, radiological studies, dynamic pedography, and foot and ankle focused was performed.

RESULTS

One hundred cases were included until April 11, 2008 (ankle correction arthrodesis, n=12; subtalar joint correction arthrodesis, n=14; arthrodesis without correction midfoot, n=15, correction arthdodesis midfoot, n=26, correction forefoot, 33). The mean preoperative scores were as follows: AOFAS, 49.1±24.6; VAS FA, 45.3±21.2; SF36, 43.1±31.2. No score, age or gender distribution differences between the two groups occurred (scores, age, t-test, $p>0.05$; gender, chi2-test, $p=0.8$). Fifty-two patients were randomized for the use of IP. The mean interruption of operative procedure for the IP was 321±39 seconds. In 24 of the 52 patients (46%), the correction was modified after IP during the same operation. The changes were done most likely in midfoot correction arthrodeses (64%), and least likely in subtalar joint arthrodeses (25%). A following IP then did not lead to further changes in the correction. No malfunctions of the IP system occurred. No complications related to the use of IP such as infection were observed.

All patients completed follow-up of more than 20 months (21 - 40 months, 28 months on average). The follow-up scores were AOFAS, 91.2±15.7; VAS FA, 90.8±12.7; SF36, 91.5±16.9. The scores at follow-up were significantly higher than the preoperative scores (t-test, $p<.05$). The scores from the IP group were significantly higher than the scores from the group without IP (t-test, $p<.05$).

DISCUSSION

At mean follow-up more than 20 months in a level 1 study, additional use of IP as the only difference between two groups with correction and/or arthrodesis at foot and/or ankle led to improved clinical outcome scores. Low scores (less than 69 points in all scores) were avoided with IP. This has to be critically re-analyzed when longer follow-up, higher case numbers, and data from other study centers are available.

Three-Dimensional Analysis of the Ludloff Osteotomy

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OBJECTIVE

To study the characteristics of the Ludloff Osteotomy.

MATERIALS

A wooden cylinder 86.56 mm in length and 30.65 mm in diameter was used to imitate Ludloff Osteotomy. A vernier caliper was used in measurement.

METHODS

①Fixed one end of the cylinder after Ludloff Osteotomy on another cylinder which was fixed on one board with a nail so that the distal of the former was more than 5 mm above the board. ②Made a hole at the intersection of the 0° and 90° on the protractor. Then fix the protractor through the hole with a nail on the center of the distal cross-section of the cylinder. And kept the 0° line on a parallel with the horizontal plane. ③Measured the length of the cylinder and the distance from the lowest point of the remote end to the flat board with a vernier caliper separately. ④Pushed the distal of the cylinder externally and made sure the moving angle of the protractor was 5°, 10°, 12°, 15° respectivly. Measured the angle between the distance and proximal end after pushing the distal of the cylinder outward with another protractor. ⑤Made the distal reduction. Rotated the whole cylinder around the screw and made sure the rotated angle of the distal protractor was 5° and 15° respectively.

RESULTS

As the extent of the orthopedic increased, the length of the cylinder became shorter and shorter, and the distal of the cylinder became pronated constantly. When the angle of the section is deviate to plantar, the lowest point of the cylinder lowered, then rose. However, if this angle is perpendicular to the sagittal plane or deviate to dorsal, the lowest point rose constantly.

CONCLUSIONS

①The reduction of the metatarsal head must be influenced by the constant pronation. ②A section of 15~20°angle deviate to plantar was suggested for patients with serious metatarsalgeail.

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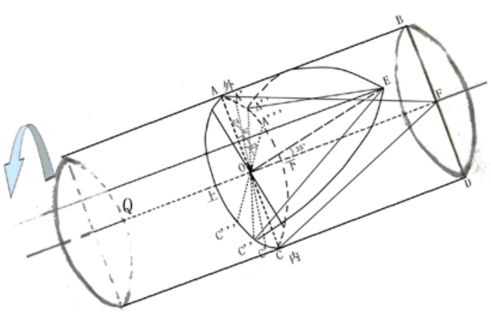

Figure 1. Ludloff Osteotomy section plane drawing

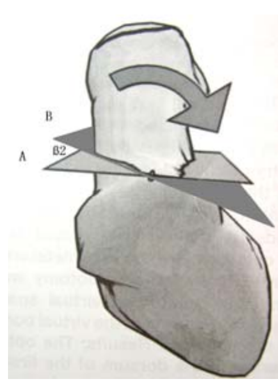

Figure 2. Behind view

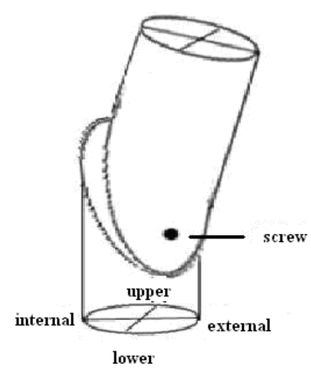

Figure 3. Back view

Improvement in Dynamic Foot Pressure in Patients After Minimally Invasive Percutaneous Distal Metatarsal Osteotomy for Hallux Valgus

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INTRODUCTION

Hallux valgus (HV), a common forefoot deformity with a marked female preponderance of about 9:1 [1], would change foot biomechanics. A rise of plantar pressure of the lateral forefoot was found correlating to the aggravation of the HV deformity [2, 3]. Minimally invasive percutaneous distal metatarsal osteotomy (MIPDMO) is newly recommended for correction of HV [4, 5], but no study has reported the effects of MIPDMO on the biomechanics of HV foot. The aim of this study was to investigate the changes in dynamical plantar pressure distribution in patients after MIPDMO for HV.

METHODS

Ten feet in five females with bilateral HV (age: 46.8±8.5 years, body weight: 62.5±5.3 kg, body height: 156.4±4.3 cm) were analyzed using radiography and foot pressure before and 4.9 months on average after MIPDMO. Dynamic plantar pressure during walking at a cadence of 110 steps/minute was measured and analyzed by a pressure platform (RSscan International, Belgium). Differences between pre-operation (Pre) and post-operation (Post) in the peak force (N) and pressure (N/cm^2) at ten foot regions, namely the big toe (T1), 2nd-5th toes (T2), 1ST -5th metatarsals (M1-M5), mid foot (MF), medial and lateral heel (MH & LH), were analyzed by non-parametric Wilcoxon signed ranks test with $\alpha = 0.05$.

RESULTS

Both hallux abductus angle (Pre: 32.4$\pm$4.4°, Post: 10.7$\pm$3.9°) and 1st/2nd intermetatarsal angle (Pre: 13.8$\pm$3.2°, Post: 8.1$\pm$1.0°) were significantly improved after MIPDMO (p<0.001). The peak force and pressure were found at M1 and M2 before MIPDMO, while the peak force and pressure were observed at heel and M2 after MIPDMO. Significant decrease in peak pressures at T2 (p=0.037) and M2 (p=0.045) were found post-operation (Figure 1).

DISCUSSION

The findings of the peak plantar pressures at the medial metatarsals in patients before MIPDMO were similar to previous studies [6, 7]. In addition to the satisfied correction of abnormal deformities in HV, decreased peak pressure at T2 showed that MIPDMO could effectively improve the abnormal rise of plantar pressure in the lateral forefoot which was commonly found in aggravated HV deformity. Abnormal elevation in peak plantar pressure was also modulated after MIPDMO.

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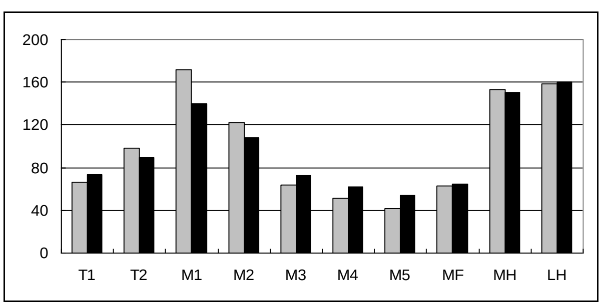

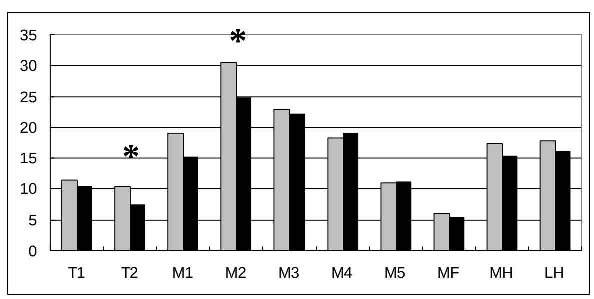

Figure 1. The peak force (N, left figure) and peak pressure (N/cm^2, right figure) pre- (gray bar) and post-operation (black bar). ***** indicated significant differences between pre- and post-operation.

Significant Reduction of Abnormal Forces and Impulses of the Lesser Toes in Patients after Chevron Osteotomy for Hallux Valgus

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INTRODUCTION

Hallux valgus (HV) is a common disease with abnormal bunion and cross-over toe deformity. The deformity is usually manged with osteotomies. Chevron osteotomy has been a main treatment for HV, with advantages of high union rate and low complication rate. However, few studies have reported the biomechanical efficacy of this particular procedure. The purpose of the study was to quantify the changes of foot pressure of patients with HV after Chevron osteotomy.

METHODS

Two groups of 20 subjects were included in the current study, namely normal subjects and patients with HV who were treated by Chevron osteotomy. The patient group was selected according to radiological findings. The foot pressures of the subjects during level walking were measured using a pressure plate (RS scan, Belgium, Fig 1). Means, standard deviations of the forces and impulses under 7 areas were calculated. Comparisons between conditions were made using independent t-test with =0.05.

RESULTS

Means and standard deviations of the measured parameters pre- and post-operation are given in Tables 1 and 2. Compared to normal subjects, forces and impulses of the lesser toes were significantly greater in the HV pre-OP subjects group ($P<0.05$). Compared to pre-OP, forces and impulses of the lesser toes were significantly lower post-OP in the HV group ($P<0.05$) and were similar to the normal subjects ($P>0.05$).

DISCUSSION

A HV deformity can be associated with abnormal foot mechanics, inflammatory change and so on. Patients suffer from bunion, subluxation of the first MTP joint, 2nd MTP metattarsalgia and cross-over deformity. Patients treated for HV by Chevron procedure showed significant reduction in the abnormal forces and impulses of the lesser toes during gait, as measured by the pressure plate. This suggests that Chevron procedure was effective in relieving abnormal pressure under the lesser toes, in addition to the restoration of the normal alignment of the big toe. While pressure plate was effective in providing foot pressure for patient evaluation, its application to subjects with gait problems, foot deformity, lower extremity deformity, etc. may be limited.

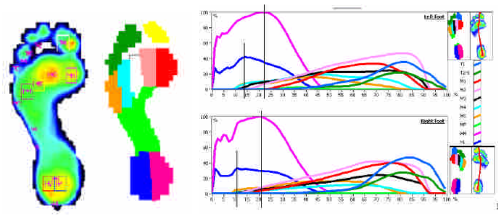

Figure 1: Normal subject on Foot Scan.

Table 1. Foot pressure variables in hallux valgus group, pre-operation.

	Start Time	End Time	Total Time	Peak Time	Force	Impulse
Med heel	0	61±10	61±10	26±7	93±11	59±6
Lat heel	0	58±1	58±11	25±6	67±18	39±7
M1	39±14	91±7	52±16	74±6	44±20	18±13
M2	29±12	95±8	66±16	78±7	55±19	25±7
M3	27±12	96±3	69±13	79±6	53±17	27±9
M4	31±15	94±4	63±16	76±8	43±16	20±8
M5	34±20	84±16	49±20	69±15	28±20	8±5

Table 2. Foot pressure variables in hallux valgus group, post-operation.

	Start Time	End Time	Total Time	Peak Time	Force	Impulse
Med heel	0	65±9	65±9	34±5	98±2	60±4
Lat heel	0	64±13	64±13	31±8	77±2	42±3
M1	43±9	89±3	45±7	70±6	39±6	22±4
M2	44±9	98±2	54±11	80±1	45±3	28±1
M3	47±13	98±2	51±13	83±3	44±6	26±6
M4	48±18	97±2	47±19	81±2	40±12	19±5
M5	63±17	91±6	28±18	81±4	23±20	5±4

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Gait Changes in Patients After Minimally Invasive Percutaneous Distal Metatarsal Osteotomy for Hallux Valgus

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INTRODUCTION

Hallux valgus (HV), a common forefoot deformity with a marked female preponderance of about 9:1 [1], would change foot biomechanics and up to the ankle, knee, and hip joints. Minimally invasive percutaneous distal metatarsal osteotomy (MIPDMO) to correct HV had potential advantages of reduced operating time, decreased surgical exposure and early weight-bearing [2, 3], but no study has reported the biomechanical effects of MIPDMO on HV foot. The aim of this study was using gait analysis to investigate the gait changes in patients after MIPDMO for HV.

METHODS

Six females with bilateral HV (age: 46.8±8.5 years, body weight: 62.5±5.3 kg, body height: 156.4±4.3 cm, pre/post HV angle: 32.4±4.4°/13.8±3.2°) were analyzed before and 4.9 months on average after MIPDMO. A control group of 8 healthy females were recruited for comparison. Each participant walked for 3 times at a self-selected pace while their kinematic data measured by a 7-camera motion capture system (Vicon 512) at 120Hz and ground reaction forces (GRF) from two AMTI forceplates at 1080Hz. Group and surgical effects on the joint angles and moments at the beginning and the end of single leg stance (bSLS, eSLS) and toe-out angle were analyzed by Mann-Whitney U test and Wilcoxon signed ranks test with $\alpha = 0.05$.

RESULTS

Although gait speeds, cadence, and stride length were statistically the same for 2 groups, both pre- and post-operative HV group had significant toe-in (p=0.022). When compared to control group, preoperative HV group had significantly greater ankle abduction and internal rotation (IR) at bSLS and eSLS, greater vertical GRF at mid-stance, and decreased medial GRF, knee and ankle abductor moments. When compared to pre-operation, postoperative HV group had reduced vertical GRF at mid-stance, hip adduction and hip abductor moments at eSLS and increased medial GRF.

DISCUSSION

Increased ankle IR and abduction were highly related to increased foot pronation and toe-in gait commonly found in feet with HV. Greater toe-in resulted in smaller medial GRF [4] and decreased knee and ankle abductor moments [5]. Increased vertical GRF at mid-stance in preoperative HV group indicated walking with an impaired shock absorption mechanism. Flatfoot with poor compliance of the foot arch had been considered as one of the major causes of HV and would alter the shock absorption mechanism because of exaggerated eversion of the subtalar joint [6, 7]. After MIPDMO, patients could restore their ability of shock absorption by correcting the abnormal alignment of HV and flatfoot, but they still walked with greater foot pronation and toe-in. Walking in such deviated pattern for a long time could still be anticipated to aggravate HV in future and the follow-up in patients after MIPDMO for HV should be noted.

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Detection of forefoot pain based on plantar pressure parameters.

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INTRODUCTION

Plantar pressure measurement is a helpful tool to evaluate patients with foot complaints. Recently, new methods are developed that spatially scale the plantar pressure image to a standard foot size and angle [1]. This method opens the door for the use of more sophisticated analysis techniques and makes it possible to compare pressure patterns between groups.

As, one of the most common foot complaint is forefoot pain, the purpose of this study was to indicate differences in plantar pressure pattern between subjects with forefoot pain and controls. Subsequently, neural networks were used to detect forefoot pain based on parameters of plantar pressure measurements.

METHODS

297 subjects with and without aspecific forefoot pain walked 10 times alternating with the left and right foot over the pressure plate (Rsscan Int, Belgium). All subjects walked barefoot at their preferred walking speed according to the three step protocol. Forefoot pain and subject characteristics were assessed by a questionnaire and a physical therapist.

Plantar pressure patterns were normalized for foot angle and foot size using the method of Keijsers et al. [1]. Pressure images of pressure time-integral (PTI), peak pressure (Peak), mean pressure (Mean), activation (PixOn) and deactivation (PixOff) in relation to first contact, and total contact time per pixel (PixCT) were calculated. Subsequently, pressure images were preprocessed by principal component analysis. Finally, a forward selection procedure of the principal component scores with neural networks was used to classify forefoot pain. The performance of the network was evaluated on feet, which were not used to develop the neural network

RESULTS

Body weight, BMI, age, foot progression angle, and contact time were not significantly different between subjects with and without forefoot pain. Figure 1 indicates that PTI and Mean were significantly larger under metatarsal II and III for subjects with forefoot pain. Subjects with forefoot pain showed later heel off (PixOff) and a larger contact time under the metatarsals (PixCT).

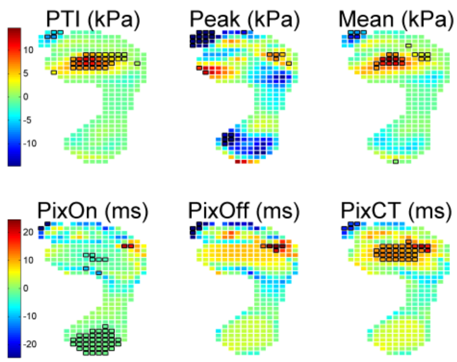

Figure 1: Differences in pressure images between subjects with and without forefoot pain. Pixels with borders indicate significant difference

A neural network with 14 input parameters based on pressure distribution, correctly classified forefoot pain in 70.4% of the feet. It was not possible to classify forefoot pain using pressure parameters of regional data.

DISCUSSION

Although significant differences were found in plantar pressure, these differences were too small to distinguish between subjects with and without forefoot pain. However, the reasonable performance of the neural network in detecting forefoot pain indicates the valuable information of the pressure pattern. Therefore, forefoot pain is more related to the way the pressure is distributed under the foot than absolute pressure values.

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The impact of an actuated ankle-foot orthosis on the walking performance in healthy subjects and spinal cord injured patients: A systematic review

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INTRODUCTION

Spinal lesions, characterized by a partial or complete transection of the ascending and descending neural pathways, often lead to severe and long term motor deficits [1]. The loss of plantar and dorsiflexor strength, particularly of dynamic strength, is a major component leading to motor impairment of the lower limb in incomplete spinal cord injured patients. Recovery after a complete or partial spinal lesion depends on multiple cooperative mechanisms modifying neural connectivity and function, defined as neuroplasticity [2]. Technological evolution has enabled the design of powered exoskeletons, stimulating these mechanisms, by providing appropriate afferent input, such as proprioceptive information related to repetitive, alternating lower limb loading and kinematics consistent with walking [3],[4]. This review provides an overview of robot assisted rehabilitation devices developed for the ankle-foot complex and their influence on gait performance.

METHODS

A comprehensive search of the entire English, German, French and Dutch literature was conducted through multiple databases (Medline, ISI Web of Knowledge, PEDro & Cochrane Controlled Trails Register). Appropriate studies were included based on a three-step selection procedure.

RESULTS

The state of the art in actuated ankle-foot orthotics (AAFO) can be classified by two methods, driving the ankle-foot complex through a predetermined physiological gait pattern:

1. Actuated ankle-foot orthosis included in a driven gait orthosis

Despite the viable role of the ankle-foot complex in human gait, only two research groups have incorporated automated ankle-foot actuation in their step rehabilitation robot.

2. Isolated actuated ankle-foot orthosis

Numerous research groups have designed and clinically tested AAFO independent of knee and hip actuation for gait rehabilitation. The AAFO currently designed and clinically tested confine to plantar and/or dorsiflexion support, limiting additional freedom of movement in the frontal and transversal plane. By restricting additional degrees of freedom, one ignores movement and shock absorption as it occurs at the level of the interconnecting bones of the ankle-foot complex. In turn, disabling the transfer of large loads and restricting degrees of freedom, necessary for normal gait.

DISCUSSION

Ankle-foot actuation is mainly developed for gait rehabilitation, independent of knee and hip actuation, in order to enhance neuroplasticity and encourage motor learning. So far studies on AAFO mainly focus on the description of the design, the actuator type and the control strategy. In addition, most of these studies report initial test results of ankle-foot actuation for healthy subjects only. In the future, randomized clinical trials should focus on the short and long term effect of gait rehabilitation through AAFO in spinal cord injured patients expressed in biomechanical outcome measures.

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Characterizing multisegment foot kinematics, kinetics and plantar pressure during gait of severely deformed feet in Rheumatoid Arthritis: a case study.

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INTRODUCTION

A multisegment approach to foot motion analysis is pursued in rheumatoid arthritis (RA), since this inflammatory disease causes foot deformities and changes in joint motion and alignment [1]. While multisegment foot (mf) kinematics [1], plantar pressure (PP) have been widely investigated, none analyzed foot [2] subsegments' ground reaction forces (GRFs) in RA subjects. The latter have been shown to provide a more complete description of foot motion abnormalities in diabetics subjects [3-4]. So far this study evaluated the effectiveness of such a comprehensive methodology on one RA subject with highly deformed feet (Figure 1).

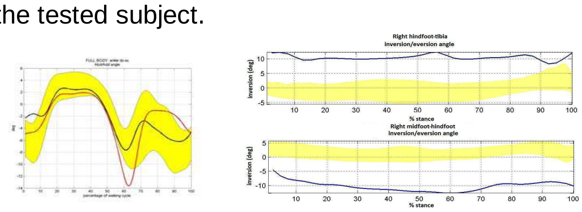
METHODS

Data were collected with a motion capture system (6 tvc Smart-BTS, 60-120 Hz) synchronized with 2 force plates (Bertec), 2 PP systems (Imagortesi), an electromyographic (EMG) system (BTS-FreeEMG). Both the mf kinematics and kinetics models described in [3-4] were applied. Gastrocnemius medialis (GM), tibialis anterior (TA), peroneus longus EMG activities were recorded. Ten normal and one RA subjects, with evidence of feet deformities, were analyzed. The following variables were evaluated [2-4]: local sub-segment vertical (V), anterior-posterior and mediolateral GRFs (normalized to body weight); center of pressure displacement during gait, the peak pressure curve, the mean pressure curve, the loaded surface curve; 3-dimesional (3D) mf kinematics.

RESULTS

When compared with normative bands RA subject 3-D mf kinematics (Figure 2) revealed excessive inversion of the left foot during initial swing, increased dorsiflexion at initial contact and increased external rotation during the whole gait cycle on both feet. Furthermore a hindfoot excessively inverted, a midfoot excessively everted, externally rotated and plantarflexed during the stance phase were observed. The forefoot resulted excessively everted, externally rotated and dorsiflexed. When considering kinetics parameters, the right foot displayed increased ankle internal-rotation and dorsiflexion moments. Also excessive foot subsegmets' tangential forces were registered. An opposite trend was observed for the left forefoot V GRFs' component. Reduced EMG activity was registered at GM and prolonged at TA.

Figure 1:3D multisegment foot protocol applied on the tested subject.



(a) (b)

Figure 2: (a) Right (black) and left (red) Inversion-Eversion- ankle angle; (b) hindfoot-tibia (top) and midfoot-hindfoot (bottom) inversion/eversion angles over the normative bands.

DISCUSSION

Based on these results both a specific insole and a footwear were prescribed, which stabilize the heel by increasing the hindfoot bearing surface and reduce the excessive forces on the metatarsal area.

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Foot and ankle joint kinematics in rheumatoid arthritis cannot only be explained by alteration in walking speed

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INTRODUCTION

Rheumatoid arthritis (RA) manifests itself in the feet and ankles of RA patients. The foot and ankle joint kinematics of these patients differ from that of healthy subjects [1,2,3]. However, the factors that lead to these differences are not yet fully understood. The aim of this study was to analyse the effect of walking speed on foot and ankle joint kinematics of RA subjects.

METHODS

Gait recordings of 21 RA and 14 age-matched healthy subjects were performed. Stance phase kinematics of their foot and ankle joints were analysed and compared. The RA group included subjects with average disease duration of 9 years (SD 7y) and moderate to severe disease activity and progression. The healthy subjects walked at 100% (Vc), 75% (V75) and 50% (V50) of their comfortable waking speed. Differences between the stance phase kinematics of the two groups caused by the factors walking speed and the RA disease process were analysed using a multi-linear model.

RESULTS

The ankle joint dorsi-flexion (figure 1), medial arch motion and hallux abduction were significantly influenced by walking speed alone. Hallux flexion (figure 2), navicular bone motion, mid foot supination and leg rotation were influenced by both walking speed and the disease process. Hind foot eversion was solely influenced by the disease process.

The RA joint kinematics were comparable to those of the healthy subjects walking at lower speeds for those joints that were influenced only by walking speed. Abnormal RA joint kinematics compared to healthy subjects walking at lower speeds were observed for those joints that were influenced by the disease progress.

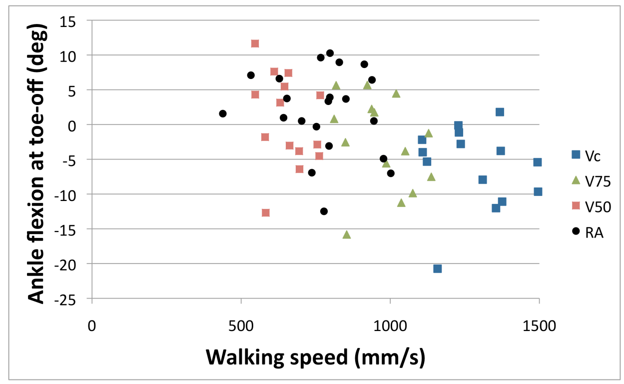

Figure 1: The ankle flexion at toe-off of healthy and RA subjects.

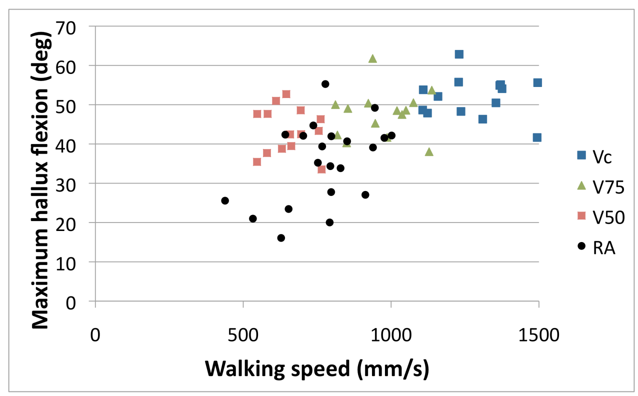

Figure 2: Maximum hallux flexion of healthy and RA subjects during stance phase of gait.

DISCUSSION

Walking speed alone cannot explain all differences in the foot and ankle kinematics between RA and healthy subjects. On the other hand, not all observed differences in RA joint kinematics are pathological compared to healthy subjects. Future studies should focus on determining the causes of abnormal hallux flexion at toe-off and reduced mid foot and hind foot eversion during mid-stance in RA subjects.

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Effects of Kinesio-Taping of the Gastrocnemius on Muscle Activity Patterns of the Lower Leg

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INTRODUCTION

Kinesio-taping is advocated for the treatment of a variety of musculoskeletal problems even though literature reports are rare [1,2]. According to the manufacturer, the elasticity of the material is being used to stimulate either muscle activation or relaxation, depending on the application technique. While the subjective feedback of patients often is positive it has not been shown whether this effect can also be objectively demonstrated.

The aim of the present study was to investigate whether any effects can actually be observed in changes of muscle activity patterns as observed by surface electromyography (sEMG).

METHODS

The present study was carried out with 20 healthy subjects (8 males, 12 females, average age 26 years, height 177 cm and weight 73 kg). Subjects were measured without and with tape applied over the gastrocnemius (Fig 1). Muscle activities were recorded during free walking from the tibialis anterior and both gastrocnemii with bipolar electrodes (blue sensor). Signals were amplified and processed with Myosystem hard- and software (Noraxon, USA).

Measurements were performed three times: 1. before tape application; 2. immediately after taping; 3. after 30 minutes of walking. At least 15 gait cycles were recorded and stored. The raw data was root-mean-squared and averaged across repeated trials to obtain mean and peak EMG activities for the investigated muscles in each condition.

Differences between conditions were tested for significance with the Friedman test ($p<0.05$).

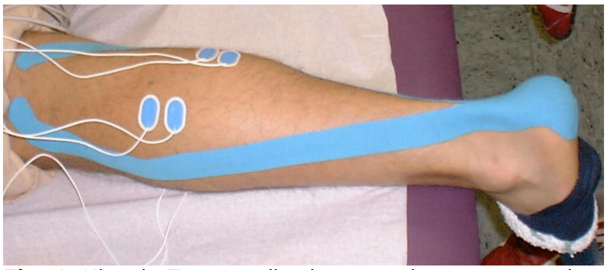

Fig. 1: Kinesio-Tape application over the gastrocnemius; also shown are the bipolar electrode over both heads.

RESULTS

A slight but significant decrease was seen in both, mean and peak EMG amplitudes (Tab. 1). However, the contralateral limb that was not taped showed a similar trend. The magnitude of changes was between 4% and 16%.

DISCUSSION

The results do not clearly demonstrate the proposed effect of muscle relaxation as reflected in lower EMG amplitudes because the same trend was seen also in the leg that was not taped. Therefore, it might be mainly due to general changes in the recording conditions over time even though the electrodes were kept in place throughout the experiments in order to prevent problems due to electrode repositioning. Furthermore, EMG differences were seen between the limbs even before tape application so that final conclusions are difficult.

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ACKNOWLEDGEMENT

The tape was provided free of charge by KinesioTex.

Table 1: Average values of the mean and peak EMG during walking before, immediately after and 30 minutes after Kinesiotape application (n=20; n.s. = not significant, p>0.05)

EMG Amplitude (µV)	Before taping		After taping		After 30 minute walk		P-Level	
	Mean	Peak	Mean	Peak	Mean	Peak	Mean	Peak
Tibialis anterior	36.3±11.3	93.0±34.4	34.6±10.8	91.3±32.3	34.8±9.4	91.6±32.5	n.s.	n.s.
Medial gastrocnemius	30.3±15.0	107.7±58.7	29.2±14.3	107.3±58.8	27.5±11.9	98.0±51.3	.0260	.0043
Lateral gastrocnemius	17.8±10.2	51.3±25.8	17.8±10.3	50.3±23.0	17.8±11.6	43.0±17.9	.0043	.0037

Effect of Elastic Taping on Motions of Ankle Syndesmosis Joint

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INTRODUCTION

Compared with patients with other ankle sprains, ankle syndesmosis injury takes more period of time to get recovered.[1-3] The common intervention of ankle syndesmosis injury is to apply either athletic or elastic taping onto the injured joint in order to bind the tibia and the fibula together.[4] Although the confinement effect of such taping is appreciated by the injured athletes, there are no solid evidences provided. Therefore, the purpose of this study was to examine the effect of either athletic or elastic taping on motion of the ankle syndesmosis.

METHODS

Thirty young healthy adults were recruited in this study, 15 men and 15 women with mean age of 24.3 ± 3.1 years. All participants were tested in three conditions in a random order: non-taping, elastic taping, and athletic taping. The participant was asked to perform a deep squat from quiet stance and then stand up with the fastest speed. Three trials were collected for each condition using FASTRAK motion tracking system.[5] An average of three trials of bilateral malleoli distance was used to represent the mobility of the ankle syndesmosis, and the angle of the talocrural joint at the most deep squat moment was served as covariate. After completing the test of each condition, all participants were asked to point out the degree of comfortability on a 5-point Likert scale during performing deep squat, heel-off, one-leg stance, walking, and running when elastic or athletic taping was applied. ANCOVA with repeated measures was used to compare the differences of the changes of bilateral malleoli distance from quiet stance to deep squat among three different conditions. The Wilcoxon Signed Ranks Test was used to examine the comfortability in performing deep squat, heel-off, one-leg stand, walking, and running between two tapes used. All statistical analyses were executed using SAS v.9.13. The statistically significant level was set at $\alpha = 0.05$ while the power was at 0.8.

RESULTS

The results showed changes of bilateral malleoli distance was 4.8 ± 1.3 mm for non-taping condition, 4.4 ± 1.3 mm for elastic taping, and 4.8 ± 1.5 mm for athletic taping. Although the change of bilateral malleoli distance tended to be less than the other two conditions, there was no significant differences among these three condition (p= .084). Comfortability in performing deep squat, heel-off, one-leg stance, walking, and running were lower for athletic taping as compared to elastic taping ($p<0.001$).

DISCUSSION

Although the elastic taping failed to have a significantly better restriction to ankle syndesmosis motion than other conditions, it was more comfortable than athletic taping, indicating another better treatment alternative in clinic. Since only healthy subjects were included in this study, further study will be directed to include the patients with ankle syndesmosis injury.

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Time of day influences the diameter of the plantar fascia

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INTRODUCTION

Time-dependent tendon conditioning has been observed with repeated muscle actions of extended duration [1,2], and with the short yet repetitive loads in walking [3]. Additionally, an increase in patella tendon axial strain over the course of the day has been observed [4]. It is important to know if time of day has an influence on the intraday variation in the diameter of the human tendon in order to make the correct diagnosis and evaluate the treatment effect. The purpose of this study was to investigate fluctuations in the diameter of the human plantar fascia (PF) during 24 hours of normal daily living.

METHODS

Ten healthy volunteers participated in the study (4 women, mean age 25,1, BMI 22,5). They were instructed not to alter their normal activity level. All had a substantial amount of walking from 8.00 am to 4.00 pm. An experienced observer scanned the subjects at 8.00 am, 12.00 pm, 4.00 pm, 8.00 pm and the next day at 8.00 am. A GE LOGIQe ultrasound scanner was used in the study. Long-axis scans were obtained at the insertion onto the calcaneus. The inbuilt software was used to measure the diameter of the PF. The mean value of 3 successive measurements for both right and left foot was applied in the statistics.
The diameter of the PF was normalized to the first measurement. This allowed inter-subject comparison across different diameters of the PF. Repeated measures one-way ANOVA with time of day as a factor was applied to investigate the effect of time on the diameter.

RESULTS

The mean diameter of the 20 PF changed during the 24 hours (Table 1). Sphericity was assumed (p=0,95). Repeated measures one-way ANOVA showed a significant effect of time on the diameter of the PF ($F_{36,4}$=3.88, p=0.01).

DISCUSSION

Grigg et al. found a significant effect of calf exercise on tendon diameter [2]. The diameter decreased after exercise and slowly returned to pre-exercise diameter. The same authors demonstrated that the repetitive loads during walking had a significant effect on the diametral strain of the Achilles tendon [3]. Compared to previous studies this indicates that tendon properties are influenced by activity and exercise [1,4]. As the PF primarily consists of the same type of collagen fibre as the Achilles tendon similar changes were expected in this study. A significant effect of time on the diameter of the plantar fascia was observed. However the changes were small and therefore we cannot exclude that the observed differences are caused by measurement uncertainty.
Further research with larger sample sizes are required to establish evidence concerning the intraday variation of the PF.

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Table 1: Relative diameter of the plantar fascia (mean±SD) in relation to time of day.

	Relative diameter of the plantar fascia (mean±SD)				
Time of day	**8.00 am**	**12.00 pm**	**4.00 pm**	**8.00 pm**	**8.00 am**
Mean ± SD	100±0	93,16±5,45	97,24±6,09	94,06±4,54	96,81±6,23

Running has no effect on the diameter of the plantar fascia

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INTRODUCTION

Tendon conditioning has been observed with repeated muscle actions of extended duration [1,2,3,4]. But the continuous repetitive loading of dynamic activities could affect tendon properties and result in tendon conditioning in the lower extremities. A previous study has shown that the short yet repetitive loads in walking are sufficient to induce time-dependant conditioning of the Achilles tendon [2]. Since the effect of short duration loading of the plantar fascia (PF) in dynamic activities has not yet been investigated the purpose of this study was to examine the effect of running on the diameter of the PF.

METHODS

11 healthy male and female runners (3 women, 22 feet, mean age 32,8years, body mass 68,3kg, height 177cm) refrained from physical activity prior to the baseline measurement. An experienced observer scanned the subjects just before, immediately after and 2 hours after a 15km run. A GE LOGIQe ultrasound scanner was used in the study. Long-axis measurements of PF thickness were obtained at the insertion onto the calcaneus. The inbuilt software in the scanner was used to measure the diameter of the PF. The mean value of 3 successive measurements at all 3 scheduled scans was applied for the statistics. No differences between right and left foot were found using paired t-test and therefore right and left PF was pooled in the analysis. Repeated measures ANOVA was used to evaluate the data for the 22 tendons.

RESULTS

The mean values (Mean±2SD) of the 22 tendons were respectively 38mm±17 before the 15km run, 37mm±17 after the run and 37mm±15 2 hours after the run (Figure 1).
Sphericity was assumed (p>0,05). The repeated measures ANOVA did not show significant interaction between time point and diameter of the PF, ($F_{42,2} = 1.359$, $p = 0.27$).

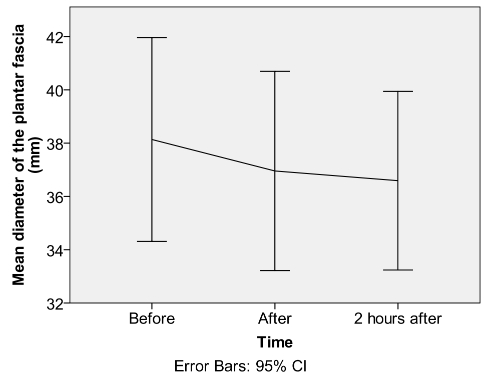

Figure 1: Mean diameter of the plantar fascia and 95% CI.

DISCUSSION

Contrary to the results found in a previous study concerning the Achilles tendon [2], the results of this study demonstrated no time-dependant conditioning. The short but repetitive loads generated during running had no statistically significant effect on the diameter of the PF. These conflicting results could be due to the duration and timing of the loading, which are extended in walking compared to running. Another explanation could be the differences in tendon properties and loading between the Achilles tendon and the PF and the fact that there is no direct muscle-tendon junction to the PF. Further research with larger sample sizes are required to establish evidence concerning the effect of continuous repetitive loading of dynamic activities on tendon properties.

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Foot and Ankle Research Northern Denmark.

Tibia internal rotation measured with a kinematic sensor -a reliability study

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INTRODUCTION

Excessive pronation of the foot is thought to cause overuse injuries in the lower extremities. Pronation with calcaneal evertion leads to a significant internal rotation of the tibia. There are individual variations of the degree to which evertion affects tibiarotation [1]. The extent to which proximal and distal forces affecting tibiarotation is also not clear. How motion control shoes and how foot orthotics affect tibiarotation and mechanisms behind the effects of the orthotics and shoes are inadequately understood. The theoretical basis for the recommendation of foot orthotics is also not clear [2].

We have not found other studies that have measured tibiarotation in clinical and everyday environments. We therefore wanted to develop a method to measure maximum internal rotation of the tibia starting at initial heel strike using mobile equipment. Thus the purpose of this study was to investigate whether a portable kinematic sensor could be used to measure tibiarotation.

METHODS

Four healthy subjects walked and ran on a treadmill with and without shoes at speeds 2.5, 4.0, 6,0 and 9.0 km/h. A 10 gram MTx kinematic sensor was used to measure accelerations and orientation along three orthogonal axes. The sensor was attached with a neoprenorthoses on the right tibia 2/3 distance between the medial malleol and tuberositas tibiae, and was associated with a battery-powered Bluetooth transmitter of 200 grams which was placed in the subject's trouser pocket. Heelstrike was identified from vertical acceleration signals and orientation from yaw gyroscope signals.

Data were transmitted in real time to a PC, and digital signal processing was conducted with a taylor made application under Matlab 7.0.4. Tibiarotation of five consecutive steps were analyzed for each of the four speeds. Statistical analysis was done with PASW 17. Reliability was assessed by ICC statistics.

RESULTS

Results in Table 1 showed reliability of tibiarotation for mean of five streps as ICC(1.5) > 0.90 in 6 of 8 data series, and reliability of tibiarotation for a single step as ICC(1.1) > 0.70 in 5 of 8 data series.

DISCUSSION

Measured tibiarotation in this study was slightly higher than in other studies [3]. Likely error sources in this study were skin movements and noise in the yaw curves during the reading of maximal tibiarotation. Further research should include more steps in the analysis to further reduce the influence of random error.

The results indicate that body fixed kinematic sensors can be used to measure tibia internal rotation. The method will be developed further to provide better measurements.

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Table 1. Velocity (km/h), mean tibia internal rotation (deg), standard deviation (SD), intraclass correlation (ICC) and intrasubject standard deviations (Sw) with and without shoes for four different speeds (n=4).

	Without shoes					With shoes				
Vel	Mean	SD	ICC(1.1)	ICC(1.5)	Sw	Mean	SD	ICC(1.1)	ICC(1.5)	Sw
2.5	8.6	3.2	0.70	0.92	2.3	11.3	1.5	0.08	0.29	3.2
4.0	14.1	2.3	0.41	0.78	2.8	12.4	3.7	0.75	0.94	2.5
6.0	14.7	4.5	0.67	0.91	3.5	14.4	5.9	0.94	0.99	1.7
9.0	11.7	3.3	0.83	0.96	1.7	8.7	4.4	0.90	0.98	1.7

Is a neutral foot posture optimal for runners: A comparative study of different foot postures on injury survival.

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INTRODUCTION

Prevention of Running Related Injuries (RRIs) is necessary to ensure adherence to running. In the clinical setting neutral foot posture is believed to be optimal for the locomotive apparatus. This is supported by expert opinions from the 1980s (1). As a consequence, pronated foot posture is still to be particularly associated with increased risk of RRIs. Therefore, "motion control" shoes and insoles are commonly used in the clinical setting as intervention to prevent running injuries (2). Despite this conviction of clinicians, runners, and shoe manufacturers, we found no clear evidence published showing a higher incidence of RRIs in persons with pronated foot postures compared to neutrals. The purpose of this study was to compare the injury survival between runners with a neutral foot posture compared to runners with a pronated foot posture.

METHODS

A seven week follow-up study was conducted. 40 healthy, recreational runners were included (24m/16f, 45.8 ±8.9 years, BMI 24.7 ±2.1). The main outcome measure was RRI. An RRI was defined as any musculoskeletal complaint in the lower extremity or back causing a restriction of running for at least one week. Foot posture was measured with Foot Posture Index (FPI)(3). Based on FPI participants were divided into two exposure groups: Neutrals (FPI=0 to 5) or Pronators (FPI 5 to 12). The RRI proportion as a function of follow-up time was estimated in each group using the Kaplan-Meier curve.

RESULTS

FPI assessment divided the participants into 24 pronators and 16 neutrals. Three persons with pronated foot posture and five with neutral foot posture sustained an injury during the follow-up period (figure 1). No significant difference in injury survival was observed between persons with a neutral foot posture and pronated foot posture ($p=0.26$).

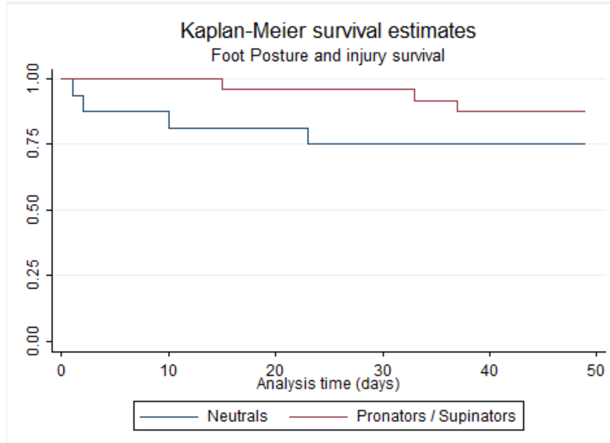

Figure 1: Foot posture and injury survival. Blue curve represents the neutrals and the red curve the pronators.

DISCUSSION

No significant difference in injury survival was observed between runners with pronated and runners with neutral foot posture. This result should be interpreted with care as the sample size was too small to include the influence of confounding factors into the analysis. Future studies should quantify foot posture dynamically and control for various confounders, especially running shoes and insoles.

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ACKNOWLEDGEMENTS

Foot and Ankle Research Northern Denmark (FARND)

The Effect of Invertor/Evertor and Plantar-/Dorsiflexor Fatigue on Plantar Pressure Distribution

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INTRODUCTION

Altered loading patterns of the foot, due to fatigue and thus unattenuated impact loading, is an aetiological factor in various running injuries [1, 2]. Treadmill running at anaerobic threshold speeds induces sufficient muscular fatigue to increase forefoot pressure [3]. This transfer of load to the forefoot reflects a diminished capacity of the fatigued muscles to stabilise and control the foot [3, 4]. Fatiguing treadmill runs however do not identify which plantar pressure distribution changes are direct results of localised muscular fatigue. The purpose of the present study was therefore to determine and compare the effect of selectively induced fatigue of the shank muscles on plantar pressure distribution.

METHODS

Twelve male participants were assessed on the RS Footscan system prior too and following dominant leg concentric isokinetic ankle invertor/evertor (InEv-F) and plantar-/dorsiflexor (PD-F) fatiguing. Maximum pressure (kPa) for ten plantar regions, of the dominant leg, was measured. Differences between the data were evaluated with a repeated-measures ANOVA with Tukey HSD tests for post hoc analysis ($p<0.05$).

RESULTS

Table 1 displays pre- and post-fatigue maximum pressure data according to the plantar regions.

DISCUSSION

Significantly less pressure was observed at T1, HM and HL due to fatigue. Fatigue is associated with increased forefoot pressure and decreased toe and heel pressure [2, 5], because dorsiflexor muscle fatigue especially is associated with increased impact loading [1]. Fatigue of tibialis anterior and posterior (primary foot dorsiflexor and evertor respectively) achieved during both fatigue protocols attributed to the observed similarities following fatigue compared to pre-fatigue. Tibialis anterior and posterior fatigue furthermore attributed to significantly decreased heel pressure due to decreased dorsiflexion. Significantly less pressure, due to InEv-F, was measured at M1, M2 and M3 compared to PD-F. Invertor fatigue has a significant effect on loading rate and ankle joint motion [1]. Significantly altered pressure distributions at the forefoot, mid-foot and heel due to InEv-F, compared to PD-F, could be related to the compound effect of impaired inversion/eversion control, shock-wave attenuation at heel strike and decreased muscular power during the loading phase [5, 6].

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Table 1: Maximum Pressure (kPa) Before and After Fatigue

	T1	T2-5	M1	M2	M3	M4	M5	MF	HM	HL
Pre-F	92±26	18±8	109±50	180±38	193±34	129±28	58±28	21±9	189±33	165±31
InEv-F	71±28*	12±6*	85±38*	157±35*	178±29	118±27	53±29	18±8	160±21*	133±22*
PD-F	78±22*	14±8	105±43†	178±44†	204±43†	136±32†	56±26	22±9†	176±28*†	145±33*†

Pre-F - Pre-fatigue; InEv-F - Invertor/evertor fatigue; PD-F - Plantar-/dorsiflexor fatigue; T1 - hallux; T2-5 - toes 2-5; M1 - metatarsal 1; M2 - metatarsal 2; M3 - metatarsal 3; M4 - metatarsal 4; M5 - metatarsal 5; MF - midfoot; HM - heel medial; HL - heel lateral

* Statistically significant difference between pre-fatigue and invertor/evertor and/or plantar-/dorsiflexor fatigue ($p<0.05$)

† Statistically significant difference between invertor/evertor fatigue and plantar-/dorsiflexor fatigue ($p<0.05$)

Soleus T-reflex amplitude modulation when standing humans adopt a challenging stance

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INTRODUCTION

While it is known that soleus Hoffmann reflex amplitude decreases when a person performs a difficult standing task requiring increased challenge in ankle position control [1], far less is known about how the more physiologic soleus stretch reflex is modulated during such motor tasks. Available stretch reflex studies utilize only small increases in balance challenge, such as hands-free standing [2,3]. This study's objective was to determine if the amplitude of the sort latency EMG response to soleus muscle stretch (T-reflex) is modulated when a person changes from a stable stance to a challenging stance.

METHODS

Nineteen young adult subjects stood in a stable stance (feet apart, left foot heel in line with toes of right foot, hands on the top of support boxes), and a tandem stance (left foot directly in front of right foot, arms at sides). Ten subjects needed to place 2-3 fingers of both hands gently against the sides of the support boxes to perform the tandem stance without a high level of soleus and TA co-contraction. A mechanical tendon tapper struck the right Achilles tendon when background (b) soleus EMG levels were similar between the two tasks. Soleus T-reflex amplitude was compared across the two stances within each subject at a similar level of motoneuron excitability (soleus bEMG) and TA bEMG. Measures in the tandem stance were expressed as a percentage change from the measures in the stable stance in the same individual, then averaged across subjects.

RESULTS

Most subjects (14 of 19) decreased mean T-reflex amplitude when in the tandem stance position, compared to stable stance (mean(s)), -13.7%(31%) (Figure 1). All 9 subjects performing the tandem stance without touching the support boxes decreased T-reflex amplitude in the tandem stance, -31%(20%), p=0.002, Cohen's d=1.5. Of the 10 subjects who lightly touched the support boxes while in tandem stance, 5 decreased & 5 increased reflex amplitude, overall: 2.0%(32%). For all subjects, and for the subgroups of those that could, or could not perform the tandem stance without touching support boxes, there was no significant change in soleus or TA bEMG levels.

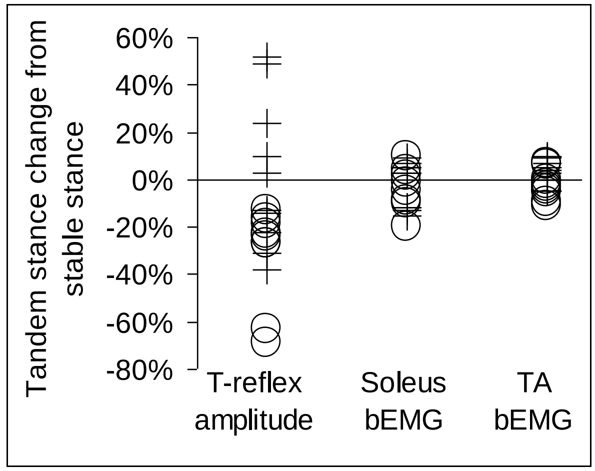

Figure 1: Changes in T-reflex amplitude, soleus & TA bEMG when subjects performed a tandem stance, compared to stable stance. Some subjects performed tandem stance with hands at sides (circles) while others lightly touched the sides of support boxes (crosses).

DISCUSSION

The decrease in T-reflex amplitude was large (Cohen's d) and significant for subjects performing the tandem stance without finger touch. This reduction is consistent with the idea that during a challenging motor task, Ia afferent mediated reflex size may be reduced to prevent the Ia signal from being strong enough to excessively activate spinal motor neurons [4]. All of the subjects that increased T-reflex amplitude, compared to stable stance, were ones that lightly touched the boxes. Haptic stabilization by finger touch diminishes the reduction in H-reflex amplitude seen when standing, or standing on one foot [1].

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Investigation of spread of foot deformity by using digital podoscope device

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INTRODUCTION

Measuring rate of spread of four main deformities of sole (hallux valgus , pes pelanus , pes cavus) is considered in this paper that are gained by analyzing images from digital podoscope device.

Deformities of sole have an important function in human health and forces on limbs and spinal column.

METHODS

In this research 83 women , 32 men in range of 19-23 year were chosen and images were gained in two form : free and bearing weight . Free mode is form of sitting in chair (that sole was on the page of image recorder of podoscope). Bearing weight mode is form of standing on the podoscope device.

Images were processed with designed code in MATLAB software and then result of these images where shown in excel. In sols ,24 points ,12 points corresponding to right foot and 12 points corresponding to left foot are important for us that by using this program, points position , points length , width and important angles are known (Figure 1). As a result deformities of sole are found.

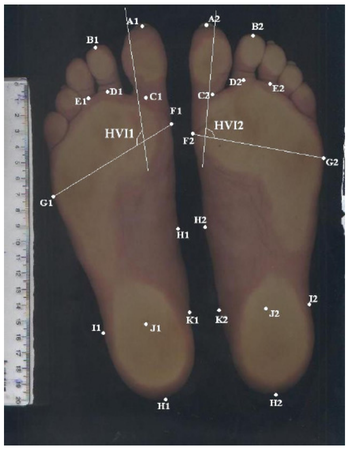

Figure1: As illustrated in figure, 24 points that are important in sole and also Halux Valgus index are shown.

RESULTS

Results show that 17% of these persons are involved with pespelanus, 30% of this persons are involved with pes cavus and 10% of this persons are involved with hallux valgus and other persons in this exprement have normal sole.

DISCUSSION

In this research, analyzing current disease of sole and finally diagnosis of this kind of deformity by using digital podoscope device were considered.

Ankle injuries and their impact in the community

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INTRODUCTION

Despite many people injuring their ankles, little is known about the long-term prevalence or impact of an ankle injury in everyday life. Therefore, the purpose of the study was to conduct a pilot survey of the prevalence and impact of ankle problems in the community.

METHODS

Randomised computer-assisted telephone interviews were undertaken of 751 people aged 18-64 years living in the community in Australia. The interview consisted of questions about past ankle injury and chronic ankle problems from any cause. Impact was assessed by; type and duration of symptoms, physical activity limitations, and healthcare utilisation.

RESULTS

The average age of the participants was 46.0 ±12.5 yrs (mean ± SD) and the ratio of males to females was 1:2. These who had a chronic ankle problem had similar characteristics. Almost 70% of the people interviewed had injured, or experienced problems with, their ankles (Table 1). Of the people who reported a history of ankle injury with no residual problems, the most common injury was ankle sprain (72%) followed by muscle strain (15%). Chronic ankle problems were reported by 178 people, 121 of which were due to an injury. Chronic musculoskeletal problems were most commonly due to ankle sprain (n = 63), arthritis (n = 30), and fracture (n = 26). The most commonly reported symptoms were pain (76%), weakness (71%) and instability (61%), with half the participants reporting symptoms that had persisted for more than 10 years. Physical activity had been modified or limited in 96 people. The activities most commonly limited were walking (23%) and sport (15%). These two activities were also the most commonly nominated as the physical activity that people could not do, but wished to. Most people (n = 84) had sought no help within the last year. Interventions that people used to reduce symptoms were heat (29%), tape or brace (28%) and medication (21%).

Table 1: Prevalence of ankle problems

	Overall	%
Never injured and no problems	231	30.8
Ankle injury - no problems	342	45.5
Chronic ankle problems	178	23.7
-non musculoskeletal ankle problems	31	4.1
-chronic musculoskeletal ankle problems	147	19.6

DISCUSSION

This is the first study to undertake a population based examination of ankle injury and chronic ankle problems. The prevalence of ankle injury is high and chronic ankle problems are common with the most frequent cause being a previous ankle sprain. The majority of people with a chronic ankle problem reported physical symptoms and activity limitations. Little is known about risk factors for the development of long-term musculoskeletal ankle problems particularly following the most common injury, an ankle sprain. Future research should be directed towards determining predictors of chronic ankle problems.

An advanced kinematic model for enhanced calculation of foot bone kinematics.

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INTRODUCTION

In depth understanding of the articulations of the tarsus and midtarsus with adjacent bones is of great importance in orthopaedics and biomechanics. However, accurate kinematic models describing movement coupling at the level of the tarsus and midtarsus are still rare in the biomechanics field. Models consisting of combinations of simple hinge joints to represent tarsal and midtarsal joints are often used to reconstruct bone kinematics from measured marker movement. However, their validity is limited[1]. This study reports on the performance of an advanced kinematic model that describes the joint coupling between segments of the foot. This model therefore has the potential to increase the accuracy of estimated kinematics of the foot.

METHODS

The advanced kinematic foot model consists of 7 segments (tibia/fibula, talus, calcaneus, midfoot, metatarsals, hallux and toes), 8 degrees of freedom (DOF) and 25 ligament fibres. The Lisfranc joint, the hallux-metatarsal joint and the phalangeal-metatarsal joints were modelled as simple hinge joints with one DOF. The kinematic coupling at the joints connecting the hindfoot segments is modelled based on previously applied principles to model the ankle[2] and knee[3,4] kinematics, by implementing joint contact and ligamentous constraints. Geometrical parameters for these joint models are calculated from CT-data of the foot.
As part of the experimental validation, hindfoot motion was measured in vitro during the stance phase using a gait simulator. This allows accurate measurement of 3D bone motion for comparison with the modelling results. Therefore, the depending kinematic variables (Table 1) for the hindfoot are calculated for the DOF values that were measured by optimization of the total ligament strain energy, and were then compared with the measured values of these variables. To improve the correspondence between the model calculation and measurement, ligament properties were manually fine tuned.

RESULTS

The foot kinematics calculated by the model show acceptable correspondence to the in vitro measurements (Table 1).

DISCUSSION

Better correspondence between model and measurement is to be expected after bounded optimization of the ligament properties. However, in its current form, the model has strong potential for improved estimation of foot kinematics.

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Table 1: Coefficient of multiple determination[5] (CMD) (right number) and average difference (left number) between the average of ten in vitro simulations of the hindfoot kinematics during the stance phase of gait, and the model simulation thereof. Underlined numbers are CMD and average difference for DOF values in the model, other values are values of CMD and average difference for the depending kinematic variables.

	Tal vs Tib	Calc vs Tal	Midf vs Tal	Cub vs Calc	Cub vs Nav
POST/ANT [mm]	0.12 / 0.83	5.99 / 0.95	3.24 / 0.94	1.48 / 0.94	3.03 / 0.97
DIST/PROX [mm]	1.48 / 0.71	1.04 / 0.21	1.87 / 0.93	0.45 / 0.27	0.40 / 0.01
LAT/MED [mm]	2.99 / 0.91	2.35 / 0.06	3.85 / 0.99	3.45 / 0.87	3.27 / 0.97
EV/INV [°]	1.59 / 0.46	*0.00 / 1.00*	2.16 / 0.30	21.07 / 0.96	8.45 / 0.95
ABD/ADD [°]	0.57 / 0.55	*0.00 / 1.00*	2.67 / 0.54	14.07 / 0.90	4.67 / 0.94
DF/PF [°]	*0.00 / 1.00*	*0.00 / 1.00*	*0.00 / 1.00*	1.38 / 0.18	9.32 / 0.99

A closed-chain 3D finite element foot-ankle model - biomechanical perspective on forefoot plantar stress redistribution following Tendo-Achilles Lengthening

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INTRODUCTION

The closed-chain effects in normal foot, first described by Hicks as the Windlass mechanism, is the synergic work of the Achilles tendon, plantar fascia and various internal articulating joints, which can render the foot a mechanical-advantageous lever during the late-stance of gait. Certain foot pathologies, such as neuropathic diabetes, can result in thickening and stiffening of Achilles tendon and plantar fascia, which may alter the normal closed-chain foot mechanics (i.e., the occurrence of an early Windlass mechanism), with a consequence of plantar stresses being elevated in the forefoot where ulcers are most common[1]. Clinically, Tendo-Achilles Lengthening (TAL) procedure has been implemented in an attempt to counteract this altered Windlass effect. However, its efficacy on forefoot stress-relieving is still controversial.

METHODS

A 3D, closed-chain, finite element foot-ankle model has been developed. The baseline model consists of 30 bony parts inter-articulated via internal joints. The forefoot skeleton was embedded into a continuum of the soft tissue which was modelled as a hyper-elastic material. Detailed 3D structures of the plantar fascia and Achilles tendon were modelled and they were assumed to behave as an isotropic, linear elastic material. The ligaments and extrinsic/intrinsic foot flexor muscles were represented as “spring” element and active force-actuated “connector” elements respectively. Forefoot kinematic data were used to assist in replicating foot pose reflecting push-off. Muscle contraction forces were applied to the Triceps surae muscle via Achilles tendon and other major extrinsic muscle groups through long flexor tendons to produce ankle joint plantar flexion and Metatarsophalangeal (MTP) dorsiflexion at push-off (Fig. 1).

In order to simulate the triceps surae muscle performance following TAL, a series of clinical model were constructed by varying the length-tension relationships of triceps surae muscle associated with Achilles tendon of the baseline model, by adjusting the following two parameters: Achilles tendon modules of elasticity (ATE) and Achilles tendon forces (ATF). Forefoot plantar stress distributions, ankle plantar-flexion/dorsiflexion and MTP dorsiflexion range of motion (ROM) (i.e., closed-chain effect due to windlass mechanism) were compared with reduced ATE (816, 612, 408 and 204 MPa) and ATF (100%, 90%, 80%, 70% and 60%).

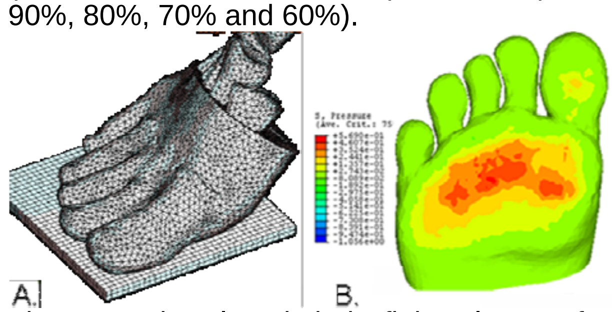

Figure 1. The closed-chain finite element foot model (baseline case) under large deformation (A) with predicted planar pressure distribution (B) during push-off.

RESULTS

With full ATF being applied, reduction in ATE only showed minimal effects on the MTP ROM and forefoot peak plantar pressure. In contrast, a local elongation of the Achilles tendon up to 48.1% was observed due to incremental reductions in ATE. Secondly, a reduction in ATF seems to have a more significant impact on peak forefoot pressure and MTP ROM. With normal ATE being maintained, a 30% decrease in ATF causes a reduction in peak plantar metatarsal pressure up to 18.4%.

DISCUSSION

Current muscle-driven, 3D FE foot model successfully simulated a closed-chain effect, indicating that forefoot plantar pressure peak and its distribution could be sensitively manipulated by changing the tension generation capacity of triceps surae muscle, but not the Achilles tendon elasticity [2].

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A novel approach using a FE-Foot Model for clinical applications

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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 were done. A modified FE Modell from the Website ww.ulb.ac.be/project/vakhum/ which is based on CT Scan was used. Contacts between bones were considered as bonded because 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 contact pair. 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 the force plate were used as boundary conditions. Even though the muscle forces we computed with the ANYBODY model.

Results

The results of the reaction moment in the subtalar joint, as an example, is shown in figure 1. The red lines indicates the mean and two convidence intervals of 10 healthy subjects (10 trials of each subject) in a period of one stance phase. The blue line shows the reaction moment of the subtalar joint in a patient suffering on an instability of the ankle joint.

Simulating a lateral slide of the calcaneus in our FE Model we see a significant decrease of this reaction moment in the subtalar joint. This indicates that a lateral slide in this case could reduce the reaction moment in x direction (means in the direction of Gait) significantly and could be a possible solution for surgical treatment.

Fig. 1: Reaction Moment in x before virtuel surgery x

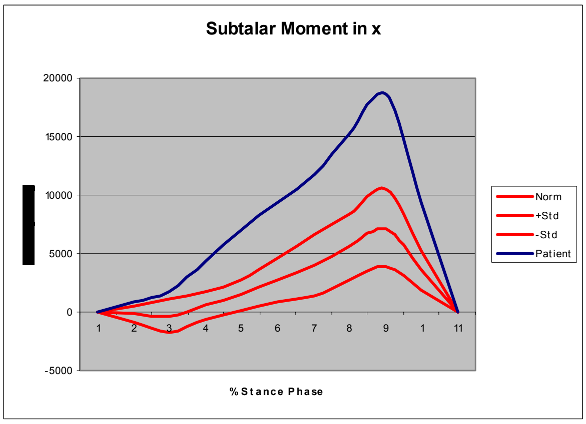

Fig. 1: Reaction Moment in x after virtuel surgery

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. There are further investigations necessary, especially to validate the used FE- and ANYBODY-Model.

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A novel approach using a FE-Foot Model for clinical applications

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Simulating a lateral slide of the calcaneus in our FE Model we see a significant decrease of this reaction moment in the subtalar joint. This indicates that a lateral slide in this case could reduce the reaction moment in x direction (means in the direction of Gait) significantly and could be a possible solution for surgical treatment.

Fig. 1: Reaction Moment in x before virtuel surgery x

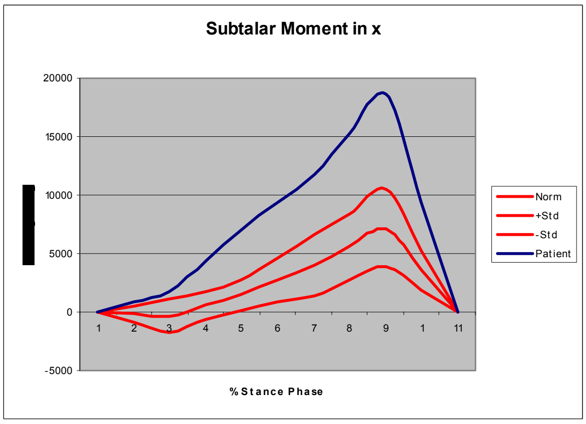

Fig. 1: Reaction Moment in x after virtuel surgery

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. There are further investigations necessary, especially to validate the used FE- and ANYBODY-Model.

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A Comparison of the Performance of Hexahedral and Tetrahedral Elements in Finite Element Models of the Foot

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INTRODUCTION
Characterization of foot-ground or foot-shoe contact stresses provides significant insight into the biomechanics of the normal and pathological foot. The finite element (FE) method is widely used in foot biomechanics for predictive simulations of plantar pressures in barefoot and shod conditions [1]. The objective of the present study was to evaluate various types of FE meshes that can be used to model the interaction of a bone-soft tissue construct in contact with the rigid surface under compressive and shear loading conditions in a heel-pad analog model.

METHODS
A simplified geometric representation of a human heel pad comprised of bone and soft tissue was used in the present study (a hollow sphere of inner and outer diameters of 20 mm and 30 mm respectively). The bone and floor were meshed using 2D shell elements (triangular and quadrilateral) and were modeled as rigid bodies. Soft tissue was meshed using 3D continuum tetrahedral (linear and quadratic) and hexahedral (linear) elements and was modeled as an incompressible hyperelastic material (Ogden material [2]). A mesh convergence study was performed to ensure accuracy of the solution. Tied contact was defined between the bone and the soft tissue while surface-to-surface contact was defined between the soft tissue and the floor with a frictional coefficient of 0.3. Two loading conditions: 1) compressive load (300N) and 2) compressive load (700N) combined with shear force (100N) were applied to the bone. In all simulations, the influence of the mesh type on the contact pressure predictions between the soft tissue and the rigid floor as well as the solution time were assessed. Abaqus 6.10 beta [2] was used for FE analysis.

RESULTS
The peak pressure values and CPU times under given loading conditions for various FE meshes are reported in Table 1. Plantar pressure distributions for loading condition 2 are shown in Figure 1.

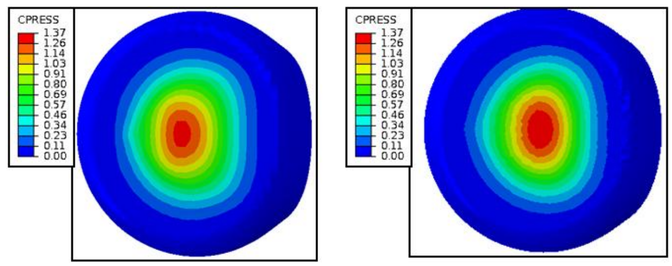

Figure 1: (a) Pressure distribution in hexahedral mesh as a result of compressive load (700N) combined with shear force (100N) and contact with friction (0.3) at full incompressibility. (b) Corresponding pressure distribution in quadratic tetrahedral mesh.

DISCUSSION
Given the large amount of time required for meshing complex anatomical structures using hexahedral elements, use of quadratic tetrahedral meshes appears to be a feasible alternative even in the setting of large-deformation hyperelastic contact. It should be noted that other FE solvers may have varying formulations to accommodate material and geometric nonlinearities, and results from a test problem such as those presented here should always be examined.

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Table 1: Influence of mesh type on peak pressure prediction and CPU times

Mesh type	μ	ν	Load	Contact Pressure (KPa)	CPU Time(Sec)
Hexahedral	0.3	0.5	300 (C)	440.4	13779
Tetrahedral (Quadratic)	0.3	0.5	300(C)	450.4	19459
Hexahedral	0.3	0.5	700(C) + 100(S)	1303	64444
Tetrahedral (Quadratic)	0.3	0.5	700(C) + 100(S)	1371	64390

Medial Ankle Instability Assessment Using Heel Varus in an External Rotation Stress Test

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INTRODUCTION

The presence of joint asymmetry with medial clear space widening on an AP/oblique radiograph (i.e., a positive external rotation stress (ERS) test) has long been regarded as definitive for syndesmotic and medial ankle ligamentous injury in rotational ankle fractures. [1,2,3] Typically, the external rotation stress in this test is applied to the foot held in valgus (and slightly dorsiflexed). In this manner, tension of the superficial (tibio-calcaneal) ligament fibers may mask presence of deep (tibio-talar) ligament fibers. This study was designed to define the effect of hind foot positioning on the ERS test.

METHODS

Six cadaveric legs without ligament injury were subjected to ERS testing, with either hindfoot valgus or varus positioning. Metallic bead markers were inserted into the tibia, fibula, and talus. Specimens were fixed in a locked Taylor spatial frame in neutral. CT scans were obtained. The frames were then unlocked, and stress was applied for (1) ERS with heel valgus, and then (2) ERS with heel varus. For each test, the foot was manually forced to the hard endpoint of rotation, and then the frame was locked to hold that position. Additional CT scans were acquired for the stressed specimen. The deep deltoid (DD) and syndesmotic ligaments were then in turn sectioned, and the same testing and CT scans were repeated.

The distances between bone markers identified on CT were measured to quantify 3D displacement changes for each ligament transection, between ERS with heel valgus and with heel varus, and compared with the ankle neutral position.

RESULTS

Following DD ligament transection, the medial gutter space was widened in ERS with heel varus, as compared to with heel valgus (p=0.005)(Figure 1). After DD combined with syndesmotic ligaments transection, the medial gutter space was significantly further increased from 3.97 mm in ERS with heel valgus, to 9.06 mm in ERS with heel varus (p=0.015). The medial-lateral separation of the tibio-fibular space was slightly increased in both hindfoot displacements with ERS, while the AP separation was significantly increased with heel varus, as compared to with heel valgus (p=0.044).

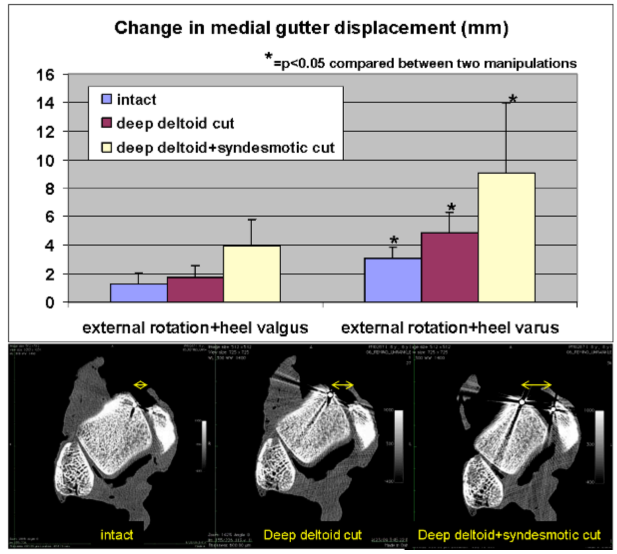

Figure 1. Change in medial gutter displacement following each manipulation.

DISCUSSION

The results showed greater changes in tibiotalar and tibiofibular displacement when coupling ERS with heel varus, as compared to with heel valgus. This finding suggests that heel varus should be coupled with ERS to increase the likelihood of identifying syndesmotic and medial ankle ligamentous injury in rotational ankle fractures

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Biomechanical study of combination of fixing the posterior malleolus and mending anteroinferior tibiofibular ligament to reconstruct syndesmotic stability

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INTRODUCTION

Syndesmotic complex include anterior-inferior tibiofibular ligament, the interosseous ligament, the interosseous membrane and the posterior syndesmotic ligament.In pronation-external rotation type (PER)Stage 4 ankle injuries syndesmotic injuries are severely and often with a posterior malleolar fragment and anterior-inferior tibiofibular ligament rupture . Objective ：To evaluate the biomechanical characteristics of combination of fixing the posterior malleolus and mending anteroinferior tibiofibular ligament to reconstruct syndesmotic stability.

METHODS

A PER Stage 4 fracture pattern with a posterior malleolar fragment was created in 24 fresh lower extremity cadaver specimens. And the specimens were divided into three groups randomly and each specimen was mounted on an MTS Bionix 858 test system to test the stability to external rotation[1]. In A group internal fixation of lower tibiofibular ligament union was performed by screws; Fixation of posterior malleolar fractures was performed in B group; Combination of fixing the posterior malleolus and mending anteroinferior tibiofibular ligament was performed in C group. Then statistical analysis was carried out carefully.

RESULTS

B group was superior to A group, there was a difference between groups, $P < 0.05$. C group was superior to A group, and there was a difference between groups also, $P < 0.05$. While between B and C group there was no significant.

DISCUSSION

To PER Stage 4 the traditional treatment has been trans-syndesmotic stabilization. However,syndes-motic screws are not without occasional inherent morbidity. Gardner MJ had confirmed that internal fixation of the posterior malleolus could provide better stability than using syndesmotic screw in his biomechanics research[1].In this study, though the difference between B and C groups showed no statistical significance, the operation in C group can provide better stability which is closer to the stability of anatomical reconstruction. Besides, this operation is relatively simple and can bring patients minor trauma. Therefore, the authors consider that it is a feasible method to repair the wound of lower tibiofibular ligament union.

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Table 1: Stiffness Date After Fixation

	A Group			B Group			C Group		
	Intact Stiffness	Stiffness after Fixation	Decrease	Intact Stiffness	Stiffness after Fixation	Decrease	Intact Stiffness	Stiffness after Fixation	Decrease
1	0.482	0.344	28.6%	0.442	0.345	21.9%	0.708	0.509	18.1%
2	0.703	0.152	78.4%	0.493	0.304	38.3%	0.478	0.197	58.7%
3	0.512	0.295	42.4%	0.758	0.327	56.9%	0.515	0.416	19.2%
4	0.540	0.201	62.8%	0.398	0.238	40.2%	0.308	0.175	43.2%
5	0.452	0.294	34.9%	0.374	0.295	21.2%	0.467	0.299	35.9%
6	0.643	0.309	52.0%	0.524	0.348	33.6%	0.594	0.461	22.4%
7	0.312	0.171	45.3%	0.289	0.197	31.8%	0.349	0.178	49.0%
8	0.465	0.206	55.6%	0.430	0.243	43.5%	0.454	0.298	34.4%
	Average		50.0%			35.9%			35.1%
	Stndard deviation		15.9%			11.7%			14.7%

＊Stiffness values are reported in Nm/°

Kinematics of Adult Acquired Flatfoot and Correction of Adult Acquired Flatfoot with Talonavicular Joint Fusion

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INTRODUCTION

Adult acquired flatfoot (AAF) is a common pathology presenting to a foot and ankle physician. Abnormal motions of the subtalar and talonavicular joints have been demonstrated in the kinematic studies that have been performed [1,2]. Therefore, fusing the talonavicular joint has become a surgical procedure for treatment of the abnormal kinematics associated with AAF [3-5]. The purpose of this study was to examine the change in height of the navicular and the angle between the calcaneus and the 1st metatarsal during normal cadaveric walking, simulated AAF walking and simulated AFF walking with talonavicular fusion.

METHODS

Seven fresh frozen midtibial amputated cadaveric specimens were placed in a dynamic gait simulator. K-wires with 3 markers were driven into the tibia, talus, calcaneus, cuboid, navicular, medial cuneiform, 1st metatarsal and 5th metatarsal. Kinematic data were collected using an 8 camera Vicon motion analysis system at 160 Hz. Each specimen was walked for 10 trials during 3 conditions (normal, AAF and talonavicular joint fusion). AAF was simulated by detaching the posterior tibial tendon from the simulator and surgically releasing the plantar fascia and the entire spring ligament. Navicular displacement and calcaneal - 1st metatarsal angular displacement were calculated between heel contact and maximum values. Effect sizes (ES) were calculated between conditions.

RESULTS

The vertical navicular displacement averaged 2.2±1.1 cm from contact to maximum displacement during normal walking. This increased to 3.3±0.8 cm during simulated AAF (ES=.50). After talonavicular fusion the navicular displacement was reduced to 2.5±0.7 cm (ES=.74). The change in angle between the calcaneus and the first metatarsal was 10.3±2.4^{o} during normal walking. This increased to 12.2±2.9^{o} during simulated AAF (ES=.76). After talonavicular fusion the range of motion decreased to 11.3±3.7^{o} (ES=.35).

DISCUSSION

The simulated AAF model in the study created a significant decrease in navicular height as well as a significant decrease in the calcaneal-1st metatarsal angle associated with the collapse of the medial longitudinal arch as has been previously observed [1,2]. Fusion of the talonavicular joint was noted to significantly restore the integrity of the medial longitudinal arch.

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RF Score as a Reliable Measurement to Assess Improved Dynamic Plantar Loading Improving Post Foot Reconstruction Surgery

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Introduction

Ulceration risk associated with anatomical deformities is generally assessed by measuring plantar pressure magnitude (PPM). However, as PPM is partially dependent on gait speed and treatment interventions may impact speed, the use of PPM to validate treatment is not ideal. This study suggests a novel assessment protocol, which is speed independent and can objectively 1) characterize abnormality in dynamic plantar loading in patients with Charcot foot and 2) screen improvement in dynamic plantar loading after foot reconstruction surgery.

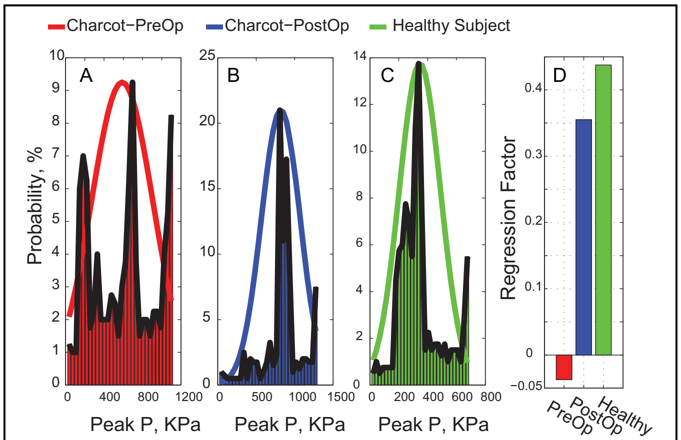

Figure 1: RF value represents self-similarity between distribution of PPM with a normal histogram.

Methods

To examine whether the plantar pressure distribution measured using EMED platform, was normal, we proposed a novel score named regression factor (RF), which represents the similarity of the actual pressure distribution with a normal distribution[1]. RF values may range from negative 1 to positive 1 and as the value increases positively so does the similarity between the actual and normalized pressure distributions (Fig.1). We tested this novel score on the plantar pressure pattern of healthy subjects (N=15), while walking with different speeds as well as patients with Charcot deformity pre operation (N=6) and one Charcot patient post foot reconstruction.

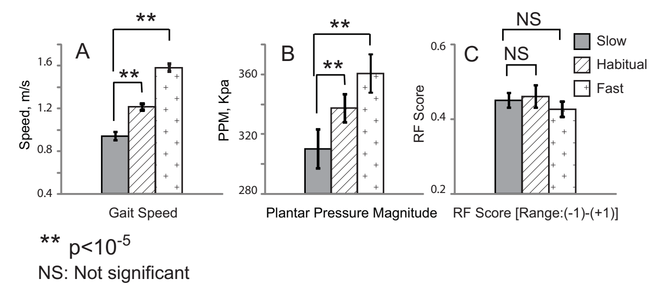

Figure 1: Plantar pressure magnitude is increased by increasing gait speed. While RF score is independent of gait speed.

Results

In healthy subjects, the RF was 0.46±0.1. When subjects increased their gait speed by 29%, PPM was increased by 8% ($p<10^{-5}$), while RF was not changed (p=0.55), suggesting that RF value is independent of gait speed (Fig 2). In preoperative Charcot patients, the RF=-0.13±0.2. However, RF increased post surgery (RF=0.42±0.12), indicating a transition to normal plantar distribution after Charcot reconstruction

Discussion

The RF appears to be independent of gait speed and offers a potentially sensitive and global interpretation of peak pressure distribution before and after foot surgery. Further larger studies are needed to observe if these relationships remain consistent. Furthermore, further testing involving both barefoot as well as in-shoe conditions may elicit a clinical outcome measure and possibly be integrated into clinical practice and telemonitoring using intelligent insoles for early recognition of Charcot neuroarthropathy.

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Different foot kinematics, kinetics and plantar pressure patterns within the gait of diabetic subjects: cluster analysis

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INTRODUCTION

Several authors found gait pattern alteration in diabetic patients both with and without diabetic neuropathy [1]. However most of these studyes were based on the assumption that only one gait pattern exists within the diabetic subjects. If different patterns, or gait strategies, exist and if the different styles are analysed together in a single group, statistical errors will result. So far the aim of this study was to determine if diabetic subjects displayed other than a typical gait pattern, and to identify which parameters best describe them using three-dimensional (3D) gait data. The methodology adopted herin was *k-means* cluster analysis.

METHODS

Simultaneous kinematics, kinetics and plantar pressure (PP) analysis were performed on 35 subjects (14 normal (C), 9 diabetics (D), 12 neuropathics (DN)). Mean age and BMI were respectively 59.9 ± 5.1 years in C, 62.9 ± 6.3 years in N, 70.2 ± 6.3 years in D and 24.6 ± 2.8 kg/m^2 in C, 25.6 ± 3.4 kg/m^2 in N, 27.0 ± 1.6 kg/m^2 in D. Six cameras BTS motion capture system (60-120 Hz) synchronized with 2 Bertec force plates and 2 PP plates (Imago) were used to collect the data. A 3D multisegment foot protocol [2] was applied and gait analysis performed: 3D subsegments ground reaction forces (GRFs), PP, contact areas (CA) and joint rotation angles were evaluated. *K-means* cluster analysis (MatlabR2008b) was used to classify the samples into 2 or 3 clusters [3]. The data of the 3 groups of subjects were used and each parameter at the time was considered. Standard Euclidean distance was chosen in defining the clusters' centroid. The final clustering quality was evaluated using the Silhouette. Selected clusters were then analysed in terms of biomechanics and clinical parameters.

RESULTS

When considering the kinematic data, cluster analysis led to definition of 2 well separated clusters for the hindfoot-tibia dorsiflexion (DP) angle, the midfoot-hindfoot DP, inversion-eversion (I/E) and internal-external rotation, the forefoot-midfoot I/E. At variance, 3 well defined clusters were obtained for the DP hindfoot-tibia angle. In terms of kinetics parameters 2 well separated clusters were obtained for the following variables: each subarea AC, whole foot anteroposterior (AP) GRF, hindfoot vertical (V) GRF, midfoot V GRF and mean PP, forefoot AP and V GRFs. Finally 3 well separately clusters were defined by whole foot both mean PP and AC, hindfoot mean PP, midfoot both V GRF and mean PP, forefoot peack PP. For instance the clinical characteristics of the 3 clusters defined by hindfoot mean PP were reported in Table 1.

DISCUSSION

K-means showed the presence of more than one gait pattern for each biomechanics variable. Indeed none of the parameters considered in the present study separated completely the C from the D or the DN group. Impressively some of the clusters grouped subjects with other similar clinical characteristics rather than D or DN. For instance the cluster reported in Table 1 grouped subjects according to type foot. However, further analysis is needed including larger samples of subjects.

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Table 1: Clinical characteristics of the subjects in the 3 clusters defined by hindfoot mean PP (% of the total number of subject of the cluster). Toes deformities includes hammer/clawed/overlapped toes and hallux valgus.

Group	**D**	**DN**	**Foot type** Normal/Planus/Cavus	**Calcaneal position** Normal/Valgus/Varus	**Toes deformities**	**Plantar callosities**
1 (n=1)	100	100	0/0/100	0/100/0	100	100
2 (n=24)	66.4	37.5	4.2/8.3/87.5	62.5/37.5/8.3	37.5	29.2
3 (n=10)	40	20	60/20/20	50/50/0	40	40

The Biomechanical Foot Differences Between Subjects with Diabetes and Hallux Valgus Compared to Asymptomatic Non-diabetic Individuals

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INTRODUCTION

Diabetic foot complications, such as neuropathic foot ulcers and amputation,. are a serious problem. Aberrant foot biomechanics (eg. hallux vaglus) are associated with neuropathic foot ulcers[1]. The purpose of this study was to examine the pedal biomechanical function of subjects with diabetes and hallux valgus compared to healthy asymptomatic subjects with rectus, planus, and cavus feet.

METHODS

Twenty diabetic subjects with hallux valgus (D-HV), and 61 asymptomatic healthy non-diabetic subjects (22 planus, 27 rectus, and 12 cavus) were examined. First metatarsophalangeal joint (1st MTPJ) flexibility was measured using a custom-designed jig as previously described[2]. Plantar pressures during self-selected speed walking was measured with the EMED-X system (Novel Inc, St. Paul) at 100 Hz. Group comparisons were made using mixed effect ANOVA with a $p<0.05$ significance level.

RESULTS

As shown in Table 1, D-HV subjects were older, had larger Body Mass Index (BMI), and walked slower than asymptomatic healthy subjects. Malleolar Valgus Index (MVI) demonstrated that D-HV subjects had foot structure similar to asymptomatic subjects with pes planus[3]. Late 1st metatarsophalangeal (MTP) joint flexibility while standing was significantly different across foot type ($p=0.02$), the D-HV exhibiting the lowest flexibility. Dynamic peak plantar pressure (sub 1st MTP joint) was also different across foot type (Figure 1).

DISCUSSION

D-HV most closely resembled the alignment and function of healthy pes planus subjects. However, significant differences in 1st MTP joint flexibility, 1st MTP joint pressure, and walking speed were noted between D-HV and healthy planus subjects. Additional studies are needed to discern if these differences are associated with diabetes or with hallux valgus deformity.

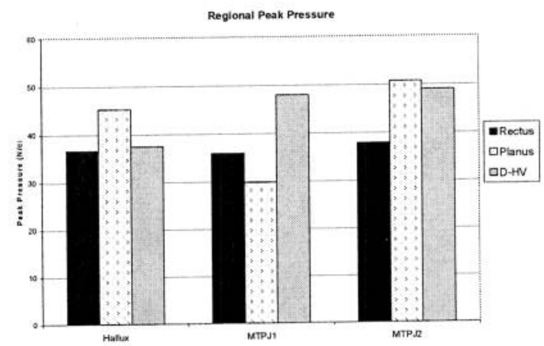

Figure 1: Peak barefoot plantar pressure demonstrated significant differences

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ACKNOWLEDGEMENTS

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Table 1: Results - Means (SE)

	Planus	Rectus	Cavus	D-HV	p-value	Post hoc*
N (Female)	22 (12)	27 (19)	12 (6)	20 (14)		
Age	35 (2.2)	34.1 (2.0)	41 (4.9)	59.9 (2.8)	0.000	4,5,6
BMI	22.2 (3.2)	24.4 (4.1)	24 (3.5)	33.2 (7.1)	0.000	4,5,6
Malleolar Valgus Index (%)	13.3 (0.8)	7.4 (0.7)	6.7(1.0)	15.3 (1.2)	0.000	1,2,4,6
1st MTPJ LateFlx Stand (°/N-cm)	27.8 (3.2)	16.1 (2.8)	19.0 (4.2)	8.2 (7.3)	0.02	2,5
Walking speed (m/s)	1.33 (0.02)	1.28 (0.02)	1.31 (0.02)	0.86 (0.02)	0.000	4,5,6
PP_MTPJ1 (N/cm^2)	29.7 (1.8)	35.9 (1.6)	33.6 (2.4)	47.9 (2.1)	0.000	4,5,6

*Bonferonni post-hoc significance set if $p < 0.0083$; 1 = Cavus vs. Planus; 2 = Rectus vs. Planus; 3 = Cavus vs. Rectus, 4 = Diabetic vs Rectus; 5 = Diabetic vs Planus; 6 = Diabetic vs Cavus

Association of Diabetic Foot Ulcer Healing and Offloading Compliance: Initial Findings

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Introduction

Offloading excessive pressure is believed to be essential to healing of diabetic foot ulcers (DFU). Previous studies indicate that patient adherence to prescribed offloading devices is low[1] and thus may result in delayed healing. However, there is little empirical research investigating this relationship. The purpose of this study is to directly assess the association between offloading compliance and DFU healing.

Methods

Subjects enrolled into a multicenter trial were provided a removable cast walker (RCW) for their neuropathic DFU. Compliance with offloading was assessed using a validated dual activity monitor method[2]. Specifically, a concealed activity monitor was attached to the RCW, and subjects were instructed to wear a second activity monitor at the hip. Over the 6 week protocol, the activity data was uploaded to a centralized server via the internet. The time stamped hip activity data was coded for compliancy using the synchronized walker activity data.

Results

Initial findings (N=16) indicate the RCW were used during 58% ±20 of subjects' activity. A positive correlation (R=0.265) was found between compliance and healing (Figure 1). This correlation was not significant, however, when wounds with an initial size <2cm^2 were excluded this correlation became significant (R=0.778 p=0.039) (Figure 2).

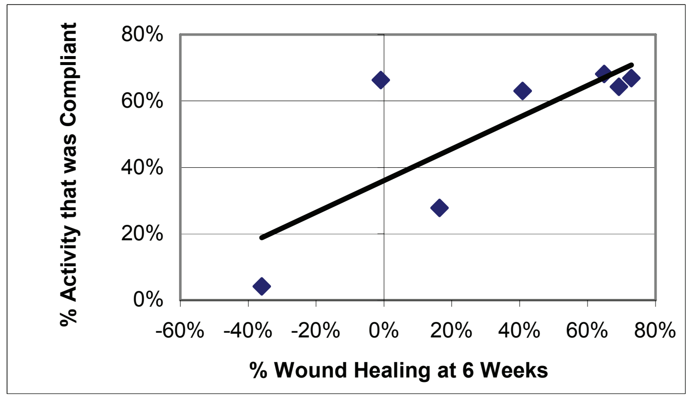

Figure 2. Wounds > 2cm^2 : Wound Healing vs Offloading Compliance

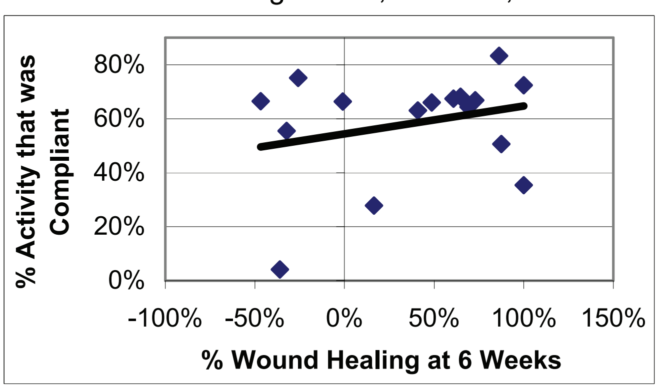

Figure 1. All wounds: RCW Compliance vs. % Wound Healing

Discussion

These initial results are the first to directly and objectively associate removable cast walker compliance with healing. The finding of divergent results for the different sized wounds was not expected and may warrant further investigation.

Acknowledgements

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Offloading the Foot by Varying Fiber Density Within Diabetic Hosiery

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Introduction

Offloading excessive pressure regions of the foot is key to both healing and preventing diabetic foot ulcers (DFU) and subsequently reducing the rate of lower extremity amputations. Much work has been done investigating offloading properties of footwear [1]. However, compliance with footwear is poor [2]. Limited work to date has focused on the offloading capacity of socks. The purpose of this study was to determine whether varying the fiber density of a diabetic sock influenced plantar foot loading.

Methods

Thirty healthy subjects walked on a treadmill at a fixed velocity of approximately 3mph in a standardized pair of athletic shoes. Four trials per subject were performed in a randomized order with each trial measuring a different sock condition. Three pairs of diabetic socks differed only in fiber density. The forth pair of socks were a standard athletic sock. Plantar pressures were measured with Pedar-X inshoe pressure insoles during 20 mid-gait strides of each trial. Data was collected at 100Hz. Ten left foot steps were averaged for each trial. Repeated measures ANOVA were used to compare the total foot and nine regional (Figure 1) peak pressures (PP) and pressure time integrals (PTI).

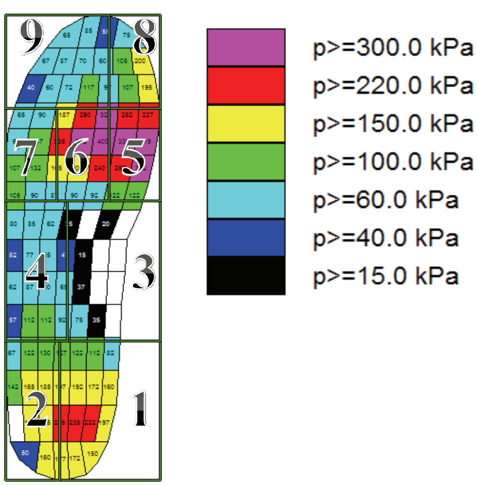

Figure 1. Masking scheme to divide foot into 9 regions.

Results

Results yielded statistically significant ($p<0.01$) differences between sock conditions in a number of regions for both peak pressure and pressure-time integral. Typically, the medium and high density socks yielded lower PP and PTI values than the low density and for some regions were also lower then the control sock.

Discussion

The results of this study suggest that altering the fiber density on the plantar surface of a sock may significantly change the peak pressure as well as the pressure-time-integral during normal gait. These results suggest that individuals at risk of diabetic foot ulceration should choose socks with medium to high densities in order to receive some offloading benefit. Socks with low densities, such as thin dress socks, should be avoided. It should be noted, however, that the reductions in PTI and PP associated with dense socks are no substitute for appropriate shoes or other offloading devices. Having conducted this initial study in healthy subjects, future studies should investigate the impact of high density socks on ulcer prevention and offloading of high risk diabetic feet.

Acknowledgements

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- Drymax Inc. donated the Drymax socks, however, had no role in the conduct or interpretation of this study.

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Histomorphological Evaluation of Diabetic and Non-diabetic Plantar Soft Tissue

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INTRODUCTION
The plantar soft tissue is subjected to repeated shear and compressive stresses, particularly in the regions of the heel, metatarsal heads and hallux [1-3]. The complex anatomic configuration of this tissue enables such forces to be endured. However, diseased tissue is often unable to adapt to and withstand these loading regimes in the same manner as non-diabetic tissue, resulting in breakdown (i.e., ulceration). Diabetes is a serious healthcare problem that can often result in significant secondary complications of the lower extremity. Diabetic foot ulceration has a complex and multi-factorial etiology and can involve changes in the pathophysiology of the plantar soft tissue. In the current study, histomorphological analyses of diabetic and non-diabetic plantar tissue were performed.

METHODS
Plantar soft tissue samples were taken from fresh cadaveric feet of older diabetic and older non-diabetic donors. Two locations of the foot (the heel and the first metatarsal) were examined, both of which have been reported to be locations with a high incidence of ulceration.

Stereological methods and quantitative morphological techniques were used to evaluate the skin thickness, interdigitation index, elastic septae thickness and adipocyte cell size.

RESULTS
The analysis demonstrated that the diabetic tissue had significantly thicker elastic septae (Figure 1) and dermis. These increases were accompanied by observations of fragmentation/fraying of the elastin fibers within the septal walls of the diabetic tissue. In addition, the septal walls of the diabetic tissue contained collagen bundles that were thicker in sections, frayed on others, more disordered and without distinct band periodicity. However, no significant difference was observed in the interdigitation index or adipocyte size.

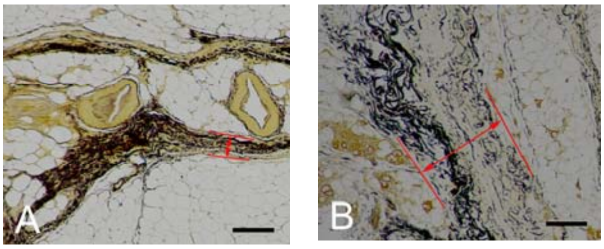

Figure 1. Non-diabetic (A) and diabetic (B) fat tissue showing elastic septae stained with Modified Heart's stain. Diabetic elastic septae were thicker (red arrow) and contained frayed elastic fibres. Scale is 200 μm.

DISCUSSION
These findings demonstrate that morphological changes can be evaluated histologically to give a better understanding of the pathological changes in the plantar soft tissue with diabetes. These evaluations can then be used to understand the biomechanical changes that occur in diabetes and the resulting susceptibility to tissue breakdown.

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Target Strain Errors in Plantar Tissue Testing

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INTRODUCTION

Accurate quantification of soft tissue properties, specifically the stress-relaxation behavior of viscoelastic tissues such as plantar tissue, requires precise testing under physiologically relevant loading. However, limitations of testing equipment often result in target strain errors that can contribute to large stress errors and confound comparative results to an unknown extent. Previous investigations have modeled this artifact [1, 2, 3] but they have been unable to obtain empirical data to validate their models. Moreover, there are no studies that address this issue for plantar tissue. The purpose of this research was to directly measure the difference in peak force for a series of small target strain errors within the range of our typical stress-relaxation experiments for plantar soft tissue.

METHODS

Five plantar tissue specimens were tested to seven incremental target strain errors of -0.9%, -0.6%, -0.3%, 0.0%, 0.3%, 0.6%, and 0.9%, so as to undershoot and overshoot the target displacement in 0.3% increments. The imposed strain errors were accurately attained using a special compensation feature of our materials testing software that can drive the actuator to within 0% (1- 2μm) of the target level for cyclic tests. Since stress relaxation tests are not cyclic, we emulated the ramp portion of our relaxation tests with 5Hz triangle waves. Stress variation was calculated according to (1):

$$\text{stress variation (\%)} = \frac{|\sigma_{max} - \sigma_{min}|}{|\sigma_{ave}|} \times 100 \qquad (1)$$

RESULTS

The average total stress variation for all specimens was 25±5% (Table 1), with the highest and lowest stresses corresponding to the largest and smallest strain errors of 0.9% and -0.9% respectively. A strain overshoot of 0.3%, the strain error observed in our typical stress relaxation experiments, corresponded to an average stress overshoot of 3±1%.

DISCUSSION

Plantar tissue in compression is sensitive to small target strain errors that can result in stress errors that are several fold larger. The extent to which overshoot may affect the peak stress will likely differ in magnitude for other soft tissues, loading modes, and materials testing machines. However, with our testing protocols, errors will be on the order of 3%.

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Table 1: Stress error and stress variation with target strain error (standard deviation)

ε error (%)	Peak σ for each specimen (kPa)					σ error (%)
	1	2	3	4	5	
-0.9	135 (0)	61 (0)	89 (0)	58 (0)	78 (0)	-16 (2)
-0.6	147 (0)	67 (0)	100 (1)	62 (0)	83 (0)	-9 (1)
-0.3	155 (1)	72 (0)	106 (1)	66 (0)	87 (0)	-4 (1)
0	161 (1)	75 (0)	109 (0)	69 (0)	89 (0)	0
0.3	164 (1)	79 (0)	110 (1)	71 (0)	92 (1)	3 (1)
0.6	168 (1)	82 (0)	111 (1)	73 (0)	94 (0)	5 (3)
0.9	170 (1)	86 (0)	112 (1)	75 (0)	97 (0)	8 (4)
Total σ variation (%)	22	33	22	25	22	25 (5)

Relationships between foot strike patterns and foot pronation in running

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INTRODUCTION

Excessive rearfoot eversion (pronation) that might occur during the support phase of running induces tibial internal rotation. This movement is thought to be one of kinematic risk factors of overuse running injuries [1,2]. There exists a lack of information on the relationships between this risk factor and foot strike patterns [3]. Recently, a "pronation control" function is advocated in developing running shoes. Therefore, the purpose of this study is to clarify the relationships between foot strike patterns and the above risk factor of running injuries.

METHODS

Thirty-one male recreational runners (22.9 ± 3.6 yrs, 174.4 ± 4.5 cm, 63.4 ± 5.7 kg) participated in this study. The subjects were asked to run along a 25m runway at a speed of 4m/s. All subjects wore identical running shoes (adizero Boston, adidas) with a moderate cushioning property. Three-dimensional marker positions were recorded using a motion capture system with 8 infrared cameras (Motion Analysis Corp.) sampling at 240 Hz, and ground reaction forces were simultaneously recorded at 2400 Hz using a force plate (Bertec Corp.) Reflective markers were placed on the pelvis, thigh, shank and shoe. Ten acceptable trials were collected per subject. Kinematic and kinetic variables were analyzed for the stance phase of the right leg. Foot strike patterns (shoe tilt, shoe obliquity and shoe rotation) were calculated as a foot segment movement with respect to a laboratory coordinate system. Joint angles were expressed in a joint coordinate system. Joint resultant moments were calculated with an inverse dynamics approach. Resultant joint moments were expressed in the segment coordinate system embedded to the distal segment. Pearson product-moment correlations were calculated to assess the relationships between foot strike patterns and kinematic and kinetic variables of interest.

RESULTS

The shoe tilt was showed a significant negative correlation with ankle eversion velocity and tibial internal rotation velocity ($p<0.05$). Shoe obliquity and rotation were found to be positively correlated with ankle in/eversion excursion, tibial internal rotation angle, ankle eversion velocity, tibial internal rotation velocity and knee external rotation impulse ($p<0.05$). Similarly, shoe obliquity velocity was found to be positively correlated with ankle in/eversion excursion, ankle eversion velocity, tibial internal rotation velocity and knee external rotation impulse ($p<0.05$), but not with tibial internal rotation angle. The shoe tilt and obliquity were significantly correlated with shoe rotation ($p<0.05$).

DISCUSSION

The results of this study demonstrated that shoe tilt, obliquity and rotation were associated with the kinematic risk factor of running injuries. Therefore, the running styles might have substantial influence on running injuries. The shoe tilt, obliquity and rotation are influenced by subject's static body alignments and running styles. Appropriate manipulation of foot strike patterns, e.g. reducing shoe obliquity to decrease ankle eversion velocity and tibial internal rotation stress, will play an important role for preventing running injuries. Because it would be difficult to control these angles at heel-strike with running shoes, a function to reduce shoe obliquity velocity during a touchdown should be in equipped in running shoes.

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An investigation into lower limb muscle activation during the golf swing

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INTRODUCTION

Previous research assessing muscle activity in golfers has focused on the shoulders and trunk [1,2,3]. Lower limb analysis, however, is scarce. Consequently there is little understanding into muscle activation during the swing process and the influence that footwear has on lower limb muscles.

METHODS

Twelve right-handed male golfers (age 24.8 ± 7.0 yrs, height 184 ± 12 cm, weight 79 ± 6.8 kg, handicap 4 ± 4) played 5 shots with a 5 iron on natural grass at specified target wearing three different shoe conditions. Shoes incorporated the same uppers but differed in their sole traction (metal spike, alternative spike and flat sole). Shoe conditions were randomised and golfers were not informed of any sole differences. Muscle activity was assessed within the tibialis anterior (TA), vastus lateralis (VL) and medial head of the gastrocnemius (MG) of the front (nearest the hole) and back legs using surface electromyography (EMG) (Noraxon USA, Scottsdale, AZ). Raw data was rectified and smoothed. Peaks and means were identified and normalised to percentage of peak during the swing. The full swing and foot movements were recorded on two cameras (Casio Exilium Ex-F1, 1200Hz, Japan & Basler A602F-2, 100Hz, Germany). Each swing was broken down into the backswing (BS), down-swing (DS), ball impact (IMP), early follow-through (EFT) and late follow-through (LFT).

RESULTS

Front leg findings indicated no difference in muscle activity between the shoe conditions within the VL muscle. The MG was found to be more active throughout the whole swing process within the flat sole condition when compared to alternative and metal spikes. The highest MG activity was identified during the downswing which then decreased at ball impact and increased again as the golfers bodyweight was transferred and decelerated around the front foot. Greater TA activity was also identified within the flat sole condition during the BS, DS, IMP and EFT. *Back leg findings* did not identify any trends in MG muscle activity between shoe conditions during the BS and DS. However, notably lower MG and TA muscle activity was found within the flat soled shoe after the DS, while greater activity was identified within the metal spiked shoe condition after ball impact. Comparable VL muscle activity patterns were found between shoe conditions as identified within the front leg, however, the level of activity was 10-20% lower.

DISCUSSION

Muscle activity adaptations were identified within the MG and TA during the swing process. The findings indicate that golfers were able to adapt to unknown differences in shoe traction by adjusting their foot movements during the swing to enhance traction. It was evident that the flat sole developed traction by golfers slightly inverting and everting the shoes onto the medial and lateral sole edges. This was possibly enhanced due to the lower profile of the flat sole to the ground. Back leg MG and TA muscle activity was identified to be higher during the EFT and LFT stages of the swing within the metal and alternative spiked shoes. Post ball impact the back foot is required to rise and rotate anticlockwise onto the forefoot, facilitating the golfers' body rotations and to maintain balance. It is probable that this increase in muscle activity was a result of the metal and alternative spiked soles impeding the natural forefoot rotation of the back foot, while the flat sole did not restrict this movement resulting in lower muscular activity. It was also noted that less turf wear was associated with the flat soled shoe at the back foot.

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The influence of sprint spike stiffness on sprinting performance and Metatarsophalageal joint kinematics

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INTRODUCTION

Increasing the longitudinal bending stiffness of sprint shoes has been suggested to improve sprinting performance [2]. Stefanyshyn and Nigg showed that sprint performance improved on average by 0.7% when a carbon insole was inserted into sprinters own spiked shoes [2]. The purpose of this investigation was to determine the effect of bending stiffness of sprint spikes on both sprinting performance and the kinematics of the Metatarsophalangeal joint (MPJ), a joint which has been shown to be a large dissipater of energy during sprinting [1].

METHODS

Twelve club level sprinters (age 21.9 ± 4.0 yrs, height 175 ± 10 cm, weight 67 ± 8.2 kg) performed maximal 40m sprints on an indoor track in four different shoe conditions, a standard Puma sprint spike and three conditions where carbon/ glass insoles increased the bending stiffness of the shoes. The longitudinal bending stiffness was quantified mechanically using a two point bending test. Sprint times were recorded for a 10 m section (30 m to 40 m) using single beam timing cells at hip height. Additionally for four subjects, kinematic sagittal plane data was collected for one stance phase of the left lower limb using a 600 Hz video camera. MPJ range of motion and angular velocity were calculated following digitising in Quintic. All subjects completed a subjective questionnaire post-testing. One way repeated measures ANOVA was performed ($p < 0.05$)

RESULTS

Average bending stiffness of the standard conditions and three insole conditions were 276 (± 30), 329 (± 20), 388 (± 29) and 518(± 29) N/mm. Average 30 m to 40 m sprint time for all trials was 1.18 s (± 0.08 s), corresponding to a mean velocity of 8.50 m/s (± 0.57 m/s). There were no differences in sprint performance between the four stiffness conditions and the best stiffness condition for each subject was subject-specific with 7 out of 12 subjects demonstrating improved sprint performance win shoes with a higher stiffness than the standard condition (mean improvement 0.02s ± 0.01s). Mean MPJ angular range of motion was reduced in the stiffer sprint spike conditions (36.7° ± 5.8° in stiffest condition versus 42.1° ± 3.8°), although not significant.

DISCUSSION

Stefanyshyn and Fusco [2] reported performance improvements of 0.02s over 20m between the athletes' standard shoe and the optimal stiffness condition. However in this study typical variation between 2 trials in the same condition was 0.02s, therefore any potential differences may have been masked by trial to trial variability. Individuals responded differently to the stiffness conditions and the best stiffness condition was subject-specific, agreeing with the notion that sprint spike design needs to be tuned to individual characteristics [2]. Reasons for these differences warrant future study. Increasing the bending stiffness appeared to have controlling influences over the kinematics although MPJ kinetics and energetics may be more influential to performance and will be explored in the future.

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Effect of footwear on the gait of children: A systematic review

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INTRODUCTION

The purpose of this systematic review was to establish what is currently known about the effects of footwear on the gait of 'normal' children. Primary outcomes were lower-limb biomechanical variables between barefoot and shod gait. Adverse effects and participant opinions were secondary outcomes.

METHODS

Inclusion criteria were barefoot versus shod gait, participants 1-16 years of age, '*normal'* participants; sample size n>1. Novel footwear types were excluded.

Search terms and Medical Subject Headings were adapted to electronic databases. Hand-searching of journals, conference proceedings and article reference lists was undertaken. No language, year or publication restrictions were applied. Known experts were contacted to identify additional data. Two authors independently assessed all papers for inclusion. Methodology of studies was assessed using the validated and reliable Quality Index [1].

Mean differences and 95% CI were calculated for continuous variables using inverse variance analysis in Review Manager 5.0 (The Cochrane Collaboration, Copenhagen, Denmark). Heterogeneity was measured by I^2 [2]. Where I^2>25%, a random-effects model analysis was used and where I^2<25%, a fixed-effects model was used. A conservative approach to meta-analyses was applied to homogeneous studies. Alpha value was P<0.05.

RESULTS

Of the 1640 studies identified 10 fulfilled the inclusion criteria. Studies investigated walking (n=6), treadmill walking (n=1) and running (n=3). Sample size ranged from 4 to 898. Median Quality Index was 20 out of 32 (range 11-24). Three studies randomised conditions. Five studies did not standardise footwear.

Compared to barefoot walking shod walking increased: velocity (P<0.00001); stride and step length (P<0.00001); stride and step time (P< 0.00001); base of support (P<0.00001); toe-off (%) of gait cycle (P<0.00001); double support (P<0.00001); stance time (P<0.00001); ankle range of motion (ROM) (P=0.04); subtalar rotation (P=0.03); knee sagittal plane ROM (P< 0.003); tibialis anterior EMG amplitude.

Compared to barefoot, shod walking decreased: cadence (P<0.00001); single support (P<0.00001); hallux ROM (P<0.00001); medial arch length (P<0.00001); foot torsion (P<0.00001); forefoot supination (P=0.02); forefoot width (P<0.00001); midfoot ROM (P ≤0.002).

Compared to barefoot running shod running decreased: maximum tibial acceleration (P=0.01); rate of tibial acceleration (P=0.0009); shock wave transmission as a ratio of maximum tibial acceleration (P=0.0001); ankle and plantar foot angle at foot strike (P<0.05); knee angular velocity (P=0.03); swing-back velocity (P=0.03).

DISCUSSION

Children walk faster in shoes, taking longer steps, with greater ankle and knee motion and tibialis anterior activity. Increased double limb support time and base of support size in footwear are suggestive of less stability. Shoes might have a splinting effect on foot joints by reducing motion. Swing phase leg speed is reduced during shod running possibly due to increased weight on the foot. Shoes attenuate some shock transmission and encourage a rearfoot foot strike pattern during running. Future research could investigate these effects on long term growth and development.

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In-vivo investigation of the existing shear reducing diabetic insoles

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Introduction:

Emerging evidence suggests that plantar shear plays a major role in diabetic ulceration. Besides the orthotics that are claimed to reduce peak shear stress, a number of insoles were designed to reduce the net shear force experienced by the plantar surface. However the biomechanical efficacy of these devices was not examined in vivo due to technical challenges. It is known that an increase in plantar shear forces results in increased step lengths [1-3]. Therefore spatio-temporal characteristics of the gait can be used to assess the biomechanical efficacy of shear reducing diabetic insoles. The use of such insoles is thought to result in reduced step length and thus an increase in the number of steps required for traversing a certain distance. The aim of this study was to quantify and compare spatio-temporal characteristics of the gait when wearing two different brands of shear reducing insoles and standard control insoles.

Methods:

Eighteen healthy volunteers were recruited. Informed consent was obtained before the study. Subjects walked along a 100ft line while wearing three types of insoles, in randomized order. The tested insoles were Vasyli Armstrong® Diabetic Orthotic (Vasyli Medical, Labrador, Australia) and the Spenco RX® Diabetic Support Foot Beds (Spenco Medical, Waco, TX, USA). Generic EVA insoles served as control. Gait speed, step length (SL), cadence and number of steps to cover the 100ft distance were calculated. Three trials for each pair of insoles were conducted. The mean values of the trials were obtained for each product. Data was analyzed using paired t-tests.

Results:

Mean SL values for control EVA, Vasyli Armstrong (VA) and Spenco (SP) shear reducing insoles were 0.74, 0.73 and 0.74 meters, respectively. No significant difference was reported in the four parameters of interest among the paired datasets (Table 1).

Discussion:

According to a number of earlier reports, a reduction in SL values is expected if net plantar shear force can be decreased. Since we have not observed such a decrease in SL values, it is thought that shear reducing diabetic insoles do not function in vivo (i.e. shod) as well as in vitro conditions.

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Table 1. Spatio-temporal characteristics of gait while wearing three different type of insoles*

	Control (EVA) Insoles	VA Insoles	SP Insoles
Gait speed (m/s)	1.42 (0.16)	1.40 (0.17)	1.43 (0.13)
Step Length (m)	0.74 (0.06)	0.73 (0.07)	0.74 (0.06)
Cadence (steps/s)	1.92 (0.11)	1.91 (0.13)	1.93 (0.10)
# of steps to cover 30.5 m	41.98 (3.52)	41.96 (3.20)	41.65 (3.17)

* Values are Mean (Standard Deviation). No significant difference was observed.

Foot Orthoses Contribute to Better Gait Initiation

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Introduction

Many studies have attempted to better elucidate the effect of foot orthoses on gait dynamics. However, to our knowledge, most previous studies exclude the first few steps of gait and begin analysis at steady-state walking. The dynamic process of gait initiation is much more complex since the human body needs to accelerate often in a limited amount of time. As a result, the skills necessary to maintain stability, weight transfer, foot clearance, etc., become more critical during this transition phase than during the steady-state conditions [1]. Such requirements become even more significant in patients with neurological disorders, lower limb complications, and in older adults with potentially inherent difficulties with postural stability and gait The purpose of this study was to quantify gait initiation period and determine how many steps were required to reach steady state walking under three footwear conditions: barefoot, habitual shoes, and habitual shoes with a prefabricated foot orthoses.

Methods

An innovative body-worn technology was used to assess spatio-temporal parameters of gait in free condition and over a distance of 50 meters on 15 healthy-young subjects. A novel algorithm was designed to explore the gait-initiation phase (Figure 1) based on statistical inter-cycle variability of the first 20 steps, in three randomly assigned conditions; 1) barefoot, 2) wearing habitual shoes, 3) wearing prefabricated foot orthoses with habitual shoes.

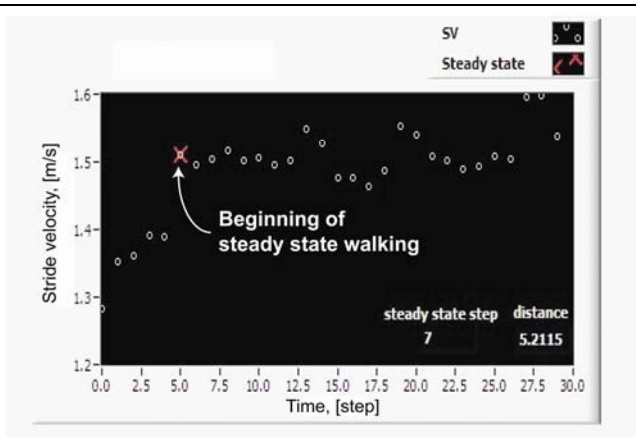

Figure 1: Identification of the beginning of steady state based on assesing inter-cycle variability of stride velocity.

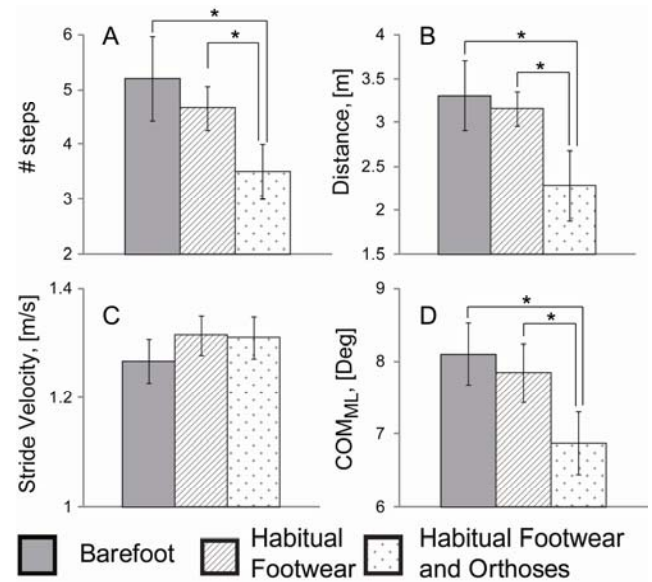

Figure 2: Gait modification as a function of the type of footwear.

Results

Wearing habitual shoes with the prefabricated orthoses enabled subjects to reach steady state walking in fewer steps (3.5 steps ± 2.0) compared to the barefoot condition (5.2 steps ± 3.0; $p=0.02$) as well as compared to the habitual shoes condition (4.7 steps ± 1.6; $p=0.05$). Interestingly, the subjects' dynamic medial-lateral balance was significantly improved (22%, $p<0.05$) by using foot orthoses compared to other footwear conditions (Figure 2).

Discussion

The results suggest that foot orthoses may help individuals reach steady state more quickly and with a better dynamic balance in the medial-lateral direction, compared to other foot type wear conditions. Additionally, these results put into context other studies that have described the minimum steps necessary to achieve stable estimates of plantar pressure.

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Assessment of the Body's Adaptation To Prescribed Foot Orthoses Using Dual Task Paradigm and Gait Initiation

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Introduction

Human gait is an extremely complex phenomenon that, until recently, was considered to be a largely automated task, requiring little or no attention from the walker. Several studies have cast doubt on this notion through use of dual task methods of adding a cognitive distraction to walking. With practice, humans with healthy sensory motor systems consistently learn and adapt in dynamic and challenging environments. However, motor performance may deteriorate under these same conditions when a novel cognitive task is introduced to the walker. In this study we examined whether the gait performance may be deteriorated for the naïve users of foot orthoses.

Methods

Fifteen healthy subjects walked 50 meters barefoot, shod, and shod with a prefabricated foot orthoses under dual task (DT) and single task (ST) conditions. DT was created by having subjects count backward out loud from a given number by sets of three while walking. A novel algorithm was designed to explore the gait-initiation phase based on statistical inter-cycle variability of the first 20 steps, measured using an ambulatory system based on body worn sensor technology.

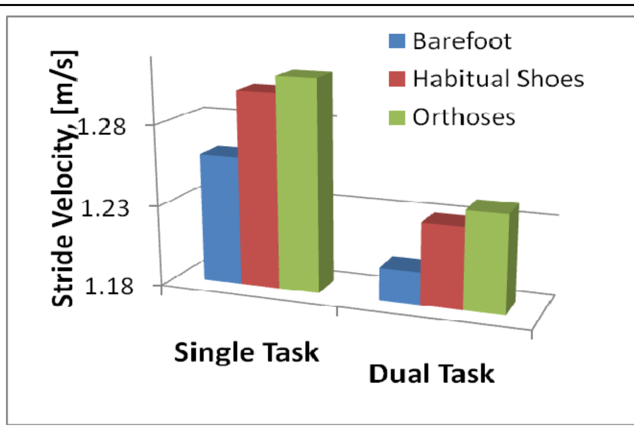

Figure 1: Gait speed during gait initiation phase.

Results

Foot orthoses improved subject's gait and balance compared to other footwear conditions (center of mass improvement in medial-lateral direction by >18% and >7% respectively during ST and DT conditions, gait speed improvement on average by >3% and >0.8% respectively during ST and DT conditions). Additionally, foot orthoses allowed subjects to initiate walking with a higher gait speed compared to other footwear conditions despite of speed decrease during DT (Fig 1). Although wearing prefabricated arch support orthoses enabled subjects to reach steady state walking in fewer steps ($p<0.05$) during ST, under dual task condition, the situation was reversed (5.7±0.4, 4.7±0.5, and 4.7±0.4 steps respectively for orthoses, habitual shoes, and barefoot conditions, $p<0.05$, Anova-2 way, Fig.2).

Discussion

These findings may implicate more central nervous system (CNS) involvement in the body's adaptation to footwear than previously thought. The body and CNS have "learned" to function without impairment under dual task conditions in shod and barefoot conditions from many years of practice. Although, orthoses produced a better gait performance during steady state, introducing the naïve orthoses condition appears to challenge the CNS due to lack of experience during gait initiation. This effect appears to be masked in the single task condition.

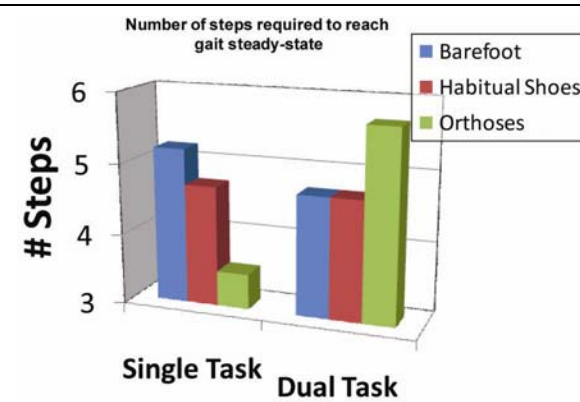

Figure 2: For the naïve users of foot orthoses, gait initiation may be detrioated when subject's attention is distracted by a cognitive task suggesting that a full adaptation is required for an adueqte functinoning of foot orthoses during activity of daily living.

Biomimetic Ankle-Foot Prosthesis with Compliant Joints and Segmented Foot

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INTRODUCTION

Biomechanical studies indicate that human foot is not a single rigid body with no intrinsic motion [1]. The segmented foot with toe joint has several advantages compared to the single rigid foot in the following aspects: walking step, walking speed, range of joint angle and the changing of angle velocity and joint energy-output. Inspired by the biomechanical studies, we start to analyze and design powered ankle-foot prosthesis with segmented foot.

METHODS

Based on the functionality of human toe and ankle joints, we design of the powered joints. As shown in Fig.1, the basic architecture and prototype of the proposed ankle-foot prosthesis are two series elastic actuators (SEA) [2], which are used to drive the ankle and toe joints respectively. Each SEA comprises a DC motor, a ball screw transmission and a spring structure. Because human toe joint only outputs net positive work at the moment of toe-off, the toe joint is designed to rotate counterclockwise passively, and to rotate clockwise actively. When toe joint is forced to rotate counterclockwise, Spring 2 will be extended to store energy. At the moment of toe-off, Spring 2 will release the stored energy and Motor 2 will drive toe joint to rotate clockwise via Transmission 2 and Spring 2. Spring 2 comprises four drawsprings set in parallel and the stiffness is 200*N=cm*. Motor 2 used in the current design is a 30*W* DC motor.

RESULTS

Before the subject wears the proposed prosthesis, we have done several preliminary experiments to evaluate the safety and the functionality of the prototype. The results show that under the load of 650*N*, a little more than the subject's weight, the ankle-foot prosthesis can operate well at the bandwidth of 2.5*Hz*.The video frame sequences captured in a walking cycle beginning with heel strike are shown in Fig. 2. Compared with the human walking gait, we can see that the ankle-foot prosthesis can well reproduce the human gait.

In the future, we will let the subject wear the prototype and optimize the control system. In addition, we will compare the prototype with the existing passive prosthesis by measuring the

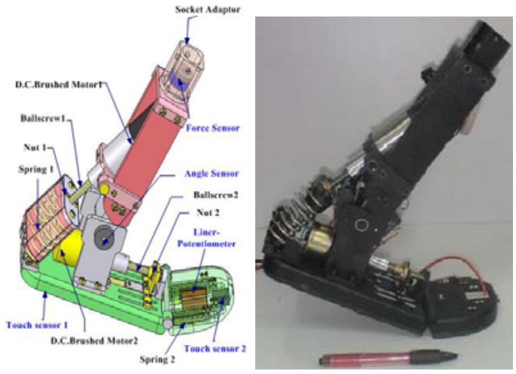

Figure 1: The ankle-foot prototype. The total weight including battery is 2.47 kg. The full angles of ankle and toe are 45° and 90° respectively.

metabolic cost of the subject. The hypothesis that the toe joint can decreases the ankle torque will also be evaluated.

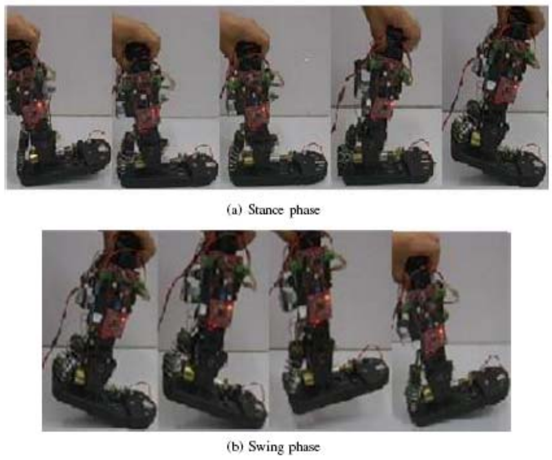

Figure 2: Sequenced pictures captured from the experiment of the prototype in a walking gait cycle beginning with heel strike.

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Effect of Heel Height on the Distributions of Plantar Triaxial Stresses

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INTRODUCTION

Foot calluses are common for high-heeled shoe wearers, which are associated with excessive pressure and shear stress. Shear stress may be important on the declined surface of high-heeled shoe. Due to lack of instrumentation, existing shod stress distribution information is not sufficient. The purpose of this study was to investigate the contact pressures and shear stresses simultaneously between plantar surface and high-heeled shoes with different heel heights using in-shoe triaxial force transducers.

METHODS

Four healthy female subjects were volunteered in the pilot study. Five in-shoe triaxial force transducers (June Sport product Co. Ltd, Anhui, China) were mounted under the hallux, the 1st, 2nd, 4th metatarsal heads (MTH) and the heel centre (Figure 1). The triaxial stresses were measured in shoes with 3 inches and 0.5 inches heel heights. Subject was asked to walk at 110 steps/min. The sampling frequency was 300 Hz.

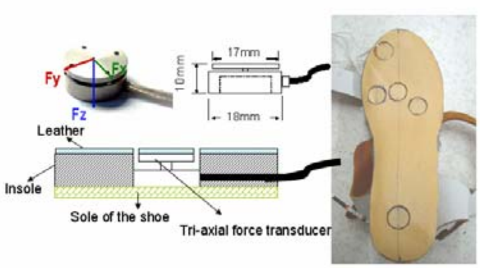

Figure 1: Transducers used in this study and their locations

RESULTS

As the heel height increased, the regions of peak pressure and shear stress had a tendency to shift from the 4th MTH to the 1st and 2nd MTHs (Table 1). The maximum peak shear for 3 inches occurred over heel region (75.8 kPa), while the maximum peak shear occurred over the 4th MTH (50.9 kPa) for 0.5 inches. Figure 2 shows the shear stress pattern from one subject for 3 inches heel height. The patterns were similar for 0.5 inches heel height, although the magnitudes were different.

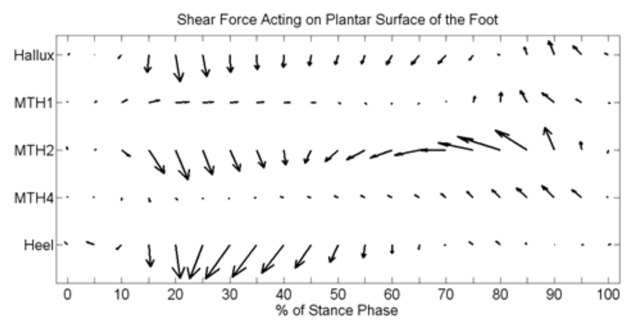

Figure 2: Shear forces from one subject for 3 inches heel height

DISCUSSION

In-shoe triaxial force transducers were used to measure the shod plantar pressures and shear stresses over specific regions simultaneously. The information obtained is useful for understanding the biomechanics of foot with high-heelded shoe and the mechanical effect of high-heeled shoe support on foot health.

ACKNOWLEDGEMENTS

This study is supported by grant from the Research Grant Council of Hong Kong (Project PolyU5352/08E) and a Research Studentship from The Hong Kong Polytechnic University.

Table 1: Mean (standard deviation) pressure and shear stress (kPa) for two heel heights.

	Hallux	MTH1	MTH2	MTH4	Heel
Peak Pressure					
0.5 inches	373.6 (152.7)	227.5 (180.3)	325.8 (137.7)	114.3 (81.2)	359.0 (32.6)
3 inches	311.1 (47.6)	306.3 (205.7)	363.4 (182.9)	28.9 (13.4)	369.0 (61.2)
Peak Resultant shear					
0.5 inches	50.8 (8.6)	29.5 (12.8)	35.8 (5.3)	50.9 (27.0)	36.3 (15.8)
3 inches	59.0 (3.5)	66.8 (34.4)	68.5 (5.1)	17.1 (6.0)	75.8 (13.5)

Hard Sole or Soft Sole? A study into shoe-specific ground reaction model parameters.

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INTRODUCTION

Several mathematical models have been developed to describe footwear kinematics and kinetics [1,2,3]. The ground reaction based model parameters relate the compression and rate of compression of the sole unit to the ground reaction force. These parameters have had implications in simulation studies. Up until now these parameters were determined for the two typical categories namely the hard and soft soled shoes. To quantify these effects the shoe was modelled as a mechanical suspension system represented by a nonlinear spring and damper, and the vertical ground reaction force was formulated according to the following equation [1].

$F=a.\,(x)^{b}+c.\,(x)^{d}.\,(v)^{e}$ Equation 1

Where F represents the force, x represents the deformation and v represents the deformation rate of the sole. While there has been an agreement on the value for *e* in the existing literature [1,2,3], there is a disagreement on the value of other parameters which can provide a realistic ground reaction force [2,3]. Whilst these parameters for a categorical hard and soft shoe can be useful in general simulation analyses, they do not reveal the case-specific ground reaction force model parameters. The purpose of this study was to find the shoe-specific ground reaction model parameters and to compare it with the categorical values for soft and hard shoe reported in the literature.

METHODS

A 3 KN universal testing machine (Lloyds Instrument, UK) was used for dynamic loading of the shoe heel area, to a maximum loading of 2.5 times body weight of a runner of 10 N weight at the deformation rate of 0.5 m/s. The function representing the force-deformation (Eq.1) was fitted to the force-deformation data using parametric curve fitting technique.

RESULTS

The results are shown in table 1. The R^2 values were 0.99 for the stiffness and 0.96 for the damping terms.

DISCUSSION

Results indicate that the parametric curve provides a good fit to the force–deformation data indicated by the high R^2 value indicating the goodness of the fit. The stiffness and damping parameters found here for the shoe tested in this study were considerably different than what was reported in the literature [1, 2, 3]. The stiffness parameters of 20×10^{6} and 2.24 were much higher than the highest values of 10^{6} and 1.6 respectively reported in the previous literature [1, 2]. Also the damping parameters *c* was found to be lower compared to the value of 20000, while *d* was found to be higher than the value of 0.75 reported in the previous studies [2, 3]. This emphasises that the categorical hard and soft shoe ground reaction model parameters do not adequately reflect the visco-elastic behaviour of a specific shoe during loading. The shoe-specific model parameters can have implications in quantifying the sole cushioning characteristics.

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ACKNOWLEDGEMENTS

The authors would like to acknowledge technical help from Abbas Daragheh during data collection.

Table 1: The ground reaction model parameters

a	b	c	d	e
20×10^{6}	2.24	12802	1.04	1

Effects of Temperature on the Performance of Footwear Insole Foams Subjected to Quasi- Static Compression Loading

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INTRODUCTION

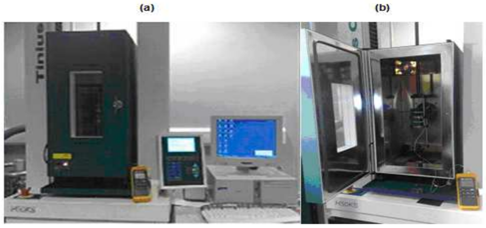
The insole is the removable insert part of the shoe that runs underneath the sole (bottom) of the foot, that is intended to correct an abnormal, or irregular biomechanics (walking pattern) [1]. They perform functions that make standing, walking, and running more comfortable and efficient, by altering slightly the angles at which the foot strikes a walking or running surface. The insoles also provide additional cushioning that attenuates the shock of impact and reduces the magnitude of localized pressure peaks by distributing the forces over a larger area of plantar surface. The current study investigates effect of temperature on the mechanical and shock absorption behavior of four different types of foams, which are commonly used as insole [1], under quasi-static compression loading.

METHODS

A testing method is introduced to subject the footwear foams to quasi-static compressive loading at varying temperatures 10°C to 40°C as shown in (Figure 1). The foam specimens were manufactured into cylindrical shapes and dimensioned at diameter=50mm and thickness=20mm.

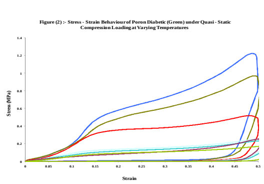
Figure (1):- (a) Environmental Chamber Shown Mounted on Frame and between Columns of Tinius Oslen Test Machine (b) Foam Testing

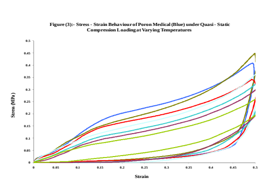
(a) (b)

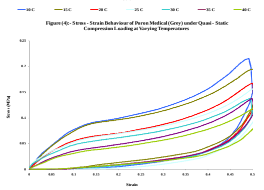


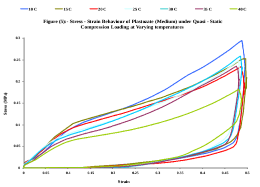
RESULTS

The stress-strain curves for the foams are shown in (Figures 2-5).





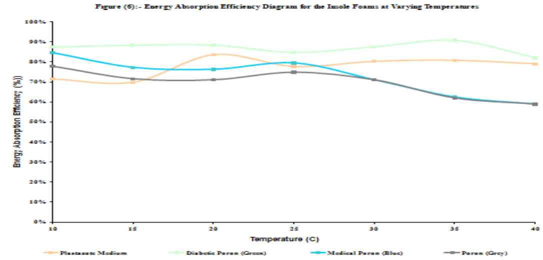




Then the non-linear hyperfoam material coefficients for N=2 number of parameters using Ogden energy relationship [2] were obtained for each case.

The shock absorption efficiency (E_e) for the midsole (F1-F4) was computed using trapezium numerical integration method from the individual data and their variations versus temperature are shown in (Figure 6).

Figure (6):- Energy Absorption Efficiency Diagram for the Insole Foams at Varying Temperatures



DISCUSSION

The conclusions drawn are:

- The mechanical properties of the foam materials were affected markedly as a function of temperature for all foam materials and exhibited some degree of sensitivity and softening with temperature elevations observed through the three phases of deformations. The softening of the foams makes them more easily deformable, with implications regarding the force transmission to the foot.
- Higher density foams exhibited less energy absorption than softer foams. Varying temperature affected the attenuation of the elastomeric foams and consequently their firmness and shock absorbability. The insole foam, Green Poron (96 grade) demonstrated the highest shock absorbing efficiency of 87%.
- The non-linear hyperfoam material parameters obtained at varying temperatures are important as they will provide an efficient FE simulation of foot-footwear for further analysis to design appropriate orthotic footwear to reduce stresses in the foot particularly for those individuals such as diabetes.

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Effect of Negative Heel Profile Shoes on Triceps Surae: A Pilot Study

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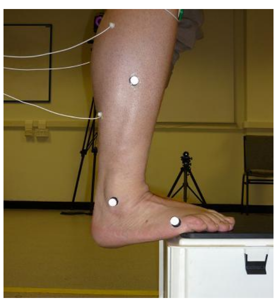
INTRODUCTION

Eccentric exercise in the treatment of Achilles tendinopathy has been shown to be the most effective conservative management. (Alfredson 2003, Mafi et, al. 2001). A recent review of eccentric exercise concluded that the main variance in success of such a treatment is the compliance of the patient to repeat the exercise as frequently as described. (Meyer et, al. 2009) The main objective of this study was to assess if the negative heel profile shoes eccentrically load the triceps surae group and could therefore be a treatment option for Achilles tendinopathy.

METHODS

A convenience sample of 10 subjects (mean age 38 years, mean height 170cms, mean weight 80 kg) with no known lower limb injury were recruited to participate in the study after giving fully informed consent. Eccentric loading exercise (Figure 1) was performed in a barefoot condition whilst the muscle activity was recorded (Portilab, TMSi ,Netherlands) for gastrocnemius and soleus. An optoelectronic movement analysis system (Vicon, OMG,UK) was employed to collect and analyse kinematic data of ankle during dorsiflexion.EMG data was sampled at 2048Hz as opposed to 100Hz for the kinematic data. MVC data was acquired at a comfortable dorsiflexed position. The participants were asked to perform a few walking trials before the actual data collection. Once accustomed to the environment, they were asked to walk at a self selected pace in a circular motion around a 10m walk way. Data was collected in a standard heel profile shoe. Kinematic, kinetic and EMG data of the gastrocnemius (GM & GL) and soleus (Sol) muscle was collected. The processed data was exported and the maximum dorsiflexion angle at heel strike was calculated for the walking trials and the same angle was calculated for the static exercise (Fig



1). EMG data was analysed to compare the time domain parameters for each condition.

RESULTS

Preliminary results indicate that the average dorsiflexion angles increased to 115.3 degrees as opposed to 105.4 degrees during static exercise. Figures 2, 3 and 4 indicate representative results from one subject.

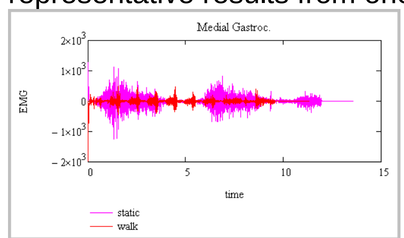


Figure 2: Raw EMG (Medial Gastroc)

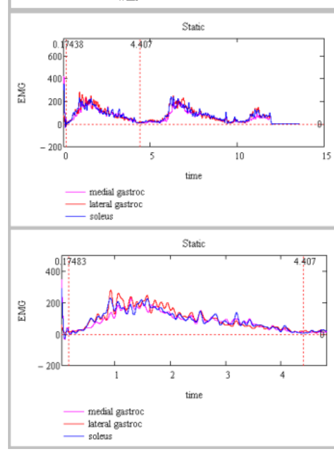


Figure 3: EMG during static Exercise

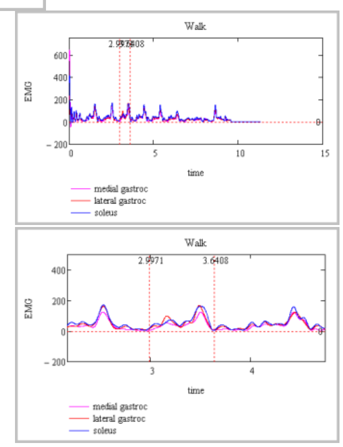


Figure 4: EMG during walking on negative heel profile shoes

DISCUSSION

The results highlight that the negative heel profile shoes does affect the kinematics and the muscle activity in the lower limbs. Whilst initial results indicate clear differences in both timing and magnitude of Triceps Surae group activation, further studies are warranted to substantiate, if the muscles behave in the same way at Heel Strike in these shoes when compared to static eccentric exercise.

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The effect of memory elastic alloy reinforced shoes on come-off distance

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INTRODUCTION

In these days the usage of women's heels has increased. The shape and size of shoes are main determinants for consumers to choose their high-heel shoes. However, practically it is difficult to completely satisfy each individual's feet shape and size as a result of ready-made shoe size. For the purpose of comfort, the slightly over-sized shoes are preferred. However, these induce over-used muscle activations to prevent heel from coming off during locomotion and inefficient walking patterns. Previous study [1] demonstrated over-pressure at fore foot area and subsequent deformation of fore foot in prolonged usage of high heel shoes.

This purpose of this study was to investigate the effect of built-in memory elastic material on heel coming-off distance in order to relieve toe pressure and heel off in wearing high heels. This elastic memory was placed in the shoe's filer where is found in between phalanges and tarsals.

METHODS

Five healthy female subjects, free from foot injuries or neuromuscular disease, participated in the experiment (age: 22.4±1.7 yrs, mass: 51.6±2.4 kg and height: 160.1±2.4 cm). Their foot size was limited to 230 mm. Three-dimensional motion data were collected with 6 cameras (VICON, MX-T40, UK), 1 high speed camera (Basler, A602fc, Germany). The locomotion was performed on the treadmill (MOTUS, M901TA, Korea) at specified speeds. Those ones were 0.9 m/s (3.5 km/h), 1.2m/s (4.5 km/h) and 1.5m/s (5.5 km/h), respectively.

Conventional high heel shoes and high heels with memory elastic alloy reinforced shoes (MEAS) are used (Figure 1). They were no different in outsole geometry. For each type of shoes, two different sizes of shoe 230mm (actual fit) vs. 235mm (slightly over-sized fit) were tested.

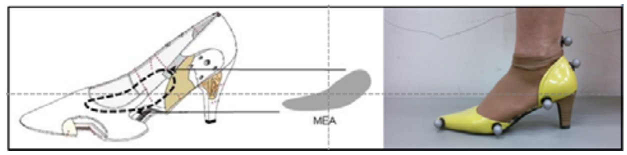

Figure 1: The position of MEA and Marker attached

RESULT

In the case of the conventional shoes, 235mm sized shoes had a greater amount of come-off distance at the heel than 230mm did, and for the MEAS, both sizes of shoes had no difference statistically (Table 1).

DISCUSSION

The conventional shoes increased come-off distance at the heel. Unexpectedly, the slightly over-sized shoes with faster speed, the heel coming-off distance was reduced. We speculate that intentional force was applied to the fore foot to prevent the heel off at high speeds. Applying this kind of force might cause the overused muscle activations of feet area and deformation of fore foot for prolonged usage. The MEAS showed smaller coming-off distance. Therefore the supplemented MEAS would be useful to reduce the removal of shoes and effective to prevent overused defects of feet.

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ACKNOWLEDGEMENT

This research project was supported by the Sports Promotion Fund of Seoul Olympic Sports Promotion Foundation from Ministry of Culture, Sport and Tourism.

Table 1 : Come-off distance at the heel. (Type A: conventional shoe, Type B: MEAS, M±SD, Unit: mm) .

	230mm		235mm	
Speed(m/s)	**Type A**	**Type B**	**Type A**	**Type B**
0.9	31.5±3.5	30.9±2.9	43.0±18.2	32.7±3.0
1.2	32.4±4.1	30.9±4.3	38.7±6.2	32.9±4.9
1.5	33.2±5.0	30.7±6.6	37.3±5.4	30.6±3.6

Kinematic Study of Load Response in the Human and Non-Human Primate Foot: Clues for the Evolutionary Development of Human Bipedalism

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INTRODUCTION

The human foot is frequently cited as one of the more distinctively human anatomical features [1,2]. Human feet are special, although not in the loss of pedal prehensile capabilities and the adoption of bipedal postures and locomotion. Yet, pedal skeletal anatomy is fairly uniform among the primate order, and different from other mammals [1,3]. This research asks: are uniquely human features of the foot merely due to functional application, or are they rooted in anatomy? This study addresses that question by comparing how the primate foot responds to load under experimental conditions.

METHODS

Data are derived from the legs of human, chimpanzee, and baboon cadavers. Legs were prepared by removing all soft tissue, so that only ligamentous structures remained to sustain limb integrity. Each specimen was subjected to a vertical load via the tibial shaft. Positions of the calcaneus, talus, cuboid and navicular were monitored as rigid clusters using an active-marker tracking system. These results are summarized as plots of joint angular motion versus percentage of the load cycle.

RESULTS

Figure 1 shows rotation about the three examined joints. There is a similarity in the rotational responses of humans and chimps. At the same time, the baboons, which are monkeys and thereby more distantly related to humans and chimpanzees, show a distinctively different response. Baboons respond more to load at the naviculo-cuboid joint, which suggests a flattening of the transverse arch. Humans and chimps respond more at the talo-navicular joint (medial longitudinal arch) and the calcaneo-cuboid joint (lateral longitudinal arch). These results are contrary to expectations, inasmuch as chimps are said to not possess a medial longitudinal arch [1,2,3].

DISCUSSION

The present study shows a unity of pedal function among the hominids (humans and African apes) to the exclusion of cercopithecoids (Old World monkeys). The results suggest that the distinctive load bearing features of the human foot are best viewed as extensions, rather than innovations, of the hominid pedal condition. Consequently, the foot of the stem hominin (exclusive human ancestry) may not have been recognizably distinct from the ancestral hominid condition.

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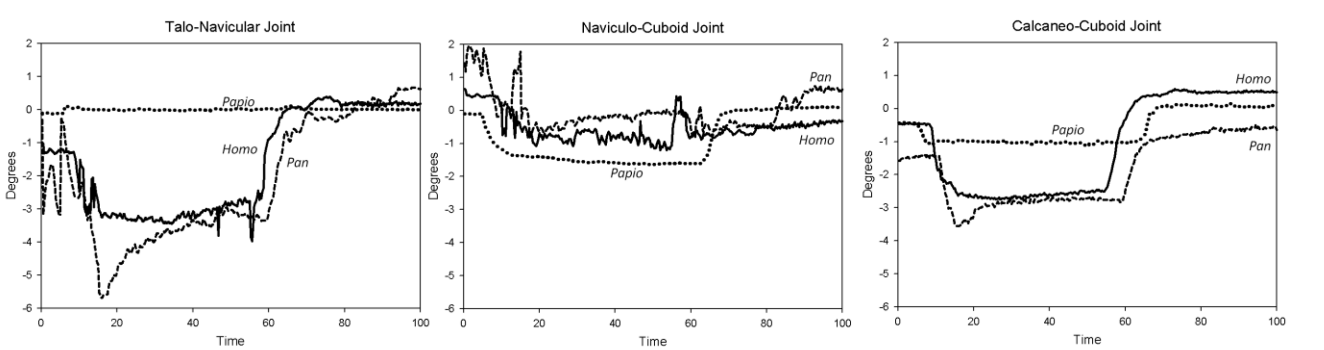

Figure 1: The rotation responses across the midtarsal joints for three representative individuals. Note the similarity of curves curves between *Homo* and *Pan*, and the distinctly different response displayed by *Papio*. Key: *Homo* = Human, *Pan* = Chimpanzee, *Papio* = Baboon.

Quantification of Foot Bone Motion Using Biplane Fluoroscopy

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INTRODUCTION

Tracking the motion of the bones of the foot is technically challenging. Methodologies such as computerized infrared camera systems with surface markers, X-ray stereophotogrammetry with implanted tantalum balls [1], magnetic resonance imaging (MRI) [2], retro-reflective markers with bone pins [3], single plane fluoroscopy [4], and dual C-arms [5-7] have been used. However these techniques suffer from one or more of the following issues: skin motion artifact, invasive markers, static or quasi-dynamic motion, bone occlusion, or inability to walk unfettered. The ability to use a biplane fluoroscopy system to study foot bone motion remains incompletely explored to date.

METHODS

Our biplane fluoroscope has been modeled after an existing system [8]. We have purchased two Philips BV Pulsera C-arms with custom designed synchronization boards and positioned them to optimally image a foot phantom; this was v1.0 of our system. With v2.0, we will make the following changes: 1) replace the existing Philips cameras with high-speed cameras, and 2) remove the X-ray generators and image intensifiers from the C-arms and mount them to four custom designed support structures. All data reported here are from v1.0, however, v2.0 is nearly complete. There are four phases of custom software development. Phase I encompasses calibration, including distortion correction, bias correction and 3-D calibration. Phase II includes the generation of digital reconstructed radiographs (DRRs) from a CT scan. Phase III entails the implementation of the similarity and comparison measures between the fluoroscope images and the generated DRRs. Phase IV is speed and memory optimizations.

RESULTS AND DISCUSSION

We have completed Phase I and II, and begun Phase III of the software development. To achieve Phase I, we have constructed a precisely machined aluminum grid for distortion correction and 3-D calibration frame with 36 metal balls. We have collected data with v1.0 of the biplane fluoroscope with a foot phantom with metal balls on 3 bones. We have successfully calibrated the system (Fig. 1).

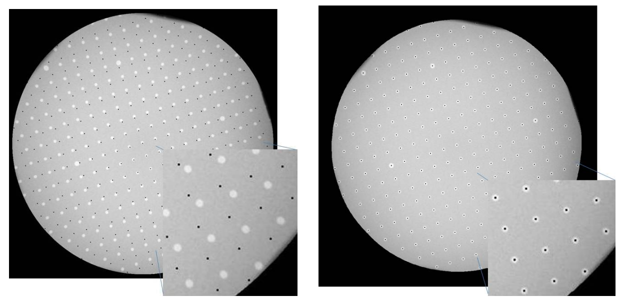

Figure 1: Distortion correction.

As a preliminary validation, we compared the distortion corrected data points to the aluminum grid and found an average RMS error for both C-arms of 0.31mm and 0.32mm. We then conducted the 3-D calibration by boot strapping each of the 36 points; the average RMS error was 0.31mm. Next we generated DRRs of the 3-D calibration frame and compared the position in the DRRs. The average RMS error of each fluoroscope was 0.56mm and 0.43mm. DRRs of the foot phantom have also been generated. Our preliminary validation demonstrated sub-millimeter accuracies, which compares well to other existing biplane systems. In the immediate future, we will complete the analysis of the v1.0 data by tracking the metal balls attached to the foot phantom bones and comparing the bony positions to our model-based approach.

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ACKNOWLEDGEMENTS

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Design of a 1st Metatarsophalangeal Hemi-Arthroplasty Implant Based on Morphological Data

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INTRODUCTION

When conservative treatments of MTPJ1 pathology fail, a number of surgical options are available including hemi-arthroplasty [1,2]. It is important that the geometry of any metatarsal head replacement be designed with close fidelity to the original metatarsal head. In this study, we present a quantitative approach to the design of replacements for the articular surface of the 1st metatarsal head based on 3-D scans of osteological specimens.

METHODS

Specimens of 1st metatarsal bones from the Hamann-Todd collection at the Cleveland Museum of Natural History were scanned using a NextEngine 3D laser scanner. A total of 97 adult bones were scanned (48 male and 49 female; age range: 30-50 years). A two-stage process was employed to transform each data set into a meaningful reference frame for ensemble analysis. An initial local reference frame for each bone was created using three anatomical landmark points. Template bones were then chosen for men and women separately. Local reference frames were aligned with optimal fit ellipsoids for each template bone determined using an unconstrained non-linear optimization method which minimized the sum of the squared distances between each point on an ellipsoid and the closest point on the bone. For the secondary alignment, the anterior 40% of individual target bones were aligned to the appropriate template bone local reference frame (after a mirror reflection of the target bones from the contralateral side) using the optimization method described above. The coordinates of each entire target bone were then transformed according to the parameters calculated from the optimization.

Slices of each metatarsal head (MTH) were obtained from cutting planes rotated at 1 degree increments around a mediolateral axis to form a patch describing the articulating surface above the crest of the MTH. Best fit ellipsoids were obtained for 3 size groups of males and females using the optimization method described above. Finally, a region of the ellipsoid surface containing the median patch was extracted for each group.

RESULTS

The semi-axes of the optimal ellipsoids for the medium size male and female groups are presented in Table 1. The resulting implant is shown on a typical specimen in Figure 1.

Table 1: Semi-axes (in mm) of ellipsoids for medium male and female specimens.

	x	y	z
Medium ♂	8.11	11.99	9.12
Medium ♀	7.88	10.61	7.12

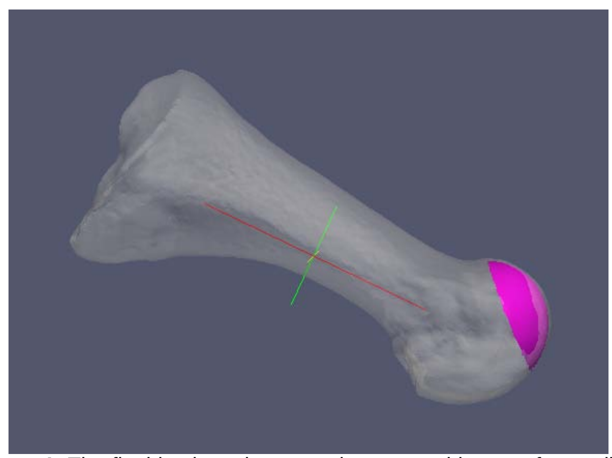

Figure 1: The final implant shown on the scanned image of a medium male specimen.

DISCUSSION

The approach to implant design described here is applicable to other joints where a hemi-arthroplasty is being considered.

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Real-time Statistical Analysis of Plantar Pressure Data: A Preliminary Feasibility Study

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INTRODUCTION

Clinical analysis of plantar pressure data is largely experience-driven, consisting mainly of qualitative judgements of a small number of images. This is statistically non-ideal because five or more trials are required for reliability [1]. Topological approaches [2] can handle this distribution variability by producing statistical images at the same resolution as the original pressure images. This preliminary study explores the feasibility of using automated topological statistics for routine clinical assessment.

METHODS

A single 'patient' subject performed five trials of each of adducted and abducted walking (Fig.1). Data were collected using an RSscan 0.5m system (Olen, Belgium) and were saved to disk in a rapid-access format [3]. Two 'clinician' subjects, with no prior experience in either plantar pressure imaging or topological statistics, were then recruited to analyse the data offline using a custom interface (Fig.1) after receiving ten minutes of instruction using an unrelated dataset.

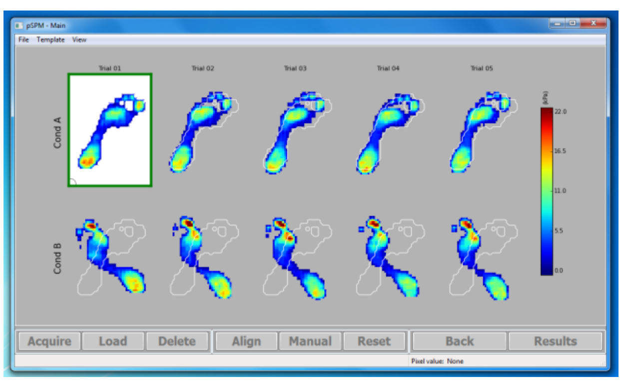

Figure 1: User interface and experimental data.

RESULTS

The three main computational tasks (data loading, image registration [4], and statistical computation) were conducted with clinically feasibly speed (Table 1). The 'clinician' subjects were able to successfully process the data and provide a satisfactory basic functional interpretation of the statistical images (Fig.2).

Table 1: Computational durations (mean ± st.dev.).

Task	**Duration** (ms) (ten images)	**Plotting** (ms) (ten images)
Image Loading	835.1 ±6.7	91.3 ±1.0
Image Registration	1061.0 ±4.0	92.2 ±1.2
Statistics	332.9 ±2.8	113.2 ±3.7

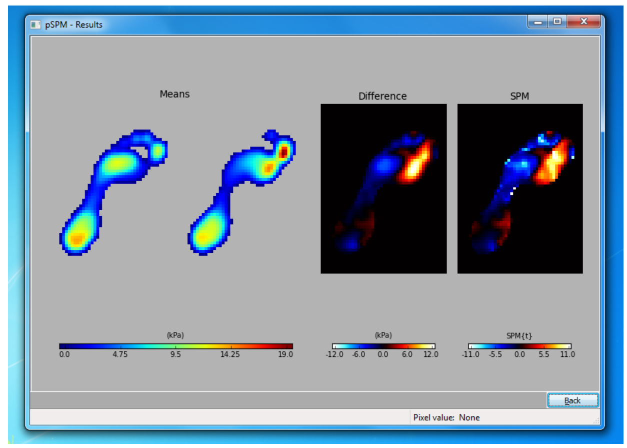

Figure 2: Statistical results screenshot.

DISCUSSION

Successful registration-based analysis and interpretation of topological results by non-experts, together with rapid and automated data processing, suggests that topological statistics [2] offer a feasible means for integrating statistical and anatomical objectivity into routine clinical assessment without sacrificing the visual richness of the original images. Computational tasks were performed rapidly despite non-optimal implementation (in an interpreted language: Python). An online real-time implementation is currently being developed for laboratory and clinical trials.

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Talocrural Joint Contact Modelling with Thin-plate Splines

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INTRODUCTION

The geometry of articular surfaces, required for contact conditions during joint modelling, may be obtained from MRI slice edges. The techniques based on tensor products of curve-fitting splines requires that the surface data be nominally gridded and arranged in a non-sparse grid as well as requires administration of the surface patches, which is not suitable for data that are distributed randomly. Thin-plate splines (TPS) were proposed to model joint surfaces based on unordered, scattered experimental data points [1] and has the advantage of avoiding surface patches and modelling scattered data points without experimental data preparation. To describe the ankle joint complex, in this study, thin-plate splines are adopted as surface modelling of the articular surfaces of the joints.

METHODS

The thin-plate spline contributes to morphometrics in two ways: as an interpolator, it optimizes a global figure of merit that often has a useful biological interpretation; as a statistical structure, it is linear in the data, and it embodies much of what we mean by large-scale versus small-scale biological variability. In this study, the sampling points for constructing the thin-plate spline models of articular surfaces were measured by using a 'Flock of Birds' electromagnetic tracking system (Ascension Technology Inc., Burlington, Vermont, USA). The passive motions during the dorsiflexion-plantarflexion were measured without considering weight and muscle forces. Ten below-knee amputation cadaver specimens were tested.

RESULTS

The trajectory of the contact points on the articular surfaces on the distal part tibia and on the proximal articular surface of the talus were

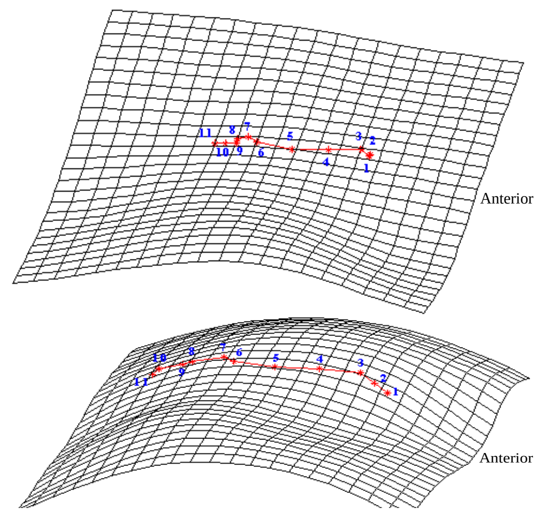

Figure 1: the contact points on the distal articular surface of the tibia and on the talus dome during dorsiflexion-plantarflexion. The articular surfaces are meshed with a spacing of 1mm.

located respectively. Table 1 lists the values of the flexion-extension angles of the calcaneus which correspond to each of these positions.

DISCUSSION

While the foot moving from the maximal dorsiflexion to the maximal plantarflexion, the articular contact point on the distal articular surface of the tibia moves from the anterior to the posterior and the corresponding articular contact point on the talus dome also shifts from the anterior part to the posterior part.

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Table 1: Value of flexion-extension angle of the calcaneus at each joint position.

Position	**1**	**3**	**5**	**7**	**9**	**11**
Ankle Flexion Extension Angle (deg)	-20	-10	0	10	20	30

Mechanical Testing of the Lapidus Plate for First Metatarsocuneiform Joint (MCJ-1) Arthrodesis

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INTRODUCTION

MCJ-1 arthrodesis, a.k.a. Lapidus arthrodesis is a treatment option for severe hallux valgus deformity with first-ray hypermobility [1,2]. The aim is to correctly align and fix the first ray. One available implant is the so-called Lapidus plate developed by E. Orthner in order to achieve a rotationally stable osteosynthesis with temporary fixation of the 1st and 2nd metatarsal. It comes with different step sizes between 0 and 6 mm. The present study investigated the implant's mechanical stiffness in relation to its step size and the orientation.

METHODS

The present study used a mechanical model to ensure comparable testing conditions. A custom plug was used to fix the plate in various orientations from 0˚ (horizontal, plate facing upwards) to 45˚, 90˚, 135˚, 180˚ (Fig 1).

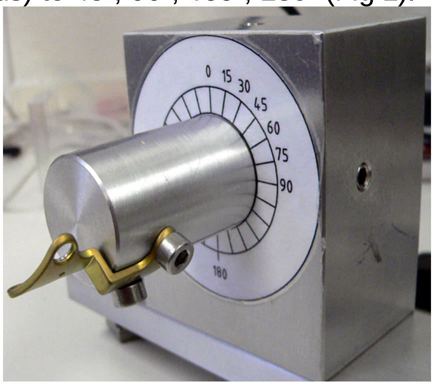

Figure 1: Experimental set-up with plate fixed at 135°

Specimens were tested with step sizes of 0,2,4,6 mm (5 each). Plates were screwed to the plug on the proximal end. A force of 80 N was applied to the plate between the screw holes for the distal fixation with a material testing machine (Zwick model 005, Ulm, Germany). The force was determined after preliminary load-to-failure tests and was applied in 10 cycles; the 11th cycle was a load to failure test to measure the ultimate stiffness

RESULTS

The implant stiffness highly depended on step height as well as plate orientation with 90° as the stiffest position. Stiffness tended to decrease with increasing step height.

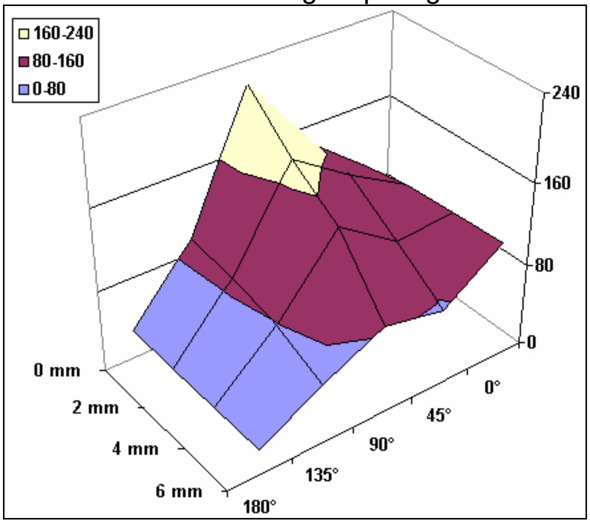

Fig. 2: Relative stiffness changes [%] (0°/0mm= 100%)

DISCUSSION

The stiffness decreased with increasing step height and varied with orientation. Therefore, implantation position should be taken into account as an additional parameter that influences the overall stiffness of the plate-bone system and could partly counteract the reduced stiffness of larger step heights.

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ACKNOWLEDGEMENT

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Table 1: Absolute [N/mm] / *relative [%]* stiffness values depending on step size and plate orientation.

â Step size/ Angle à	0°	45°	90°	135°	180°
0	336 / ***100***	489 / *145*	**739 / *220***	317 / *94*	93 / *28*
2	370 / *110*	476 / *141*	597 / *178*	301 / *89*	89 / *26*
4	379 / *113*	363 / *108*	492 / *146*	266 / *79*	86 / *26*
6	387 / *115*	252 / *75*	288 / *86*	196 / *58*	**83 / *25***

Low-cost bi-axial shear stress transducer: a step closer to shear distribution measurement

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INTRODUCTION

Plantar pressure imaging is widely used to evaluate human gait and has become a standard tool for biomechanical analysis. However, there is currently no commerically available system capable of evaluating shear stress distribution beneath the foot. Traditional strain-gauge multi-component load transducers are expensive and time consuming to build, especially when several hundred sensors are required to fully characterise the load distribution beneath the foot. This study presents a novel, low-cost, miniature bi-axial shear stress transducer suitable for plantar shear distribution measurement.

METHODS

The utilisation of magnetic based systems for the measurement of plantar shear stress has been demonstrated previously [1]. This study has extended on this work, by using magnetic based sensor integrated circuits (ICs) in a novel and cost-effective design. The mechanical structure of the transducer was simple, easy to fabricate and provided a rigid base for housing the sensor ICs. An elastic layer (Malaysian Rubber Board, UK) [2] located between the sensing surface and the transducer housing permitted minute shear movement of a permanent magnet held in close proximity to the sensor ICs.

The housing and the sensing surface were made of aluminium alloy and steel, respectively. The elastic medium was 0.5mm in thickness. The completed prototype transducer measured 8.38x13x13mm. The transducer assembly was calibrated using a 6-channel reference load cell (Nano25, ATI Industrial Automation, USA). Force outputs from the reference load cell and voltage outputs from the magnetic sensor ICs were sampled simultaneously at 200Hz. Dynamic calibration tests for vertical and shear axes were conducted at 1Hz over the range of ±40N: this corresponds to a shear stress of 237kPa on the sensing surface, which is higher than previously measured during walking [1].

RESULTS

As anticipated, the transducer was insensitive to vertical loads applied during calibration. Thus, cross-talk between vertical and shear axes was negligible. The hysteresis for shear axes was <6.9% over the applied load range, while the full-scale non-linearity was <4.2% (Figure 1). Dynamic performance of the sensor revealed average differences of <1.1N between the transducer and the shear outputs from the strain-gauge based reference load cell.

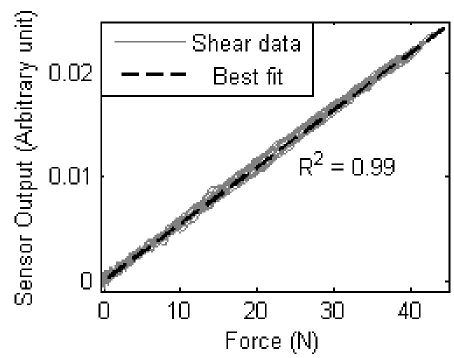

Figure 1: Calibration curve for a typical shear axis of the bi-axial shear stress transducer.

DISCUSSION

This bi-axial shear stress transducer can be assembled easily using off-the-shelf sensor ICs and has the capacity to form arrays of various sizes, making it suitable for load distribution measurement. The cost of components to manufacture the one-off prototype transducer was <US$8, whereas batch production could be considerably less. This low-cost transducer design has the ability to measure shear distribution under the entire plantar surface of the foot effectively and economically. Due to the size of the sensor ICs, the transducer area could be further reduced to 10x10mm for a high-resolution plantar shear imaging system.

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A Systematic Review of Kinematic Models used in Foot & Ankle Biomechanics

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INTRODUCTION

Over the past decade our understanding of foot function has increased significantly [1,2]. Our understanding of foot and ankle biomechanics appears to be directly correlated to advances in models used to assess and quantify kinematic parameters in gait. These advances in models in turn lead to greater detail in the data. However, we must consider that the level of complexity is determined by the question or task being analysed. This systematic review aims to provide a critical appraisal of commonly used marker sets and foot models to assess foot and ankle kinematics in a wide variety of clinical and research purposes.

METHODS

An electronic search of the following databases was performed in March 2010: MEDLINE, Embase, Cinhahl, ISI Web of Science, Scopus and SportDISCUS. The search strategy used was "foot model* AND human* AND kinematic* AND (gait* OR ergonomic* OR automotive*)". The secondary snowball method was applied to identify literature not identified during the electronic database searching process.

Titles and abstracts of identified articles were assessed by a single reviewer (CB). Articles were only included if they were published in the English language, were full text and original publications. Further, only three-dimensional kinematic models were included. No reviews of the literature were included. Data was extracted based on standardised protocol. The quality of studies was assessed by two reviewers (CB and DT) based on a modification of the method established by Peters et al [3]

RESULTS

The search strategy and data reduction method is presented in Figure 1. The initial search identified 287 articles. Inclusion/exclusion criteria were applied by one examiner (CB). This process excluded 224 articles. A secondary snowball search identified a further 4 articles. 27 original articles were included in the final review.

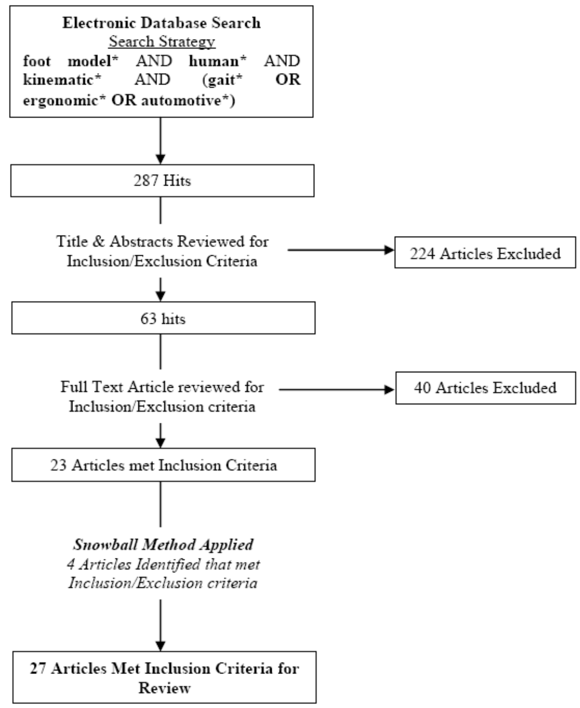

Figure 1 – Flowchart of Systematic Search Process

DISCUSSION

This paper presents a systematic overview of current techniques used in analysing foot and ankle kinematics in clinic and research throughout the world. It remains important that biomechanists appreciate that the level of complexity of the model required is determined by the question/task to be analysed.

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A Comparison of Two Multi-Segment Foot Models in High-Arched and Low-Arched Athletes

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INTRODUCTION

High-arched (HA) and low-arched (LA) feet have been associated with relatively different kinematic, kinetic and injury patterns [1, 2]. Recently several multi-segment foot models have been developed and utilized to investigate motions within the foot [3, 4]. However, no previous research has investigated the analogous nature of different multi-segment foot models. The purpose of the current study is to examine the intersegmental kinematics calculated using two different multi-segment foot models. These models will be compared in two distinctly different foot types (HA and LA). It was hypothesized that there would be no significant difference between the two models in either foot type the variables of interest.

METHODS

Arch index was used to determine arch height [5, 6]. Ten HA (AI>0.375) and ten LA (AI<0.290) female recreational athletes participated in the current study. Each performed five barefoot walking and five barefoot running trials at a self-selected pace while three-dimensional kinematics (240Hz, Vicon*PEAK*) and ground reaction forces (1200Hz, AMTI) were collected simultaneously. The foot was modelled using the Leardini [2] and Oxford [3] multi-segment foot models. The Leardini model is a 4 segment model (lower leg, rearfoot, midfoot and forefoot) while the Oxford model is a 3 segment model (no midfoot). Kinematic and GRF data were filtered at 8Hz and 50Hz, respectively. A one-way ANOVA was used to compare frontal plane kinematics of HA and LA athletes in each of the multi-segment foot models. Alpha level was set at $p<0.05$.

RESULTS

The Leardini model detected differences in peak eversion angle at the mid-forefoot joint during running ($p=0.03$). There were no significant differences when using the Oxford model in peak eversion angle at the rear-forefoot joint in walking ($p=0.07$) or running ($p=0.07$). When using the Leardini model, no difference in eversion excursion in the rear-midfoot (walking: $p=0.82$; running: $p=0.12$) or mid-forefoot joints was found (walking: $p=0.46$; running: $p=0.18$). With the Oxford model, significant differences were detected in eversion excursion during walking ($p=0.02$)

DISCUSSION

The multi-segment foot models examined in the current study reveal unique differences in HA and LA athletes. The Leardini model is more sensitive to differences in joint position due to the definition of an independent midfoot segment. Conversely, the Oxford model is more sensitive to differences in excursion values compared to the Leardini model as the motions within the rearfoot and midfoot are summed generating greater calculated differences in excursion values. A limitation of these data is that only a single plane of motion is presented. Further investigation into sagittal plane motions within the foot would lend greater insight into the differences between these two multi-segment foot models.

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A Reliability Study of Biomechanical Foot Function in Psoriatic Arthritis Based on a Novel Multi-segmented Foot Model

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INTRODUCTION

Psoriatic arthritis (PsA) is an inflammatory joint disease associated with foot-related impairment and disability [1]. The aim of this study was to determine the within-and between-day reliability of spatio-temporal, plantar pressure, kinematic and kinetic measurements based on a novel, seven segment foot model applied in patients with PsA.

METHODS

Nine PsA patients and matched healthy adult controls underwent three-dimensional gait analysis on two occasions, one week apart using a seven segment foot model. Standard clinical/disease metrics were measured to assess disease status. A core-set of functional variables including inter-segment kinematics, kinetics, spatio-temporal and plantar pressure distribution were analysed using the coefficient of multiple correlation (CMC), intraclass correlation coefficients (ICC) and the standard error of measurement (SEM).

RESULTS

No change in disease status was observed between time points ($p>0.05$ for all clinical/disease metrics). Excellent within- and between-day reliability was found for intersegment kinematic and kinetic data patterns with CMC values typically greater than 0.950. Between-day reliability ranged from poor to excellent for absolute CMC values (Figure 1). Corrected CMC values were consistently higher across all variables ranging from fair-to-good to excellent. ICC values indicated excellent reliability for discrete spatio-temporal, plantar pressure, and ankle moment and power variables for both groups. Reliability for ground reaction forces and kinematic discrete variables ranged from fair-to-good to excellent. SEM values ranged from 0.7 to 3.0° for discrete kinematic variables across both groups with greater variability in the PsA patients.

DISCUSSION

In disease stable PsA kinetics, spatio-temporal parameters and plantar pressure distribution can be reliably measured. Overall, intersegment kinematic reliability was consistent with that reported elsewhere for healthy and inflammatory arthritis cohorts [2,3]. However, some discrete kinematic variables (absolute values) have poor reliability and should not be used in prospective cohort and intervention studies.

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ACKNOWLEDGEMENTS

The work of Semple, Hendry and Turner is funded by the UK Arthritis Research Campaign.

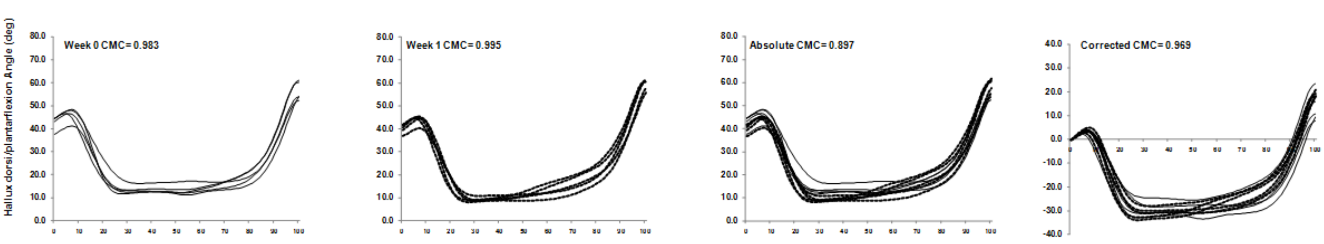

Figure 1: Typical example of within-day and between-day (absolute and corrected) reliability CMC values for hallux dorsi/plantarflexion in a single psoriatic arthritis patient

A clinically applicable six-segmented foot model

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INTRODUCTION

The aim of this study is to develop a clinically easy applicable model based upon the anatomical structure of the foot which includes the lower leg, hindfoot, midfoot, medial and lateral forefoot and the hallux using surface markers. To our knowledge, thus far there is no such model available which includes all these segments and which is clinically easy to use.

METHODS

Ten healthy volunteers (5M,5F) participated in the study (72.8 ± 9.7 kg; 1.77 ± 0.09 m). Their age varied between 21 and 61 years. All subjects walked barefoot over a 12 m long instrumented walkway. Kinematic data were collected with a 6 camera opto-electronic system (Oqus 3, Qualysis, Sweden) at 500 Hz. In the middle of the walkway a force platform (Accugait, AMTI, USA) and a plantar pressure platform (Footscan®, RSscan, Belgium) were build in. Force data were collected at 1000 Hz and pressure data at 500 Hz. A midstep protocol was used and trials with obvious targeting for the platform were excluded from further analysis. The subjects used a self selected comfortable walking speed, which was controlled for steady state by means of a laser at 1000 Hz (CMP3-30, Noptel OY, Finland). A variation of 5 % in speed was accepted between the different trials. Data collection continued till 7 usable trials were registered.

Fig.1: marker placement on the foot

To identify and track the segments 23 reflective markers (7mm Ø) were used. They were placed on anatomical landmarks on the foot and the lower leg and some supplementary tracking markers were added (Fig.1).

RESULTS

For each foot segment, rotation about the latero-medial axis is defined as plantar-/dorsiflexion. In-/eversion is rotation around the longitudinal axis of the foot (proximo-distal). Rotation around the plantar-dorsal axis is called ad-/abduction (Fig.2).

Duration of the stance phase was derived from the kinetic data.

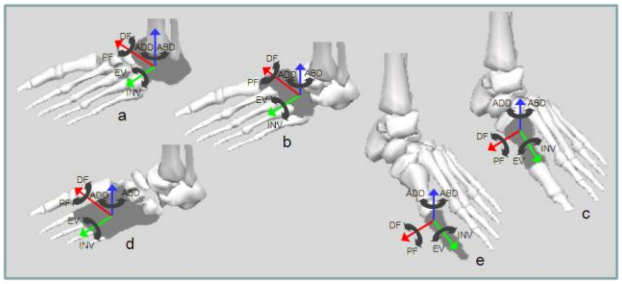

Fig. 2: the 5 segments of the foot with indication of all axes of rotation and motions.

Besides the low intra- and intersubject variability characterized by a low standard-deviation, the observed movement patterns comply with previous descriptions in literature, when comparison with other models is possible [1].

DISCUSSION

The innovative part of this study is that it consists out of 6 segments to represent all the functional units of the foot. It describes the motion between the hindfoot and lower leg (Fig.2a), between midfoot and hindfoot (Fig.2b), between first metatarsal and midfoot(Fig.2c), between lateral forefoot and midfoot(Fig.2d), between the hallux and first metatarsal (Fig.2e) in a clinically applicable model, using surface markers. Moreover we were able to track the motion of the foot structures at 500 Hz with only six cameras.

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Repeatability of the Salford multi segment foot model.

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INTRODUCTION

Multi segment foot models are tools for measurement of normal and abnormal foot biomechanics. Several different models have been proposed without clear criteria for their construction or segment selection. Based on prior cadaver and in vivo work [1-4] we have adopted a 5 segment foot model and report here the between day and between observer repeatbility of the kinematic data.

METHODS

Data was collected on two separate days from the five asymptomatic subjects (3 male). External markers on heel, midfoot, medial and lateral forefoot and hallux plates were tracked during 10 gait trials. On each day, one set of gait data was collected for each of two assessors. Each assessor followed an agreed protocol for attachment of the plates and was blind to the placement by the other assessor.

RESULTS

The table below identifies the mean differences (°) between assessor 1 day 1 and 2 (A1D1 - A1D2) , assessor 2 day 1 and 2 (A2D1 - A2D2), assessor 1 and 2 on days 1 (A1D1 - A2D1) and day 2 (A1D2 - A2D2).

DISCUSSION

The repeatability of foot kinematics is likely less sensitive to the choice multi segmental foot model used compared to the validity of the kinematic data. As such the data reported here only provides context for use for the model and interpretation of its data rather than confirmation of model validity. The differences observed were largely comparable to those from other models and reports of within and between day differences in lower limb kinematics.

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		A1D1 - A1D2	A2D1 - A2D2	A1D1 - A2D1	A1D2 - A2D2
Segments	Variables	Mean Diff	Mean Diff	Mean Diff	Mean Diff
Calcaneus relative to tibia	Initial Contact	-0.81°	-0.57°	0.94°	1.18°
	Total range of motion	0.39°	0.97°	-1.1°	-0.6°
	Toe off	-0.4°	0.89°	0.64°	1.94°
Midfoot relative to calcaneus	Initial Contact	-2.14°	2.26°	1.85°	6.26°
	Total range of motion	-1.18°	-0.57°	0.59°	1.19°
	Toe off	-1.85°	2.32°	0.74°	4.91°
Lateral Forefoot relative to Midfoot	Initial Contact	-0.29°	3.84°	-1.0°	3.11°
	Total range of motion	-2.14°	-1.54°	0.04°	0.63°
	Toe off	-2.18°	3.44°	0.29°	5.92°
Medial Forefoot relative to Midfoot	Initial Contact	0.89°	1.79°	-0.3°	0.57°
	Total range of motion	1.17°	-0.6°	0.01°	-1.8°
	Toe off	4.07°	3.06°	4.08°	3.08°
Hallux relative to Medial Forefoot	Initial Contact	3.11°	3.29°	-1.3°	-1.1°
	Total range of motion	3.45°	-2.13°	2.78°	-2.8°
	Toe off	4.05°	3.62°	-2.8°	-3.2°

Classification of Mid-foot Break Using a Multi-segment Foot Model

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INTRODUCTION

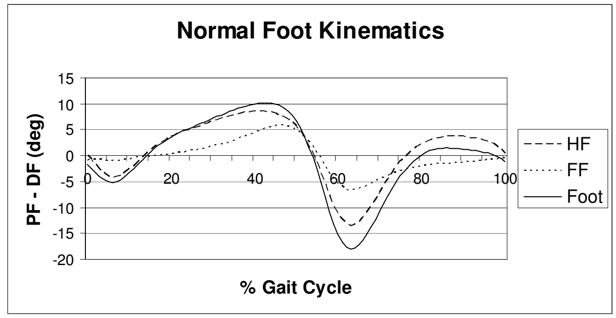
A mid-foot break (MFB) typically occurs when dorsiflexion (DF) of the ankle is reduced, causing spurious DF of the foot through the mid-foot. The literature is currently lacking an objective measure of MFB during walking. Conventional gait models examine the foot as a single rigid segment. This traditional method is not suitable in the case of foot deformities where dynamic motion can occur between the forefoot (FF) and hindfoot (HF). Recently, researchers have begun to use multi-segment foot models to adress this issue[1-5]; however, no studies have examined foot kinematics in children who have developed a MFB. Studying the FF and HF kinematics in this population may lead to earlier identification of MFB, and assist with preventative treatments such as Botox injections or surgery. The purpose of this study was to characterize the kinematic patterns of FF and HF motion during gait in children with MFB.

METHODS

Study participants were divided into 2 groups: (a) children with unilateral or bilateral MFB as determined by an orthopaedic surgeon or registered physiotherapist (Left: n = 14, Right: n = 15), and (b) children with no evidence of MFB or any gait abnormalities (Left: n = 7, Right: n = 7). A twelve-camera Motion Analysis system (Motion Analysis Corporation, Santa Rosa, CA) was used to record the 3-dimensional positions of reflective markers placed according to the Shriners Hospital for Children Foot Model, using the Smart Surface[4]. Foot motion was evaluated in two different ways: (1) motion of the foot modelled as a single rigid segment (conventional), and (2) motion of the HF and FF modelled as separate rigid segments.

RESULTS

As shown in Figure 1a, the HF angle mimics the foot angle in a rigid (Normal) foot. In the MFB group (Figure 1b), there is a 2-fold increase in DF of the FF and diminished DF of the HF segment compared to Normal.



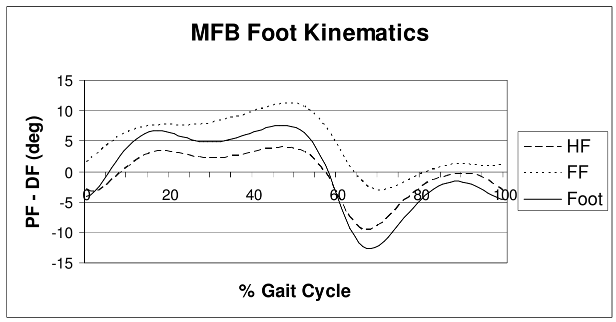


Figure 1a & b: Sagittal plane foot kinematics for Normal vs. MFB group. HF is measured with respect to (wrt) Shank, FF wrt HF, and Foot wrt Shank.

DISCUSSION

Comparing the two models in the MFB group, the conventional foot model misrepresented the true foot motion during stance phase by underestimating the equinus across the ankle. In the MFB group, dorsiflexion occurred primarily at the mid-foot (FF wrt HF) rather than the ankle joint (HF wrt Shank). By using this model it is now possible to quantify the presence of MFB in children in the sagittal plane.

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Use of multiple calibration in multisegment 3D foot kinematics

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INTRODUCTION

The current state of the art is that several foot models have been published in sufficient detail to allow them to be implemented by others and whose performance in terms of reliability is well documented [1-2]. However no one reported about soft tissue artifact (STA) propagation to foot kinematics. STA are commonly considered the most troublesome source of error in measurements of human motion carried out using stereo-photogrammetry. While researchers attempted to estimate the amount of STA during various motor tasks, multiple calibration (MC) resulted a successful method in compensating for STA propagation to knee kinematics [3].

METHODS

A 3 dimensional (3D) multi-segment foot protocol [1] was applied on the same subject first by means of direct skin (DS) marker placement (Figure 1a) and second in a modified version which entails calibrating each anatomical landmark with respect to a local cluster of marker (Figure 1b). Six cameras BTS S.r.l. motion capture system (60-120 Hz) synchronized with 2 Bertec force plates (FP4060-10) were used to acquire at least 3 walking trials together with a static one. During the static acquisition the subject was asked to stand in upright position with the heels jointed, and with the feet 30° apart (measured between the two medial parts of the feet). Anatomical calibration (AC) was performed [3] in 3 different positions: neutral position (NP), maximum passive dorsiflexion (MPDF, Figure 1c), maximum passive plantarflexion (MPPF). MC was performed by 2 physicians in order to test the inter-rater reliability. Anatomical reference frames were computed and joint angles estimated with each method (DS, AC, MC) during the stance phase of gait. MC was performed using, as control variable, the ankle dorsi/plantarflexion (DP) angle evaluated with the single AC acquired in NP. *Intra-rater* (each subsegment angle evaluated during three different walking trials), *inter-rater* and *inter-methods* reliability were assessed with the coefficient of multiple correlation (CMC) [4].

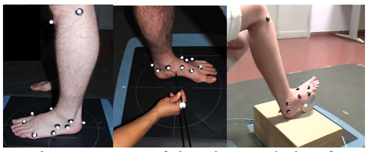

Figure 1: The protocol in the original version [1] (a) and with technical clusters during MC in NP (b) and in static MPDF (c).

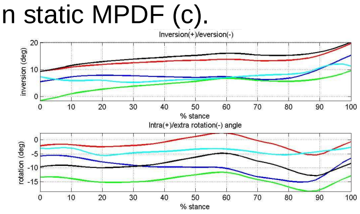

Figure 2: Hindfoot vs tibia angle evaluated with the different methods: mc (blu), mpdf calibration (green), np calibration (red), mppf calibration (black) and ds (light blue).

RESULTS

Mean and SD value of each sub-segment relative angles were evaluated over the stance phase of gait as in [1] (Figure 2). The worst reliability was observed for the forefoot-midfoot DP with DS (CMC=0.75±0.1) and hindfoot-tibia DP both with single AC and DS (CMC<0.7).

DISCUSSION

Our findings suggest a better intra-rater than inter-rater reliability for each method. However MC showed the best performance in term of inter-rater reliability (CMC=0.99±0.0). The main drawback of the present study is the lack of a gold standard, which was overcome performing the synchronous acquisition of stereo-photogrammetry and fluoroscopy of an in vivo foot kinematics, reported in another abstract.

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A Novel Technique for Improving Marker Placement Accuracy

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INTRODUCTION

Repeatability of marker placement has been acknowledged as a source of error affecting the reliability of multi-segment foot models [1]. A novel technique intended to reduce marker placement error is proposed and its effect on the reliability of intersegment kinematic data of the foot is investigated.

METHODS

Reflective markers (7mm diameter with flat base (Qualysis AB, Gothenburg, Sweden)) had 1mm holes drilled centrally, perpendicular to their base. These markers were then threaded on to a 1mm diameter flexible polystyrene wire which was bent over at each end to prevent the markers falling off (Figure 1).

Figure 1: Novel marker placement device

When attaching the markers to the foot, a piece of transparent double sided tape was adhered over the pen marks on the subject's foot. Then, the end of the plastic wire was placed on the target mark and the wire used to guide the marker down to the skin where it adhered to the tape.

The technique was tested by comparing replacement accuracy and reliability for static and walking trials in eight subjects who twice had markers attached by a podiatrist using the standard approach (and blinded to the purpose of the trial), and twice by a researcher using the novel technique.

RESULTS

The mean marker placement variability using the standard placement method was 1.4mm (SD 0.23). Using the novel device this error was reduced to 1.1mm (SD 0.28), a statistically significant improvement ($p=0.03$).

In general, between test coefficient of multiple correlations (CMCs) tended to be greater and had lower ranges for trials using the novel technique (Figure 2). Overall these improvements were also shown to be significant ($p=0.02$).

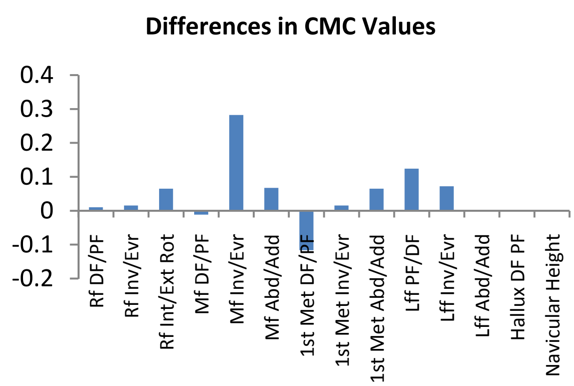

Figure 2: Differences in CMC values for each segment. Positive columns represent instances where the novel technique has improved the CMC value, negative where the standard placement method was better. Rf - Rearfoot; Mf - Midfoot; 1st Met - 1st Metatarsal; Lff - Lateral Forefoot; DF/PF-Dorsiflexion/Plantarflexion; Inv/Evr - Inversion/Eversion; Int / Ext Rot: Internal/External Rotation; Abd/Add – Abduction/Adduction.

DISCUSSION

The novel technique is a simple and inexpensive tool for improving the consistency of skin mounted marker placement. Results suggest that reductions in the error related to marker placement tended to improve the overall reliability of multi-segment kinematic data from the foot model.

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ACKNOWLEDGMENTS

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Two Foot Marker Placement Methods Reveal Different Coordination Patterns During Running

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INTRODUCTION

It has been suggested that differences in the timing of peak knee flexion and peak rear foot eversion may contribute to knee injuries in runners [1]. While a three dimensional gait analysis could potentially prospectively screen runners for injury risk, tracking markers placed on the shoes do not quantify true rear foot motion. As a solution some authors have advocated the use of windows in the shoe heel counter with tracking markers attached directly to the foot [2]. However, no comparison has been made between the ability of the two marker placement methods to detect potentially injurious mechanics. Therefore, the purpose of this study was to examine differences in time between peak knee flexion and peak rear foot eversion in a group of recreational runners, measured using both shoe based and heel windows markers.

METHODS

Thirteen individuals who ran at least 20 miles per week were recruited for this study. For each marker condition, subjects ran approximately 40 laps around a 25 meter track in the laboratory, with data being collected over a 5 meter segment of each lap. An 8-camera motion capture system (Motion Analysis Corp) recorded marker position data at 200 Hz while three AMTI (Advanced Mechanical Technology, Inc.) force plates recorded ground reaction forces at 1000 Hz.

For each trial the time between peak knee flexion and peak rear foot eversion was calculated using custom LabView (National Instruments) software. Left and right feet of each subject were analyzed separately and 10 trials per foot were used to create an average performance for that foot. Changes in timing between marker conditions were assessed using a dependent observations *t* test. Since a prospective assessment of injury risk should focus on deficits or changes in a single individual, a single subject design was also used to compare differences between marker conditions within an individual subject using an independent observations *t* test. Alpha was set to 0.05 a priori for all statistical tests.

RESULTS

Due to subject drop out only 17 feet were used in the final analysis. The group analysis showed no significant difference in the length of time between peak knee flexion and peak rear foot eversion when measured with the different marker placements (p = .24). However, in the single subject analysis 8 feet had significantly longer times between peak knee flexion and peak eversion under the heel windows markers (Figure 1). Three feet had significantly shorter times and 6 feet showed no significant difference between conditions.

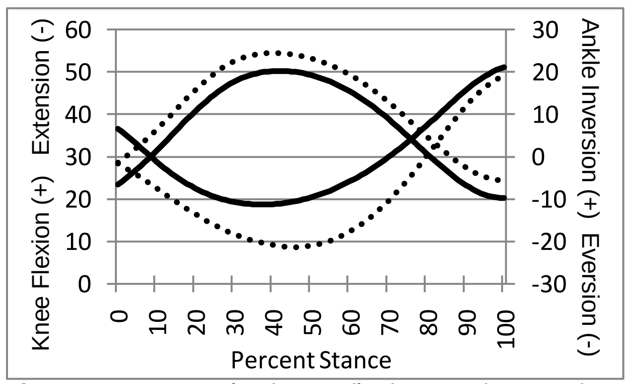

Figure 1. Example knee flexion and rear foot eversion curves for one subject. Solid lines show shoe marker data while dotted lines show heel windows marker data. Note the shift in timing between the knee flexion and rear foot eversion peaks under the heel windows markers.

DISCUSSION

The results of this study suggest shoe based marker placements and traditional data analysis methods may mask biomechanical markers potentially related to running injury. This has important implications for prospectively identifying individuals at risk for running injuries.

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A combined helical axis and Euler angle approach to calculate the foot torsion axis

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INTRODUCTION

Torsion of the foot (rotation of the forefoot relative to the rearfoot in the frontal plane) is relevant for shoe development [1]. To calculate a torsion axis during dynamic movements, markers are needed on the rearfoot and the forefoot. The finite helical axis approach allows one to calculate a rotation axis. For athletic movements, forefoot flexion is usually the most prominent movement while torsion and ab/adduction have smaller ranges of motion. Calculating the helical axis of such data would therefore result in an axis mainly influenced by flexion. Hence, a method is proposed to remove flexion from three-dimensional kinematic data and calculate the helical axis so that the foot torsion axis can be defined.

METHODS

A rearfoot coordinate system (CS) was established according to the ISB recommendations [2] while for the forefoot CS the z-axis was defined as originating at the medial metatarsal head and pointing to the lateral metatarsal head. This line is assumed to be the forefoot flexion axis (Figure 1). To remove forefoot flexion from three-dimensional data, the rotation of the rearfoot about Z_{ff} was determined using the Euler approach with rotation about Z as the primary rotation. This rotation was then subtracted from the initial rotation matrix. The helical axis method was then used on this new data set.

For an initial validation of the method, a virtual data set was created. A rotation matrix devised from random Euler angles (Table 1) was applied to a virtual marker set to simulate rearfoot rotation relative to the forefoot CS. To compare the input and output angles for abduction and torsion, the rotation about the helical axis was projected onto the axes of the forefoot CS.

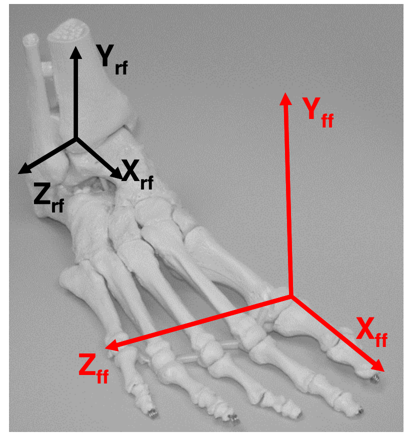

Figure 1: coordinate system for rearfoot and forefoot

RESULTS

The input and resulting output angles are listed in table 1. The orientation of the helical axis is (-0.74, -0.67, -0.0234) and the location is (0, 0, 0).

DISCUSSION

A method to calculate the midfoot motion axis without forefoot flexion has been described. An initial validation indicates that this may be a feasible method. The location of the resulting helical axis is at the origin of the forefoot CS because the virtual rotations occurred about these CS axes. The main orientation of the axis is in the XY-plane and therefore does not incorporate rotation about the Z-axis. However, further steps of validation with more complex data that also incorporate noise are needed. For real data, it is assumed that abduction is small therefore the resultant helical axis could be assumed as the torsion axis. However, further validation is needed to determine applicability to real data.

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Table 1: input Euler angles and resulting rotations about the axes of the coordinate system

	Θ_1 (flexion) [°]	Θ_2 (abduction) [°]	Θ_3 (torsion) [°]
input angles	18.390	-3.608	-4.023
output angles	18.390	-3.607	-4.021

First Metatarsal (1-Met) Helical Axis and Angle Measurements in Subjects Having Bunion as Compared to Controls

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INTRODUCTION

Collapse of the arch was theorized to orient the 1-Met axis vertical leading to bunion [1]. This study measured orientation of the 1-Met axis with helical parameters and compared angular kinematics of controls to subjects having bunion.

METHODS

15 females participated (8 controls; 7 bunions). Bunion ranged from mild to moderate based on hallux valgus angle. Subjects were scanned while standing in a .6 Tesla MRI, with their foot positioned (Fig) in a sequence of 3 gait event postures (MS, HO, TS). The 1-Met, navicular, and other bones were reconstructed using Mimics. Data sets were linked together in the MRI frame of reference for each subject. Embedded coordinate axes were directed positive Z lateral; Y up; X anterior. Helical parameters described orientation of the 1-Met axis with respect to the navicular from midstance (MS) to heel off (HO) and from heel off to terminal stance (TS). Axis parameters were expressed in the vertical reference frame. Cardan angles described rotations [2]. Helical parameters were assessed descriptively; a two by two ANOVA assessed for difference between groups and gait events.

RESULTS

Table. Mean ± SD (range) for 1-Met helical axis orientation between gait events, expressed about Z, Y, X component axes.

Midstance to Heel Off gait events			
Group	**Z(c)**	**Y(c)**	**X(c)**
Control	**-0.14 ± 0.65**	**0.06 ± 0.34**	**-0.27 ± 0.69**
	(-0.97 to 0.97)	(-0.62 to 0.48)	(-0.95 to 0.99)
Bunion	**0.06 ± 0.51**	**0.46 ± 0.32**	**-0.27 ± 0.67**
	(-0.81 to 0.76)	(-0.17 to 0.88)	(-0.89 to 0.93)
HO to TS Helical Axis Parameters			
Group	**Z(c)**	**Y(c)**	**X(c)**
Control	**-0.11 ± 0.32**	**0.20 ± 0.14**	**-0.91 ± 0.04**
	(-0.44 to 0.44)	(-0.07 to 0.36)	(-0.98 to -0.85)
Bunion	**-0.40 ± 0.37**	**-0.06 ± 0.42**	**-0.53 ± 0.57**
	(-0.73 to 0.42)	(-.61 to 0.68)	(-0.90 to 0.61)

Group mean helical axis orientation parameters (Table) reflect the change in direction of mean angular position of the 1-Met across gait events in the respective anatomical planes (Figure).

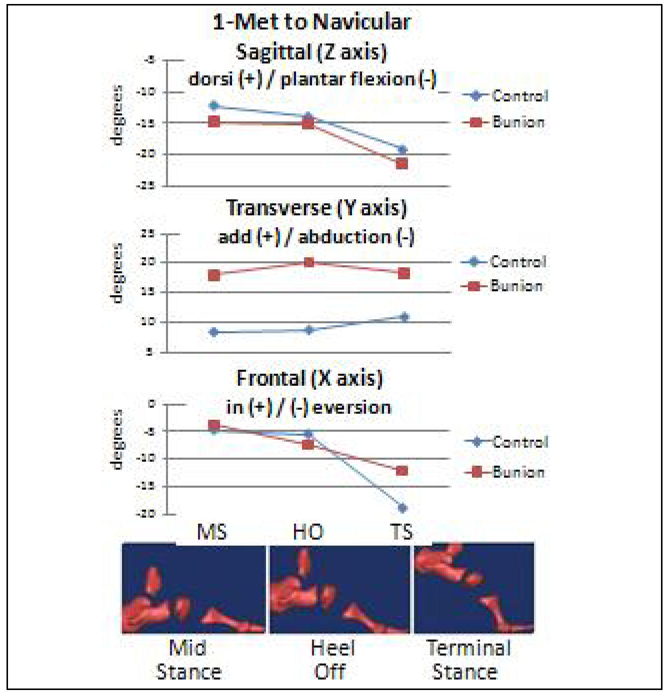

Figure. Mean angle position of 1-Met between groups and across events (MS, HO, TS) reported in the anatomical planes.

Between groups mean comparisons found difference (P<.001) in 1-Met angular position for adduction (Y axis). Across gait events, 1-Met adduction tended to decreased from MS to HO for the bunion group while it was unchanged for the control group.

DISCUSSION

1-Met kinematics differ in subjects with bunion as compared to controls. 1-Met angular rotation in relation to the navicular can be expressed with helical axis parameters and in general, reflect the Cardan angle computations. Future study can determine whether the 1-Met axis orients more vertically in subject having bunion.

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Comparison of Two Methods to Calculate the Helical Axis Parameters in the Subtalar Joint

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INTRODUCTION

The location of the subtalar joint axis varies across subjects, which makes it difficult to calculate moments and forces about the subtalar joint. To analyze subtalar joint kinematics, the Helical axis (or screw axis) method is commonly used as it may estimate the subtalar joint axis. Two similar methods have been presented to calculate the helical axis parameters in biomechanics. Kinzel et al. [1] described a body motion as a rotation around an instantaneous axis and a translation along this same axis. In 1980, Spoor et al. [2] presented a method to calculate a rigid body motion from spatial co-ordinates of markers by considering the screw motion.

There are some mathematical differences in these two methods, and they have both been used in the literature. The purpose of this study was to determine if differences exist in the helical axis parameters between these two methods.

METHODS

Seven fresh-frozen cadaveric lower extremities were obtained. Kinematic data were collected in a custom positioning and loading device from the tibia, talus and calcaneus with a Polhemus Liberty (Polhemus, Colchester, VT) and The MotionMonitor (Innovative Sports Training, Chicago, IL). Data were collected throughout the complete range of motion in plantar/dorsiflexion, inversion/eversion, internal /external rotation, supination/pronation, and inversion/eversion while the ankle was dorsiflexed. These motions were collected when the ligaments were intact and after the anterior talofibular ligament, calcaneofibular ligament, cervical ligament and interosseous talocalcaneal ligament were serially sectioned

The helical axis unit vector [U_x, U_y, U_z], a point on the helical axis, amount of translation along the helical axis and the angle of rotation (ϕ) around this axis for both methods were calculated using a custom Matlab (The Mathworks, Natick, MA) program for each motion and condition.

RESULTS

With both methods, the maximum and minimum angles of rotation occurred at the same frame with the exact same amount of rotation for each condition and motion. The unit vector varied when $|U_x| < 1E-4$ and where the U_y and U_z change in sign (Figure 1). This occurred in approximately 50% of the datasets.

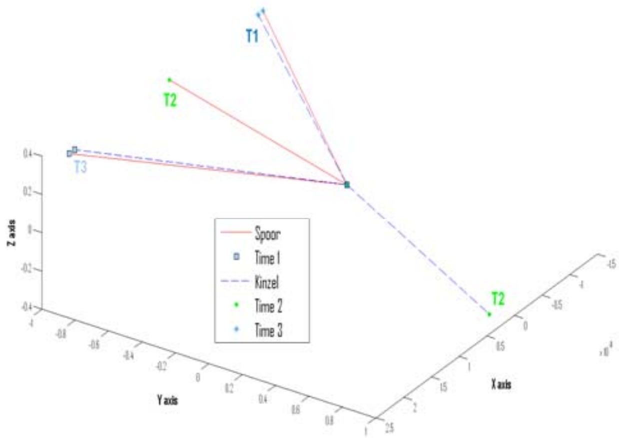

Figure 1: Unit vectors for 3 consecutive frames where U_x is small and U_y & U_z are changing signs. Note that the methods are different at time step 2.

DISCUSSION

The few differences between the two methods may be due to the fact that the method used by Kinzel defined U_y and U_z in function with U_x. In this function, U_x is calculated with a square root, which does not allow for zero values. This study showed that the rotations were identical and in most conditions the helical axis parameters calculated with either method were identical. Therefore, you only need to consider the differences in methods when examining the orientation of the helical axis when U_x is small.

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Is Bi-Axial Modelling Appropriate for the Ankle Complex?

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INTRODUCTION

The ankle complex is comprised of the complex talocrural joint and the talocalcaneal (subtalar) joint. Kinematic recordings of the rearfoot with respect to the leg frequently report three degree-of-freedom motion, hence the term "tri-planar ankle." Many researchers explain this using a bi-axial model [1,2,3]. The subtalar joint with its "mitered-hinge" [4] is presumed responsible for both foot adduction/abduction and inversion/eversion; whereas the talocrural joint is thought to provide plantar-dorsiflexion. We suggest the talocrural contribution is not adequately represented; therefore the bi-axial model may not be appropriate.

METHODS

Observations were drawn from cadaveric specimens (leg, ankle and foot). A rigid cluster was inserted into the tibia, talus and calcaneus. An active marker camera system recorded bone motion during three driving actions:

PD – Plantar / Dorsiflexion,
IE – Inversion / Eversion,
ML – Medial / Lateral rotation.

Results were processed using the Functional Alignment algorithm [5] to identify mean best-fit axes of rotation. Analyses focused on the ROM and joint axis orientations of the:

(1) Talus to Calcaneus – talocalcaneal joint,
(2) Tibia to Talus – talocrural motion,
(3) Tibia to Calcaneus – ankle complex.

Motions among the joints were compared within the each driving action. Analysis followed the axis triangle method [6].

RESULTS

Results (Fig. 1) show close agreement between tibia-to-calcaneus and talocrural responses for PD motion, general agreement between the tibial-to-calcaneus and subtalar responses to IE motion, and significant ($P<.05$) differences among the responses to the ML driving action. Within the ML driving action, the tibia-to-calcaneus exhibits almost twice the ROM of the subtalar joint. The extra ROM was found to be transverse plane rotation at the talocrural joint.

DISCUSSION

Conventional descriptions of the ankle complex do not allow for significant transverse rotation at the talocrural joint. Therefore, when these motions are found they are uncritically attributed to the subtalar joint. Our investigation suggests, however, that a new tri-axial model is needed – one that builds upon the functional concepts of the bi-axial model by including transverse plane rotation at the talocrural joint.

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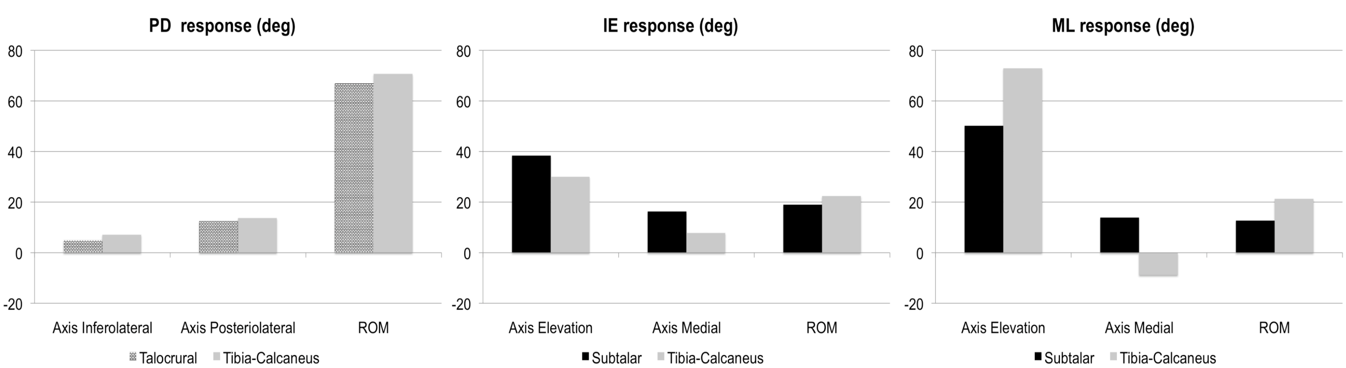

Figure 1. The effects of driving actions on joint axis orientation and Range of Motion (ROM).

Aligning the Foot and Knee

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INTRODUCTION

Osteoarthritis (OA) is widely believed to result from local mechanical factors acting within the context of systemic susceptibility. Research into OA requires the development of systematic methods for quantification of the influence of therapeutic advances on quantifying joint mechanics. In our previous research, we found the instantaneous motion of the knee ($) is reciprocally connected with acting restraint forces ($') on the foot [Kim W and Kohles S. A reciprocal connection factor for assessing knee joint function. Computer Methods in Biomechanics and Biomedical Engineering (in press)]. By connecting the motion of the knee with the forces acting on the foot reciprocally, we found that the reciprocity connection factor (RCF) may be easily measured by coupling the $ to a ground reaction force (GRF), and the RCF may, thus, perhaps be used as an overall measure of joint instability.

METHODS

When the force acing at the foot during the stance phase of gait intersects the instantaneous axis of rotation (ISA) of the knee motion, it was stated that the line of force must be incapable of causing a change in the relative displacement in ISA—a reciprocity condition of the healthy knee joint. Capture of the knee in such circumstances i.e., within “a stationary configuration” for measure of overall condition may prove to be a valid alternative to radiograph imaging, for static limb alignment. Previously reported experimental gait data collected from an adult male subject implanted with an instrumented knee replacement (mass 65 kg. Height 1.7m, right knee) were used for this study [1]. Coordinates in shank and thigh markers were used to generate screws and synchronized to ground reaction forces with associated COP data obtained by two gait trials.

RESULTS

We found that GRF considered as external forces aligned each reciprocal to ISA, indicating that GRF providing one component of restraining reciprocal connection (Figure 1). The migration of screw axes during one stance phase indicates typical screw patterns [2] of

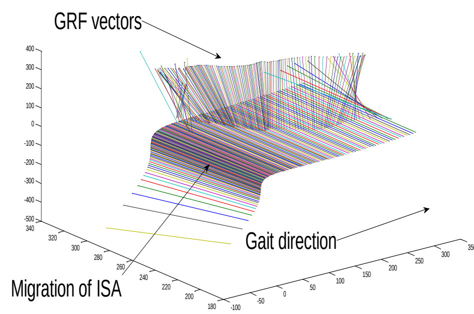

Figure 1: ISA are regularly intersecting with GRF vectors, indicating reciprocal connection between the motion of the knee joint and forces acting on the foot.

this subject, observed as two vectors regularly intersecting with 10 cm apart.

DISCUSSION

By linking ISA ($) of the knee to GRF ($′) on the foot, a newly developed measuring technique will produce a crucial information for the improvement of implant design and for the increased longevity of the implant components. Another area of applicable is to design of orthotic/Inserts prescription to correct gait abnormalities,

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We would like to thank Dr. Benjamin J. Fregly at University of Florida for providing us experimental data.

Correlation Between Forefoot and Rearfoot Intersegmental Joint Angles

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INTRODUCTION

Given the high incidence of running related injuries a consistent recommendation made by clinicians and researchers is an improved understanding of intersegmental foot motion.[1] The aim of this study was to investigate the relationship between midfoot and rearfoot intersegmental angular displacements.

METHODS

Informed consent was obtained prior to subject participation. For this study, 67 (134 feet) asymptomatic subjects (51 females, 16 males, mean age 22.6 ± 6.8 and 25.7 ± 13.8 years, respectively) were recruited.

Each subject had navicular height and foot length measured bilaterally while non weight and weight bearing. Retroreflective markers were then attached over specific bony landmarks on the subjects' bilateral lower legs and feet. An 8 camera motion analysis system, sampling at 120 Hz, recorded a 3 second static standing and 10, 6 second walking trials for each subject.

The marker displacement data was filtered using a 4th order, low pass Butterworth filter with a cut-off frequency of 10 Hz, the stance phase extracted and then time normalized to 100%. Using the filtered data and a 3 segment kinematic model consisting of a shank, rearfoot, and forefoot, intersegmental three dimensional (3D) joint angles were obtained using an Euler decomposition method. The maximum, minimum and range (maximum – minimum) of the angular displacements for the rearfoot and forefoot joints were obtained. Pearson correlation coefficients were obtained for the range (ROM) of the 3D angular displacements.

RESULTS

The correlation coefficients for the 3D ROM intersegmental data are presented in Table 1.

DISCUSSION

Several of the correlation coefficients were statistically significant, none with a value above .40. The majority of the coefficients were below .20. The low correlations noted in this study are in agreement with previous findings in the literature and possibly explaining the difficulty encountered in using this measure as a means to type feet.[2,3] Our findings are in agreement with Nielsen et al. who noted that static measures are not predictors of midfoot motion with walking.[3] This study's findings help to underscore the complex nature of the intersegmental motions of the foot and the limitations of kinematic foot models currently in use.

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Table 1. Pearson correlation coefficients for the 3D intersegmental angular displacement range of motion.

		Midfoot			Rearfoot			SND[d]
		In/Ev[a]	Ab/Ad[b]	P/D[c]	In/Ev[a]	Ab/Ad[b]	P/D[c]	
Midfoot	In/Ev[a]	1	0.237*	0.270*	0.001	0.004	0.085	0.098
	Ab/Ad[b]		1	0.395^	0.178#	-0.052	0.016	0.090
	P/D[c]			1	0.133	0.068	0.156	0.187#
Rearfoot	In/Ev[a]				1	0.204#	-0.019	0.258^
	Ab/Ad[b]					1	0.256*	0.302
	P/D[c]						1	0.213
SND[d]								1

[a] – Inversion / Eversion, [b] – Abduction / Adduction, [c] – Plantar / Dorsiflexion, [d] – Standardized navicular drop ((non weight bearing navicular height – weight bearing navicular height/ foot length) X 100)).

* - $p < .01$, ^ - $p < .0001$, # - $p < .05$

In-vivo foot kinematics: definition of a fluoroscopic gold standard for the evaluation of marker-based protocols

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INTRODUCTION

The quantification of foot kinematics is an issue in a large number of pathologies. The great clinical interest is demonstrated by the number of protocols recently proposed. These protocols are prone to accuracy limitations like any marker-based protocol, but even more due to the deformability of the foot throughout the gait cycle. Therefore, a validation or at least a quantification of the accuracy of foot protocols is required to assess their adequacy to fulfil the requirements for clinical applications. Therefore, a method for the accurate quantification of foot kinematics in physiological conditions, without limitations to range of motion and skin sliding, is necessary.

3D fluoroscopy (2-3) can provide the required accuracy. This technique was previously applied to the different joints. It can accurately (in the order of 1° and 1mm) reconstruct the relative kinematics of bony segments using an anatomical model of the relevant bony segments, but requires function-related models for the quantification of accurate relative kinematics of compound bony segments (3-4).

In foot kinematics, the rear-, mid- and forefoot are compound segments, intrinsically deformable. The purpose of the present study is the definition of a fluoroscopic gold standard based on a functional-anatomical model for the assessment of marker-based foot protocols.

METHODS

One subject (female, 26 years, 174 cm, 61 kg) was analysed (Fig. 1). The kinematics of foot and ankle was synchronously acquired using stereophotogrammetry (SMART, BTS, Italy) and fluoroscopy (Sirecon 40hd, Siemens). Reflective markers were positioned according to Sawacha et al. (1). Flex-extension and Inv-eversion cycles were acquired, together with neutral and maximal flexion, extension, inversion and eversion static postures. Simplified movements were used to analyze the specificity of the modeling approach.

Bone models of the foot were reconstructed from RMN scan, and the anatomical-model based alignment used to estimate bony segments kinematics. A function-based model was then adopted for the reconstruction of the kinematics of the 3 foot segments. Functional axes of each segment were associated to specific anatomical features of the relevant bony segments, e.g. for the forefoot, the antero-posterior axis was associated to the long-axis of the first ray and the vertical one to the plane containing the first and the fifth ray.

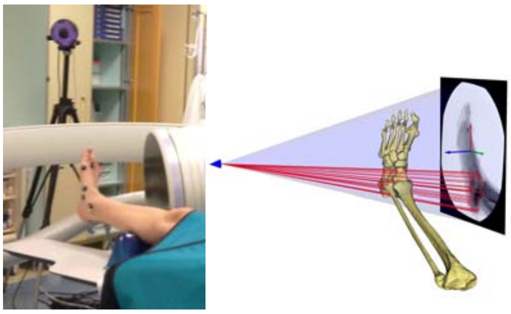

Figure 1: Synchronous acquisition of foot kinematics with fluoroscopy and stereophotogrammetry.

RESULTS AND DISCUSSION

The preliminary results show that the function-based model constrained to bony segment kinematics provides a good specificity in describing the relative kinematics of foot sub-segments, while the kinematics of the different bony segments within the sub-segments can hardly be generalised, without the proposed functional approach. The present methodology is ongoing further evaluation, resulting a promising tool for the evaluation of marker-based foot protocols.

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Can Heel Rocker Dynamics During Stance in Gait be Driven via Shank Kinematics?

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INTRODUCTION

During the heel rocker mechanism at the beginning of the stance phase of gait, the heel becomes a deformable fulcrum about which the foot rolls to the ground as the body weight is transferred to the stance limb [1]. We hypothesized that when ankle motion is restricted, natural heel rocker dynamics can be achieved by combined shank rotation and custom orthosis or prosthesis heel rocker curvature. The purpose of this study was to explore the heel rocker mechanism to examine the feasibility of driving heel rocker dynamics via shank kinematics.

METHODS

Bilateral, lower extremity movement analysis data were collected on five healthy subjects (age: 24.6±2.9 yrs, height: 1.73±0.1 m, mass: 70.6±12.2 kg) walking barefoot at a scaled normal walking velocity of 0.8 statures/sec. Kinematic data were collected at 120Hz using a 6-camera optical motion analysis system. Kinetic data were collected at 360 Hz from four forceplates. Kinematic and kinetic data were filtered at 6 Hz and 25 Hz, respectively, using a zero-lag low-pass Butterworth filter. Data during the stance phase of the right leg were analyzed. Specifically, shank, ankle and foot angles and ankle moment were examined.

RESULTS

The data revealed two distinctly different heel rocker intervals: a kinematic interval from heel strike (HS) to foot flat (FF) and a kinetic interval from HS to the termination of the ankle joint dorsiflexion moment (M0). FF occurred at an average of 14% of stance. During the kinematic interval, the foot and shank rotated through 15±2.5° and 12±2.2°, respectively (Figure 1). The ankle reached peak plantarflexion (PF) of 13±2.9° before the end of the kinematic interval (Figure 2). A dorsiflexion (DF) ankle moment peaked at 0.15±0.03Nm/kg half way through the interval and reached a 0.06±0.04Nm/kg at FF (Figure 3). M0 occurred at an average of 16±2.4% of stance. The shank continued to rotate after FF, for a total of 15±2.4° during the kinetic interval (Figure 1). From FF to M0, the ankle dorsiflexed (Figure 2) and the ankle moment went to zero as it transitioned from DF to PF (Figure 3).

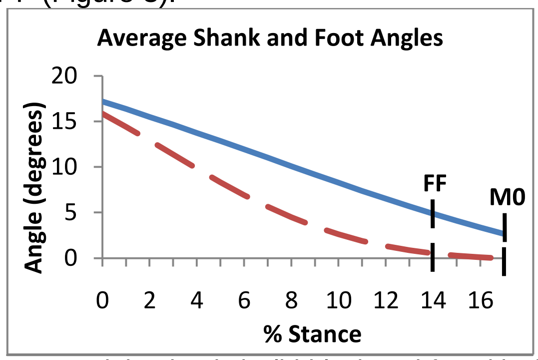

Figure 1: Right shank (solid blue) and foot (dashed red) angles relative to the lab from right HS to M0.

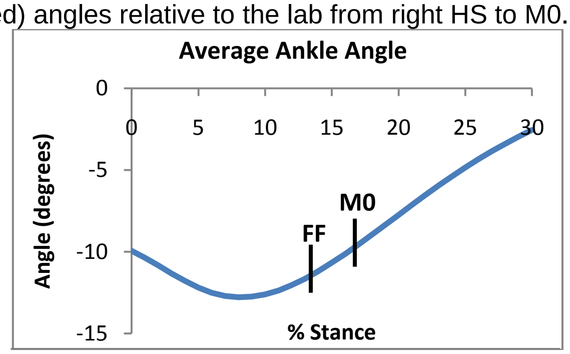

Figure 2: Right ankle angle during beginning of stance. FF and M0 marked.

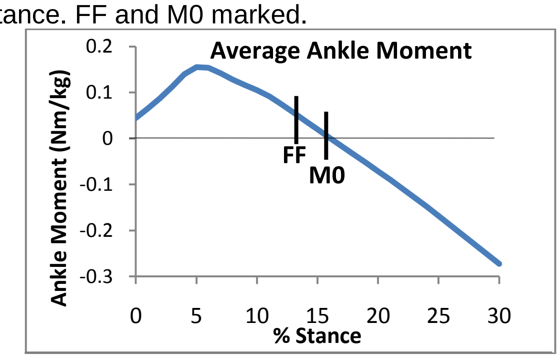

Figure 3: Right ankle moment during beginning of stance. FF and M0 marked.

DISCUSSION

The results indicate ankle motion contributes minimally to the heel rocker mechanism, demonstrating the feasibility of driving heel rocker kinetics with shank rotation.

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The effect of two types of lateral wedged insole on the knee and ankle.

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INTRODUCTION

Lateral wedge insoles have been found to reduce the coronal moments affecting the knee in patients with medial osteoarthritis (OA) of the knee joint [1]. Reports have indicated that due to the increased ankle valgus moment caused by the lateral wedge, some subjects complained of pain in their ankle joint. Therefore, a new design of lateral wedge insole with an arch support was designed with the purpose of the study to assess the effect of two different types of lateral wedged insole on the kinematics and kinetics of the knee and ankle.

METHODS

15 healthy subjects participated in the research. All the subjects were asked to wear standard shoes (ECCO Zen) and three conditions were investigated: (a) standard shoes only (b) Traditional Lateral Wedge: TLW (c) Arch Supported Lateral Wedge: ASLW. A ten-camera motion analysis system (Vicon 612) capturing at 100 Hz and two bilateral force plates (Kistler 9286) capturing at 3000 Hz were used for the data collection. Marker placement was based on the Calibrated Anatomical System Technique [2] with cluster plates over the shank, thigh, pelvis and foot. Anatomical reference frames were modeled to represent the lower limb as segments in the local coordinate system. A Repeated measures ANOVA with the Bonferroni adjustment with the significance level set at 0.05.

RESULTS

Both of the lateral wedged insoles showed significant reductions in the peak value of the External Knee Adduction Moments (EKAM) and the Knee Adduction Angular Impulse (KAAI) with larger ankle eversion moments as expected ($P<0.05$). The arch supported lateral wedge showed smaller ankle eversion angles and a higher foot progression than the traditional lateral wedged insole ($P<0.05$).

DISCUSSION

Both lateral wedges were effective at reducing the EKAM but a reduced ankle eversion and increased foot progression was seen in the ASLW. These results suggest that the ASLW whilst reducing the loads on the knee also helps to create a better foot function and may alleviate some of the ankle problems found in other studies. The new design of lateral wedged insole will now be taken forward into a group of OA patients to determine the clinical and biomechanical effects in this population.

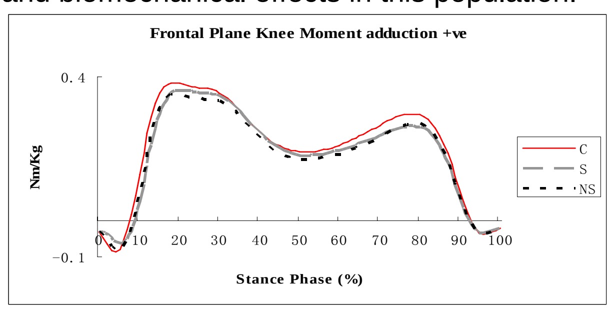

Fig 1: External knee adduction moment in three conditions. C=control condition S=Arch supported lateral wedge NS=traditional lateral wedge (non-supported)

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Table 1.0 – Biomechanical outcome measures for the Knee and ankle for the three conditions

Parameter	Group (C) Mean	Group (S) Mean	Group (NS) Mean	P value C vs S	C vs NS	S vs NS
Knee adduction moment peak (early stance) (Nm/kg)	0.406±0.026	0.374±0.023	0.372±0.023	**0.011**	**0.003**	1.000
Knee adduction moment peak (late stance) (Nm/kg)	0.316±0.021	0.281±0.021	0.284±0.021	**0.027**	**0.037**	1.000
Knee adduction angular impulse (Nm/kg/s)	0.140±0.010	0.127±0.010	0.123±0.010	**0.001**	**0.000**	0.317
Ankle peak eversion (Degrees)	2.889±0.313	-3.084±0.296	-3.853±0.328	**1.000**	**0.004**	**0.002**
Ankle progression (Degrees)	0.143±1.048	-2.073±1.004	-3.163±1.054	**0.000**	**0.000**	**0.009**

The Effect of Foot Invertor Fatigue on Rearfoot Motion and Tibial Rotation During Running and Turning

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INTRODUCTION :

Many overuse injuries in running are associated with excessive pronation of the foot during the stance phase, although the etiology of leg injury is uncertain and certainly multi-factorial. Fatigued or weak foot invertors would theoretically be less effective in controlling these calcaneal motions during locomotion and other dynamic activities [1]. Excessive calcaneal pronation can lead to excessive internal rotation of tibia [2] and in turn, possible problems at the knee joint. It is hypothesised that invertor fatigue reduces control of calcaneal pronation and increases tibial internal rotation during initial contact of running and turning. Objective: To assess the effects of foot invertor fatigue on frontal plane movement of the hind foot and tibial transverse plane movement during running and side step cutting.

METHODS : 10 healthy, recreationally active individuals (Mean age = 23.9 yrs, Ht = 175.68 cms and Mass 69.7 kgs) performed 10 trials of both barefoot running and a 90 degree cutting task and then 5 trials after a foot invertor fatiguing protocol. Fatigue was induced by repeated stretching and controlled relaxing of a rubber band (Thera-Band ®) using a combined movement of resistive plantar flexion and inversion movement during cross-legged sitting. During the straight running and cutting tasks 3D kinematic data of the foot and shank were captured using six opto-electronic cameras (Qualisys ProReflex system). 3D coordinate data were filtered with a fourth order zero-lag, low pass butterworth filter at 12Hz cut-off frequency. All the data were exported to Visual 3D (C-Motion Inc) to calculate calcaneal eversion and tibial rotation during the initial landing phase of the right foot.

RESULTS : The patterns of movement of the rearfoot and tibia were very similar after the fatigue protocol. The rearfoot and tibial rotation range of motions during running and cutting before and after fatigue was not statistically significant. Foot invertor fatigue did not increase peak calcaneal eversion angle and peak tibial internal rotation angle during running. The only significant change observed with fatigue was an increase in the average peak tibial internal rotation angle during cutting of about 0.9 degrees. This is a relatively small average group effect but the influence of fatigue varied between subjects. Half the subjects demonstrated a clear increase in peak tibial internal rotation angle during cutting. Average tibial rotation range of motion during cutting before and after fatigue was 10.1 and 11.3 degrees, respectively, which was a non-significant change but, again, larger individual changes could be observed with fatigue.

DISCUSSION: The results revealed that foot invertor fatigue did not change either rear foot eversion or tibial internal rotation motion during running. This is in agreement with previous work by Christina et al [3] during shod running after using an Elgin leg/ankle exerciser to induce fatigue and the ranges of rearfoot motion were similar to that study.
One possible explanation for the lack of change in rearfoot motion may be possible compensation by other muscles or muscle groups. The fatigue-induced change in tibial internal rotation motion during cutting may indicate dissociation between the coupled motion of the rearfoot and tibia. This coupling is typically linked to a distal to proximal control but perhaps with the cutting maneuver the tibial movement is under greater influence of proximal segmental movements. This notion is supported by the work of Besier et al [4] and McLean et al [5] who found an increased internal rotation moment associated with increased activation of the quadriceps in the flexed knee during cutting.

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Walking Cadence Affects the Kinematics of the Joints of the Foot

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INTRODUCTION

Pedobarographic analyses [1,2] have shown that some important kinematic modifications must occur along the longitudinal arch of the foot when walking at different walking speeds. These studies revealed significant increase in peak plantar pressure at the heel and forefoot and decrease of pressure at the midfoot region with increasing walking speed. Aim of this study was to assess whether walking speed significantly affects the kinematics of the main joints of the foot during barefoot walking.

METHODS

The shanks and left feet of ten subjects were instrumented with 15 reflective markers attached to the skin with double-sided adhesive tape, according to Leardini et al. [3]. A six-camera motion capture system (Qualysis, Gothenburg, Sweden) recorded the kinematics of the main joints of the foot when walking at three self-determined walking cadences: slow, normal and fast. A Kistler force plate (Kistler Instruments Ltd., Hook, Hampshire, UK) was employed to record ground reaction forces at 500Hz. Statistical differences between time-history of the rotations at each sample within the normalized stance phase duration and between relevant ranges of motion (ROM) from different walking speeds were tested using the non-parametric Mann-Whitney U-test [4].

RESULTS

The slow- and normal-cadence datasets showed similar profiles of joint rotation in the three anatomical planes, but significant differences were found between these and the fast-cadence. At all joints, frame-by-frame statistical analysis revealed increased dorsiflexion from heel-strike to midstance ($p<0.05$) and increased plantarflexion from midstance to toe-off ($p<0.05$) with increasing cadence (Figure 1). The fast-cadence kinematic data showed decreased ROM in the sagittal plane between forefoot and rearfoot (3.2°±1.2° at fast cadence; 2.0°±0.8° at slow cadence; $p<0.05$) during midstance.

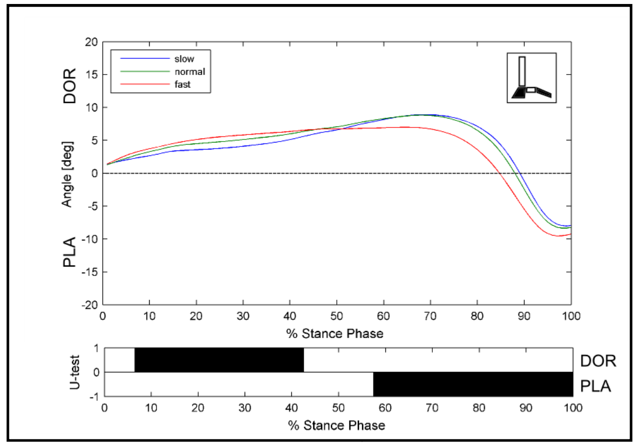

Figure 1 Top: intersubject mean sagittal plane rotations [deg] between forefoot and calcaneus at three walking cadences. **Bottom**: frame-by-frame statistical analysis reveals differences between samples of rotations from slow- and fast-walking datasets.

DISCUSSION

Walking cadence affected consistently rotations at the ankle, midtarsal, and tarsometatarsal joints, i.e. increased cadence resulted in increased dorsiflexion at early stance and increased plantarflexion at late stance, accompanied also by a general increase of inversion. The present detailed kinematic analysis of the foot joints supports the inference of some sort of arch-stiffening mechanism at increasing walking cadence, as previously highlighted by pedobarographic analyses.

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The design of an artificial foot for simulating the unroll of the foot on a robotic gait simulator

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INTRODUCTION

Over 2,5 billion pair of shoes are sold in Europe Annually. A justified selection of shoes, based on static and dynamic shoe properties and taking into account the intentional utilization is essential: inappropriate shoes lead to an increased risk of (stress) injuries.
The ultimate goal of this study is to develop an innovative gait simulator, including a parametric artificial foot, enabling the simulation of the natural behaviour of the human unroll of the foot during walking and running. With this simulator, the influence of different types of footwear on the foot and gait kinematics can be determined.
This abstract discusses the development of such an artificial foot. In the literature, most of the described foot models exists of 4 or 9 segments. [1, 2]

METHODS

Subjects: 10 healthy adult male participants with foot size 43 were selected and gave written informed consent.
Apparatus: Force and kinematic data are collected using an AMTI 1m force plate (1000Hz) and a Codamotion system with 4 CX1 cameras and active markers (400Hz).
Procedures: In a first fase, a 4 segmented model is selected. 12 active markers are attached on the right leg following a predefined marker setup, i.e. 3 markers per segment (Figure 1). Participants are asked to run barefoot (3 à 3,5 m/s) across a 21,5m Tartan track. Data are collected from 10 consisted measurements.
Data analysis: Kinematic and force data were collected and averaged.

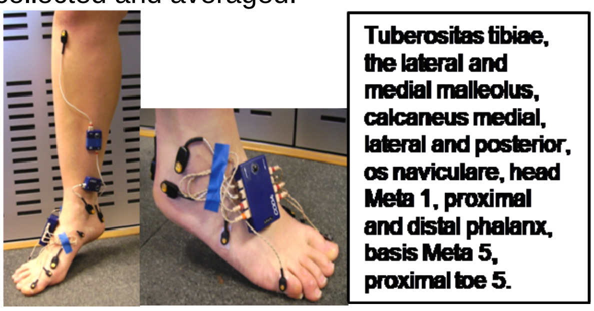

Figure 1: Marker setup

RESULTS

The 4 segments of the first model of the artificial foot are: the lower leg, the rear-foot (calcaneus and talus), the mid-foot and the toes.

The position of the axes is defined based on the literature of foot models and the anatomical bone geometry, allowing all movements necessary to simulate a natural unroll behavior of the foot.
The axis between lower leg and rear-foot is oriented from the lateral to the medial malleolus to allow flexion and extension. The axis for inversion and eversion is derived from literature [3].
An axis between rear-foot and mid-foot is defined with respect to the foot symmetry axis to allow pronation and supination in the mid-foot.
The last axis between mid-foot and toes is oriented from MTP1 to MTP 5.
For validation purposes, data obtained by the measurements are correlated with the behaviour of this 4 segmented foot model.

DISCUSSION

The geometry and functional behavior of the 4 segmented foot model will be optimized based on the measurements. Consequently a first model is built. This artificial foot will be installed on the robotic gait simulator. These results will be used to further optimize the foot model and the artificial foot.

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Kinematic Comparison of Three Ankle-Foot Orthoses for the Conservative Management of Stage II Posterior Tibial Tendon Dysfunction

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INTRODUCTION

Posterior tibial tendon dysfunction (PTTD) is a progressive disorder (stages I – IV), thought to be a leading cause of acquired flatfoot deformity.[1] Surgical management is common however, conservative care with ankle foot orthoses (AFOs) is recommended before surgery is considered. Limited data is available comparing the effects various devices making clinical management difficult.[2] The objective of this study was to compare correction of flatfoot deformity across three commonly used AFOs. The correction of flatfoot deformity (hindfoot inversion, forefoot adduction, higher MLA) is thought to unload the posterior tibialis tendon and supporting ligaments to prevent progression of the dysfunction.

METHODS

Ten subjects (age 63.6±6.8 years) with stage II PTTD walked in the lab under four conditions: (1) shoe only (control condition), (2) shoe with a custom solid AFO (Arizona Co, Mesa, AZ), (3) shoe with a custom articulated AFO (Arizona Co. Mesa, AZ), and (4) shoe with an off-the-shelf AFO (AirLift, DJ Orthopedics). Kinematic data were collected at 60Hz using a 10 camera Vicon 512 motion analysis system (Vicon Inc. CO, USA). The shoes and AFOs were modified with windows to allow placement of reflective marker clusters for tracking a 3 segment kinematic model (1st metatarsal, calcaneus, and tibia).[3] The kinematic dependent variables of interest included; hindfoot inversion, forefoot plantar flexion (reflective of raising the MLA), and forefoot adduction. A two-way repeated measure ANOVA with two factors (orthoses – 4 levels, and stance phase – 4 levels) was repeated for each dependent kinematic variable.

RESULTS

The effect of the AFO on hindfoot inversion (p=0.02) and forefoot plantar flexion (p=0.05) was dependent on the phase of stance (interaction effect) while there was no effect on forefoot abduction (p=0.5).

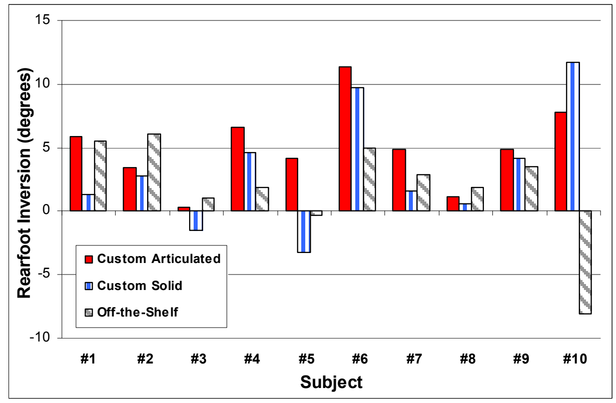

Figure 1. Difference between each AFO and shoe condition on Hindfoot Inversion at midstance phase of gait. Positive values indicate improvement.

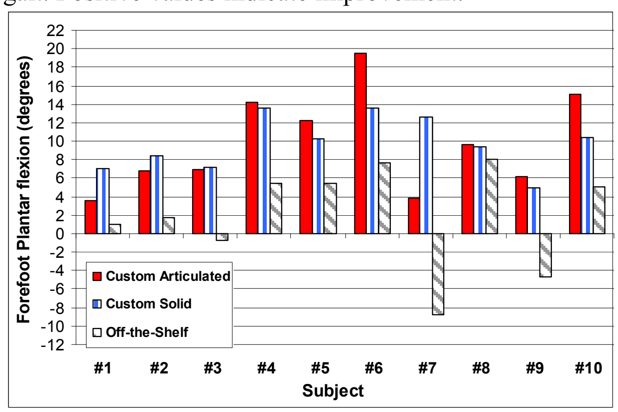

Figure 2. Difference between each AFO and shoe condition on Forefoot Plantar flexion at terminal stance phase of gait. Positive values indicate improvement.

DISCUSSION

Based on these results, the clinical recommendation to use a custom articulated AFO would provide the greatest correction of flatfoot deformity while maintaining ankle motion and push-off function. However, all of the tested AFO designs failed to adduct the forefoot suggesting improved designs are needed.

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Arizona Company and DJ Orthopedic for providing AFO's used in the testing.

Changes in foot kinematics when getting older

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INTRODUCTION

As has been shown in literature the static foot structure and foot mobility changes with ageing [1]. The elder people show flatter foot arch, and a decreased range of motion [1]. A prevailing opinion is that the elderly show more toe-out during walking. Those two gait characteristics could result in differences in subtalar eversion and/or eversion velocity during walking [2].

To our knowledge there is yet no study comparing walking of young with elder people using a multi-segmented foot model. The aim of this study is to describe foot kinematics of younger and older people during walking at self-selected speed using a 6 segmented foot model. In this abstract differences between young and old adults in static foot posture (6 segmented foot) is examined. A flatter foot arch is expected for the elderly. The toe-out position of the foot is examined during walking. For the elderly, more toe-out is hypothized.

METHODS

20 healthy volunteers (10M:10F) participated in the study. Their age varied between 21 and 24 for the younger population (67.2±10.5 kg; 1.72±0.11 m) and between 60 and 68 years for the older group (72.7±9.0 kg; 1.71±0.06 m).

All subjects walked barefoot over a 12 m long instrumented walkway at self-selected speed (young: 1.41±0.15 ms^{-1}; old: 1.27±0.09 ms^{-1}). Kinematic data was collected with a 6 camera opto-electronic system (OQUS 3, Qualysis, Sweden) at 500 Hz. A midstep protocol was used with subjects walking at a self selected comfortable walking speed. The latter was controlled for steady state by means of a laser at 1000 Hz (CMP3-30, Noptel OY, Finland).

A 6 segmented foot model was used, consisting of the lower leg (tibia+fibula), the hindfoot (talus+calcaneus), the midfoot (naviculare+ cuboid+cuneiforme), the medial forefoot (metatarsal 1), the lateral forefoot (metatarsal 2-5) and the hallux (De Mits S et al., submitted i-Fab 2010).

RESULTS

Table 1. A significant difference is found in the orientation of the midfoot segment to the rearfoot segment indicating that the older people have a flatter foot arch. A trend to signifance was found for the orientation of both forefoot segments to the midfoot segment.

		Toe-out angle
Young	Mean	5.82*
	SD	2.97
Old	Mean	11.27*
	SD	3.06

Table 2. Toe-out angle during walking *significant difference young versus old

Table 2. The elderly place the foot 4.5° more in toe-out during walking.

DISCUSSION

These results confirm the findings of Scott et al. as a flatter foot arch is found for the older people. During walking more flatting of the arch was not found. The widespread opinion that older people place their foot in a more toe-out position during walking is proved.

Furthermore, the movement of the 6 foot segments during walking will be examined to give insights in the dynamic functionality of the foot (segments) for young and old people.

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		Rearfoot-Tibia			**Midfoot-Rearfoot**			**Lateral Forefoot-Midfoot**			**Medial Forefoot-Midfoot**			**Hallux Medial Forefoot -**		
		X	*Y*	*Z*	*X*	*Y*	*Z*	*X*	*Y*	*Z*	*X*	*Y*	*Z*	*X*	*Y*	*Z*
Young	Mean	99.5	**14.06**#	3.53	**-49.39***	15.72	12.51	**27.89**#	2.66	-3.42	**24.31**#	-11.92	10.44	18.12	10.11	-4.13
	SD	6.39	7.67	7.42	11.15	5.61	7.30	10.80	7.65	5.93	8.38	5.43	6.12	8.39	4.80	4.52
Old	Mean	96.59	**9.87**#	2.81	**-62.20***	14.56	12.20	**34.70**#	1.00	-0.29	**33.31**#	-10.34	17.88	16.68	13.83	-3.21
	SD	7.22	7.20	5.76	20.41	4.60	5.04	15.49	6.68	9.37	18.65	5.66	11.42	6.99	8.45	2.56

Table 1. Static footposture : joint angles for young and old persons in a 6-segmented foot model
X: dorsiflexion (+) plantar flexion (-); Y: abduction (+) adduction(-); Z: eversion (+) inversion (-)
* significant difference between young and old persons ($p<0.05$); # trend to significance ($0.05<p<0.085$)

Personalised Orthotics for Rheumatoid Arthritis (po*4*ra): Development of 3D Kinematic and Plantar Pressure Based Design Rules for Commercial CAD Software

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INTRODUCTION

The aim of this study was to develop a design rules framework based on the therapeutic goals for orthotic management in *early* rheumatoid arthritis (RA). This was based on matching design options in a commercial foot orthotic CAD software platform with personalised 3D intersegment foot kinematics and plantar pressure distribution.

METHODS

Orthomodel CAD software (DELCAM, Birmingham, UK) was used to design personalised orthotics based on 3D surface scans of neutral plaster casts. Orthomodel design options permit customisable functional elements including arch height, heel pitch and lift, rear- and fore-foot posting, skive, heel cups and cut-outs. Video-based 3D gait analysis was undertaken using an 8-camera MOCAP system (Oqus, Qualysis, Gothenburg, Sweden) in 5 non-RA adult subjects who were current foot orthotic users ('first-onto-man' phase I trial design). Plantar pressure measurement was performed using an EMED platform (Novel-X, Novel GmbH, Munich, Germany). Intersegment kinematics were analysed using a multi-segment foot model in Visual3D software (C-Motion Inc., MD, USA). Peak rearfoot eversion, peak forefoot inversion, navicular height and peak pressure at metatarsophalangeal (MTP) joints I-V were used to develop rules for 4 design elements: rearfoot posting, forefoot posting, arch height and forefoot cushioning. Rules were developed from archive kinematic/ pressure data for n=52 healthy adult subjects [1].

RESULTS

An example of a design rule for rearfoot posting and application in a single subject is shown in Figure 1. Peak eversion angles created extrinsic medial posts ranging from 0-12°. All 5 devices had intrinsic forefoot posts at 0° (neutral). Navicular height motion during stance created medial arch profiles with heights from 26.4mm to 28.6mm. Peak MTP pressures recorded in excess of 2 standard deviations above normal values indicated forefoot cushioning in 4/5 subjects. Abnormal pressure values varied from 565kPa at MTP V to 1125kPa at MTP II.

DISCUSSION

In early rheumatoid arthritis (RA), inflammatory destruction of joints and soft-tissues of the foot and ankle indicates the use of custom-made rigid foot orthoses [1]. The approach described herein uses personalised biomechanical data to drive design features within a commercial software platform. These prototype devices will be tested in phase II and III clinical trials.

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ACKNOWLEDGEMENTS

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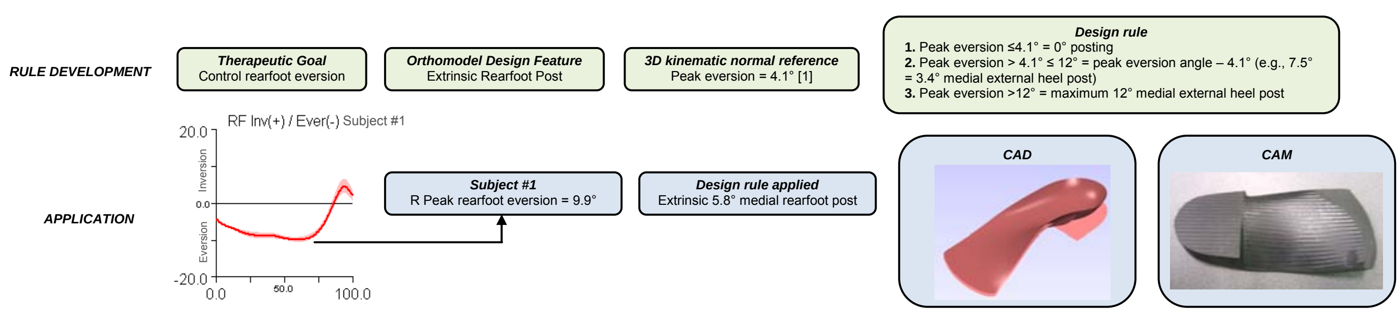
Figure 1: Design rule development and application



Gait Analysis after Bilateral Total Ankle Replacement – A 12-month Follow-up Case Study

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INTRODUCTION

In recent years, total ankle replacement (TAR) becomes the alternative of arthrodesis to treat severe ankle osteoarthritis (OA) to reduce pain and restore ankle joint function. The Salto Talaris™ Anatomic Ankle (STAA) (Tornier Inc. France) is a fixed bearing TAR that could mimic the anatomy and flexion/extension movement of the natural ankle joint [1]. Although the previous generation of the Salto Talaris™ TAR had received high satisfaction from patients [2], it is not clear if the STAA TAR system could help patients regain normal ankle strength and restore normal ankle function during daily activities. Laboratory gait analysis is a valuable method used to assess the function of artificial joints and help identify potential mechanical problems associated with the joint replacements [3]. The purpose of the study was to evaluate gait parameters and ankle joint mechanics during walking at different time points after the surgery.

METHODS

This is a single-subject design case study. A 74-year-old male (body mass: 66kg; body height: 173cm) had advanced right ankle OA and underwent a TAR with an STAA system. The same surgeon had installed a different TAR system (Agility™ Depuy orthopedics Inc.) in his left ankle three years prior to the right ankle TAR (2005). The patient performed level walking at self-select pace on five different occasions in a gait laboratory: one week before the right TAR surgery, three-month, six-month, nine-month, and twelve-month after the TAR surgery. 3D kinematics and kinetics data were collected. Gait parameters such as walking speed, single-support time, and step width were analyzed. For both the involved (right) and non-involved (left) limbs, ankle joint mechanics such as range of motion during stance and plantar-flexor moment at push-off were analyzed.

RESULTS

Prior to the surgery, compared to the non-involved limb, the involved limb had significant less ankle range of motion (5 vs. 15 deg), plantar-flexor moment (0.99 vs. 1.07 Nm/kg), and single-support time (0.32 vs. 0.42 s). The walking speed was slow at a pace of 0.65 m/s. Three months after the surgery, single-support time of the involved limb was increased to the level of non-involved limb (0.42 vs. 0.41 s); the step width was reduced to 0.24 m from 0.32 m; the ankle range of motion of the involved limb was increased to 18 deg. The ankle plantar-flexor moment of the involved limb continued to increase until 6-month post-op and was greater than the non-involved limb after 9 months of surgery (1.40 vs. 1.23 Nm/kg). Walking speed continued to increase to 1.5 m/s at 9 months post-surgery.

DISCUSSION

Prior to the surgery, the patient depended on his non-involved limb to perform level walking with greater plantar-flexor moment and longer single-support time. The increased step width seemed to be a strategy to gain stability during walking. Just three months after the surgery, the involved ankle joint had gained normal range of motion and plantar-flexor moment during stance, which helped increase the single-support time and improve the walking speed. In addition, the reduced step width indicated that stability was improved during walking. Interestingly, the patient started to rely on his involved limb after 9 months of surgery as a greater plantar-flexor moment and longer single-support time were presented by the involved limb. In conclusion, the STAA TAR system can help patients regain ankle strength and restore ankle joint function during level walking. The Agility™ TAR system also showed satisfactory mid-term outcomes after three years of surgery.

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Patients with Medial Tibial Stress Syndrome are characterized by low foot kinematics complexity and high muscle activation complexity

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INTRODUCTION

Non-linear analysis has previously been applied to various biological signals and has revealed a decrease in the complexity of the physiological system output in relation to ageing and disease. A general decrease in signal complexity in relation to pathology has led to propose the "loss of complexity hypothesis" [1] where high variability has been described as healthy and low variability associated with pathology. The purpose of the study was to compare the structure of complexity in foot kinematics and EMG-signals obtained from m. soleus and m. tibialis anterior during walking between patients with medial tibial stress syndrome (MTSS) and a healthy control group.

METHODS

Fourteen patients diagnosed MTSS and 12 healthy controls were sequentially included from an orthopaedic clinic. MTSS was defined as continuous pain in the tibial region, exacerbated during repetitive weight bearing activity, and localized pain detected by palpation along the distal two thirds of the posterior medial tibia. EMG signals were recorded from the m. tibialis anterior and m. soleus using the recommendations of SENIAM project. EMG signal corresponding to the swing phase from the gait cycle was excluded from the EMG analysis. Twenty consecutive gait cycles of EMG and kinematics were recorded while subjects were walking on a treadmill at a self-selected pace. The number of gait cycles along the whole recording time was in the EMG data divided into 4 intervals representing (0-25-50-75-100%) of the contact time. The EMG signal were digitally band-pass filtered (Butterworth, 2nd order, 10-400 Hz) and a notch filter (2nd order Butterworth band stop with rejection width 1 Hz centered at power line frequency (50 Hz)) was applied. A 3D multi video sequence analysis procedure was employed to assess midfoot and rearfoot kinematics during the stance phase. Kinematic data from the stance phase was divided into 1 interval representing (0-100%). Permuted sample entropy (PeSaEn) was used as a measure of complexity of the time series from the EMG data and kinematic data. Two-way ANOVA was applied for PeSaEn values introducing time and subject groups (healthy, MTSS) as factors in analysis of EMG data.

RESULTS

The results showed that the EMG signal from both the m. soleus and m. tibialis anterior in patients with MTSS was characterized by higher structural complexity than healthy controls ($p<0.001$). The opposite was found in midfoot and rearfoot kinematics. The kinematic data from subjects with MTSS was characterized by a lower structural complexity than healthy controls ($p=0.01$).

DISCUSSION

In healthy subjects an increasing complexity at the kinematic level was associated with a decrease at the muscle level. In subjects with MTSS a decrease in complexity was observed in terms of movement but an increased complexity was found at the muscle level. These findings fall within the conceptual framework of complexity tradeoffs between the macroscopic and microscopic levels of a system [2]. The study indicates that MTSS alters system organization of muscle activity and foot kinematics. Future studies should investigate if the inverse relationship between the structural complexity of kinematics and muscle activation change following rehabilitation programmes.

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ACKNOWLEDGEMENTS

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Kinematic and Electromyographic Asymmetries of Ankle Joint During Drop Landing

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INTRODUCTION

Landing is a dangerous activity to ankle joint among many intensive high-risk sports. To our knowledge, biomechanical asymmetries have not been studied during double-leg landing. The objective of this study was to research on the side-to-side differences in ankle kinematics and Electromygram (EMG) during drop landing.

METHODS

16 healthy adult participants dropped from three different heights (low: 0.32 m; medium: 0.52 m; and high: 0.72 m). Position data of the lower-extremities were collected by motion capture system. With the software Visual3D, angular displacement and velocity peaks of ankle dorsiflexion, inversion and abduction were calculated. Surface EMG signals were measured in tibialis anterior (TA) and lateral gastrocnemius (LG) of both limbs. The average EMG amplitudes were calculated over a period of 100 ms before and after touch-down.

Two-way ANOVA and Tukey's HSD post hoc analysis was used to determine significant effects lateral (2 levels) and dropping height (3 levels) on biomechanical variables. Statistically significant difference was defined with $p<0.05$.

RESULTS

The comparisons of the measured variables were listed as Table 1.

DISCUSSION

Generally speaking, both factors of lateral and dropping height affect more greatly on the peak angular velocities than the displacements. All kinematic variables are greater in the right ankle. This finding is very meaningful, since greater joint motion has a potential correlation with greater injury risk.

Dropping height affects more greatly on LG, while the lateral factor affects TA more greatly. Left TA produced significantly greater EMG activities both before and after touch down during landing. TA was one of the main ankle flexors, and played important role during dorsiflexion. Therefore, it can be concluded that the left lower-extremity during drop landing provides greater ankle flexor activities to effectively control the ankle motion. Considering that most people have stronger left lower-extremity [1], the right ankle should be in more injury risk.

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ACKNOWLEDGEMENTS

Supported by NSFC Grant 10925208 and the National Key Lab of Virtual Reality Technology.

Table 1: The comparisons of the measured variables with two-way ANOVA (mean ± standard deviation, n=16)

VARIABLE		Left			Right		
		Low	Medium	High	Low	Medium	High
Peak Angular Displacement (°)	Dorsiflexion	11.2±6.3	12.7±9.1	16.4±10.1	11.5±9.9	12.9±10.6	17.9±10.6
	Inversion	16.5±5.4	18.9±6.6	21.0±8.2	18.9±7.2	20.4±8.6	21.6±6.8
	Abduction†	16.2±9.9	17.6±7.5	22.8±7.6	17.0±9.1	20.3±9.0	24.8±7.1
Peak Angular Velocity (°/s) †	Dorsiflexion*	993±390	1209±458	1304±553	1151±431	1457±441	1615±535
	Inversion	288±105	318±104	415±192	307±106	351±100	481±197
	Abduction*	363±142	437±169	474±182	432±130	534±134	562±164
TA EMG Amplitude (μV)*	Pre-landing	250±250	278±259	301±219	133±99	165±152	222±240
	Post-landing	567±524	804±723	911±598	365±278	416±292	511±350
LG EMG Amplitude (μV)†	Pre-landing	76±64	120±84	148±128	73±54	113±93	159±106
	Post-landing	237±204	322±280	421±414	259±195	303±270	418±263

* indicates significant bilateral difference with $p<0.05$ and mean SA>5%, and † indicates significant difference across three dropping heights with $p<0.05$. No significant interaction is found between factors of lateral and height with $p>0.05$.

Center of Pressure Trajectories During Landings With a Secondary Redirection Task

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INTRODUCTION

During sport-specific landings, athletes effectively distribute large reaction forces experienced immediately after contact (impact) and redirect their vertical momentum at touchdown as quickly as possible [1-4]. The purpose of this study was to determine how systematic modifications of a secondary task at the whole body level affect preparation and execution of a sport-specific land and go task. We hypothesized that changes in secondary task direction would result in modifications of the roles of the right and left legs during impact, transition, and push phases. Block, land and go tasks with increasingly more posterior secondary task direction requirements were hypothesized to involve redistribution of the ground reaction forces (GRF) between legs and modifications in the center of pressure (CP) to accommodate for increasingly more rotation of the whole body prior to push.

METHODS

Six female collegiate Division I volleyball players (height: 1.89 m (0.03); weight: 775.5 N (43.1) cleared for participation provided written informed consent in accordance with the Institutional Review Board. Participants performed a series of volleyball block landing tasks along a net followed immediately by a secondary task (STOP; LEFT: 90˚/parallel to net, DIAG: 135˚ to back and left, BACK: 180˚/ perpendicular to net) using standard three-step footwork at a self-selected speed. Three-dimensional kinematics (200 Hz, NAC Visual Systems, Burbank, CA, USA) and reaction forces (dual force plates, 1200 Hz, Kistler, Amhurst, MA, USA) for each foot were measured. Markers on the second metatarsal and calcaneus were used to define the longitudinal axis of the foot as a projection of the foot vector onto the force plate.

RESULTS

Initial foot placement during the landing was determined during the flight phase and foot orientation prior to the push was determined

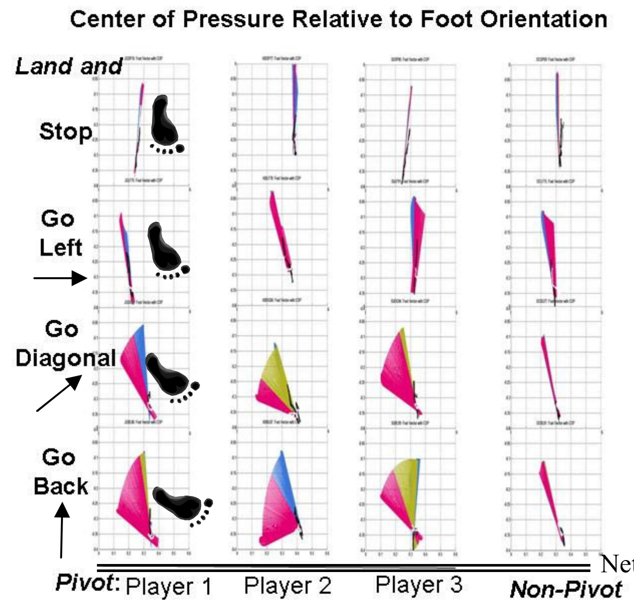

Figure 1. Right foot vector during each phase (impact: blue, transition: green, push: pink) and center of pressure trajectory (impact: black; transition: white; push: grey). Note players who pivot reorient foot during different phases of the movement.

after the impact phase. Players (1-3) using a pivoting action to reorient the foot did so during different phases of the movement. (Figure 1).

DISCUSSION

Between task differences in CP relative to the foot and reaction forces between legs [3] will likely influence the mechanical demand imposed on the ankle (net joint moment) as well as the knee and hip[1,4].

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Foot loading after several days of long distance walking.

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INTRODUCTION

The popularity of long distance walking (LDW) has increased rapidly over the last decades. It is a common observation that during LDW many participants develop overuse injuries and foot complaints. Intense running significantly increases peak pressure under the forefoot during walking [1] but the effects of LDW, particularly after several days of walking, are unknown and may be very different. Therefore, the aim of this study was to investigate the influence of LDW on plantar pressure distribution.

METHODS

Sixty two participants of the International Four Day Marches of Nijmegen (IFDM) with no prior history of foot complaints participated in this study. On average, the male participants (N=30) walked a total distance of 199.8 km in 4 consecutive days (161.5 km for the females). Plantar pressure was measured when walking barefoot over a pressure plate (Rsscan International Belgium) three times per foot. Measurements were done one day before the IFDM (baseline) and each day after finishing the walking (post-test I-IV). The normalization method of Keijsers et al. [2] was used to analyse plantar pressure distribution per sensor of each foot per measurement day. Significant differences between pre and post-tests were detected by a paired t-test with adjusted p-level (Bonferroni correction for day and sensors).

RESULTS

Eighty eight percent of the 50 participants who finished the IFDM developed foot complaints. The most common foot complaints were blisters (26%), forefoot pain (22%), heel pain (7%) and a combination of forefoot and heel pain (18%). The short-term effect of LDW on plantar pressure is a decreased loading of the forefoot and increased loading of the metatarsal head III-V ($p < 0.001$). At all stages, but especially near the end, there was significantly more heel loading ($p<0.001$) (Figure 1). Furthermore, the COP significantly displaced in posterior direction but not in mediolateral direction after LDW. Contact time increased slightly from 638.5 (+/-24.2) to 675.4 (+/- 22.5)($p < 0.05$).

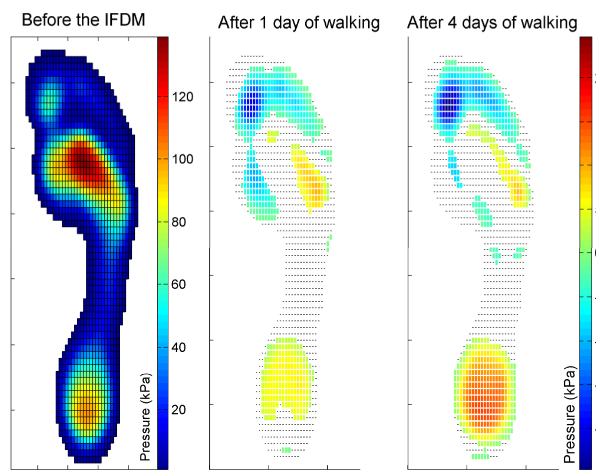

Figure 1: Distribution of the mean plantar pressure for the total group before the IFDM and after 1 and 4 days of walking. The 2nd and 3rd figure show sensors that were significantly different ($p < 0.001$) from the baseline test. The small black lines in these figures indicate sensors with no significant difference in plantar pressure

DISCUSSION

The increased heel loading and decreased loading of the toes found after LDW indicates a change of walking pattern with less roll off of the foot. It is assumed that these changes reflect avoidance of loading of those parts of the foot that are most vulnerable to damage due to overloading. Therefore, wearing shoes or insoles which decrease pressure underneath the forefoot might prevent for foot complaints when walking long distances.

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Validation and relevance of methodologies for anatomical and geometrical identification of plantar sub-regions in baropodometry

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INTRODUCTION
A study has previously been presented, assessing an anatomically based selection of foot sub-regions, especially in presence of pathologies which alter foot contact with the ground [1]. The methodology, described in [2], relied on the integration of the Oxford Foot Model (OFM) with pressure measurements obtained from a prototype piezo-dynamometric platform. The present study attempts to generalize the methodology to more widely available technology. A validation study to compare the proposed anatomical selection with commonly used geometrical selection techniques is presented. Data acquisition and processing is still in progress. Some preliminary results of the study are reported.

METHODS
A Vicon MX system (Vicon, Oxford, UK) was used to track markers on the foot, placed according to the OFM protocol. A Novel Gmbh EMED platform and software package was used to obtain and process pressure data. An algorithm based on previous validation studies [3] was implemented within EMED software. Co-ordinates of marker positions on the foot were projected onto the footprint at mid-stance. Five foot regions were then defined based on marker positions: medial and lateral hindfoot, midfoot, medial and lateral forefoot (toes included). The geometrical selections of sub-areas which best corresponded to the anatomical method (AM) where identified. Two geometrical selections were implemented: G1 was based on the bisecting line of the foot and the perpendicular lines at 27% and 55% of foot length; G2 was based on the line from the center of the heel to the center of the second toe, and on the perpendicular lines at 27% and 55% of foot length. Up to now data were acquired from 5 healthy children and 3 children with foot deformity, with at least 3 footprints per foot. The time curves of vertical force (F) and peak pressures (P) were assessed to determine differences between the three selection methods for each of the 5 sub-areas.

RESULTS
% differences between AM-G1 and AM-G2 ranged 5-70% (mean 27%) for P and 2-50% (21%) for F for pathologic footprints, and 1-20% (7%) and 0-25% (9%) respectively for healthy footprints. For the latter, greater variability was found in the midfoot and forefoot regions, and for AM-G2 compared to AM-G1.

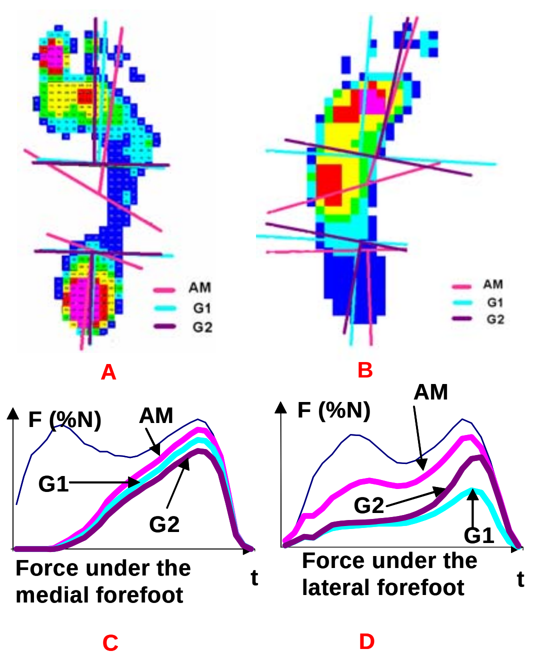

Figure 1: Healthy (A) and pathologic (B) feet. Force curve obtained with the 3 methods for healthy medial forefoot (C) and pathologic lateral forefoot (D).

DISCUSSION
Acceptable differences were found between the AM and the different G methods for healthy footprints. This suggests that G methods can accurately identify sub-regions in healthy footprints when the G method is optimized. For example, it appears G1 is better than G2 for the forefoot, while G2 is better for the hindfoot. As expected, AM is more meaningful for footprints where total foot contact is not present.

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FORTY-YEAR PERSPECTIVE ON FOOT AND ANKLE SURGERY
An Attempt to Have Biomechanics or Functional Anatomic Reconstruction Introduced to Foot and Ankle Surgeons

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My first interest in biomechanics probably goes back almost 60 years. I began Whitman College in 1953 on a so-called 3-2 Engineering program with Cal Tech or Columbia and began as a math/physics major. In my second year of taking a theoretical math course, I could see that "theoretical" was going to be a major stumbling block. My mind did not work in the abstract, only with what I could see and touch. A change to Biology brought me into a course on Comparative Anatomy and the theory of evolution. The evolutionary changes were striking in the foot. Later, in medical school but particularly in residency, the foot and ankle became an increasing interest because of the complex mix of evolutionary modifications to understand the biomechanical and functional anatomic needs of being bipedal.

My first foot surgery was 51 years ago as a surgical extern at Doctors Hospital in Seattle in about 1959. I had an internship at Harborview in 1961-62 and then a Navy practice as a General Practitioner with a great amount of treatment of outpatient minor foot and ankle problems. Residency at the University of Washington followed from 1965-69 with 1967-68 spent at the Shriners Hospital in Spokane, where we saw congenital problems in the foot and lower extremity. Later I took a Fellowship in children's neuromuscular surgery in Sheffield, England, with the world's authority on polio and neuromuscular surgery. The pertinence of the latter is that I was exposed to biomechanical principles and problems of muscle imbalance. Basically, any muscle couple out of balance would cause a progressive deformity in the foot when it happened before or during growth and could cause deformity in the bone in addition to the deformity through the joints.

In 1973 I took my first AO course in Davos, Switzerland, and thereafter have functioned as a faculty member in the Davos courses. The AO courses at that time were very fixated on biomechanics and internal fixation of fractures, but that's a different story. I had begun a foot clinic at Harborview in 1970 when I returned from Sheffield and have had both pediatric and adult foot and ankle clinics going essentially since that time. I was a little distracted for the first 20 years of practice by running the Trauma Service at Harborview and dealing with the biomechanics of long bone fractures. However, the application of biomechanical principles to foot and ankle reconstruction remained my primary interest and for the last 15-20 years has been almost my exclusive practice.

My guiding principle in foot and ankle surgery is that "nothing happens for no reason". In reality this means that we must look for an anatomic or biomechanical abnormality as a cause or at least a contributing factor in functional disorder. Such factors include symptomatic fallen arches, cavovarus feet, tendon disorders, muscle imbalance, plantar fasciitis, chronic instability, etc. The two most common disorders in my experience are clearly evolutionary faults or atavistic traits. These are gastrocnemius equinus and hypermobile first metatarsal (or first tarsometatarsal joint). Recognizing and treating these traits are essential for dealing with the secondary clinical problems. Clearly, other biomechanical variations, including malalignment, malrotation, and unequal length in the lower extremities also qualify as inherent problems to be resolved.

Several clinical examples of the problems mentioned above and the application of biomechanical principles and their solution will be shown in the full presentation.

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Functional Gait Analysis of Ankle Arthrodesis and Arthroplasty

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INTRODUCTION

Ankle osteoarthritis (OA) affects 6% of the general population [1]. Standard surgical treatment for ankle OA has been tibiotalar arthrodesis. However, OA in surrounding joints has been reported during follow up [2]. Arthroplasty may be a functional alternative to arthrodesis. One study compared outcomes of arthrodesis and arthroplasty, however their study design did not contain a within-subject paired sample [3]. The purpose of this study was to compare functional gait outcomes in a paired sample pre-/post-surgery design. It was hypothesized that at 12 months post surgery, arthroplasty would have improved temporal-distance parameters, and increased ankle range of motion (ROM), peak internal moment, and power compared to pre-surgery condition, and compared to arthrodesis.

METHODS

Sixteen patients (58.4 +/- 10.0 years, 1.7 +/- 0.1 m, 89.7 +/- 17.1 kg) were recruited. The protocol was approved by the Institutional Review Board. Eight patients received arthrodesis (FUSE), and eight received arthroplasty (TAR) (Salto® Talaris, Tornier, Edina, MN). All patients were evaluated prior to surgery and 12 months post surgery. Patients walked across a 10m walkway at a self-selected walking speed, while barefoot. Ground reaction force data were recorded at 1200 Hz (AMTI, Watertown, MA; Bertec, Columbus, Ohio). Marker trajectories were recorded at 120 Hz with a 12-camera Vicon MX system (Vicon, Lake Forest, CA). The Plug-In-Gait model was used to estimate joint kinematics and net internal joint kinetics using standard inverse dynamics. A mixed effects linear model was used to compare between pre- and post-surgery conditions for both groups.

RESULTS

Temporal-distance measures of gait improved across both groups at 12 months post-surgery, with no significant group effect (Table 1). Compared to pre-surgery conditions, TAR patients showed a trend of increased ankle ROM (3.4 deg; p=0.051). Pre-/post-surgery changes were different between groups for peak internal plantar flexor moment (p=0.032); increasing 0.18 Nm/kg for FUSE, and decreasing 0.14 Nm/kg for TAR. With gait velocity as a covariate, both groups increased ankle power absorption from -0.42 to -0.56 W/kg (p=0.039), with no significant group effect.

DISCUSSION

Though plantar flexor moment decreased in the TAR group, increased ROM and subsequently greater angular velocity resulted in large power generation. The combined effect allows sustained gait velocity and improved gait function overall; in agreement with previous findings [3]. It appears that overall gait function is improved at 12 months post-surgery for both surgery types. However, tibiotalar arthroplasty appears to retain more natural ankle joint function. Long term follow up should reveal more significant functional outcomes.

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ACKNOWLEDGEMENTS

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Table 1: Temporal-distance parameters significantly improved after surgery ($p < 0.05$); mean (SD).

	Gait Velocity (m/s)		Stride Length (m)		Cadence (steps/min)	
	Pre	**Post**	**Pre**	**Post**	**Pre**	**Post**
FUSE	0.99 (0.29)	1.19 (0.18)	1.10 (0.21)	1.28 (0.13)	105 (13)	109 (10)
TAR	0.94 (0.25)	1.15 (0.18)	1.08 (0.24)	1.25 (0.15)	103 (11)	113 (10)

Reliability and Clinical Utility of Active Foot Range of Motion in Individuals with Midfoot Arthritis and Matched Controls

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INTRODUCTION

Limited foot motion has been postulated to contribute to the development of symptoms in a variety of clinical populations.[1] A variety of weight-bearing and non-weightbearing techniques have been proposed to quantify ankle dorsiflexion,[2] subtalar motion,[3] and 1st metatarsophalangeal (MTP) joint motion.[4] However, limited objective data are available examining the reliability of techniques of measuring foot motion in clinical populations. The purpose of this study was to examine the reliability of peak active ankle dorsiflexion, subtalar eversion, and 1st MTP joint dorsiflexion in patients with midfoot arthritis and matched controls.

METHODS

30 patients with midfoot arthritis and 20 control subjects, matched in age, gender and BMI, participated in this study. *In vivo* foot motion was assessed using a multi-segment model.[5] Sensors were placed on the subject's skin over the hallux, 1st metatarsal, calcaneus and tibia. Peak active motion was assessed in the barefoot condition with the subjects seated during three motions: active dorsiflexion, subtalar eversion, and 1st MTP dorsiflexion. Data were acquired over 10 seconds as subjects actively performed the test motions. Kinematic data were analyzed using MotionMonitor™ to obtain peak and total peak-to-peak range of motion (ROM) calcaneal eversion, ankle dorsiflexion and 1st MTP joint dorsiflexion.

RESULTS

Peak active peak active ankle dorsiflexion, subtalar eversion, and 1st MTP joint dorsiflexion showed excellent reliability (Table 1). No differences in peak active peak active ankle dorsiflexion, subtalar eversion, and 1st MTP joint dorsiflexion were noted between individuals with midfoot arthritis and matched controls. (Figure 1)

	ICC (2,k)	Mean ± SD,°
Ankle dorsiflexion	0.993	5.5±4.9
Ankle plantarflexion	0.872	26.7±12.4
Subtalar inversion	0.958	9.2±9.5
Subtalar eversion	0.980	4.8±10.8
1st MTP dorsiflexion	0.983	42.2±15.2
1st MTP plantarflexion	0.936	12.9±10.4

Table 1. Summary of Intraclass correlation coefficients in the active motions tested.

DISCUSSION

The findings of our study suggest that peak active motion may be used to reliably measure of ankle dorsiflexion, subtalar eversion and 1st MTP dorsiflexion in patients with midfoot arthritis and asymptomatic control subjects.

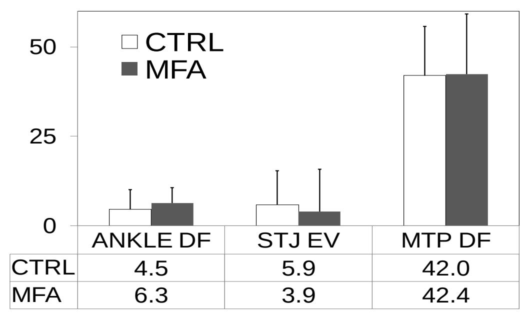

Figure 1. Peak active motion in individuals with midfoot arthritis (MFA) and matched controls (CTRL). Error bars indicate standard deviation.

The values for peak active motion reported in our study are in agreement with previous reports. [2-4] The lack of difference in active motion in patients with midfoot arthritis compared to control subjects may indicate that our patient group represents a relatively high functioning cohort of patients with early degenerative changes.

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In-vivo kinematics of the three components of an innovative ligament-compatible total ankle replacement: a fluoroscopic study

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INTRODUCTION
An innovative total ankle replacement was designed with the aim to establish full compatibility between the shape of articulating surfaces and the retained ligaments throughout the flexion arc [1]. This was achieved with a convex spherical tibial and a talar component with a radius of curvature in the sagittal plane longer than that of the natural talus, unlike most of the current three-part designs. A fully conforming meniscal bearing is interposed, free to move backwards/forwards on both metal components during plantar-/dorsi- flexion (PlDo). This design followed mathematical model predictions from measurements on specimens in virtually unloaded conditions [2]. We assessed quantitatively whether the predicted mechanisms really occur in patients.

METHODS
A total of eight patients implanted with the BOX Ankle (Finsbury Orthopaedics, UK) were analyzed at 6, 12, 24 and 36 month Follow-Up (FU) when available. The acquisitions were performed during flexion against gravity (FaG) and in static double leg stance in maximum plantar- and dorsi- flexion (MaP-MaD), by means of a standard fluoroscope (CAT Medical System, Rome, IT) at 10Hz. 3D motion of the metal components was obtained by stereo-fluoroscopic analysis [3], that of the meniscal bearing by using 3 tantalum beads stuck on the polyethylene in known positions.
Tibial and talar reference frames were defined onto relevant component models. PlDo, inversion/eversion (InEv) and internal/external (InEx) rotation, in the sagittal, frontal and transverse planes respectively, and antero-posterior (AP) translation of the meniscal and talar components with respect to the tibia, were expressed by a standard joint convention [4].

RESULTS
In FaG, InEv and InEx were significantly coupled to PlDo ($p<0.05$). This was found also for AP translations in both motor tasks ($p<0.01$). Over the 4 FUs, AP translation of the meniscal bearing for each 10° flexion was 1.7-1.4-1.6-1.5 mm in FaG and 2.5-2.1-2.0-2.0 mm in MaP-MaD. Mean overall ranges are in Table 1.

DISCUSSION
The considerable AP motion of the meniscal bearing, together with the coupled rotations revealed natural motion is restored at the replaced joint according to the original biomechanical design. This is somehow maintained over the FUs. These observations support the main features claimed for this design, implying also full congruence between the three components and restoration of ligament natural function for good longevity and function, as preliminary reported elsewhere.

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Table 1: Mean and standard deviation of the kinematics variables over the patients for four follow-ups

	Flexion against Gravity				**Maximum plantarflexion-dorsiflexion**			
(n° patients)	**1FU (5)**	**2FU (7)**	**3FU (5)**	**4FU (3)**	**1FU (5)**	**2FU (7)**	**3FU (5)**	**4FU (3)**
PlDo [°]	17.2±7.5	17.0±6.1	16.0±7.1	15.1±6.3	19.4±9.5	14.5±6.3	14.3±9.2	14.0±5.1
InEv [°]	6.7±4.4	6.1±4.0	4.6±2.3	4.8±2.0	3.2±2.8	2.3±2.4	1.0±0.4	1.3±1.3
InEx [°]	7.4±5.6	6.5±4.9	4.0±1.9	4.4±1.9	2.0±1.7	2.4±3.1	2.2±2.3	3.0±4.3
Meniscal AP [mm]	3.4±0.8	3.6±1.1	2.9±1.0	3.3±2.0	3.8±0.9	3.2±2.1	2.9±2.5	2.5±1.3
Talar AP [mm]	4.4±2.3	4.8±1.6	4.6±1.4	4.5±3.0	3.4±3.5	3.7±1.6	4.5±2.0	4.2±2.8

Center of Rotation Position in Prosthetic Feet

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INTRODUCTION

Joint kinetics have been estimated via inverse dynamics to analyze the performance of non-articulated energy storage and return (NA-ESR) prosthetic feet [1] and the motor control strategies and compensations adopted by lower limb amputees [2]. A central assumption of a commonly used link-segment model is that the ankle's axis of rotation acts as a fixed hinge; yet in NA-ESR prosthetic feet, no true ankle articulation exists, and the extent to which any axis remains fixed is unknown. Therefore, reports of estimated kinetics in NA-ESR prosthetic feet, and proximal joints may be in question [3]. Two groups have attempted to address this concern [3,4], yet while reporting prosthetic foot kinetics, neither reported actual axis location or proximal joint kinetics. The primary focus of this paper is the quantification of the center of rotation position in NA-ESR prosthetic feet.

METHODS

The position of the center of rotation for eight commonly prescribed NA-ESR prosthetic feet was estimated from kinematic gait data collected at 120 Hz from one unilateral transtibial amputee walking at self-selected walking speed. The sagittal plane position of the tibia's helical axis of motion was estimated. Differences between the assumed fixed ankle axis and helical axis position were examined over stance phase.

RESULTS

Using a helical axis of motion to estimate the center of rotation position resulted in notable differences from the traditionally assumed fixed axis position. Tibial angle at the point of peak displacement (difference between fixed and helical axis positions) differed across directions and in some cases between feet within a direction. Peak anterior displacement of the helical axis compared to the assumed fixed axis position for eight NA-ESR prosthetic feet and the tibial angle at peak displacement are shown in Figure 1.

DISCUSSION

Displacements of 10 mm in the anterior direction occurring near mid-stance (foot #1), would have little effect on estimates of power at the ankle, while

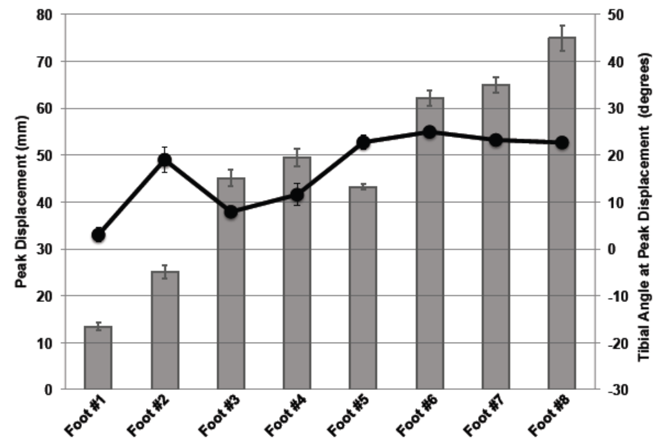

Figure 1. Peak anterior displacement of the helical axis from the fixed axis position for eight prosthetic feet and corresponding tibial angle; columns represent peak displacement, line represents tibial angle.

a displacement of 80 mm occurring later in stance (foot #8), may grossly misrepresent power at the ankle. We are currently working on comparisons between joint kinetics estimated using the assumed fixed axis and helical axis positions respectively.

This paper represents a first effort to estimate the center of rotation position in NA-ESR prosthetic feet. Based on these results, the validity of calculating joint kinetics for NA-ESR prosthetic feet and the proximal joints may be in question if the ankle axis is assumed to act as a fixed hinge. This raises questions about how unilateral transtibial amputees utilize the energy storing capacity of these feet and how they should be trained during rehabilitation.

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Brain, Body, and Foot:
A Multidomain Approach for Predictive Computational Modelling

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INTRODUCTION

The foot is a complex organ with many mechanical degrees of freedom. Finite Element (FE) models have been developed to simulate multidimensional load-deformation responses, and have reached the point that they are sufficiently valid to answer certain clinical questions. This, however, raises the issue of boundary conditions. FE models require boundary conditions that are either prescribed loads or prescribed displacements.

PROBLEM STATEMENT

As an illustrative example, consider the question of the effect of footwear (or surgery) on mechanical stresses in the foot during locomotion. Such questions are directly relevant to the treatment and prevention of diabetic ulcers, as well as overuse injuries in sports. Computational techniques could be extremely valuable for footwear design and treatment planning. Simulations are typically obtained by applying controlled loads at the boundary between the foot and rest of the body. These loads represent articular contact forces, ligaments, and extrinsic muscle forces. A nominal simulation, after iterative refinement, would likely produce quite realistic stresses and deformations in the foot.

The problem arises, however, in simulations where the geometry or properties of footwear (or the foot itself) is altered. A change in footwear would alter alter the distribution of stresses with the foot, but would never change the applied loads, since these are controlled boundary conditions. This is clearly inconsistent with the fact that footwear can dramatically alter the applied loads [1] via sensory and mechanical effects. One could, of course, use boundary conditions obtained from human subjects using experimental footwear (or after experimental surgery), but this greatly reduces the utility of computational modelling in the design process.

MULTIDOMAIN APPROACH

In order to overcome these limitations, we have developed a multidomain approach in which the foot is coupled to a musculoskeletal model of the rest of the body, with optimal neural control of muscles to produce movement. A musculoskeletal model, even when it is quite coarse compared to the foot model, is already an enormous improvement over prescribed loads or displacements at the boundary.

Two major challenges had to be overcome. First, dynamic musculoskeletal models are typically simulated in very small time steps, and it is not computationally feasible to solve an FE foot model at each time step. This problem was addressed by adaptive surrogate modelling, in which local regression models gradually replace the FE model calculations. We were able to reduce computation time by 95% without loss of accuracy [2].

Second, movement optimization in dynamic musculoskeletal models is extremely time consuming, even when not coupled to an FE model [3]. We have addressed this problem by developing collocation-based optimization methods, which require far fewer time steps and far fewer iterations than traditional methods [4]. With these methods, we were able to predict gait adaptations due to pressure related forefoot pain [5].

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Patient-Specific Finite Element Modeling Techniques

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INTRODUCTION TO IA-FEMesh

To address the meshing challenges associated with anatomic models, an interactive multiblock approach to mesh development has been implemented in our open-source software package, IA-FEMesh [1]. The mesh generation procedures are intuitive and are general enough to be applied to a range of skeletal structures (Figure 1). Novel mouse interactions, introduced via new 3D widgets, allow the user to sculpt building block structures with ease. Thereafter, the building block structures act as the foundation for the subsequent steps of mesh seeding, mesh morphing, and interior node calculation. The capability of IA-FEMesh to directly couple the mesh with image data facilitates not only the geometric definitions but the material property assignments as well. A mesh quality viewer provides a means to evaluate the quality of the mesh prior to exporting it for analysis.

EASING MESH DEVELOPMENT

As the complexity of the models increases, the majority of the mesh development time is expended creating the multiblock structure. Consequently, our efforts have been aimed not only at improving the mesh definitions using the current techniques, but further automating/easing the meshing process in hopes of making large scale patient-specific studies a reality.

- Mapped Building Blocks
- Automated Building Block Definitions
- Interactive Tracing Capabilities
- Accommodate for Multiple Surfaces

Ideally, the modeling capabilities available in IA-FEMesh will enable the surgeon working alongside the engineer to plan patient-specific care using objective analysis-based tools. Thus, providing the means to better evaluate alternate surgical protocols and treatments, thereby contributing to improved surgical outcome.

MESHING THE BONES OF THE FOOT

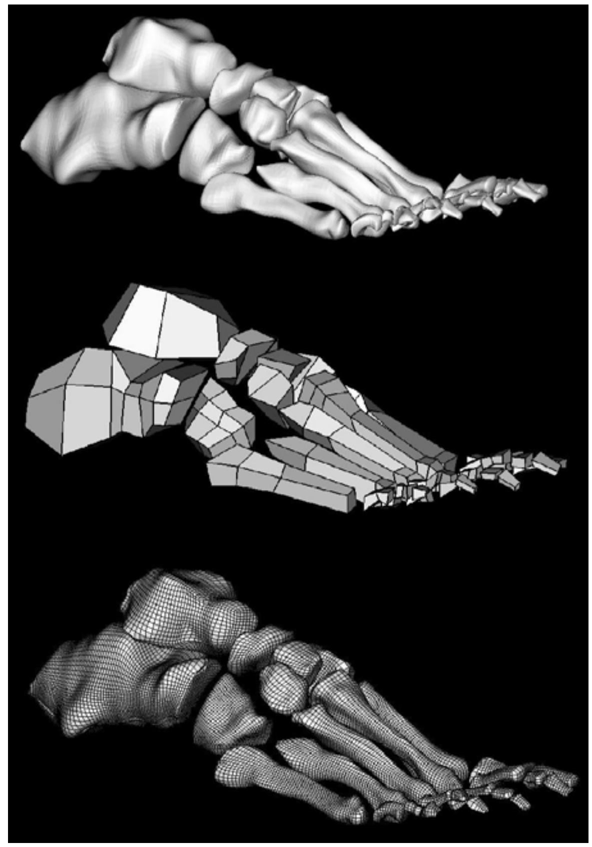

Figure 1. Bones of the foot: (a) the bony surfaces segmented directly from CT image data, (b) building blocks, and (c) the resulting hexahedral meshes of the individual bones.

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Equivalence of Elastic Contact and Finite Element Models of Patient-Specific Contact Stress Exposure in the Human Ankle

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INTRODUCTION

Prior assessment of chronic contact stress exposure in the ankle following articular fracture reduction has relied upon patient-specific finite element (FE) analysis [1]. FE contact analysis is expensive both in computational and in analyst time, limiting its usefulness in multi-center studies involving large numbers of subjects. The purpose of this project was to establish the equivalence of ankle contact stresses computed using more expeditious elastic contact modeling methods, to those computed using FE analysis.

METHODS

Model geometries used were from prior patient-specific FE studies of the relationship between contact stress exposure and post-traumatic arthritis development after tibial plafond fracture [1]. Eleven FE models of intact ankles from CT data were used, with the cartilage assumed to have a 1.7 mm uniform thickness.

An elastic contact modeling approach [2] was used to compute ankle contact stress over the gait cycle. The contact modeling, implemented as custom MATLAB functions, treated the apposed cartilage surfaces as a system of linear springs, with an underlying rigid bone foundation. Surface facets were deemed to be in contact when the two surfaces overlapped. Force and moment boundary conditions were enforced to simulate a full gait cycle, driving the ankle through a flexion-extension arc. Site-by-site comparisons of contact stress were used to establish the degree of agreement between the two computational approaches.

RESULTS

The eleven ankle loading simulations were completed in approximately one hour, compared to 10 to 20 hours per model for FE. Similar contact stress distributions were obtained with the two methods (Figure 1). The mean difference in the computed values of contact stress was 0.42 (±0.41) MPa, averaged across the 11 cases. The majority of the disagreement between the two methods was at the edges of contact, likely attributable to the continuum-vs-discrete contact modeling.

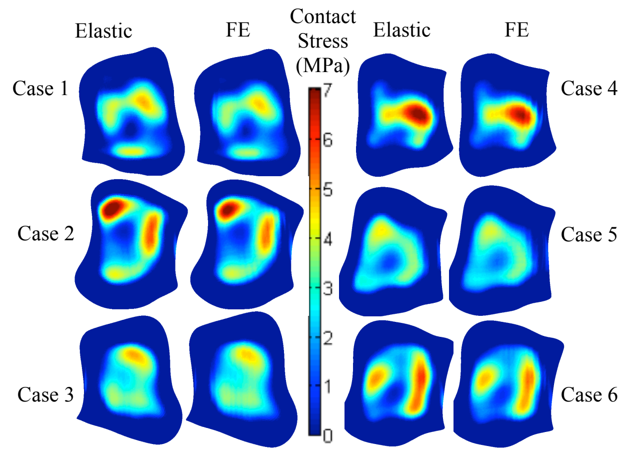

Figure 1: Elastic contact model stress compared to FE contact stress for 6 of 11 cases in the distal tibia. Instantaneous contact stresses were averaged over the gait cycle for comparison.

The mean translational differences between elastic and FE contact model computations in the inferior-superior, medial-lateral, and anterior-posterior directions were 0.02(±0.01), 0.14(±0.07) and 0.49(±0.16) mm, respectively. The mean rotational differences in inversion/ eversion and internal/external rotation were 0.11(±0.02) and 0.05(±0.09) degrees.

DISCUSSION

The elastic contact modeling approach provided contact stress and kinematic solutions reasonably close to FE analysis, while drastically reducing the amount of computational and investigator time required.

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Direct determination of toe flexor muscle forces based on sub-metatarsal/toe pad load sharing by using finite element method

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INTRODUCTION

Direct quantitative measurement of foot muscle forces is difficult [1]. Only few invasive techniques, such as buckle-type implantable force transducer, are available but only accessible to Achilles tendon as they often require a specific volume/length of tendon for attachment. In this study, forces in the toe flexor muscles were determined noninvasively by using finite element (FE) models of the 2nd ray of the human foot.

METHODS

The FE model comprises a planar section representative of the 2nd ray of the human foot (Fig. 1). The basic anatomical geometry was obtained by slicing a 3D CT-scan reconstructed foot volume through the 2nd metatarsal bone in the sagittal plane. The metatarsophalangeal (MTP) joint was embedded into a continuum of the forefoot soft tissue which was modelled as a hyper-elastic material. Relative articulating movements of the bony joints were included. A frictional contact condition was defined at forefoot/ground interface. Major flexor muscles and ligaments were modelled as force-actuated "connector" elements and passive "spring" element respectively. A rigid beam construction was connected with the superior end of the tibia bone. Pre-determined resultant forefoot forces (see below) were applied through the body's center of gravity located at one nodal point of this beam structure. The ground is fully constrained.

The vertical and shear forces acting at local sub-metatarsal and toe pad of the 2nd ray were collected in a specially-designed gait platform [2]. Muscles of flexor digitorum brevis (FDB) and flexor digitorum longus (FDL) in the model were parameterized and varied in discrete steps until the load sharing between the sub-metatarsal pad and toe pad were similar to those measured in a normal subject. The calculated dorsal displacement of the 2nd proximal phalanx served as additional criterions for predicting muscle forces.

RESULTS

By running 56 parameterized FE models, muscle forces of FDB and FDL were determined with sub-metatarsal/toe pad load sharing of the vertical component reached the measured values of 16.5% (i.e., toe pad load sharing in percentage). Satisfactory results for the shear component load sharing still pose a challenge. In addition, it was found that tensile strains in the passive stabilizers spanning the medial-longitudinal arch were significantly increased (i.e., plantar fascia by 37.7%, long plantar ligament by 48.2%, and short plantar ligament by 27.8%) without applying the muscle contraction forces (Fig. 1).

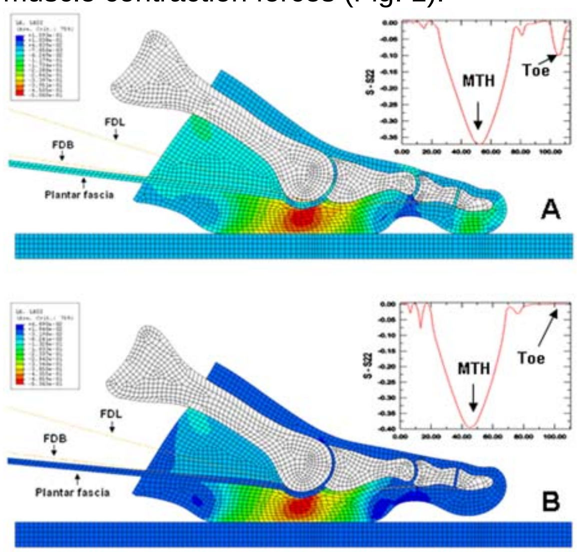

Fig. 1 Soft tissue strain/stress distributions of the 2nd-ray plane-strain FE model during ground contact with (A) / without (B) applying intrinsic/extrinsic muscle forces. Note only forefoot is shown.

DISCUSSION

From modelling point of view, it seems important to consider the exertion of toe flexor muscle forces which may have a significant impact on the internal soft tissue stress/strain distributions of the foot ray.

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Prediction of Plantar Pressure Distribution in Flat-foot Children

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Introduction

Assessing plantar pressure (PP) could provide valuable information in prescribing appropriate footwear to reduce or limiting further complication in flat-feet children. However, PP is not easily measured in most pediatric practices because the necessary equipment is expensive and requires training to operate. The main goal of this study was twofold. First we explored the dynamic of plantar loading during walking in flatfeet children and aged match control subjects. Then we proposed a simple model to predict PP in flatfeet children using two variables measurable using widely accessible and low cost tools.

Methods

This study examined PP distribution in 60 flat-foot children and 25 aged-matched control subjects. Measured parameters were included PP distribution (PPD), grand reaction force (GRF), plantar arch angle estimated from footprint (Clarke's angle–Fig1A), and Calcaneal-first metatarsal and inclination angles measured using X-ray image (Fig.1B). Shoes insoles and two force platforms were used for measuring PPD and GRF respectively. To examine PPD during walking, time-series of the magnitude of pressure measured by insole sensors were grouped into five anatomical masks. Finally, after initial evaluation and identifying the most significant descriptors related to flatfeet complication, a multi-variable model was designed to estimate the PP descriptor using independent variables including subject's body mass, Clarke's angle and the plantar area contact.

Results

Results confirmed that Clarke's angle estimating using footprint in flatfeet children has a high agreement with the results obtained from X-ray measurements (see Fig 2). In Flatfeet children, the area of contact under cuboid bone was significantly higher in average by 42% ($p<0.05$) compared to control subjects. This caused a significant reduction in plantar pressure under

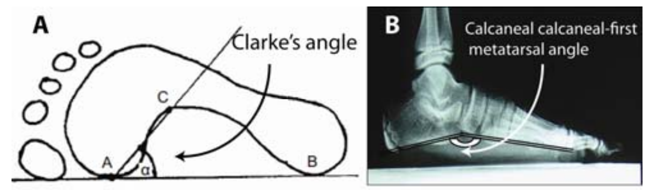

Figure 1: (A) Definition of Clarke's angle. (B) Definition of Calcaneal-first metatarsal angle

cuboid bone in average by 47% ($p<0.05$). The suggested model allows predicting PPD under the cuboid bone using subject's body mass and Clarke's angle ($r=0.97$, RMSE=0.04N/cm2 (2%), $p<0.05$). The results suggest that by increasing body mass, PP under the cuboid bone is increased, whereas Clarke's angle has a negative impact on PP under cuboid bone.

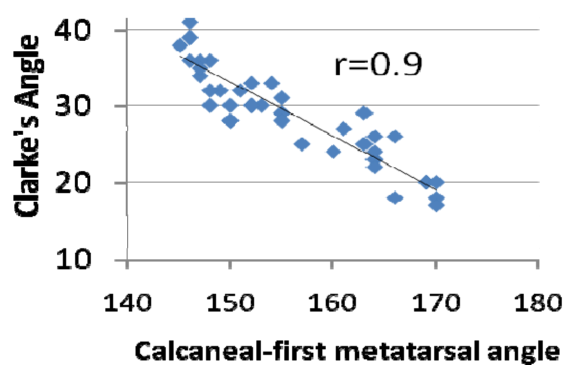

Figure 2: Clarke's angle etimated uding foot print has high agreement with measuremend obtained using x-ray.

Discussion

This study proposed a simple model to estimate plantar pressure distribution in flat-foot children without using any sophisticated technology. The model may assist podiatrist/ pediatric to reduce the consequences of high plantar pressure in flat-foot children via controlling subject's weight or prescribing appropriate footwear. The only measurable parameters for this model are Clarke's angle estimated using subject's footprint as well as subject's body mass.

Prediction of the effect of a subject-specific AFO on the gait of a healthy test subject.

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INTRODUCTION

An Ankle Foot Orthosis (AFO) is commonly used in clinical practice to assist gait of patients with different pathologies. Individualising the AFO design to patient specific requirements is still difficult.

This study investigates the possibility to predict the effect of a subject-specific AFO on the gait of a healthy test subject using a personalized musculoskeletal model. Mechanical properties of the AFO are determined using finite element (FE) analysis.

METHODS

A FE-model with a constant wall-thickness was constructed based on the scan of the inner surface of an existing subject-specific poly-propylene AFO [1]. Loads were implemented based on pressure-measurements between AFO and the lower leg- shoe [2]. Material properties of polypropylene were taken from literature [3]. The FE-model was used to calculate the stiffness of the AFO.

A musculoskeletal model of the lower limbs with 23 degrees of freedom and 92 muscles was scaled in OpenSim to match the test subject's anthropometric data [4].

Marker trajectories of a normal gait (without AFO) were used to calculate the kinematic parameters in gait (hip, knee, ankle angle) and the muscle activation of the lower limb muscles. This muscle activation pattern was then combined with a simulation of the AFO as an angle-dependent torque around the ankle [5] in a forward dynamic calculation to predict the effect of the AFO on the kinematic parameters. These results were then compared with actual AFO-gait.

RESULTS

The stiffness of the AFO as calculated in the FE-model is 175 Nm/rad. This is implemented in the musculoskeletal model as an angle-dependent torque

Figure 1 shows the ankle plantarflexion – dorsiflexion movement during gait: barefoot, with a simulated AFO and with the actual AFO.

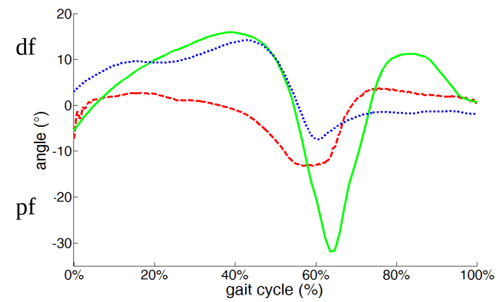

Figure 1: Ankle angle of the test subject barefoot (-), simulated AFO (- -) and with actual AFO (...).

DISCUSSION

The simulated ankle pattern shows a similar limitation of the plantarflexion movement as the actual AFO gait when compared to barefoot gait. However, some differences between the actual and the simulated AFO-gait can be observed. Further investigation is conducted to optimize the predicted kinematics.

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Mechanical and Functional Measures Reveal a Coherent Structure to Ankle Instability: A Principal Component Analysis

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INTRODUCTION

Individuals with a history of lateral ankle sprain present with functional and mechanical impairments [1]. Clinicians quantify impaired ankle function with tests that measure the impairment, but it is unclear if test results demonstrate a coherent structure of latent factors which adequately describe the constructs of ankle instability. The purpose of this study is to determine if a coherent structure exists between mechanical instability, functional instability, and self-reported function variables commonly used in ankle instability patients.

METHODS

A cross-sectional study of subjects with no history of ankle sprain (n=14), ankle sprain copers (n=14), and subjects with chronic ankle instability (CAI) (n=14) participated. Fifteen dependent variables were measured and are listed in Table 1. Ankle dorsiflexion and posterior talar glide were measured with a bubble inclinometer [2]. Anterior and inversion laxity were measured with an instrumented ankle arthrometer [3]. Manual anterior drawer and talar tilt tests were scored on a 5 point scale from 0 (hypomobile) to 4 (hypermobile). The star excursion balance test and balance error scoring test were performed using standard clinical techniques.

Data from all subjects (n=42) (22.7±3.4 years, 69.0±12.3 kg, 170.5±7.9 cm) were analyzed with a principal components analysis using orthogonal rotation to identify a simple structure of factors. Criteria for latent variable retention included a) eigenvalues (EV) >1.0, b) retained factors should account for >50% of overall variance, and c) each latent factor should load at least two variables with correlations of >0.4 and with minimal double loadings.

RESULTS

A four factor solution explained 54.5% of the cumulative variance. The factors represented the constructs of: 1) ankle dorsiflexion (EV = 2.5), 2) self-reported function (EV = 2.04), 3) dynamic balance (EV= 1.9), and 4) ankle laxity (EV = 1.7). Twelve variables loaded onto the four factors with correlations >0.4 except anterior laxity (.371) which was chosen to load along with the anterior drawer test, talar tilt test and inversion laxity (Table 1).

	Factor			
	Ankle Dorsiflexion	Self-Reported Function	Dynamic Balance	Ankle Laxity
Weight bearing ankle dorsiflexion, bent knee(°)	.955	-.032	-.099	.095
Weight bearing ankle dorsiflexion, straight knee(°)	.901	.000	-.014	.174
Active ankle dorsiflexion(°)	.690	-.004	.050	-.130
Anterior Reach Test (%)	.400	.229	.120	.024
FAAM- Sports	.188	.939	.002	-.087
FAAM- ADL	.267	.684	-.009	-.206
Posterolateral Reach (%)	.135	.075	.938	.047
Posteromedial Reach (%)	.054	.008	.857	.104
Manual Anterior Drawer Test (0-4)	.004	-.296	-.074	.910
Manual Talar Tilt Test (0-4)	.076	-.488	.139	.569
Instrumented Inversion Laxity (°)	.101	-.148	-.023	.491
Instrumented Anterior Laxity (mm)	.021	.012	.089	.371
BESS-Firm single limb	.187	-.385	-.467	-.030
Manual Posterior Talar Glide Test (°)	.014	-.056	.161	-.010
BESS-Foam single limb	.096	-.364	-.136	.067

Table 1. Exploratory Factor Analysis. Values are Cronbach α. FAAM, Foot and Ankle Ability Measure self-reported disability questionnaire; ADL, Activity of Daily Living; BESS, Balance Error Scoring System. Anterior, Posterolateral and Posteromedial reach are 3 components of the Star Excursion Balance Test.

DISCUSSION

Related measures of weight-bearing ankle dorsiflexion, self-reported function, dynamic balance, and ankle laxity loaded onto distinct factors thus demonstrating a coherent structure to mechanical and functional measures of ankle instability. The loading pattern of variables onto these factors supports the clinical framework of mechanical, self-reported, and functional impairments that represent coherent factors that help describe the spectrum of dysfunction following lateral ankle sprain.

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Muscle Activation Patterns For Functional Ankle Instability and Normal Subjects During Anticipated and Unanticipated Jump-landings

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INTRODUCTION

Some researchers have implied that improving feed-forward control strategies may be effective in treating chronic ankle instability [1,2]. The purpose of this study was to observe potential alterations in feed-forward control mechanisms in a functional ankle instability (FAI) sample during two different jump-landing conditions. We hypothesized that FAI subjects would exhibit altered muscle activation patterns, relative to matched controls.

METHODS

Ten FAI subjects (6 males, 4 females; age = 23 ± 4 yr; height = 1.80 ± 0.14 m; mass = 80.9 ± 25 kg) and ten control subjects (6 males, 4 females; age = 23 ± 2 yr; height = 1.81 ± 0.13 m; mass = 81.6 ± 27 kg) participated in this study. Electromyography (EMG) for the medial gastrocnemius (GA), tibialis anterior (TA), fibularis longus (FL), medial hamstrings (MH), vastus medialis (VM), vastus lateralis (VL) and gluteus medius (GM) was recorded. Subjects performed several jump-landings over a 20 cm barrier onto a force plate, followed by a contra-lateral hop. Three randomized landing tasks (left foot, right foot, or both feet) and two randomized conditions (anticipated and unanticipated) were used to enhance the unanticipated condition. Unanticipated trials included an in-flight visual cue that specified landing type. EMG for the involved leg was averaged across three successful trials. Onset of muscle pre-activation (Figure 1), pre-landing mean EMG amplitude (150 ms prior to impact), and mean EMG amplitude during landing (Table 1) were calculated. Between group differences were analyzed using a one-way ANOVA ($p \leq 0.05$).

RESULTS

Two statistical differences were observed during the anticipated trials: (1) mean GM pre-activation occurred 57 ms earlier for the controls ($F = 10.578_{(1,19)}$, $p = 0.004$) and (2) mean FAI fibularis longus EMG amplitude was almost twice that of the control group ($F = 5.463_{(1,19)}$, $p = 0.03$) during the landing phase (≈ 40 ms). No other significant differences were observed.

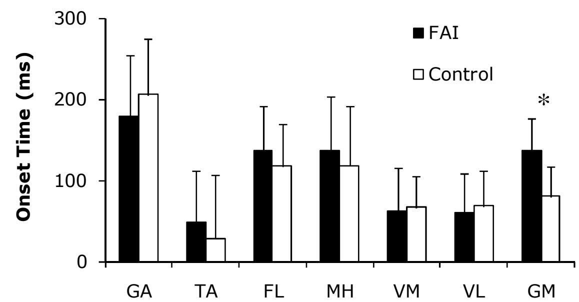

Figure1: Timing of muscle activation prior to impact for the anticipated condition. The asterisk indicates a statistical difference.

DISCUSSION

Differences in the activation patterns of both distal (FL) and proximal (GM) musculature, prior to reflexive responses, imply altered feed-forward control strategies in the FAI group. We speculate that altered GM pre-activation may affect the landing position of the involved foot and ankle. We also speculate that the lack of differences for the unanticipated trials may be due to heightened anticipation during this condition, suggesting that the unanticipated model used in this study is flawed.

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Table 1: Mean EMG amplitudes, normalized to quiet stance, during landing for the anticipated condition. The asterisk indicates a statistical difference.

Group	GA	TA	FL	MH	VM	VL	GM
FAI	2.5 ± 18.2	6.6 ± 4.8	11.6 ± 6.6*	8.4 ± 7.0	14.3 ± 15.2	8.9 ± 5.5	5.8 ± 3.3
Control	5.9 ± 10.9	5.8 ± 3.8	6.0 ± 3.8	6.0 ± 4.6	13.7 ± 12.1	7.1 ± 5.1	5.2 ± 3.8

Preparatory Muscle Activation during a Lateral Hop Before and After Fatigue in Those With and Without Chronic Ankle Instability

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INTRODUCTION

Feed-forward mechanisms are utilized to maintain dynamic postural control and both proximal and distal changes in muscle activation have been established in those with chronic ankle instability (CAI) [1]. Additionally, fatigue has a significant effect on both muscle activation and dynamic postural control [2]. The purpose of this study was to examine if muscle activation at the hip and ankle differed after fatigue in subjects with and without CAI prior to landing a lateral hop.

METHODS

Sixteen CAI subjects (8M, 8F,172.25 ±10.87cm, 69.13±13.31kg, 20.50±2.00yrs) were matched to 16 healthy subjects (8M, 8F 170.50 ±9.94cm, 69.63±14.82kg, 22.00±3.30yrs).

Surface electromyography (EMG) was collected from the tibialis anterior (TA), peroneus lognus (PL), gluteus medius (GMed), and gluteus maximus (GMax) a series of five lateral hops performed before and after a functional fatigue protocol. [3] Average EMG was collected at 1000Hz during the 200ms prior to landing. EMG signals were smoothed, filtered, rectified, and normalized to the mean peak EMG amplitude from the five hopping trials. Separate, 2x2 repeated measures ANOVAs were used to assess differences in EMG between CAI and healthy groups before and after fatigue for each muscle.

RESULTS

Significant Group main effects were found for PL and GMax muscle activation. The PL demonstrated significantly higher muscle activation in the CAI group compared to the healthy group ($F_{1,30}$=8.60, p=0.01). The GMax also displayed significantly higher muscle activation in the CAI group compared to the healthy group ($F_{1,30}$=4.20, p=0.05) just prior to landing. There were no statistical differences between Groups for the TA or the GMed as well as no influence of Fatigue on either muscle.

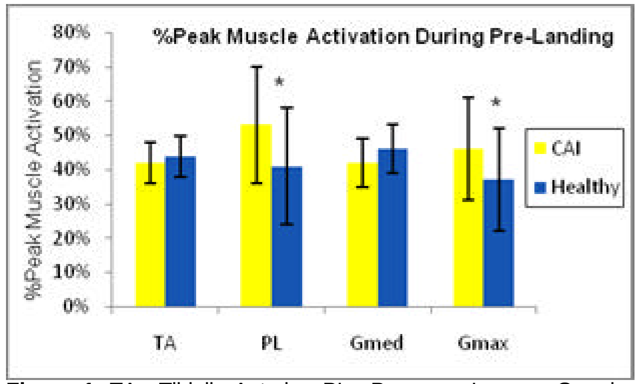

Figure 1: TA= Tibialis Anterior, PL= Peroneus Longus, Gmed= Gluteus Medius, Gmax= Gluteus Maximus, CAI= Chronic Ankle Instability, *=p=.05

DISCUSSION

These results support the presence of feed-forward neuromuscular changes not only in the muscles at the ankle joint, but also in hip muscles. These changes appear to be positioning the lower limb for landing to help prevent the motion of ankle inversion and internal rotation of the femur in active subjects with CAI. Both proximal and distal muscles should continue to be investigated in prospective studies to determine if these neuromuscular changes are present prior to development of CAI or have occurred as a result of repeated injury.

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Plantar Loading in the Cavus Foot

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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 that 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. Data were analyzed with a univariate mixed-effect analysis of variance model followed by Bonferroni post-hoc t tests. Type and replication were modeled as fixed and random effects, respectively. Significance set at $p \le 0.05$.

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 the cavus foot 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.

DISCUSSION

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

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ACKNOWLEDGEMENTS

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Table 1: Plantar loading measurements across foot types.
[a] *Post Hoc Comparison : 1 = Cavus vs. Planus; 2 = Rectus vs. Planus; 3 = Cavus vs. Rectus, p≤0.016*

Region	Peak Pressure (PP)				Maximum Force (MF)			
	Planus	Rectus	Cavus	P-hoc[a]	Planus	Rectus	Cavus	P-hoc[a]
Hallux	43.9(6.9)	37.1(5.6)	32.5(5.8)	1,2	135(17.1)	108.8(14.1)	91.5(14.5)	1,2,3
MTH1	28.1(8)	35.8(6.6)	37.3(6.8)	1,2	134.9(23.4)	148.2(19.2)	157.2(19.7)	1
MTH2	51.1(5.8)	37.7(4.8)	38.8(4.9)	1,2	177.1(16.2)	151.1(13.3)	152.1(13.6)	1,2
MTH3	40.4(4.3)	34.2(3.6)	34.8(3.6)	1,2	172.6(17.8)	150.7(14.6)	152.3(14.9)	1,2
MTH4	27.8(3.6)	26(3)	25.6(3.1)		104.6(14.9)	98.9(12.3)	100.8(12.6)	
MTH5	20.1(5.8)	26.3(5.2)	25.3(5.3)	1,2	41(10.7)	51.6(8.8)	58.4(9)	1,2
LatHeel	33.8(4)	33.6(3.3)	38.4(3.4)	1,3	221.2(19.4)	215(15.9)	207.1(16.3)	
MedHeel	37.2(4.8)	36.5(4)	38.8(4.1)		265.4(21.6)	255.6(17.7)	241.7(18.2)	1
LatArch	11.5(1.7)	10.9(1.4)	7.7(1.5)	1,3	107(25.8)	79.7(20)	52.7(20.5)	1,2,3
MedArch	20.1(5.8)	26.3(5.2)	25.3(5.3)	1,3	26.8(7.7)	13.7(6.3)	6.2(6.5)	1,2,3

Does Foot Type Affect Foot Contact Dynamics?

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INTRODUCTION
Human movement is influenced by foot structure. Pes cavus is associated with clawing of the great and lesser toes [1], and sometimes with pain. [2] Pes planus is associated with increased plantar surface contact area and can be a risk factor in the development of overuse injuries [3]. Foot type was found to affect the center of pressure excursion index (CPEI). [4] Although important when planning treatment for pes cavus and pes planus feet, the effects of foot type on foot contact dynamics and function are not well understood. Hence, the aim of this study was to develop a normative dataset of temporal sequence of loading, CPEI, and the transverse foot angle (TPFA) of healthy subjects with pes planus, rectus, and pes cavus foot types. We hypothesized that subjects with different foot types have significant different temporal sequence of loading, CPEI, and TPFA.

METHODS
Sixty-one healthy asymptomatic test subjects (22 pes planus, 27 rectus and 12 pes cavus) were recruited with no symptoms of pain, pathology, and visible pedal deformities. The foot type of each test subject was determined based on resting calcaneal stance position and forefoot-to-rearfoot alignment. Temporal sequence of loading (contact, midstance and propulsion phases of stance), CPEI, and TPFA were calculated from plantar pressure distributions. The emed X system (Novel gmbh, Germany) was employed to measure each individual's dynamic plantar pressure distribution. A custom software was developed in C++ to calculate each of these parameters. The effect of foot type was tested for each parameter, using a mixed effect analysis of variance (ANOVA) model. Significance was set at $p< 0.05$. A trend was operationally-defined at $p<0.1$. Post hoc t-tests were performed using the Bonferroni method ($P<0.0167$).

RESULTS
The temporal sequence of loading (contact, midstance, and propulsion) was not significantly different across foot type. Midstance on the left was nearly significant. CPEI demonstrated significant differences across pes planus and rectus as well as pes planus and cavus foot types. The transverse plane foot angle was significantly different across foot types on the right and was nearly significantly different on the left.

DISCUSSION
The temporal sequence of loading was not significantly different across foot types. CPEI and transverse plane foot angles did demonstrate differences between the rectus and planus and cavus and planus groups. No parameter in the study could distinguish the pes cavus from rectus foot types.

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Table 1: Foot contact dynamic results

Parameters	Pes Planus-Right/Left	Rectus - Right/Left	Pes Cavus - Right/Left	ANOVA	P-Hocs
Contact (%St)	9.74(1.69) / 9.64(1.73)	9.81(1.54) / 9.71(1.55)	9.13(2.29) / 9.19(2.30)	0.14/ 0.33	
Midstance(%St)	49.93(5.64)/ 0.74(5.71)	48.91(5.13)/ 49.03(5.14)	50.25(7.67)/ 51.21(7,59)	0.43/ 0.10	
Propulsion(%St)	39.65(5.30)/39.58(5.62)	41.36(4.82)/ 41.16(5.03)	40.50(7.21)/ 39.58(7.47)	0.18/ 0.19	
CPEI (%)	18.73(5.84)/ 18.57(5.65)	22.08(5.16)/ 21.30(5.02)	24.45(7.76)/ 24.01(7.47)	<0.001-R,L	1,2/ 1,2
Foot Angle (°)	7.36(3.94)/7.22(4.86)	9.81(3.48)/8.64(4.32)	10.03(5.21)/9.35(6.43)	<0.001/0.08	1,2/ 1,2

%ST = %Stance, CPEI=Center of Pressure Excursion Index, Bonferonni post-hoc significance set at p<0.0167, 1 = Cavus vs Planus; 2 = Rectus vs. Planus; 3 = Cavus vs. Rectus

Effect of Modified Low-Dye Taping on First Ray Mobility in Individuals with Pronated Foot

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INTRODUCTION

A pronated foot is usually associated with dorsiflexed first ray, developing foot pain, abnormal gait pattern, or injuries to other weight-bearing joints [1]. Although low-Dye taping has been clinically used to treat first ray disorders [2, 3], there is no evidence-based research to explore its effects on first ray mobility. The objectives of this study were (1) to compare differences in position and mobility of the first ray and (2) to compare differences in angle of the metatarsophalangeal joint between with and without application of elastic low-Dye taping in individuals with pronated foot.

METHODS

Twenty-three persons with pronated foot participated in this study. The position of the first ray at the subtalar neutral position as well as dorsal and plantar mobility of the first ray were measured at the sitting position using a modified first ray ruler while the peak angle of the first metatarsophalangeal joints during push-off were measured using the 3D motion analysis system. Each participant received those two measurements under two conditions, with or without elastic low-Dye taping, in a random order and with a 20-min rest interval in between. All data with normal distribution were examined using the paired *t*-tests to compare the mean differences between taped and non-taped conditions whereas those with non-normal distribution were tested using the Mann-Whitney *U* test. All statistical analyses were executed using SAS v.9.13. The statistically significant level was set at $\alpha = 0.05$ while the power was at 0.8.

RESULTS

Application of elastic low-Dye taping changed the first ray position to a more plantarflexed position in individuals with pronated foot and increase dorsal mobility in non-weight–bearing condition. For the weight–bearing condition, elastic low-Dye taping significantly increased the first metatarsophalangeal joint angle during push-off motion although the segmental angle of the metatarsal remained the same. (Table 1)

DISCUSSION

The findings of this study were based on a well control of excessive pronation by low-Dye taping which further modified the functions of the first ray. This implies that uses of elastic taping might decrease impingement of the first metatarophalangeal joint during ambulation.

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Table 1: Effects of elastic taping on the first ray mobility test and push–off test

	First ray mobility test（mm）				**Push–off test（deg）**	
	Initial Position	**Dorsal Mobility**	**Plantar Mobility**	**Total Mobility**	**First MPJ[a] Angle**	**Segmental Angle[b]**
Taped	1.2 ± 0.8	5.7 ± 1.0	4.4 ± 1.4	10.1 ± 1.8	64.2 ± 15.2	64.0 ± 14.0
Non-taped	$1.7 \pm 0.9^{\#\#}$	$6.1 \pm 1.0^{\#}$	4.6 ± 1.5	$10.7 \pm 2.0^{\#}$	$59.2 \pm 12.7^{\#\#}$	63.5 ± 12.2

[#] $p<0.05$ and [##] $p<0.005$ indicate significant differences between taped and non-taped conditions.
[a] MPJ stands for the metatarsophalangeal joint；[b] segmental angle indicates the angle of the metatarsal bone relative to the horizontal plane.

What We Have Learned From Dynamic Gait Simulation

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INTRODUCTION

The biomechanics of human locomotion are most commonly measured using non-invasive techniques that limit potential risks to subjects or patients. In this regard newer motion analysis techniques and imaging modalities have proven extremely useful in recent years, but such approaches still fail to completely elucidate the complex mechanics of the foot and ankle during dynamic events. To more fully understand internal biomechanical function researchers still must turn to either physical or computational models. Here we describe one such model that has proven useful, the Robotic Dynamic Activity Simulator or RDAS, a device that re-animates cadaver limbs to invasively examine normal and pathological human gait.

THE APPARATUS

The operation and capabilities of the RDAS and its predecessor were extensively evaluated in previous work [1, 2]. The device includes a translating specimen-carriage, a standard force platform and nine linear actuators, all controlled by custom-written software. Motion of the shank is recreated using select kinematic data previously captured from several normal men and women with a wide range of foot sizes. Kinematic profiles matching the sex and size of the experimental specimens are selected from this library of trajectories and used to program a system of three servo powered actuators to reproduce the sagittal plane shank kinematics of any desired activity. Kinematics at the foot and ankle are not directly constrained but are dependent upon motion of the shank, foot-ground interactions, and simulated muscle activity.

The Robotic Dynamic Activity Simulator

The timing and force output of the triceps surae (Achilles), tibialis anterior, tibialis posterior, combined peroneus brevis and peroneus longus, flexor hallucis longus, and flexor digitorum longus are independently controlled with stepper motor powered actuators each equipped with a force transducer for measurement and feedback control, as well as a linear displacement transducer to track musculotendinous length. Muscle force profiles are based on normalized electromyographic (EMG) data [3] modified to account for force-length and force-velocity behaviour using the approach described by Gallucci and Challis [4].

INTERESTING APPLICATIONS

In the presentation we will briefly examine how the extrinsic muscles of the foot and related soft tissue structures modulate skeletal stress and strain; explore factors implicated in the generation of metatarsal and tibial stress fractures; describe some of the kinematic consequences of malleolar fracture and repair, better define the role played the flexor hallucis longus in normal and pathologic loading of the first metatarso-phalangeal joint, and evaluate the utility of measuring inter-segmental motions of the foot with external skin-mounted markers.

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Effect of subtalar arthroereisis on tibiotalar contact characteristics in a cadaveric flat foot model

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INTRODUCTION

Subtalar arthroereisis has been shown to provide good clinical and radiographic outcome in pediatric and adult flexible flatfoot [1]. The effect of this procedure on the tibiotalar joint loading is unknown. The aim of the present study was to quantify the effects of subtalar arthroereisis on tibiotalar contact characteristics in a cadaveric flat foot model.

METHODS

Four cadaver specimens without deformities, history of trauma or arthrosis were used. Each specimen was transected at the tibial and fibular shafts, 40 cm proximal to the heel. The proximal ends of the tibia and fibula were mounted in a cylindrical form with a six-point screw fixation. A material testing machine (Zwick 005; Zwick GmbH, Ulm, Germany) was used to apply repeatable loads to the specimens. The capacitive ankle joint pressure (AJP) sensor (Novel Gmbh Munich, Germany) with a saturation pressure of 2.5 MPa was used to assess joint contact characteristics at 50 Hz (Fig. 1).

Figure 1: Experimental set-up with the cadaver foot in the Zwick machine with the AJP sensor placed inside the tibiotalar joint.

A flatfoot model was created by incision of the talonavicular capsule, the tibionavicular portion of the deltoid ligament, the spring ligament, and the plantar aponeurosis as described before [2]. To simulate the mid-stance of gait, the peroneus longus and brevis, soleus, flexor digitorum and hallucis longus, tibialis posterior tendons were clamped and connected to external weights with pre-calculated forces. Testing conditions were normal foot, flat foot and ŕcorrectedò foot with the KalixÊ implant under axial loads of 300 N.

RESULTS

In the normal foot, the mean peak pressure was higher in the antero-medial region (1457± 53 kPa) while the highest load was the antero-lateral region in the flat foot and the ŕcorrectedò foot (1081±14 kPa and 1141±49 kPa respectively).

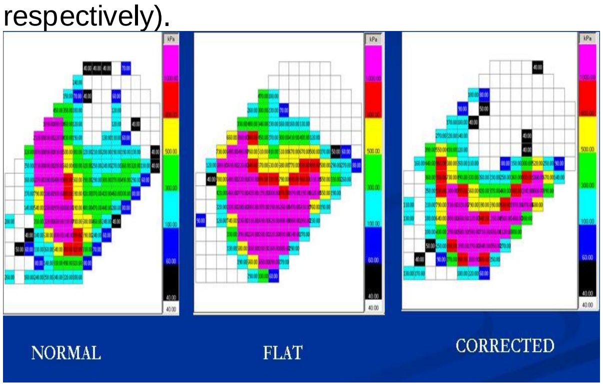

Figure 2: Peak pressure pattern in testing conditions.

DISCUSSION

In the flat foot model the maximal loading is located in the antero-medial region. Subtalar arthroereisis with the Kalix implant did not restore the normal ankle joint pressure in acquired adult flat foot model.

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The Sensitivity of Plantar Pressure and Talonavicular Alignment to Lateral Column Lengthening

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INTRODUCTION

Lateral column lengthening plays a central role in surgical treatment of adult acquired flatfoot deformity. It is thought to restore the medial longitudinal arch by reducing midfoot abduction at the talonavicular joint [1]. Moreover, small changes in the amount of lengthening can increase pressures on the lateral aspect of the foot, which may lead to lateral foot pain[1]. Therefore, it is important to understand the relationship between the amount of lateral column lengthening, and changes in both plantar pressure and the alignment of the talonavicular joint. The goal of this study was to begin to characterize this relationship.

METHODS

Ten lower leg preparations were loaded axially using a 6 degree of freedom robot (ZX165U, Kawasaki). The tibia was fixed in place, and the feet were loaded axially to 400 N, representing about half body weight. The robot was free to rotate about the long axis of the tibia to allow ab/adduction of the foot. The Achilles tendon was loaded to 311 N using a custom-made tendon loading system to transfer the center of pressure to the midfoot. Each specimen was tested with all soft tissue intact, after sectioning the spring ligament and cyclically loading 100 times to 800 N, and after performing three levels of lateral column lengthening in a randomized order. Custom-machined metal wedges that were 6, 8, and 10 mm in thickness were used to accurately lengthen the calcaneus.

Pressure measures were ratio of mean pressure between medial and lateral sides of the forefoot. They were obtained using a custom made pressure measurement system (Pliance, Novel Inc). Talonavicular alignment was described by the angle in the sagittal and transverse planes, which represented talonavicular coverage and sag, respectively. Alignment was obtained using a motion capture system (ProReflexMCU, Qualisys Inc). RMANOVA ($p<0.05$) was used to assess differences between conditions.

RESULTS

Spring ligament sectioning increased the medial/lateral ratio of mean pressure from 1.1 to 1.3 ($p<0.05$), abducted the navicular ($p<0.05$), and caused the talus to sag compared to the intact condition ($p<0.05$), (Fig. 1). Lateral column lengthening of 6, 8 and 10 mm decreased the mean pressure ratio to 0.9, 0.8 and 0.7, respectively ($p<0.05$ for each). It also increased adduction ($p<0.05$ for 8, 10 mm) and decreased sag of the talus relative to the intact foot ($p<0.05$ for 10 mm) (Fig. 1).

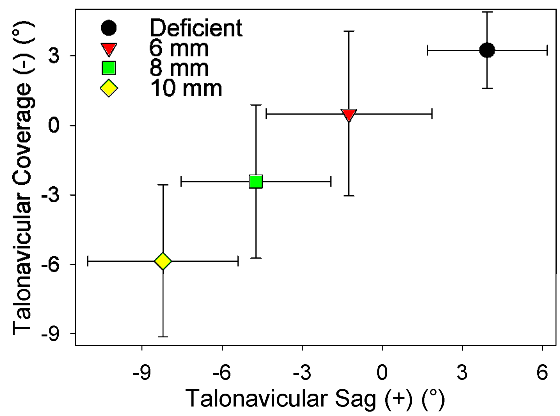

Fig. 1: Change (mean, ±SD) in talonavicular angles relative to the intact condition after release of the spring ligament (Deficient) and after lateral column lengthening using 6, 8, and 10 mm wedges.

DISCUSSION

This study supports the ability of the lateral column lengthening to independently shift plantar pressure laterally, and reduce both adduction and sag of the talonavicular joint [2]. The smallest wedge (6 mm) restored plantar pressure and talonavicular geometry most closely to the intact status, while larger wedges caused an overcorrection. We did not compromise the long plantar ligament and aponeurosis, which could mitigate the ability of the lateral column lengthening to restore plantar pressures and arch geometry, since they may transfer load from the hindfoot to the forefoot.

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A Cadaveric Robotic Gait Simulator with Fuzzy Logic Vertical Ground Reaction Force Control

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INTRODUCTION

Lower limb dynamic cadaveric gait simulators are useful to investigate the biomechanics of the foot and ankle but many systems, including earlier versions of ours, have several common limitations including, simplified tibial kinematics [1-4], greatly reduced velocities [1-4], and open loop trial and error vertical ground reaction force (vGRF) control [1, 2, 4]. To address these limitations, we have developed a high fidelity robotic gait simulator (RGS) capable of simulating the stance phase of gait by prescribing 6-degrees of freedom (DOF) tibial kinematics at biomechanically more realistic velocities, while controlling the vGRF via a novel fuzzy logic vGRF controller.

METHODS

The RGS simulated the stance phase of gait inversely in 2.7s by fixing the cadaveric tibia and fibula in place while a mobile force plate, articulated by a 6-DOF R2000 parallel robot, created the relative tibia to ground motion. Nine force controlled actuators prescribed tendon force estimated from electromyography activity. The target tibial kinematics and vGRF input into the RGS were previously collected from ten subjects. The vGRF was controlled by a fuzzy logic vGRF which altered the tibialis anterior and Achilles tendon force in real time based on the vGRF error. The fuzzy logic controller also changed the R2000's trajectory iteratively between gait simulations similar to an iterative learning controller previously developed by our group [5]. Three cadaveric specimens were used to collect three quartets of trials per foot, with each quartet consisting of three learning trials and one final trial.

RESULTS

The fuzzy logic vGRF controller performed well with an average RMS tracking error of 7.6% body weight between the target *in vivo* and actual *in vitro* vGRF (Figure 1). The sagittal, frontal, and transverse plane angles of the tibia with respect to the ground were almost entirely within ± 1SD of those found *in vivo* for all three feet. The average RMS tracking error for the seven tendons was 3.9N. The mean peak Achilles tendon force needed to attain the target vGRF at 80.4% stance was 1262N.

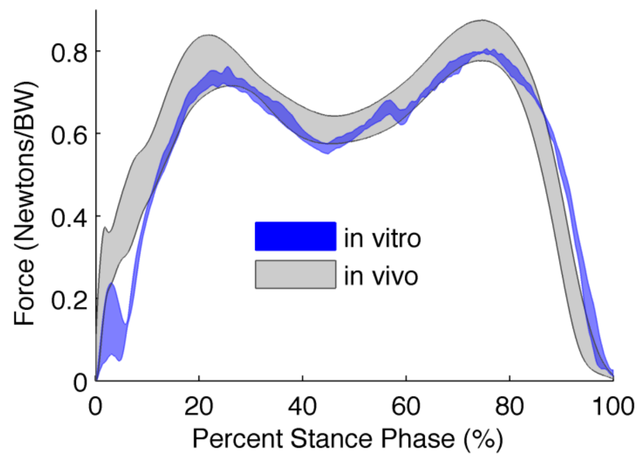

Figure 1: *In vitro* (gray) compared to the target *in vivo* (blue) vGRF.

DISCUSSION

Fuzzy logic vGRF control greatly improved the fidelity of the vGRF as compared to an earlier version of the RGS with trial and error vGRF control which had an RMS tracking error of 30.0% body weight and a 10s stance phase [1]. A stance phase of 2.7s represents a substantial improvement compared to our previous system and most other gait simulators which range from 2s [2] to ~12s [3] to 60s [4]. Performing gait simulations at reduced body weight is a common limitation for many dynamic gait simulators because of specimen frailty [1-3].

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ACKNOWLEDGEMENTS

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In Vitro Description of Foot Bony Motion Using a Cadaveric Robotic Gait Simulator

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INTRODUCTION

An accurate description of the bony motion of the foot during normal gait can aid in identification of foot abnormalities, injury prevention, and surgical correction. Kinematics from i*n vitro* models used with dynamic gait simulators [1-3] have been affected by non-physiologic ground reaction forces (GRFs) [2], low velocity [1,3], low vertical GRF magnitude [1,2], exclusion of bones [1,3], and technically (rather than anatomically) based coordinate systems [1-3]. We have developed a ten-segment foot model for use with our robotic gait simulator (RGS) that addressed the limitations and provides a more thorough and realistic description of the bony motion of the foot during gait.

METHODS

The foot model consisted of the following ten segments, each instrumented with four retro-reflective markers inserted into the bone: tibia/fibula, talus, calcaneus, navicular, cuboid, all three cuneiforms (grouped), first, third, and fifth metatarsals, and the proximal phalanx. All segments used right-handed, anatomically based coordinate systems constructed from three markers or digitized bony landmarks. Kinematic data were collected with a six-camera Vicon MX system. Three neutral foot-type cadaveric specimens were tested with our RGS, which consisted of a force plate mounted to a six-degree of freedom R2000 parallel robot. Tibia-to-ground motion and muscles forces were prescribed from average *in vivo* data. Vertical GRF was scaled to ¾ of the donor's body weight and controlled with fuzzy logic real time control of the Achilles and tibialis anterior tendon and close loop iterative fuzzy logic control of the force plate position. *In vitro* GRF was achieved to an average of ±7.6%*BW of the *in vivo* data. Three stance-phase trials were performed in 2.7 sec each for each foot.

RESULTS AND DISCUSSION

The ten-segment foot model provided repeatable kinematic data for the cadaveric specimens. Average standard deviations in sagittal plane segment angles across all three trials did not exceed 2°. The anatomical coordinate systems proved beneficial in their ability to describe the initial position of each bone; for example, specimen 2 had a more plantar flexed first metatarsal with respect to talus (Figure 1). This angle is an important measure in assessing foot type, and repeatable data such as these demonstrate the utility of the foot model.

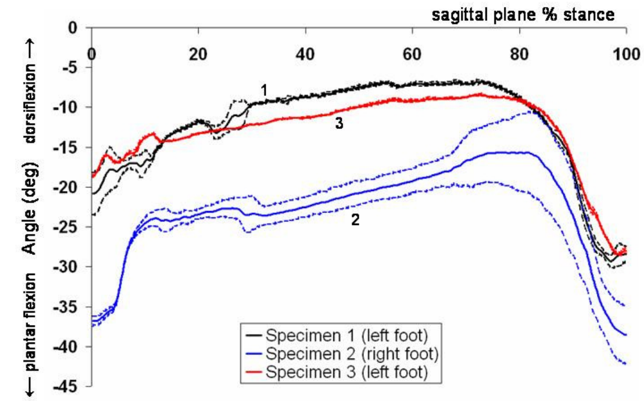

***Figure 1**: Mean ± 1SD of three trials for first metatarsal with respect to talus angle in the sagittal plane*

This study provides an improvement in the understanding of foot bony motion during gait by providing anatomically significant kinematic data during gait simulated with biomechanically realistic GRFs at both a higher velocity and load than is currently available. An extensive variety of kinematic interpretations can be made from these data.

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ACKNOWLEDGEMENTS

This work was funded in part by the VA RR&D, grants A3923 and A4843C.

Comparison of Joint Pressure Changes with Adult Acquired Flatfoot and Talonavicular Joint Fusion for Correction of the Adult Acquired Flatfoot

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INTRODUCTION

Adult acquired flatfoot (AAF) is attributed to a progressive loss of function of several structures linked to the integrity of the medial column of the foot. The pain associated with AAF is not well understood but may be due to increased joint pressures. A common surgical procedure for correction of AAF is fusion of the talonavicular joint [1,2]. While dynamic joint pressures within the foot have been previously investigated [3-5], this study presents dynamic joint pressures prior to AAF, after AAF and after talonavicular fusion for AAF.

METHODS

Five fresh frozen midtibial amputated cadaveric specimens were placed in a dynamic gait simulator. I-scan #6900 pressure sensors were placed in the posterior facet of the subtalar (ST), calcaneocuboid (CC), talonavicular (TN) and naviculocuneiform (NC) joints. Each specimen was walked for 10 trials during 3 conditions (normal, AAF and talonavicular joint fusion). AAF was simulated by detaching the posterior tibial tendon from the simulator and surgically releasing the plantar fascia and the entire spring ligament. Joint pressures were collected at 100 Hz. Peak pressures were averaged within subjects and effect sizes were calculated between conditions.

RESULTS

Medium and large effect sizes were observed when comparing the normal condition to the AAF condition as well as comparing the AAF condition with the TN fusion (table 1). ST joint pressures increased from the normal condition to the AAF condition and decreased significantly when the TN was fused. Likewise, as AAF occurred, CC joint pressures increased significantly and were significantly reduced upon fusion of the TN joint.

Table 1: Measurements of effect size.
* medium effect, ^ large effect.

	ST	NC	CC	TN
Normal vs Flatfoot	0.16	-0.76*	0.25	-0.30
Normal vs TN fusion	-0.70*	0.01	-0.57*	0.79*
Flatfoot vs TN fusion	-0.86 ^	0.77*	-0.82^	1.09^

DISCUSSION

AAF appears to increase joint pressures within the ST and CC joints and decrease the pressures in the NC joint during stance phase of gait. Fusion of the TN joint decreases the pressures in the ST and CC joints and increases the NC joint pressures to normal levels.

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The Diabetic Foot: Unresolved and Future Issues

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The issue of identifying persons at high risk for diabetic foot complications and predicting the likelihood of the development of foot ulcer and amputation are areas of active research. The ability to predict these outcomes is not completely satisfactory but is still in an early stage of development compared to the prediction of other outcomes, such as cardiovascular disease. Research on physiologic measurements associated with risk of foot ulcer will be reviewed as well as other potential causes or correlates of causes including lifestyle, weight, foot deformity, and co-morbid conditions. Measurement of plantar pressure will also be discussed included findings from the Seattle Diabetic Foot Study that collected thousands of these measurements in hundreds of patients with diabetes. Current prediction models will be reviewed with an examination of the validity of methods used in their development. Potential fruitful areas for further investigation of these problems will be discussed.

Stress Thresholds for Injury and Adaptation on the Neuropathic Foot; Moving Targets

(Part of Panel Discussion on "The Diabetic Foot: Unresolved and Future Issues")

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INTRODUCTION

This presentation will explore the unresolved issues related to how the neuropathic foot responds to physical stresses. Is there a "pressure (stress) threshold" for injury? We know that off-loading helps to heal plantar neuropathic ulcers, but how might prolonged low stress place the neuropathic foot at increased risk of injury? Can neuropathic skin adapt to increases in stress (i.e., walking program or weight-bearing exercise) without causing skin breakdown? What, if any, are the criteria for safe weight-bearing activity?

DISCUSSION

Physical stresses can combine in an assortment of ways to result in unnoticed injury to the neuropathic foot. Investigators have long sought a pressure threshold that might indicate a safe level of plantar stresses to the neuropathic foot during walking (1, 2). This threshold likely is a complex "moving target", depending upon a number of instrumentation, physiological, and behavioural factors. In general, low stresses have been considered "good" for the neuropathic foot because evidence clearly shows that neuropathic ulcers will heal by off-loading stresses, particularly through total contact casting. Protection from unnoticed physical stresses on the neuropathic foot has been a hallmark of good care for over 20 years (3). Current activity guidelines often view weight bearing exercise (i.e., walking) to be contraindicated with peripheral neuropathy and even mild deformities. Conventional clinical wisdom suggests that the less weight-bearing activity the better to prevent skin lesions on their feet for people with diabetes and peripheral neuropathy.

There is a growing body of literature, however, that indicates those patients who are least active are at greatest risk for foot lesions (4-6). One recent randomized controlled trial (6) reported modest improvements in weight-bearing activity without increased skin breakdown following a community, weight-bearing intervention. Current data from our lab will be shared showing people with severe neuropathy walking greater than 10,000 steps a day, wearing standard footwear, without developing foot problems. These data challenge our 'conventional wisdom' regarding the ability of people with diabetes and peripheral neuropathy to tolerate, and perhaps even adapt to weight-bearing stresses. Clearly, factors such as prior ulcer, vascular disease, and deformity will affect the magnitude of the stress thresholds for injury and adaptation (7). These "moving target" thresholds related to injury and tissue adaption to physical stress will be discussed within the framework of a general "Physical Stress Theory" (8).

Conclusions

Stress protection promotes healing in neuropathic wounds, but most tissues respond negatively to prolonged stress protection; i.e. they become less tolerant to subsequent stresses and are injured more easily. An important unresolved issue for treatment of the diabetic foot is how we manage the different stress levels at all stages of disease progression to minimize impairments and maximize the person's ability to participate in their home and community activities.

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The diabetic foot: unresolved and future issues

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BACKGROUND

From a biomechanical perspective, the diabetic foot has been studied since the early 1980's. Both retrospective and prospective analyses show that elevated dynamic plantar foot pressure is an important risk factor for the development of plantar foot ulcers, in particular in patients who are complicated by a loss of protective sensation due to the presence of peripheral neuropathy [1]. Foot ulcers are a common precursor of lower extremity amputation in diabetes. For this reason, the reduction of plantar pressure, named "offloading", has become an important component in the prevention and treatment of foot ulcers in this patient group.

CURRENT KNOWLEDGE

Changes in foot structure have been linked with the increases found in pressures underneath the feet in diabetic patients. Charcot midfoot arthropathy, claw toe deformity, hallux rigidus, prominent metatarsal heads and limited joint mobility are deformities that cause high foot pressures [2]. Detailed assessments of the association between claw toe deformity and elevated plantar pressures have shown that distal displacement of the plantar metatarsal fat pads is the underlying mechanism [3]. Several suggestions have been made on the causation of these deformities in diabetes. An important role is indicated for atrophy of intrinsic and extrinsic foot muscles, which is clearly present in patients who have peripheral neuropathy. However, clear evidence for this causal relationship has not yet been provided [4].

Areas of the foot that show high plantar pressures can be offloaded using different modalities. Total contact casts (TCCs), removable walkers, cast shoes and offloading shoes are often prescribed for ulcer treatment, whereas therapeutic footwear is often prescribed to patients once an ulcer has healed. Most healing devices are very effective in reducing foot pressures in diabetic patients. TCC's can reduce forefoot peak pressure by 80% compared to a control shoe [5]. This is likely an important component for the proven efficacy of these devices to heal plantar ulcers [6]. Generally, uncomplicated neuropathic foot ulcers can be healed in 6-8 weeks time.

UNRESOLVED AND FUTURE ISSUES

Many issues regarding the development of foot deformity in patients with diabetes are unresolved. In particular the role of motor neuropathy remains to be investigated in this context. Furthermore, despite several attempts, a threshold of pressure above which foot ulcers develop or below which they heal has not been identified to date [7]. Many factors influence this relationship such as patient behaviour (adherence to treatment, amount of barefoot walking, activity level) and other biomechanical factors, for example shear stress. Future research should focus on this delicate relationship between biomechanical and behavioural factors in diabetic foot outcomes.

The efficacy of therapeutic footwear to prevent (recurrence of) foot ulcers has not been clearly identified, despite widespread prescription, and should be proven in well designed prospective trials. Finally, surveys in the US and Europe show that for ulcer treatment the most effective devices are not used in clinical practice. This gap between evidence and practice needs to be bridged, for example through the adoption and implementation of evidence-based and specific guidelines that may alter expectations on what entails adequate healing [8].

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Feasibility of using plantar temperature to assess the biomechanics of the plantar foot

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Introduction

The use of plantar temperature increase as an indicator of plantar loading has been suggested by a number of investigators before [1,2]. However this idea has not been validated. The aim of this study was to reveal a potential linear relationship between plantar stresses and walking-induced temperature increase under the foot. If confirmed, thermographs can be used to assess plantar tri-axial and/or shear loading, which may lead to significant advancements in the biomechanics of the diabetic foot.

Methods

Thirteen healthy volunteers were recruited. Informed consent was obtained before the study. An infrared thermal camera was used to collect pre-exercise baseline plantar temperature profiles. Tri-axial plantar stress distributions were recorded by a custom-built platform. Enrollees were asked to walk barefoot on a treadmill for 10 minutes. Post-exercise temperature distribution was recorded. Locations of peak temperature increase and peak stresses were determined under each foot. The rate of matches between these sites was found. Plantar stress magnitudes were correlated against the peak temperature increase values (Figure 1).

Results

Peak shear location matched the peak temperature increase site in only 23% of the subjects. On the other hand, peak resultant stress occurred at the peak temperature increase site in 39% of the volunteers. A significant correlation was reported between the magnitudes of peak shear and temperature increase (R=0.78, p=0.002).

Conclusions

Only a moderate linear relationship could be established between the horizontal component of the plantar stresses and peak plantar temperature increase. A potential non-linear association between these two parameters needs to be explored by using non-linear modeling schemes. If such a relationship is present and can be modeled, it may be possible to predict plantar shear which can help assess ulcer risk in the diabetic foot. As it stands, thermographs are not yet reliable to use for shear prediction, but our results are encouraging for further investigation.

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Acknowledgment

This study was possible due to research funds from the Ohio College of Podiatric Medicine.

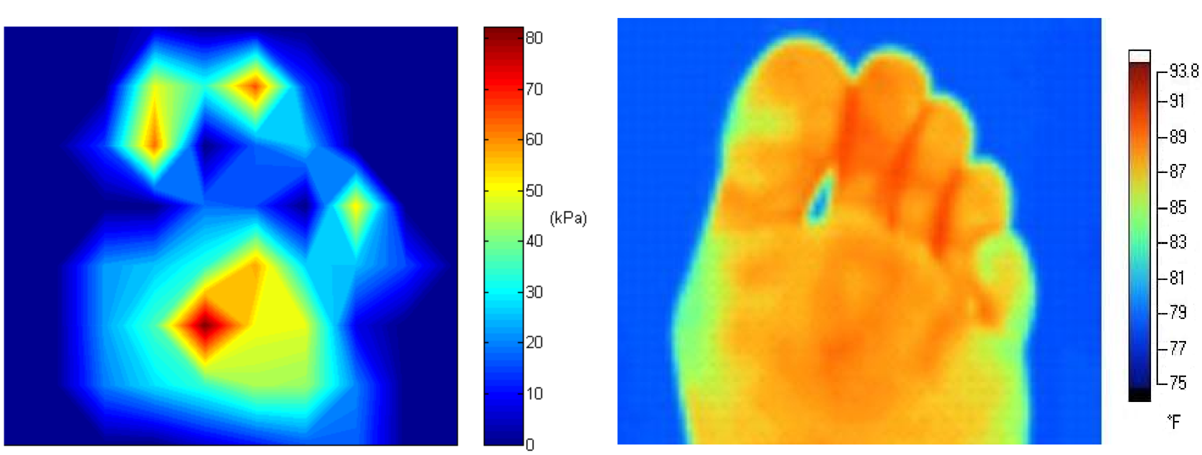

Figure 1. Plantar shear stress (left) and post-exercise temperature (right) profiles of a representative subject.

Feasibility of using plantar temperature to assess the biomechanics of the plantar foot

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Introduction

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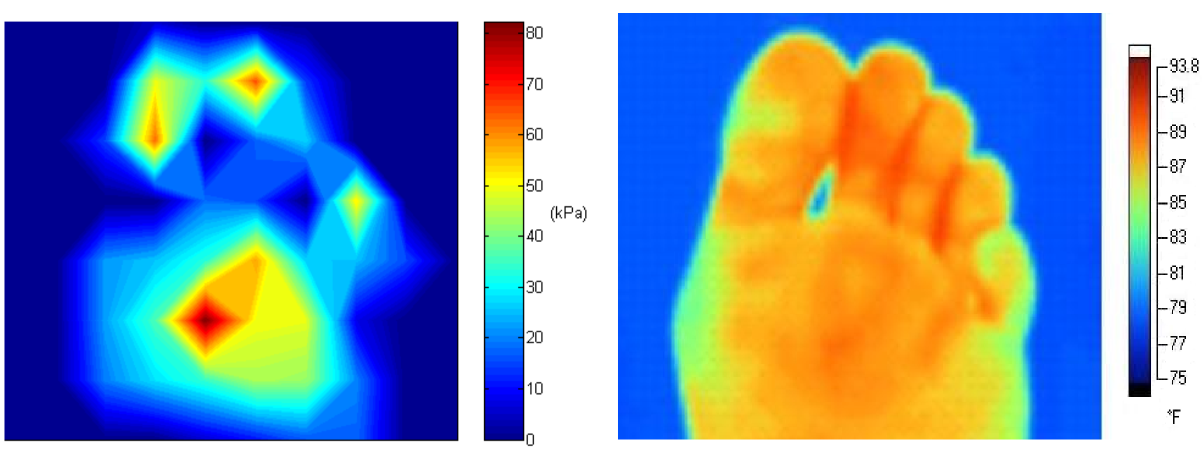

Figure 1. Plantar shear stress (left) and post-exercise temperature (right) profiles of a representative subject.

The Compressive Mechanical Properties of Diabetic Plantar Soft Tissue

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INTRODUCTION

Diabetic subjects are at an increased risk of developing plantar ulcers [1,2]. Knowledge of the physiologic properties of the plantar soft tissue is critical to understanding mechanisms of ulcer formation and improving treatment options. The purpose of this study was to determine the compressive mechanical properties of diabetic and non-diabetic plantar soft tissue from six relevant locations, namely the hallux, first, third, and fifth metatarsal heads, lateral midfoot, and calcaneus.

METHODS

Cylindrical specimens (1.905cm diameter) from these locations were excised and separated from the skin and bone from 4 diabetic and 4 non-diabetic age-matched, elderly, fresh-frozen cadaveric feet. Specimens were then subjected to biomechanically realistic strains of ~50% in compression using triangle wave tests conducted at five frequencies ranging from 1 to 10 Hz to determine tissue modulus, energy loss, and strain rate dependence.

RESULTS

Diabetic vs. non-diabetic results across all specimens, locations, and testing frequencies demonstrated altered mechanical properties (Figure 1) with significantly increased modulus (1146.7 vs. 593.0kPa) but no change in energy loss (68.5 vs. 67.9%). All tissue showed strain rate dependence (Figures 1 and 2). Tissue beneath the calcaneus had decreased modulus and energy loss compared to other areas.

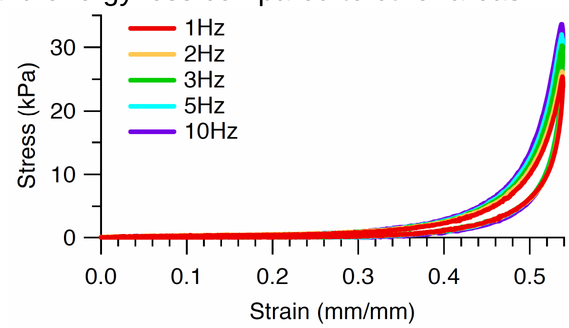

Figure 2: Stress-strain response with toe region to inflection point and an increase in modulus at higher strains and frequencies.

DISCUSSION

The results of this study could be used to generate material properties for all areas of the plantar soft tissue in diabetic or non-diabetic feet, with implications for foot computational modeling efforts and potentially for pressure alleviating footwear that can reduce ulceration.

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ACKNOWLEDGEMENTS

This work was supported by NIH grant 1R01 DK75633 and the VA RR&D Service.

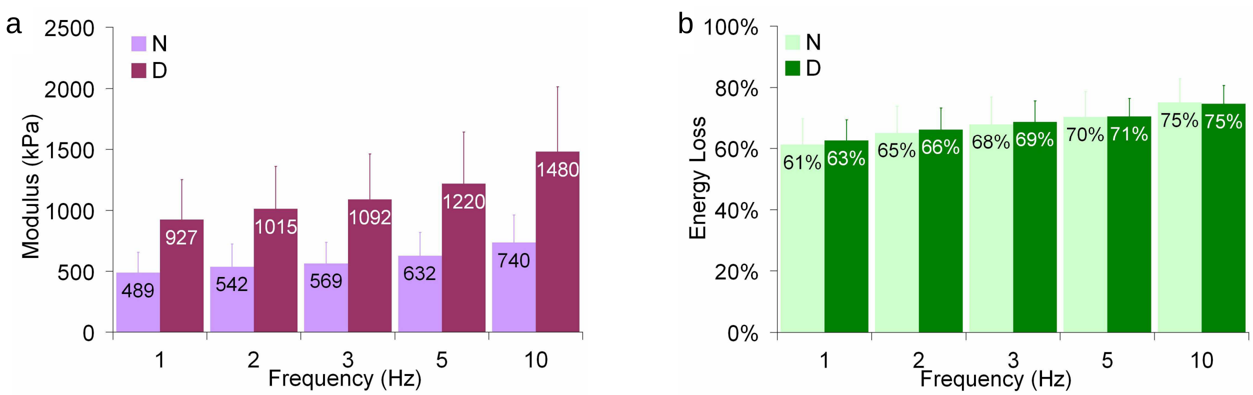

Figure 1: (a) Modulus and (b) energy loss as a function of frequency across all locations where N = non-diabetic and D = diabetic.

A longitudinal investigation into functional and physical durability of insoles used for the preventative management of neuropathic diabetic feet

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INTRODUCTION

Neuropathic diabetic foot ulceration may be prevented if plantar tissue stress can be reduced. Orthotic therapy is one practical method commonly used to maintain tissue integrity [1]. Limited longitudinal data suggests insole effect may be compromised after 6-months of wear, prior to visual insole fatigue [2]. Insole replacement is often only triggered when foot health status deteriorates or the device appears worn out. These clinical indicators suggest functional performance may already have been comprised, exposing patients to unnecessary increased ulcer risk. The purpose of this study was to investigate the functional and physical durability of insoles used for ulcer prevention in neuropathic diabetic feet over 12-months.

METHOD

As part of a larger clinical trial, 60 consecutive neuropathic individuals with diabetes were provided with insoles and footwear. The functional durability of each insole over a 12-month period was evaluated in terms of peak pressure reduction using the F-scan in-shoe pressure measurement device. Further evaluation was undertaken through the repeated measurement of material depth at two sites of interest (sub 1st metatarsal head and central heel seat) measured in millimetres (mm). ANOVA was used to assess change in peak pressure and compression across three time periods (issue of intervention, 6-month and 12-month follow-up).

RESULTS

A total of 17 were lost to follow up; 11 through illness or death, 6 withdrew or failed to attend follow-up. Analysis was conducted using two strategies; all available data n=43 and fully compliant n=18; equivalent to wearing insoles for a minimum 7 hours a day, 7 days a week for 12 months. No significant difference was found in the reduction of mean peak pressure tested across three time periods (insole issue, 6-months and 12 months) using the two analysis strategies ($p<0.05$).

For both sites significant differences in insole depth were identified between issue and 6-months (Mean difference: central heel seat, n=43, 0.76mm, p=0.008. n=18, 0.53mm, p=0.008 and Sub 1st met head, n=43, 0.33mm, $p<0.001$. n=18, 0.32mm, $p<0.001$) and issue and 12-months (Mean difference: Central heel seat, n=43, 0.59mm, $p<0.001$. n=18 0.50mm, p=0.010 and sub 1st met head, n=43, 0.35mm, $p<0.001$. n=18, 0.31mm, p=0.022), however the analysis showed no significant difference between 6 and 12-month comparisons ($p<0.05$). The greatest degree of insole compression occurred during the initial 6-months following insole issue.

DISSCUSSION

This study investigated the durability of insoles used for ulcer prevention in neuropathic diabetic feet. It found that whilst the insole condition deteriorated, particularly over the first 6-months, this deterioration did not appear to affect insole function. In contrast to other reported studies, the reduction in peak pressure continued for the full 12-month study duration [2]. Differences in insole design could account the increased durability demonstrated by our insole.

The results suggest that physical changes in insole condition through normal usage may not signify a need for insole replacement. Therefore in-shoe pressure measurement evaluation would appear the most accurate method of determining functional loss and frequency of insole replacement.

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Orthotic, inserts and shoes – aligning the skeleton

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THE FACTS

Millions of people run and jog. Some of them are injured and excessive pronation has been proposed as one major reason for some running injuries and the sport shoe is often seen as one factor responsible for these injuries.

The concept of movement (or rearfoot) control was developed, and strategies were studied to reduce "potentially harmful" foot pronation through appropriate running shoes and shoe inserts. However, results of recent studies challenged the proposed association between shoes and/or orthotics/inserts aligning the skeleton and foot pronation being a strong predictor of running injuries.

The facts are that:

- The kinematics of the lower extremities change little when changing the shoe and/or the orthotic conditions.
- Kinematic changes (if any) are typically subject-specific, small and often not systematic Thus, shoes, orthotics and/or inserts align the skeleton in normal subjects little.
- The correlation between foot "alignment" and the frequency of injuries is small.

THE NEW PARADIGM

We propose (Nigg, 2001) that for a given movement task each joint has a "preferred movement path". Appropriate muscles will be activated if a shoe/orthotic/insert intervention tries to produce a skeletal movement, which is not in the "preferred movement path" or the "minimal resistance path" with the goal to keep the movement in this standard path.

This concept would be in agreement with the experimental observations that the movement changes due to shoe/orthotic interventions are typically small. However, the criteria used by the locomotor system are not known or understood at this point in time.

SUPPORTING EVIDENCE

Differences in local and/or global energy demands (or differences in acting forces) may be seen in differences in muscle activity and/or in differences of local oxygenation or total oxygen consumption. For instance: The study for different custom made and generic inserts (Mündermann et al., 2003; Mündermann et al., 2004) provides direct evidence for changes in muscle activity for different insert conditions. The result of this study showed substantial and often significant differences in EMG intensities for the same movement task for the tested subjects, which were all "pronators". The result that in most cases the EMG intensity increased with the orthotic intervention may, at first glance, be surprising. It fits well, however with the concept of the preferred movement path.

FINAL COMMENTS

The presented "preferred movement path" concept for the functioning of inserts, orthotics and shoes has not yet been supported by enough experimental and/or theoretical evidence. There is some initial experimental evidence in support of it, which is, however, rather weak at this point in time. Most of the indicators are necessary but not sufficient conditions to support the new paradigm. As a result, more and stronger evidence must be provided to support or reject the proposed paradigm.

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Relevance ranking of features involved in modelling dorsal pressures on the foot surface

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INTRODUCTION

This work presents the use of Machine Learning techniques to analyse the most relevant features involved in modelling dorsal pressures on the foot surface. Modelling is a promising way for exploring comfort in footwear design. In previous works [1, 2] we presented an approach based on Artificial Neural Networks (ANNs) to automate prediction of pressure exerted over the foot surface by the shoe upper while walking. Those models produced accurate predictors for the dorsal pressures on the foot surface. In this work, we use the neural predictors to evaluate the most relevant features that contribute to the prediction.

METHODS

Four subjects aged 27-31 were required to walk on a platform wearing five different kinds of shoes. The shoes were manufactured from the same design but with five different types of shoe upper material. The pressures exerted by the shoe uppers over the foot surface were measured for each subject while walking using 14 sensors (TEKSCAN Flexiforce®) placed on 14 anatomical points over the foot surface, under the shoe upper (Figure 1).

Figure 1: Distribution of sensors on the foot surface.

The most widely used ANN, the so-called Multilayer Perceptron (MLP), was used to model dorsal pressures on the foot surface [1, 2]. Inputs to MLP were the 3-dimensional position of the sensors as well as the characteristics of the material, namely, Young modulus, Poisson coefficient and thickness.

The achieved prediction was accurate with correlation coefficients *r* between the prediction and the desired signal higher than 0.95 in the validation data set. We carried out a sensitivity analysis to measure the relevance of the six independent variables in the prediction.

RESULTS AND DISCUSSION

Relevance analysis showed that in those subjects for whom the prediction was most accurate (r>0.97 in the validation data set), the characteristics of the material (Poisson coefficient and Young modulus) were more relevant than the position of the sensors. Taking into account all the subjects (r=0.89), Poisson coefficient was still the most relevant feature, but the position of the sensors in the directions *x* and *y* parallel to the platform appeared as the second and third most relevant features, respectively. Nevertheless, the relevance of all six features could be considered as relevant and there was not any feature that could be easily withdrawn from the predictor without affecting accuracy. The results suggest that the experiment setup was appropriate since all the variables affected to the pressure prediction. The fact that the material properties affect prediction more for those subjects with a more accurate prediction suggests that the pressure on each sensor depends rather on these properties than on the sensor position.

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The impact of a health Flip Flop on asymptomatic gait

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INTRODUCTION

The area of health and wellbeing footwear is highly commercially valuable, however it lacks significant scientific validation. A health flip flop has been produced consisting of a multi-density midsole to produce 'micro-instability' during mid-stance. Consumers report alleviated conditions such as lower back pain, heel pain and knee pain, however the reason for these improvements is not evident. The impact of the health flip flop on gait was assessed in a group of healthy females in order to quantify gait alterations and identify potential mechanisms.

METHODS

Seventeen asymptomatic female subjects (age: 43±11.4 years, height: 162.8±5.13 cm, mass: 64.8±11.46kg) participated in the study. Subjects walked in 4 conditions: barefoot (BF), health flip flop (HF), standard flip flop (SF) and rocker soled sandal (RS). Subjects performed five walking trials in each condition at a self-selected walking pace for each condition. Data was collected from walking trials using a 12 camera Qualysis system and 2 AMTI Force Plates operating at 100 and 3000 Hz respectively. Marker placement defined the foot, shank, thigh, pelvis and thorax segments. Data was exported to Visual 3D for processing and analysis. Variables included temporal parameters, lower limb kinematics, posture, ground reaction force and lower limb joint moments. Kinetics were normalised for body and footwear mass. Statistical analysis, undertaken in SPSS, included ANOVA and Bonferonni adjustment (significance set at .05).

RESULTS

Temporal characteristics identified a non-significant faster walking speed in the HF and longer stance times than other footwear conditions and significantly longer than SF. Kinematics of the lower limb and trunk showed few statistically significant differences between conditions with gait patterns remaining relatively consistent. GRF variables indicated a lower magnitude and longer durations in loading, significant when compared to SF and RS respectively (Table1).

DISCUSSION

The functionality of the HF is evident by the increased walking speed. The increased stance time can potentially be attributed to the construction of the footbed prolonging mid-stance. The HF demonstrated a reduction in foot motion in the frontal plane compared to the SF as well as reductions in loading (Table 1), which may relate to some reported symptoms. Differences with the RS condition were consistent with those previously identified at the ankle and knee [1,2]. Future studies will involve symptomatic populations.

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Table 1: Selected variables demonstrating differences for HF on the right leg/foot.

Variable	Condition				Significance		
	BF	HF	SF	RS	HF-BF	HF-SF	HF-RS
Walking Speed ($m.s^{-1}$)	1.26±0.097	1.30±0.096	1.28±0.082	1.28±0.090	0.172	1.000	0.875
Stance Time (s)	0.616±0.0377	0.647±0.0362	0.632±0.0372	0.642±0.0377	0.001	0.117	1.000
Maximum Ankle Eversion (°)	-4.4±1.92	-3.7±2.69	-4.1±2.21	-5.7±2.21	1.000	1.000	0.011
Time of Maximum Eversion (% Stance)	46.1±10.6	42.8±11.1	41.2±10.6	26.7±9.7	1.000	1.000	0.000
Total Ankle Coronal Excursion (°)	14.8±3.46	14.2±3.58	15.9±4.54	14.2±3.64	1.000	0.042	0.147
Peak Loading GRF (BW)	1.08±0.055	1.09±0.047	1.12±0.058	1.11±0.056	1.000	0.042	1.000
Time of Peak Loading GRF (% Stance)	22.8±2.34	24.2±2.57	22.7±2.48	21.8±3.72	0.099	0.007	0.048

Five-Toed Socks with Grippers on the Foot Sole Improve Static Postural Control Among Healthy Young Adults as Measured with Time-to-Boundary Analysis

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INTRODUCTION

In Japan, five-toed socks with grippers on the foot sole are commonly worn by both athletes and physical laborers. These socks are gaining popularity because of the perceived improvements in balance that result from their use. No scientific research, however, has been conducted to examine the effect of the socks. Therefore, the purpose of this study is to determine the effectiveness of the socks in influencing static postural control among healthy young adults.

METHODS

Twenty three healthy young adults (6 males, 17 females;22.3±3.3yrs;168.3±9.2cm;67.5±11.9kg) were recruited to complete three testing sessions, separated by approximately one week. The subjects were tested under three conditions: wearing five-toed socks with grippers on the foot sole (FS), wearing regular socks (RS), and wearing no socks (NS). For each condition, static postural control was assessed on a force plate (model 4060NC; Bertec Corp. Inc., Columbus, OH) with the subject in a single-limb stance with eyes open (EO) and eyes closed (EC). The test limb was determined as the limb the subject would choose to stand on while kicking a ball. Center of Pressure (COP) data were sampled at 50Hz. The subjects completed three 15-second trials with a one-minute rest between trials. The order of the sock conditions was randomized. Motion Monitor software (Innovative Sports Training, Inc., Chicago, IL) collected COP data during the single-limb stance test. MATLAB software (Mathworks Inc., Natick, MA) was utilized to calculate Time-to-Boundary (TTB) variables in both the anteroposterior (AP) and mediolateral (ML) directions.

The TTB dependent variables, calculated for AP and ML directions separately for EO and EC trials, were the TTB absolute minimum and mean of the TTB minima, and standard deviation of TTB minima [1,2]. The independent variable was sock condition (FS, RS, and NS). For each dependent variable, a one-way repeated measures ANOVA was performed. Significance was set a priori at $p \leq 0.05$.

RESULTS

There was a significant main effect of the sock condition for the mean of the TTB minima in the ML direction during EO trials ($F_{2,20}$ =3.32; p=0.05). A Post-hoc paired t-test revealed the FS condition (1.58±0.76 seconds) had a significantly higher TTB value than the NS condition (1.45±0.70 seconds) (p=0.04), indicating that FS is associated with improved static postural control. While it was not a significant relationship, The FS condition also had a higher TTB value than RS (1.41±0.59 seconds) condition (p=0.06) (Figure 1).

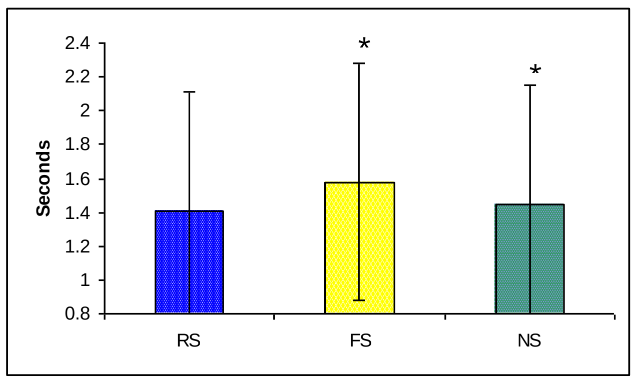

Figure 1: The mean of the TTB minima with ML direction during EO trials.

DISCUSSION

This study was aimed at determining the influence of FS on static postural control in healthy young adults. These results suggest that the FS condition is associated with increased static postural stability when compared to the RS and NS conditions. Continued research is needed to evaluate if this style of sock is able to influence balance differently in the elderly population.

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The Effect of Flexible Ski Boots on Knee Joint Loading and Muscle Activation in a Simulated Skiing Movement

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INTRODUCTION

Skiing manoeuvres such as mogul skiing technique leave participants vulnerable to injury. Many of these may be avoidable with a greater understanding of biomechanical parameters related to equipment design. It has been noted that the knee is involved in the majority of freestyle skiing injuries [1]. Harmful body postures may be avoided through appropriate adjustment of equipment and riding technique. A modified ski boot which enables greater forward lean may lead to a more advantageous position of the whole body centre of mass above the base of support.
The aim of the current study was therefore to investigate the effects of such a boot in a laboratory based simulation of skiing movements.

METHODS

Ten healthy subjects without any ankle or lower extremity restraints (age: 26+/-4.4 y; height: 176+/-3 cm; body mass: 78.4+/-6.8 kg) participated in the study. Two custom-made force platforms were mounted on a robotic perturbator which imposed cyclic movements simulating skiing on an ungroomed slope. Ground reaction forces (GRF) (1000 Hz) and 3-dimensional kinematics (Qualisys Oqus, 8 cameras, 250 Hz) were recorded. An individually scaled full body model (AnyBody Tech) was used to calculate joint reactions. The model included all muscle groups crossing the ankle, hip and knee joints (Figure 1). A skiboot with increased range of movement at the ankle joint in flexion-extension direction was compared to an unmodified standard ski boot (Head Raptor). Electromyographic (EMG) recordings of the main muscle groups crossing the ankle and knee joints were recorded simultaneously (biovision, 2000 Hz). Boot stiffness was tested in a material test and included in the model as an internal joint stiffness.

RESULTS

When using the flexible boots subjects managed to stay with their centre of mass more forward during the simulated compensation movements. This was mainly realised by a greater dorsal extension of the ankle joints. Dynamically, trunk movement was reduced in the flexible boots. There was an increased anterior posterior movement measured at the boot level which was linked to both a reduced knee and hip moment. Simultaneously, the muscle forces of the quadriceps group were reduced with the hamstrings showing higher forces.

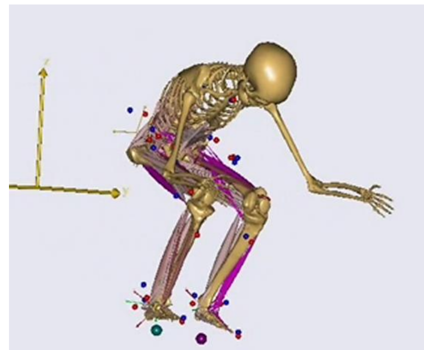

Figure 1: 3D model of a skier.

DISCUSSION

The changes found indicate a beneficial alteration of skier positioning with regard to the ski. Slightly reduced net joint moments demonstrate small effects on effective joint loading. However, the substantial changes in muscle forces indicate an improved muscular stabilisation of the knee joint possibly decreasing the quadriceps induced loading of the anterior cruciate ligament. If such changes would also apply to real life skiing needs further verification.

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The Evolution of the Human Foot and Bipedalism

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Bipedal walking and running are widely recognized as the hallmark of humanity. Compared to other primates, humans possess a number of anatomical adaptations to habitual use of an upright bipedal striding gait. Pedal adaptations include a large calcaneus, a long stiff midfoot, a longitudinal arch, robust first and fifth metatarsals, and short toes. These anatomical features have been associated with heel strike plantigrady, a unique roll-over pattern, and conservation of energy during bipedal movement. A rich fossil record and increasing numbers of studies in living animals allow us to interpret human foot function and human foot pathology in a comparative and evolutionary context. While human gait is unique, a number of other primates use bipedalism in specific facultative contexts. In particular, studies on chimpanzees have informed our interpretation of the early hominin fossil record and helped demonstrate the complex and mosaic nature of the foot and bipedalism in early hominins.

Like humans, chimpanzees and other apes heel strike at touchdown, a unique trait of large-bodied hominoids (great apes and humans) that appears to provide a longer effective limb length and stride length to animals walking with extended knees and high hip protraction. Thus great apes appear to "roll over" their feet in way not unlike modern humans. Chimpanzees, however, differ from humans in having a mobile midfoot that allows them to maintain a secure grasp during climbing but does not provide an effective lever for the foot. Humans, in contrast, have a long stiff midfoot and a longitudinal arch which prevents a midfoot break as the heel comes up. This results in a rollover pattern that subsequently loads the lateral (4^{th} and 5^{th}) metatarsals, which results in distinctly robust human lateral metatarsals. As load passes from lateral to medial across the metatarsal heads, humans experience toe-off loads on the first and second metatarsal and toes that are relatively higher than any other primates. Therefore, although apes and humans gain roll-over advantage from heel-strike, the loading environment of the modern human foot is different from that of our ape relatives, which raises the question of when this pattern appeared during human evolution

Evidence present in fossil hominin feet and footprints allows us to interpret the path by which hominins acquired these unique anatomical and functional features. Recent discoveries of *Ardipithecus* (5 mya), ancient trackways (1.5 mya), and the fossil pedal remains of the Flores hominid (13 kya), have demonstrated the diversity of early hominin bipeds and the mosaic of human and nonhuman skeletal features they possess.

Our studies of functional anatomy in humans and other primates suggest that early hominins, and even the Flores hominin, had a lateral foot column similar to humans but gracile halluces that were not close-packed in dorsiflexion. These diminutive early human ancestors likely used a roll-over pattern resembling modern humans during early stance but similar to the pattern seen in apes later in stance with the medial midfoot weight bearing, a laterally-placed toe-off, and lack of full extension or high loads on the toes during toe-off. While this type of early hominin load bearing provided a rigid lever during walking, later hominins developed a stable forefoot and stiff medial midfoot promoting stability in running. Using *in vivo* study of footprints and foot pressure, we can analyze the structure of fossil footprints, providing additional insight into actual pedal function in early hominins. Our understanding of the evolutionary history of the human foot and bipedalism is enlightened by both paleontological discovery and comparative anatomical and functional data.

Footprint-based gait reconstruction of the 3.75 Ma Laetoli hominin

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INTRODUCTION

Footprints are mechanical recordings of gait and thus constitute our most direct evidence regarding extinct species' modes of locomotion. An individual from the species *A. afarensis*, a species made publicly famous by the discovery of the 'Lucy' skeleton in 1974, left bipedal footprints in volcanic ash at the Laetoli site in northern Tanzania circa 3.75 Ma [1], but debate persists regarding the nature of this bipedal walking; was this an early and primitive form of bipedalism [2] or was this in most respects modern upright human walking [3]? The purpose of this study was to use image processing and gait simulation techniques to determine how far along this continuum of bipedalisms the Laetoli prints most likely lie.

METHODS

Two main techniques were used: (i) footprint and foot pressure image analysis and (ii) bipedal gait simulation. The former was conducted on high-resolution scans of original Laetoli print casts (Fig.1) obtained at the National Museum of Kenya and on experimental plantar pressure data collected from human and a variety of primate subjects. The latter was conducted both with a relatively simple planar foot model consisting of multiple spring contacts and in a planar finite element environment with continuous foot geometry and more realistic plastically deforming ground (Fig.2).

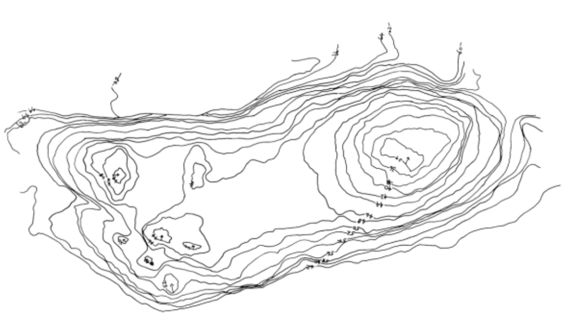

Figure 1. Depth contour plot of a single Laetoli print.

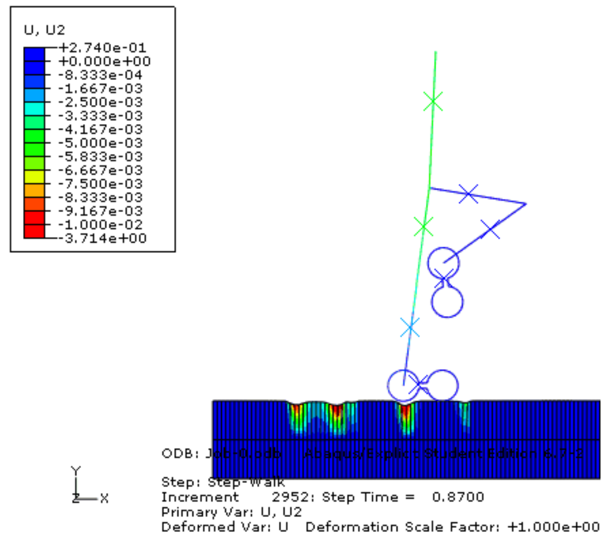

Figure 2. Example finite element gait simulation. Color contours encode vertical displacement ('U2', cm).

RESULTS AND DISCUSSION

While the results, we believe, provide strong evidence for one side of the ongoing debate, due to an embargo the details and their implications are presently withheld, but will be presented orally. We expect the embargo to be lifted by the time of the meeting.

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Postural control and tone of gastrocnemius muscle in male soccer players and endurance trained athletes

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INTRODUCTION

Masi and Hannon [1] give a new term for muscle tone description – a human resting muscle tone (HRMT) (EMG-silent). It seems there is a reason to distribute human resting muscle tone into two: passive resting and postural tone. The postural tone, which is predominantly distributed among force moments of the extensor muscles, play a dominant role in keeping erect posture. Present study was conducted to investigate postural control and tone of gastrocnemius muscle in male soccer players (SP), trained for both explosive power and endurance (n=12) and endurance trained athletes (ET) (long distance runners, cross-country skiers; n=13) aged 19-25 years with sport training period of 7 to 20 years.

METHODS

Static standing balance was assessed by centre of foot pressure (COP) sway registered during 30 s quiet bipedal standing with eyes open on force platform (stable ground; SG) (*Kistler* 9286A, Switzerland) using *Sway* software of motion analysis system *Elite* (BTS S.p.A., Italy) and on balance pad.

Myotonometry was used to investigate muscle tone, elasticity and stiffness. Natural oscillation frequency (characterizes the tone), logarithmic decrement (characterizes the elasticity) and stiffness of gastrocnemius muscles medial head (GM) of dominant leg were recorded by myometer MYOTON-3 (Müomeeteria Ltd, Estonia) in the central part of muscle belly using MultiScan mode (5 times). Mean value was accepted for analysis. Myotonometric measurements were performed at rest supine (resting tone) and during standing for both balance tests (postural tone).

RESULTS

Resting tone of GM was significantly ($p<0.05$) lower in SP than in ET athletes (mean ± SE; 12.56 ± 0.21 and 13.23 ± 0.31 Hz, respectively). SP had higher elasticity (lower decrement) of GM (1.14 ± 0.07 and 1.45 ± 0.12, respectively) at rest. Postural tone at standing did not differ between groups ($p>0.05$). ET athletes had higher elasticity of GM than SP in both balance tests. Significantly lower ($p<0.05$) COP sway range in medio-lateral direction and COP sway velocity as well as shorter COP trace length were found in SP group as compared to ET athletes (Table 1).

DISCUSSION

Main findings of our study were: (1) SP athletes had a significantly better static postural control and (2) lower resting tone of GM than ET athletes. Postural control is an important factor in active sport for prevention of injuries and in rehabilitation [2]. Previously it was found that postural sway characteristics improve according to the training type [3]. Skeletal muscle tone estimation gives additional information about postural control in athletes.

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Table 1. COP sway characteristics in SP and ET athletes (mean ± SE, * p<0.05)

Characteristics	Standing on stable ground		Standing on balance pad	
	SP	ET	SP	ET
ML COP sway range [mm]	9.63 ± 0.78	12.82 ± 1.57 *	24.53 ± 1.57	26.06 ± 1.85
AP COP sway range [mm]	18.89 ± 1.94	18.87 ± 2.09	33.78 ± 2.25	37.80 ± 3.92
COP trace length [cm]	282.32 ± 8.49	304.77 ± 9.61 *	297.44 ± 9.16	312.30 ± 10.29
COP sway velocity [mm/s]	94.12 ± 2.83	101.61 ± 3.20 *	99.16 ± 3.06	104.12 ± 3.43

Real-World Locomotor Behavior Following Clubfoot Treatment: 10 Year Outcomes

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INTRODUCTION

Treatments for congenital talipes equinovarus (clubfoot) have evolved considerably in the last decade. Posteromedial release (PMR) surgeries have been replaced with Ponseti serial casting and French functional (physical therapy) methods with excellent technical results [1, 2]. Computerized gait analysis has been utilized to assess ankle kinematics and kinetics following interventions for clubfoot [3]. The aim of this study is to complement technical gait metrics with a sensitive method of evaluating real-world locomotor endurance behavior following treatment for clubfoot.

METHODS

Seven children who were treated for severe clubfoot as infants were followed for ten years. Their parents gave informed consent to participate in this prospective study. This initial cohort had French functional treatment, but 5 of the 7 went on to have PMRs between age 7 months and 3.7 years. At ten years of age, each child wore a StepWatch Activity Monitor to count steps in each 10 second time interval for a period of one week; 10 age-matched normal children served as controls. Custom code identified each bout of continuous steps and categorized the step rate into four different levels of intensity for each bout: max step rate; high step rate; med step rate; low step rate. Steps were sorted into nine bout durations with a quasi-log scaling: <20 sec; 20-40 sec; 40-60 sec; 1-1.5 min; 1.5-2 min; 2-4 min; 4-8 min; 8-16 min; and 16+ min. These bout lengths were chosen to differentiate walking for transport from long-duration locomotor behavior.

RESULTS

Children treated for clubfoot by French functional methods and eventually PMRs appear to be able to achieve the bout length distributions and intensities for nearly all ambulation except bouts lasting more than 16 minutes. Typically developing children had 2606 steps in bouts of 16+ min versus 1345 steps in children treated for clubfoot ($p < 0.013$).

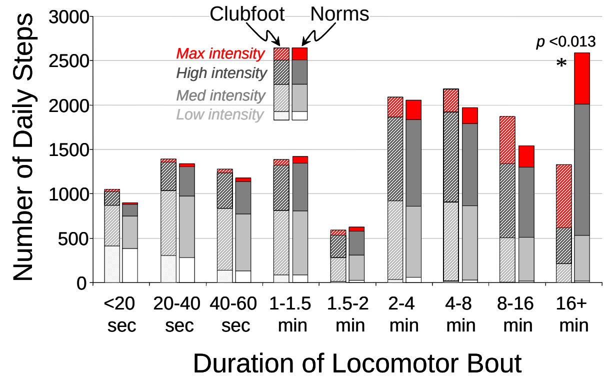

Figure 1: Ten year follow-up: Intensity, number of steps and duration of locomotor bouts in children treated for clubfoot versus typically developing children.

DISCUSSION

These data suggest that children treated for clubfoot initially with French functional (physical therapy) that progressed to PMRs likely have the ankle strength and power [2] to meet the community mobility demand for short duration locomotor bouts. However, the endurance of their ankle during longer duration activities may be compromised, limiting participation in play behavior with peers. This may be a local muscular endurance issue, or the ankle function may have resulted in reduced cardiopulmonary endurance in these children. These data compliment computerized gait analysis that accurately characterizes ankle moment and power generating capacity during walking by describing the real-world locomotor behavior that challenges endurance. This method describing walking bout intensity and duration may be useful in evaluating long-term outcomes from non-operative clubfoot treatments.

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Strength training of the foot and ankle in paediatric neuromuscular disease

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INTRODUCTION

Charcot-Marie-Tooth disease (CMT) is the most common neuromuscular disorder. It demyelinates peripheral nerves, causing progressive weakness and deformity of the hands and feet, difficulty walking, and sensory loss. Weakness of foot dorsiflexion is the most prominent symptom of the disease and contributes to foot drop, cavus foot deformity, ankle contracture, poor motor function and difficulty walking in affected children and adults. We investigated if a 12-week progressive resistance foot dorsiflexion strengthening program was feasible, safe and beneficial in a 15-year old girl with an axonal form of CMT.

METHODS

Training load was based on a dose-escalating percentage of one-repetition maximum (Table 1), completed on three non-consecutive days each week. Adjustable ankle weights contained in a neoprene sleeve were attached with velcro around the tarso-metatarsal region (Figure 1):

Outcomes included ankle dorsiflexion strength, quantified using hand-held dynamometry (Citec, The Netherlands). Secondary strength outcomes were ankle plantarflexion, foot inversion and eversion, also measured with hand-held dynamometry, according to a standardised procedure.[1] Other outcomes included gross motor function and walking ability evaluated barefoot using widely validated paediatric measures of balance, power and endurance.[1] Walking ability was evaluated using the GAITRite (CIR Systems Inc. Haverton, PA, USA) electronic instrumented walkway. Temporospatial walking gait parameters recorded included speed (cm/sec), cadence (steps/min), step time (sec), step length (cm) and stride length (cm).[1] Presence of foot drop as a sign of dynamic dorsiflexion weakness during gait was determined when the forefoot loaded the GAITRite earlier or equal to the rearfoot. This sensitive technique enabled quantification of foot drop during each step, which can be subtle, and is not always evident from clinical gait observation. Compliance, tolerability and any adverse events were recorded in a daily study diary.

RESULTS

At 12-weeks, dorsiflexion strength improved 56-72% and plantarflexion strength by 15-20%. Standing long jump increased by 16%, while balance and endurance did not. Walking ability improved for speed, cadence, step time and stride length. The 1RM increased considerably from baseline (1.83 kg) to 12-weeks (4.03 kg) (Table 1). Compliance was high and there were no adverse events.

Table 1. Details of the training load based on a dose-escalating percentage of one-repetition maximum

Weeks	1RM (kg)	Target Intensity (%1RM)	Actual Intensity (%1RM)*	Load (kgs)	Sets
1-2	1.83	60	52	0.95	2
3-4	2.71	70	68	1.83	2
5-6	4.03	80	78	3.15	2
7-12	4.03	80	78	3.15	3

*Actual training intensity was lower than the prescribed intensity due to fixed increments of the weighted satchels.

DISCUSSION

This case suggests progressive strength training might be a feasible intervention to help foot weakness and disability in childhood CMT, and provides a basis for larger trials of efficacy.

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The Effects of Foot and Ankle Strengthening with the AFX (Ankle Foot MaXimizer) on Athletic Performance in Male Varsity Basketball Players

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INTRODUCTION

The aim of this study was to evaluate the effectiveness of a new foot and ankle strengthening device on athletic performance in male varsity basketball players during their regular season. Performance measures included vertical jump (static, counter movement, one-step), agility (T-test), and dynamic balance (Star Excursion Balance Test).

METHODS

At the onset of the study, there were 12 experimental and 12 control subjects selected at random from two local universities, resulting in 6 experimental and 6 control subjects on each team, who were then matched according to their pre-trial vertical jump height relative to standing reach height for purposes of statistical analysis. In addition to the normal training program that was followed by control subjects for each team, experimental subjects were placed on a 12-week foot and ankle strength training program incorporating full range of motion exercise, high-speed concentric movements, and eccentric loading. Due to injuries sustained by some of the players (non-foot/ankle) and drop-outs, 7 experimental and 7 matched control subjects completed the study and were used in the final analysis. Experimental subjects were also given a questionnaire to obtain qualitative feedback on performance factors.

RESULTS

Statistical analysis of the data showed that: 1) vertical jump height (one-step technique) improved significantly for the experimental group (mean 7.6 cm, SD 4.0) relative to the control group (mean 2.0 cm, SD 4.0) ($p<0.01$), 2) vertical jump height (average of 3 different jump techniques) improved significantly for the experimental group (mean 4.3 cm, SD 2.6) relative to the control group (mean 1.8 cm, SD 4.1) ($p<0.05$), 3) dynamic balance improved significantly for the experimental group relative to the control group in the posterior ($p<0.01$) and lateral ($p<0.05$) directions for both left and right feet, and 4) there was no significant improvement in agility scores for the experimental group relative to the control group ($p=0.14$), although the mean score for the experimental group was 0.5 seconds faster than that of the control group, and all seven experimental subjects reported noticeable improvements in sprinting speed down the court and speed with which they could change direction.

DISCUSSION

The improvements in vertical jump height are believed to be associated with improved strength of the toe flexor and ankle stabilizer muscles, which may in part allow for a more rigid and balanced platform for improved transmission of force from proximal to distal segments in the kinetic chain, in addition to increased force production during the toe-off phase of jumping. This may have important implications for not only basketball, but other activities involving jumping, as well as activities involving explosive force production such as sprinting.

The improvements in dynamic balance are believed to be associated with improved eccentric muscle strength over the full range of joint motion for eversion and dorsiflexion. The improvements in lateral balance are an indication of improved lateral ankle stability, which may help to reduce risk of lateral ankle sprains.

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Kinematics of the foot and ankle: review of techniques and findings

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INTRODUCTION

The research history in foot and ankle kinematics is a paradigm of what biomechanical knowledge formation should be; it has developed continuously since the pioneering studies, progressively building up over the previous findings, taking advantage over time of the developments in technology and algorithm. Not a single of the original findings has been rejected by more modern evidences; rather the further understanding has thrown lights over the initial partial observations and models.

TECHNIQUES AND FINDINGS

In-vivo analysis is nowadays possible with easy to use motion capture instruments and modern protocols [1] for straightforward 3D reports and international data sharing of kinematics data-set with various resolutions, i.e. number of segments. Traditional stereophotogrammetric systems are now expanded with easier devices based on inertial sensors. Medical imaging and 3D matching procedures have speeded up the traditional time-consuming procedures for direct bone tracking, though limited by the special measure conditions. In-vitro measurements are also largely available, though in unloaded conditions; those of the daily living activities are simulated with robust mechanical testing machines, but the complexity of these motor tasks is hardly replicated. The least invasive of these techniques are utilised routinely nowadays for quantitative assessments of clinical conditions, before and after treatments.

Motion at the tibiotalar articulation [2] has been a paradigmatic process of knowledge building. After a very first assumption of a ball-and-socket joint, this was shown originally to be hinge-like, with a fixed axis of rotation passing through the tips of the malleoli. Then this axis was demonstrated, also by computer models [3], to move from position to position over the flexion arc even in passive flexion [4], but with the mean axis in fact at the location identified originally; finally, relevant angular and linear dispersion parameters for the axode, consented to consider a nearly spherical motion about a pivot point. These findings have been elucidated by the mechanical advantage of the plantar- and dorsi-flexor muscle units, and have been accounted for the good clinical results of a device designed explicitly to restore this complex kinematics mechanism [5].

CONCLUSIONS

The mechanisms which guide foot and ankle kinematics are far from being fully comprehended, also because of the enormous complexity of this anatomical system. On the other hand this knowledge is fundamental not only for this specific complex, but also to understand overall mechanisms at the lower limbs, both in physiological and in pathological conditions, as well as after surgical treatments or wearing different shoes. However, remarkable progresses are enhancing this knowledge: multi-segmental protocols are providing large resolution measurements of foot and ankle motion and deformation in-vivo under load despite the not invasive skin markers; medical imaging in special environments are offering access also to the underlying bone segments; measurements in-vitro using pin trackers make available large series of accurate segment motion and anatomical geometry. These are all enhanced by computer models, finally able to interpret the measurements and provide what hardly is directly accessible. This is revealing to be in fact a very multi-disciplinary subject, where many different competencies shall be shared; the i-FAB community seems to offer this easily.

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Medial Longitudinal Arch Deformation during Walking and Running

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INTRODUCTION

Many researchers and clinicians have attempted to quantify motion of the medial longitudinal arch (MLA) during dynamic activity. Dynamic mobility of the MLA may be more important in predicting one's injury risk than static arch type, which may help explain conflicting findings that both pes cavus and pes planus feet are at increased risk for metatarsal stress fractures [2, 1]. The purpose of this study was to compare MLA deformation during walking and running using 3-D motion analysis.

METHODS

Eleven participants (8 males, 3 females; age: 19.8±1.3 yrs; height: 1.73±0.06 m; mass: 69.4±10.0 kg; weekly mileage: 16.9-80.5 km) walked (1.37±0.21 m/s) and ran (3.28±0.35 m/s) at a comfortable speed barefoot across a force platform. Reflective markers were placed on the right first metatarsal head (2 cm from ground), navicular tuberosity, and medial calcaneus (2 cm from ground, 3 cm and 4 cm from the posterior aspect of the foot for females and males, respectively). An 8 camera Vicon motion capture system (200 Hz) and AMTI force platform (1000 Hz) were used to collect marker positions and ground reaction forces (GRF). The arch length was defined as the 3-D distance from the medial calcaneus to the first metatarsal head. The navicular height was defined as the 3-D perpendicular distance from the arch length line to the navicular. Changes in arch length and navicular height (displacement) were expressed in relation to a seated trial.

RESULTS

Maximum navicular displacement during walking was 5.1±1.9 mm and maximum change in arch length was 4.6±1.3 mm. Greater changes were seen during running, with navicular displacement of 7.3±1.9 mm and arch lengthening of 6.0±1.1 mm (Fig. 1). Both variables were statistically significant ($p<0.05$) between walking and running. Maximum change in arch length during walking occurred at 26±11% of stance phase and at 35±11% during running ($p<0.05$). Maximum navicular displacement occurred at 74±13% of stance during walking and 55±9% of stance during running ($p<0.05$).

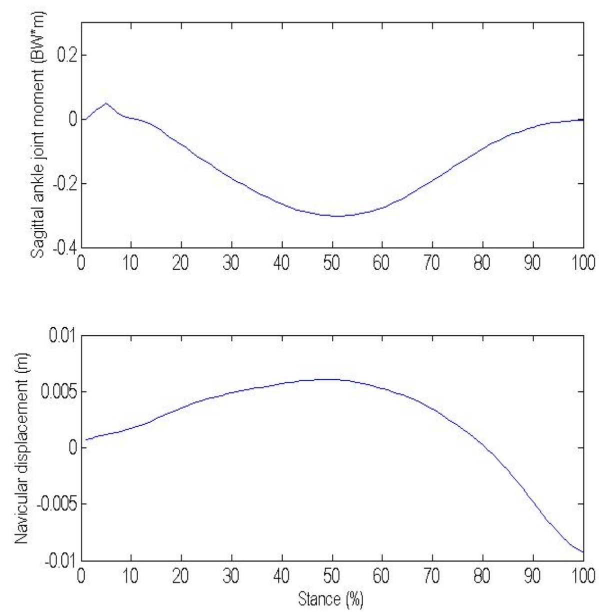

Figure 1. Example of sagittal ankle joint moment (top) and navicular displacement (bottom) during running.

DISCUSSION

Deformation of the arch is significantly greater during barefoot running compared to barefoot walking. This is likely due to increased ground reaction forces and Achilles tendon forces. Post hoc analysis found navicular displacement was highly correlated to the ankle joint moment during most of the stance phase. However during late stance of running the joint moment approaches zero while the arch becomes higher than the seated trial. This negative displacement at the end of stance is likely due to a windlass effect increasing tension in the plantar fascia.

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The Relationship Between Static Arch Rigidity and Foot Kinematics During Gait

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INTRODUCTION

Static measurements of foot posture are often used to evaluate foot function. Their simplicity allows them to be undertaken in clinical environments where time and space may restrict more functional measurements (e.g. gait analyses) from being used. Reliable and valid methods of assessing the static behaviour of the medial longitudinal arch have been developed [1]. One such measure, the arch rigidity index (ARI), provides a static assessment of the flexibility/ rigidity of the arch [2]. However, the relationship between static arch rigidity and dynamic foot kinematics during gait remains unclear. Therefore, the purpose of this study was to compare the foot kinematics in flexible and rigid arched subjects during walking. It was hypothesised that flexible-arched (FA) subjects would undergo greater foot and shank kinematic angular excursions compared to rigid-arched (RA) subjects.

METHODS

Nine FA and 9 RA subjects were recruited for the study. FA and RA subjects were defined as having an ARI that was less than 0.906 or greater than 0.938 respectively. These values fell outside of 0.5 standard deviations of the mean ARI based on a sample of 63 subjects. An Arch Height Index Measurement System was used to calculate arch height index (AHI) during both sitting and standing. The ARI was calculated as the ratio of AHI standing:AHI sitting [2]. Retroreflective markers were attached to the forefoot, rearfoot and shank of the right limb [3]. After a standing calibration trial, subjects walked at 1.2 ms^{-1} on a treadmill while kinematic data was captured at 120Hz. Three-dimensional kinematics were calculated for the forefoot (relative the rearfoot), rearfoot (relative to the shank) and shank (shank relative to the rearfoot). Mann Whitney tests were used to compare forefoot and rearfoot angular excursions (from initial contact to peak value) between groups.

RESULTS

The kinematic comparisons between FA and RA subjects are shown in Table 1. Although just short of statistical significance, there was a trend for reduced rearfoot abduction and increased forefoot dorsiflexion excursion in FA subjects. There were no differences in terms of rearfoot eversion or forefoot abduction.

Table 1. Forefoot / rearfoot angular excursions.

Values / Variables	Mean (sd) angle (°)		*P* values
	FA	RA	
Rearfoot eversion	6.5 (1.8)	6.2 (1.4)	0.500
Shank internal rotation	3.1 (1.6)	4.4 (1.3)	0.068
Forefoot dorsiflexion	9.7 (3.5)	7.2 (1.3)	0.057
Forefoot abduction	6.8 (1.8)	6.8 (1.8)	0.466

DISCUSSION

Subjects with flexible arches displayed increased forefoot dorsiflexion during walking compared to their rigid-arch counterparts. This suggests that the static measure, ARI, may be related to the dynamic behaviour of the foot during gait. Given that excessive forefoot dorsiflexion has been associated with medial longitudinal arch lowering, this is not surprising. In contrast, rearfoot eversion during gait was not greater in subjects with flexible arches, suggesting that the midfoot and rearfoot may function independently from one another. Finally, the greater tibial rotation in high-arch subjects may be a consequence of the rigid foot transferring more motion to the shank.

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Foot Bone Motion in Cavus, Neutral, and Planus Feet Using an In Vivo Kinematic Foot Model

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INTRODUCTION

Understanding *in vivo* bony motion of the foot allows for more accurate identification of atypical foot types and possible prevention of injury. Several studies [1-3] have used various multi-segment foot models (MSFMs) to explore the kinematics of the pes planus deformity, but no study has fully described the range of foot bone kinematics that can exist between the pes planus and pes cavus deformities. The purpose of this study is to provide a thorough analysis of *in vivo* foot bone kinematics during static loading and gait for pes cavus, neutrally aligned and pes planus subjects with a MSFM.

METHODS

Ten subjects were tested under static body weight and dynamic gait from each of the following four foot type groups: pes cavus (PC), neutrally aligned (NA), asymptomatic pes planus (APP), and symptomatic pes planus (SPP). The kinematic foot model consisted of the following six segments: shank (tibia and fibula), hindfoot (calcaneus and talus), midfoot (cuboid, cuneiforms, navicular), lateral forefoot (5th ray), medial forefoot (1st ray), hallux (proximal phalanx). Sixteen retro-reflective markers were placed on the skin on bony landmarks and captured with a 12-camera Vicon system at 120 Hz. Six medial markers were used only in a static trial to align anatomical axes and were removed for dynamic trials. In general, the x-axis pointed laterally (right feet), the y-axis anteriorly, and the z-axis superiorly. Coordinate systems followed the right hand rule. All feet were scanned to determine malleolar valgus index (MVI) and X-rayed to determine talometarsal (TM) angle, calcaneal pitch (CP) angle, and talonavicular coverage (TNC) angle. Ten metrics each for both dynamic and static trials were hypothesized to relate to foot type, and linear mixed effects models were used to determine statistical significance.

RESULTS AND DISCUSSION

Three dynamic metrics from the hindfoot (calcaneus-to-tibia [CALC_TIB]) were statistically different ($p<0.05$) for APP and/or SPP vs. PC (Table 1). While some trends existed in the forefoot, there was little statistical significance due to large variability in the data. Six static metrics were statistically different across the foot types (Table 1), including TM and CP, which predicted foot type for all pairs. These data provide a more thorough understanding of the kinematic differences between abnormal foot types.

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ACKNOWLEDGEMENTS

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Table 1: Statistically significant dynamic and static measures by foot type. All angles in right foot orientation.

		PC (n=10)	NA (n=10)	APP (n=10)	SPP (n=10)	p*
Dynamic data	Peak CALC_TIB angle, sagittal (°)	-15.4 [9.7]	-18.5 [6.6]	-25.6 [6.4]	-24.0 [7.4]	.011[1]
	Peak CALC_TIB angle, frontal (°)	-1.6 [9.6]	-5.8 [3.5]	-3.7 [4.3]	-10.2 [8.6]	.046[2]
	CALC_TIB ROM, frontal (°)	4.2 [4.3]	5.9 [3.7]	8.1 [2.6]	8.3 [3.9]	.0090[1,2]
Static data	Malleolar valgus index	0.2 [16.7]	10.3 [4.0]	14.7 [7.2]	20.0 [7.3]	.0008[1,2]
	Talometatarsal angle (°)	-17.5 [10.1]	0.2 [5.5]	10.1 [6.3]	18.9 [8.9]	<.0001[3]
	Calcaneal pitch angle (°)	31.7 [5.8]	24.5 [3.7]	14.3 [5.2]	14.5 [5.7]	<.0001[3]
	Talonavicular coverage angle (°)	2.6 [14.8]	13.1 [9.5]	15.0 [16.6]	27.3 [10.7]	.0029[2]
	Navicular tuberosity height (mm)	62.9 [12.1]	52.7 [6.4]	46.8 [12.6]	36.3 [12.9]	.0001[1,2,4]
	CALC_TIB angle, sagittal (°)	-5.2 [9.2]	-5.6 [6.8]	-16.1 [6.5]	-13.3 [7.0]	.0039[1,5]

[1]APP vs PC significantly different, [2]SPP vs PC significantly different, [3]All pair-wise comparisons significantly different except SPP vs. APP, [4]SPP vs NA significantly different, [5]APP vs NA significantly different; dynamic: $p<.0083$ using Bonferroni's correction for multiple comparisons; static: Tukey's HSD test, $p<.05$

Should Linear Regression Be Used to Assess the Relationship between Multi -segment Foot Kinematics and Plantar Pressures in the Pediatric Patients with Foot Pathology?

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INTRODUCTION

Plantar pressure (PP) measures can provide valuable information regarding the magnitude and location of forces under the foot, while multi-segment foot (MSF) models have been shown to quantify pediatric foot deformities.[1-3] Several authors have tried to use linear regression to correlate PP and MSF kinematics, with only moderate correlation coefficients found.[3] This study aims to illustrate the complex nature of the biomechanics of the foot, and the need for more sophisticated techniques to relate MSF kinematics and PP measures.

METHODS

Nineteen pediatric patients were classified into groups based on clinical notes from their recent orthopaedic office visits: Varus/Cavovarus (CV group, N=11) and Valgus/Planovalgus (PV group, N=8). Subjects were instrumented with a modified Helen Hayes marker set and a multi-segment foot kinematic marker set.[4] Plantar pressure data was collected using an Emed platform (4 sensors/cm^2, Novel, Munich, Germany). Descriptive statistics were determined for ankle motion and foot rotation (LE), and MSF variables. Foot prints were divided into 8 masks (medial/lateral hindfoot, midfoot and forefoot, 1st, 2nd and 3rd-5th MTs and Hallux) using a modified prc automask. Data from the subject groups were compared to 15 age-matched healthy children.

RESULTS

Although significant differences were seen across many MSF and PP measures, kinematic variations led to differing pressure patterns within groups. Pearson correlation coefficients between PP measures and MSF kinematics ranged from -0.76 to 0.56, with the highest correlations seen with transverse plane forefoot alignment. Figure 1 illustrates the coronal plane hindfoot and forefoot variation for each group as well as two individual examples of the maximum pressure maps. The planovalgus group shows two subjects with similar average hindfoot coronal alignment, however

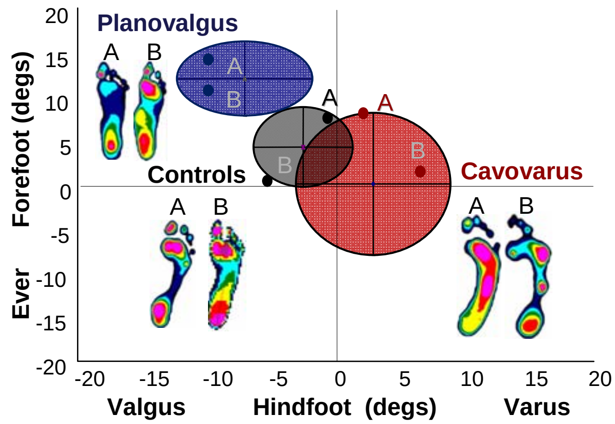

Figure 1: Coronal plane MSF variation seen within each group: Planovalgus (blue), Controls (black) and Cavovarus (red). Two representative maximum pressure maps are shown to illustrate the variation in plantar pressures.

subject A demonstrated increased compensation in the forefoot. This led to decreased medial pressure in the medial midfoot and forefoot. Similar differences are seen in the examples for the Cavovarus and Control groups.

DISCUSSION

MSF motion has been shown to have moderate linear correlation to plantar pressures. Segmental range of motion and alignment, motion in other planes, and open-vs.-closed chain kinematics can all significantly affect the plantar pressure patterns. The variation seen within each group was most likely affected by the variation in subject pathology. For example, those with increased stiffness due to neuromuscular disease or bony coalitions were unable to compensate for hindfoot misalignment, which led to abnormal pressure patterns. More complex models are needed to assess the relationship between MSF and plantar pressures in pediatric patients with foot pathology, perhaps incorporating kinematics, kinetics, plantar pressures and clinical measures.

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Peri-talar kinematics and kinetics in Psoriatic Arthritis patients with Achilles tendon enthesitis

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INTRODUCTION

The entheses are a primary target organ in psoriatic arthritis (PsA) [1]. Differential presentation at the Achilles tendon (AT) in PsA is associated with function [2]. The main aim of this study was to compare rotational movements across the ankle and subtalar joints and ankle joint moments and powers in PsA patients with and without AT enthesitis.

METHODS

Twenty nine PsA patients fulfilling the CASPAR diagnostic criteria (mean (SD) age- 47.6 (12.3) yrs, BMI- 24.7 (3.4), and disease duration- 11.3 (9.7) yrs), and 10 healthy adults (mean (SD) age- 44.7 (10.6) yrs, and BMI- 24.2 (2.5)) were studied. Standard clinical metrics including pain, impairment and disability were measured. AT enthesitis was detected using high-resolution ultrasound (Esaote MyLab 25 Gold, Esaote Genoa, Italy) and the GUESS scoring system [3]. Video-based 3D gait analysis was undertaken using an 8-camera MOCAP system (Oqus, Qualysis, Gothenburg, Sweden) and force plate (Kistler, Winterthur, Switzerland). Intersegment kinematics and kinetics were analysed using a multi-segment foot model in Visual3D software (C-Motion Inc., MD, USA). A core set of spatio-temporal, kinematic and kinetic variables was selected *a priori* and between group differences analysed using one-way ANOVA with post-hoc Tukey's test.

RESULTS

The GUESS score positively identified 20 PsA patients and 6 controls subjects with AT enthesitis. A summary of the inter-segment kinematics and kinetics are presented in Table 1. A statistically significant between-group difference was detected for ankle power only. Tukey's post-hoc test indicated a significant difference between PsA enthesitis positive and control subjects only (p=0.008). There was a trend towards slower walking, poor global health and higher global and foot pain in the PsA enthesitis positive group.

DISCUSSION

Complex multi-planar movements have been associated with the differential presentation of enthesitis in PsA, with the AT regarded as a prime site [1,2]. The findings from this cohort of patients do not support this hypothesis. Moreover, enthesitis can be detected in otherwise healthy adults with equivalent function. These findings require confirmation.

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ACKNOWLEDGEMENTS

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Table 1: Mean (SD) for discrete 3d kinematic and kinetic variables for PsA patients with and without AT enthesitis and healthy adults subjects. p-value from ANOVA and post-hoc test results are provided.

Variable	Control (n=10)	PsA Enth (-) (n=9)	PsA Enth(+) (n=20)	p-value	Post-hoc
Peak RF dorsiflexion (°)	7.5 (3.2)	5.2 (3.9)	7.2 (3.3)	0.26	-
Peak RF Eversion (°)	9.4 (6.1)	7.4 (4.1)	5.4 (7.1)	0.26	-
Peak RF external rotation (°)	4.6 (5.1)	5.2 (4.5)	4.7 (6.3)	0.96	-
Peak MF dorsiflexion (°)	1.0 (5.6)	0.7 (9.3)	5.7 (7.7)	0.16	-
Peak MF inversion (°)	9.7 (7.3)	10.1 (5.6)	7.8 (6.7)	0.61	-
Peak MF abduction (°)	0.9 (7.1)	-2.2 (8.2)	-3.3 (8.4)	0.39	-
Navicular height in stance (mm)	44.8 (6.1)	42.8 (6.8)	40.2 (6.1)	0.17	-
Peak ankle joint moment (Nm/kg)	-1.60 (0.18)	-1.51 (0.18)	-1.40 (0.21)	0.22	-
Peak ankle joint power (W/kg)	3.51 (0.51)	3.05 (1.05)	2.52 (0.74)	0.008	PsA(+)/Con, P=0.007
Walking speed (m/s)	1.34 (0.26)	1.29 (0.18)	1.16 (0.24)	0.11	-

Multi-Segment Foot Kinematics During Barefoot Treadmill Running

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INTRODUCTION

Shod runners typically strike the ground with the rear-foot, whereas barefoot runners tend to land first on the fore-foot or mid-foot [1]. Since rear-foot strikers experience higher impact forces, running with a fore- or mid-foot strike (or, indeed, running barefoot) might offer some protection from impact-related injuries [1]. In the absence of a running track or other safe surface, barefoot running could most safely be practised on a treadmill, but this might alter foot kinematics compared to over-ground running. The aim of our study was to investigate foot motion during barefoot treadmill running and to compare the results to those previously published for over-ground barefoot running [2].

METHODS

Ten uninjured female long-distance runners (age 26.8 ± 8.5 yrs, distance run 57.6 ± 18.4 km/week) participated in the study. Spherical reflective markers were located at known anatomical landmarks of both lower limbs [3]. Subjects ran barefoot on a treadmill (Activ8, Ultim8 Fitness Ltd, UK) at 3.56 m/s. Three-dimensional coordinate data were collected using a 12-camera Vicon MX Motion Analysis System at 200 Hz. Digital video was collected simultaneously at 100 Hz (Basler, A602FC-2, UK). The coordinate data were low-pass filtered using cubic spline smoothing. Heel-strike and toe-off times were detected using kinematic methods. Rear-foot and fore-foot angles were calculated in Vicon Nexus using the Oxford Foot Model [3], and five strides for each foot were used in the analysis. Subjects were classified as either rear-foot (RFS) or fore-foot (FFS) strikers based on joint angles, confirmed by visual inspection of video data. Joint angles and discrete kinematic variables were averaged for each group.

RESULTS

Six subjects were identified as FFS and four as RFS. Although sample size is too small for meaningful statistical analyses to be performed, the joint angle data suggests differences between the two groups (Figure 1).

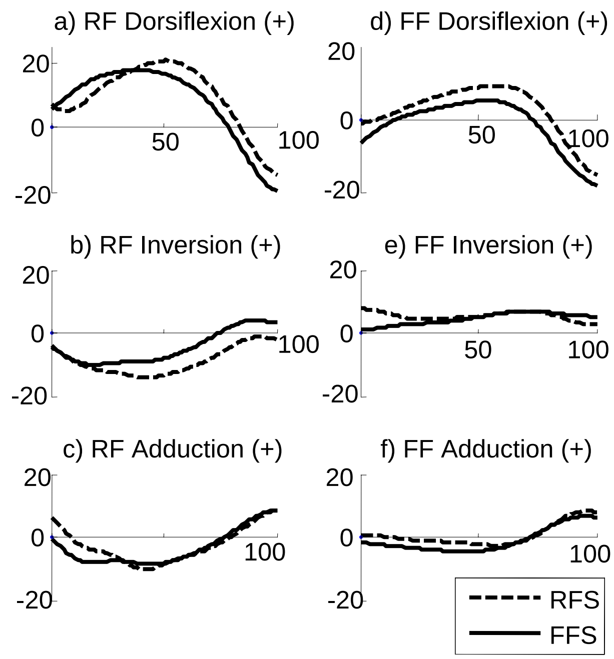

Figure 1: Mean rearfoot (RF) and forefoot (FF) joint angles for RFS and FRS versus % stance phase.

DISCUSSION

Foot kinematics for RFS and FFS during barefoot treadmill running were mostly consistent with those reported for over-ground barefoot running [2]; however, for FFS, rearfoot eversion excursion (6.5°) and time to peak rearfoot eversion (24.7% stance) were different from published results (17.9°, 42.6%). We plan to investigate these differences in more detail by comparing the kinematics of the same subjects during treadmill and over-ground barefoot running and by collecting data on a larger group of subjects.

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ACKNOWLEDGEMENTS

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Evaluation of the Contact Surface at the Ankle during Walking and Slow Running

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INTRODUCTION

Knowledge of the ankle joint contact area during physiological loading and movements is essential to understand biomechanics and pathogenesis of this joint as well as to improve prosthetic design and ligament reconstruction surgery [1]. The aim of this study was to determine the ankle joint contact area for the first time in vivo over the whole stance phase during walking and slow running.

METHODS

We applied intracortical pins to monitor shank and foot bone kinematics [2,3]. Based on computer tomography, 3D reconstructed surfaces of the tibia, talus, and related markers attached to the pins were made for two subjects. The ankle joint surfaces on tibia and talus were extracted and moved through the stance phase. Contact between the surfaces was defined on distance thresholds (1...3.5 mm) comparing each point on one surface with all points on the other, similar to [4]. Points in contact were triangulated to determine the contact area. The path of the geometric center of the contact area was plotted on a rolled out cylinder surface fitted into the talar dome.

RESULTS

Different thresholds defining contact changed only the magnitude of the contact area but not its development during stance phase. During dorsiflexion at the ankle, minimal contact areas were observed (Figure 1). The variability of the contact area is higher during walking than during slow running (Figure 2). The path of the geometric center is also more repeatable during running than during walking.

DISCUSSION

In contrast to previous studies applying static joint excursion under load [1,5], maximal contact areas were found during plantarflexion rather than during dorsiflexion. Since frontal plane motion at the ankle during walking is

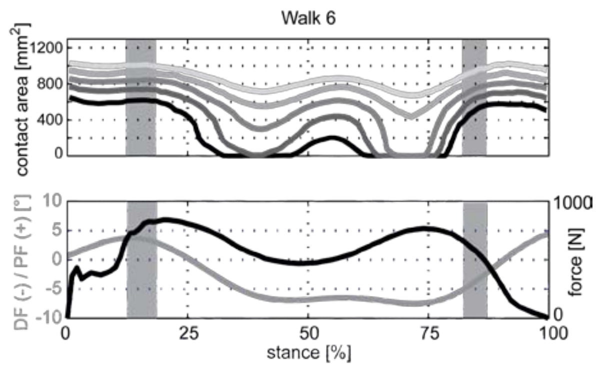

Figure 1: Upper figure: Development of contact area over stance phase. The darker the line, the lower the threshold (3,2.5,2,1.5,1mm). Lower figure: Sagittal plane motion at the ankle (grey) and vertical ground reaction force (black). Grey bands represent pushoff and heelstrike of contralateral leg.

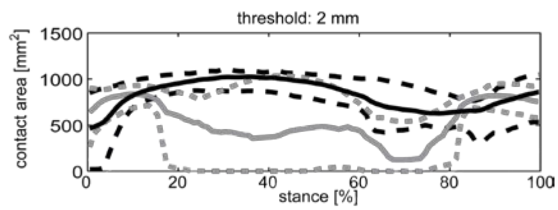

Figure 2: Developing of contact area over stance phase during walking (grey) and running (black). Straight lines represent mean, dashed lines minimal and maximal values of one subject.

greater than during slow running [2,3], it can be concluded that running results in a higher stabilisation of the ankle which is also indicated by the lower variability of contact area/path of its center during running. Data sets of five additional subjects were acquired last December, the analysis is in progress and will be presented at the conference.

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Invasive in vivo description of the effect of foot orthoses on foot kinematics.

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INTRODUCTION

Medially wedged and laterally wedge foot orthoses are frequently used in physical therapy. Apart from simple foot-leg or heel-leg interactions the biomechanical consequences of these are largely speculation. The specific effects of these orthoses on either subtalar or talocrural functions are difficult to describe *in vivo*. This paper reports on the effects of these two contrasting orthotic designs on foot kinematics using an invasive in vivo approach.

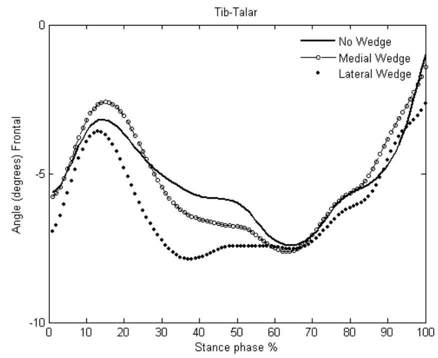
METHODS

Intracortical pins (with markers attached) were inserted into the tibia, calcaneus, talus, navicular, cuboid, medial cuneiform and first and fifth metatarsals. Four subjects walked in three conditions: shoes only, shoes with medially wedged orthosis; shoes with laterally wedged orthosis (fig 1). Cadence between trials was controlling using a metronome.

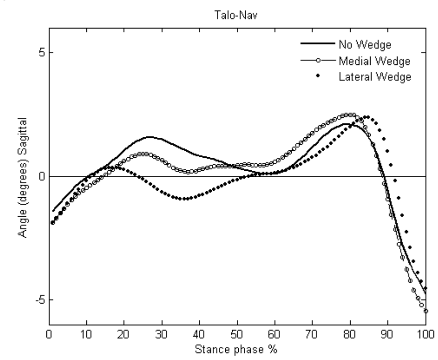
Figure 1. Modified shoe and bone pins in situ.

RESULTS

The *mean* difference between shod an orthotic conditions during the whole of stance was less than 2° in almost all cases. Maximum differences between shod and orthotic conditions were largely less than 4°. Greatest effect of the orthosis was seen at the talonavicular joint. Effects at other joints were comparable to each other. Data for the ankle (frontal) and talo-navicular (sagital) are in fig. 2 (one subject).



(+ve = inversion of talus rel. to tibia)



(+ve = dorsiflexion of navic. relative to talus)

DISCUSSION

Orthotic effect was less dramatic and systematic than clinical paradigms assume. Orthoses reduced/increased range of motion at all joints but the gross pattern of joint movement was not altered. The vast majority of effects on kinematics were small. Responses to the orthoses was highly person specific. Responses were complex and unsystematic in terms of the effect in each plane of motion, and how the effect at one joint related to that at adjacent joints. The effect of one orthoses was not accompanied by predictable and opposing effect of the alternative orthosis design. The effect of orthoses on leg vs. heel motion was largely as previously reported. However, the individual ankle and sub talar contributions was variable.

An in-shoe comparison of foot kinematics in normals versus mechanical foot pain.

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Introduction Multi-segment kinematic foot models quantify walking foot motion in normal and pathological gait but have been limited mainly to barefoot assessments as shoes confound the marker sets. Previous attempts to measure in shod states have included the use of sandals or modified boots to accommodate the foot marker sets. This aim of this study was to provide a gait shoe solution that accommodates the multi-segment foot marker set without compromising function.

Methods Fifteen pain free volunteers and 15 patients with mechanical mid-foot pain were recruited from local rheumatology and musculoskeletal podiatry departments. Motions were captured at 150Hz using an eight camera, 3D infra-red motion analysis system (Vicon Motion Systems Ltd., UK). The Vicon MX system with T40 cameras was integrated with a force plate (1000Hz, Bertec Corporation, USA) to detect foot contact events. Foot model kinematics were processed using Vicon Polygon v 3.1. All 30 participants undertook one session of gait analysis in two conditions: barefoot and shod, in a random order. The gait shoe comprised a laced fastening and canvas upper, which was customised with netting panels in order to allow clear visualisation of the foot markers. Markers were placed by a single clinician (JH) according to the Oxford foot model recommendations [1], after which all volunteers completed 8 gait cycles captured at a self selected walking speed. Consistency graphs were plotted and the most representative single gait cycle was chosen for each participant and condition. Data were processed using the Oxford foot model and filtered with a Woltring fifth-order spline-interpolating function. The gait cycle was normalized to 51 centiles.

Results The group with foot pain demonstrated slower walking than controls in both barefoot (-0.28m/s) and shod conditions (-0.25m/s). Differences between pain vs no-pain group kinematics at the hindfoot were consistent in barefoot and shod conditions. At 50% of stance in both the barefoot and shod conditions, the foot pain group showed a similar increase in hindfoot eversion compared with controls (diff -4.23° bare vs -4.85° shod) and decrease in hindfoot dorsiflexion (diff 4.66° bare vs 3.45° shod). At the forefoot, the gait shoe had some effect in reducing the magnitude of the differences between the conditions although the motion time curves were comparable. At 50% of stance, the foot pain group showed greater forefoot dorsiflexion than controls (diff -2.3° bare vs -0.61° shod) and less adduction (diff 2.6° bare vs 0.37° shod).

Discussion The results of this study suggest that our novel gait shoe is a real-world solution to measuring foot kinematics in-shoe. The results show the gait shoe has a minimal functional effect on the hind-foot as differences between groups are retained. The shoe may have some effect on the forefoot, but it did not change the kinematic pattern significantly compared to the barefoot condition.

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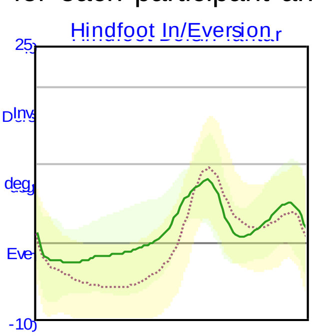

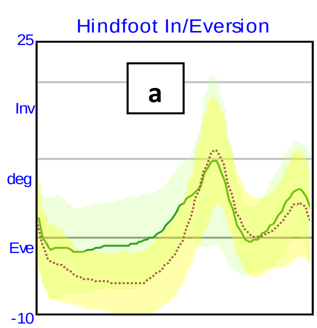

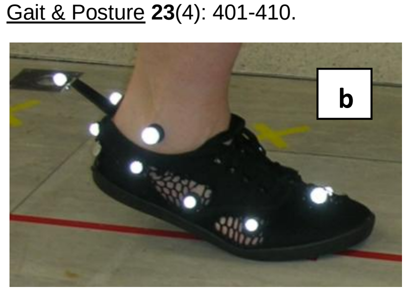

Figure shows mean hindfoot axial roations (a) barefoot and (b) shod, and (c)image of gait shoe. The graph shows the normalized mean (pink line - foot pain, green line - normal) and standard deviations (yellow shading - foot pain, green shading - normals).

antar Forefoot Ab/Adduction 20

Reliability of the Oxford Foot Model During Gait in Healthy Adults

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INTRODUCTION

The Oxford foot model (OFM) is a multi-segment model for calculating rearfoot, forefoot and hallux motion [1]. Kinematic data from the OFM may aide in the description of foot and ankle pathology during gait. Stebbins et al. [2] modified the rearfoot alignment and knee joint center, and subsequent research has adopted these changes [3, 4]. Curtis et al. [3] evaluated the reliability of this modified OFM in children at specific gait events. However, the reliability during gait in adults has not previously been investigated. Additionally, the reliability of the common practice of normalizing angles to static calibration (assuming all angles are zero in neutral stance) [4], has not been established with the OFM. Therefore, the purpose of this study is to assess the intra-tester reliability of rearfoot and forefoot kinematics obtained using the modified OFM in adults at initial contact (IC) during gait with and without normalization to static calibration.

METHODS

Seventeen healthy adults (10 males, 7 females, 25.1 ± 4.8 yrs, 1.75 ± 0.10m, 74.0 ± 12.4kg) were tested during a single visit to our Sports Medicine Research Laboratory, during which 1 examiner recorded 2 sessions. For each session the OFM marker set was applied, a static calibration trial captured, then 10 walking trials recorded at 100Hz using a 12-camera motion analysis system (Vicon, Oxford, UK). A trial consisted of walking across the capture space with IC of each foot occurring on a force plate (Bertec Corp.,Columbus, Ohio). Markers were removed between sessions. Data was low-pass filtered at 12Hz. Rearfoot and forefoot angles were calculated both with and without normalization. Intraclass correlation coefficient ($ICC_{2,k}$) and standard error of the measurement with 90% confidence bounds [$SEM_{90} = 1.64*SD*(1-ICC)^{1/2}$] were calculated on the average of 10 trials to assess intra-examiner reliability and error in the sagittal and frontal planes.

RESULTS

See Table 1. Normalization to static calibration resulted in very good reliability for rearfoot and forefoot angle in both planes, with very small error. Without normalization, sagittal plane reliability was also good and error small; however, frontal plane reliability was poor, with relatively large error (5.1-5.7°).

DISCUSSION

Our results show that the modified OFM is reliable in adults during gait. Reliability in adults is higher than in previously reported in children [3]. Additionally, when it is reasonable to normalize to static calibration, error between marker set applications can be minimized, and is up to 2° lower than previously reported [1, 3], enabling detection of smaller angular changes.

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Table 1: Intra-Tester Reliability at Initial Contact using the Oxford Foot Model (Average of 10 trials)

Plane	Joint	Normalized to static	Mean angle (º)	SD (º)	ICC (95% CI)		SEM_{90} (º)
Sagittal	Rearfoot	No	1.45	2.78	.91	(0.76-0.97)	1.37
		Yes	-1.67	3.37	.91	(0.75-0.97)	1.66
	Forefoot	No	-6.36	4.96	.92	(0.77-0.97)	2.30
		Yes	-0.39	2.82	.91	(0.75-0.97)	1.39
Frontal	Rearfoot	No	2.10	5.06	.53	(-0.37-0.83)	5.69
		Yes	0.88	2.10	.90	(0.70-0.96)	1.09
	Forefoot	No	4.49	6.43	.77	(0.34-0.92)	5.06
		Yes	-0.02	3.71	.90	(0.71-0.96)	1.92

SD = Standard Deviation, ICC = Intraclass Correlation Coefficient, SEM_{90} = Standard Error of the Measure

In vivo kinematics of two-component total ankle arthroplasty during gait

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INTRODUCTION

Good clinical outcomes have reported after total ankle arthroplasty using various types of implant design. However only limited data are available on in vivo kinematics. We have used two-component total ankle arthroplasty (TNK ankle™)[1], and the purpose of this study was to measure in vivo kinematics of the TNK ankle during gait.

METHODS

Twenty ankles of 17 patients (14 women and 3 men) with a mean age of 74 years were enrolled. Preoperative diagnosis was osteoarthritis in 15 patients and rheumatoid arthritis in 2 patients. The mean follow-up was 45 months, and the mean AOFAS score at the examination was 86 points.

Patients walked at self-selected speed, and lateral fluoroscopic images during the stance phase of gait were recorded at 10 frames/sec. Three dimensional kinematics were determined using 3D-2D model-image registration techniques [2] (Figure 1). Three-dimensional CAD models of the tibial and talar implants were obtained from the manufacture (Japan Medical Materials, Osaka, Japan), and anatomic coordinate systems were embedded in the implant models. The models were projected onto the fluoroscopic image, and three dimensional positions and orientations of the implants were determined by matching the silhouette of the implants with the silhouette of the fluoroscopic image. Kinematics were analyzed at five points of the stance phase; heel strike, early, mid and late stage of the stance phase, and toe off.

RESULTS

Dorsi-\plantarflexion was the most dominant, and consistent plantarflexion pattern was found from heel strike to toe off. The mean(SD) ranges of motion was 11.2°(5.0°). Inversion/eversion was minimal, and the mean range was 0.8°(0.4°). No ankle showed more than 2° of rotation. The mean range of internal/ external rotation was 2.5°(1.5°), and there was no consistent trend among patients.

DISCUSSION

The TNK ankle is a two-component prosthesis in which a polyethylene insert is fixed on the tibial component. It is a semi-constrained design; the joint surfaces are cylindrical, and the diameter of the talar surface is slightly smaller than that of the tibial surface, allowing the talar component some internal/external rotation. The range of inversion/eversion, and internal/external rotation was smaller than those of less constrained two-component prosthesis [3] and mobile bearing prosthesis [4], while the range of dorsi-\plantarflexion was similar. The kinematic difference may be due to the difference in the implant design.

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Figure 1: 3D-2D registration of the implant models. From the left; heel strike, early stance, mid stance, late stance, and toe off.

Soft tissue thickness under the metatarsals: Is it reduced in those with toe deformities?

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INTRODUCTION

Adequate soft tissue under the metatarsal heads of the foot is vital to protect and cushion the foot during ambulation. Despite many anecdotal reports suggesting atrophy of the soft tissue underneath the metatarsal heads in those with toe deformities, there are few quantitative *in vivo* studies to support this notion. Particular focus has been given to those patients with diabetes [1, 2] and not otherwise healthy older adults. In addition, no previous studies have reported soft tissue thickness in older people with hallux valgus. Therefore, the purpose of this study was to determine whether the thickness of the soft tissue under the metatarsal heads differed between individuals with and without toe deformities.

METHODS

A portable SonoSite® 180PLUS ultrasound system was used to measure the thickness of the (i) soft tissue and (ii) plantar fat pad (Figure 1) at the 1st metatarsal head (1MTH) and 5th metatarsal head (5MTH) on 312 randomly selected older men and women aged over 60 years. Each participant had their feet assessed by the Chief Investigator (KJM) for the presence of lesser toe deformities whilst hallux valgus severity was rated using the Manchester Scale [3]. The thickness of soft tissue and the fat pad underneath 1MTH and 5MTH in those with moderate-to-severe hallux valgus (n = 36) or lesser toe deformities (n = 68) were compared to gender-, age- and BMI-matched controls using independent *t*- tests.

RESULTS

Individuals with hallux valgus had significantly reduced soft tissue underneath 1MTH compared to controls ($p = 0.002$; Table 1). Similarly, individuals with lesser toe deformities displayed significantly reduced soft tissue underneath 5MTH compared to their controls ($p = 0.01$; Table 1). There were no differences in fat pad thickness between individuals with hallux valgus or lesser toe deformities and their controls at either 1MTH or 5MTH.

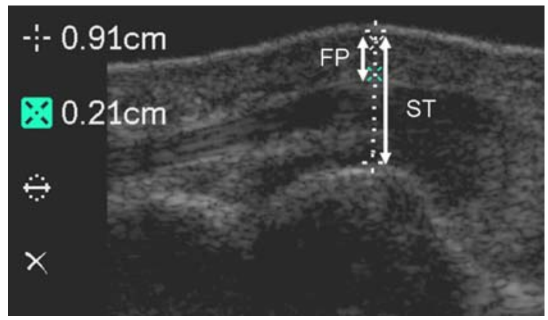

Figure 1: Ultrasound image showing soft tissue (ST) and fat pad (FP) thickness measurements.

DISCUSSION

Individuals with hallux valgus and lesser toe deformities display reduced soft tissue under 1MTH and 5MTH, respectively. As the thickness of the fat pad was not affected, this suggests that the thinner soft tissue may represent atrophy of the plantar muscles. This may partly explain previous findings that individuals with hallux valgus and lesser toe deformities have decreased flexor strength of the hallux and lesser toes, respectively [4].

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Table 1: Tissue thickness (mm) at the 1st and 5th MTH in individuals with toe deformities and their respective control participants. * indicates a significant difference between deformity and control groups.

Location	Hallux valgus (n=36)	Control (n= 36)	Lesser toe (n=68)	Control (n=68)
1MTH soft tissue	7.4 ± 1.6*	8.5 ± 1.5	8.2 ± 1.8	8.5 ± 1.7
1MTH fat pad	1.6 ± 0.6	1.8 ± 0.6	1.7 ± 0.6	1.9 ± 0.7
5MTH soft tissue	5.1 ± 1.3	5.3 ± 1.2	5.1 ± 1.0*	5.6 ± 1.3
5MTH fat pad	1.4 ± 0.5	1.4 ± 0.6	1.4 ± 0.4	1.5 ± 0.6

Distribution of Intrinsic Foot Muscles in Healthy and Plantar Fasciitis Feet

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INTRODUCTION

It has been shown that plantar intrinsic foot muscles (PIFM) and the plantar fascia play a significant role in providing support to the medial longitudinal arch[1]. The distribution of PIFM size across the foot is not known. Muscle atrophy may occur in individuals with chronic plantar fasciitis (PF), thereby compromising the supportive role offered by these muscles and thus perpetuating a state of injury. The purpose of this study was to determine the distribution of PIFM, and whether chronic plantar fasciitis is accompanied by atrophy of PIFM.

METHODS

Eight subjects with unilateral PF consented to this study. Axial bilateral foot MRIs were taken with a 1.5 Tesla MR system via a four-channel head coil positioned in the magnet's isocentre. T1 weighted images of the entire foot length were acquired perpendicular to the plantar aspect of the foot using a spin-echo sequence (relaxation time=500ms, echo time=16 ms, slice thickness=4mm, gap between slices=0mm).

PIFM cross-sectional areas (CSA) were digitized [2] from the calcaneus through to the image containing the maximum diameter of the sesamoid bones. CSA were summed across the rearfoot and forefoot segments. Forefoot and rearfoot segments were defined by splitting the total number of images containing muscle into halves, anterior and posterior.

RESULTS

The distribution profile for muscle CSA from heel-to-toe was bimodal with PIFM being larger in the forefoot than the rearfoot (Figure 1).

In comparison to healthy feet, PF feet were associated with a 5.2% reduction in PIFM CSA at the forefoot (p=0.03), but not at the rearfoot (p=0.26) (Table 1).

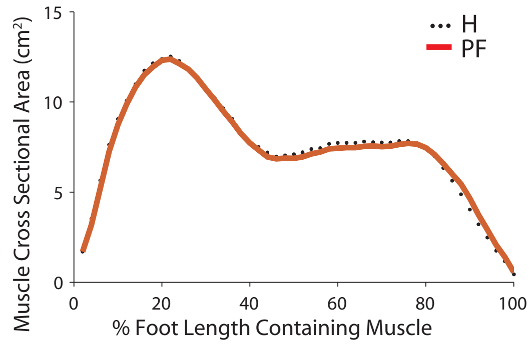

Figure 1: Mean muscle cross sectional areas across the foot length for healthy (H) and plantar fasciitis (PF) feet, from sesamoids (0% foot length) to calcaneal tuberosity (100%).

DISCUSSION

The bias toward greater muscle size in the forefoot is likely an indication of the higher degree of dexterity at the metatarsals and phalanges in comparison to the rearfoot.

Many of the muscles of the forefoot insert onto the first ray, and when atrophied may destabilize the medial longitudinal arch, and therefore delay recovery by placing a greater stress of the plantar fascia.

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Table 1: Group means for total muscle cross sectional areas (CSA) across segments (difference with respect to healthy: %H). A *p* value for a one-tailed dependent *t*-test and effect size (ES) are provided.

	Healthy (cm^2)	PF (cm^2)	Difference (%H)	*p* value	ES
Forefoot	168.8 ± 47.3	158.4 ± 37.1	-5.2 ± 6.2	0.03	0.26
Rearfoot	114.4 ± 43.6	111.5 ± 33.3	0.7 ± 8.3	0.26	0.08

Skeletal Demands on the Ankles of Female Ballet Dancers Evident from Orthopaedic Imaging

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INTRODUCTION

Few activities require as extreme a range of ankle motion as classical ballet [1]. Though the *en pointe* position (where a ballet dancer stands on her toe tips) is integral to this dance genre, systematic study of the uninjured ankle in ballet dancers is not sufficiently developed. Therefore, the purpose of this paper is to summarize the findings of our recent studies using orthopaedic imaging to investigate the ankle in female ballet dancers.

METHODS

Following appropriate ethical approval and informed consent, conventional radiography and magnetic resonance imaging (MRI) were utilized to gather musculoskeletal data from one normal ankle on each of 15 female ballet dancers (mean age=21±3.0 yr; mean experience in dance=17±3.5 yr; mean experience *en pointe*=8±4.8 yr). Eight subjects were imaged by x-ray and 9 by MRI (2 were imaged by both). Three lateral weightbearing x-rays of the ankle and foot were taken: in ankle neutral position and in the ballet positions of *demi-plié* (maximum dorsiflexion) and *en pointe* (maximum plantar flexion). For MRI, an open low-field unit outfitted with an extremity coil obtained weightbearing images of the ballet dancers' ankles as they stood in the ankle neutral and *en pointe* positions.

RESULTS

The extreme plantar flexion required of the talocrural joint in ballet dancers was confirmed on x-ray, as was the contribution of the foot bones to approximately 30% of overall plantar flexion. The posterior edge of the tibial plafond, the posterior talus, and the superior calcaneus converge in virtually all dancers *en pointe*, and when moving to this position, the posterior surface of the plafond passes beyond the articular cartilage of the talus, coming to rest on the posterior talus. One interesting finding in the MR images may help explain an increased risk of Achilles' tendinopathy in dancers: the posterior edge of the *pointe* shoe counter and the *pointe* shoe ribbons compress the tendon as it travels in a curvilinear path (Figure 1).

Figure 1: T1-weighted spin echo MR image of female ballet dancer *en pointe*. White arrow indicates impression by shoe at Achilles' tendon insertion, white bracket indicates compression over Achilles' tendon by shoe ribbons, and black arrow points to high signal area in Achilles' tendon suggestive of pathology.

DISCUSSION

Pathologic conditions in dancers have been identified by x-ray [2] and MRI [3]. However, we are not aware of prior research using orthopaedic imaging to evaluate the structure and motion of the ankles of uninjured ballet dancers. Neither could we locate research assessing dancers' ankles with these techniques apart from a healthcare setting. Our findings offer useful information about the demands of classical ballet on the ankle, especially during extreme plantar flexion, and provide insight to certain pathological conditions of this body region in ballet dancers.

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Structural polymorphisms in midtarsal bone alignment lead to focal midfoot pressures

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INTRODUCTION

Neuropathic Charcot arthropathy (NCA), most commonly seen in individuals with diabetes mellitus (DM) and peripheral neuropathy (PN), is characterized by a rapid and progressive destruction of pedal bones and joints leading to subluxation, dislocation, fracture, and fixed foot deformity. Structural foot deformities increase risk of plantar ulcers and amputation [1]. Here we report the relationship between fixed bony malalignment in midtarsal joints and elevated midfoot plantar pressures during walking.

METHODS

Six volunteers with NCA and 6 matched DM+PN controls underwent a foot-ankle CT scan and plantar pressure assessment. Foot bones were segmented from reconstructed CT data with edge-detection filtering and semi-automated bone separation software [2]. Segmented 3D voxel arrays were imported into an in-house software tool (Figure 1) [3] in which anatomic landmarks were located to define 3D coordinate axes. Cardan rotation sequences were used to measure joint angles of navicular relative to talus (Nav/Tal) and cuboid relative to calcaneus (Cub/Cal) in the sagittal (α), frontal (β) and transverse (γ) planes. Medial and lateral midfoot pressures were found by dividing the pressure profile into thirds antero-posteriorly and in half medio-laterally (Figure 2). Medial and lateral pressure patterns were defined as peak pressures $\geq$ 300 kPa in the medial and lateral midfoot regions, respectively [4].

Figure 1: Segmented foot bones (phalanges omitted)

RESULTS

NCA subjects had similar peak pressures and joint angles in the uninvolved foot compared to DM+PN controls (Table 1). In involved (NCA) feet, two patterns emerged:
(1) High medial midfoot pressure pattern was related to a more inverted Nav/Tal and a more dorsiflexed and inverted Cub/Cal.
(2) High lateral midfoot pressure pattern was related to increased Nav/Tal eversion and increased Cub/Cal plantarflexion.

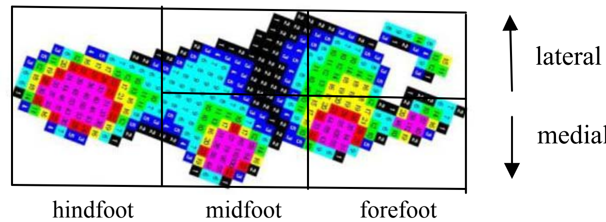

Figure 2: Pressure profile for NCA medial pattern

DISCUSSION

This pilot analysis shows the utility of our CT-based method to assess joint changes in NCA. DM+PN control subjects had consistent joint angles and normal midfoot pressure in both feet, which implies symmetrical pedal joint orientation prior to onset of NCA. The pattern of altered joint orientations and increased midfoot pressure suggests that fixed bony deformities are related to focal increases in plantar pressure which may increase risk for soft tissue damage and ulceration.

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Table 1: Cardan joint angles in sagittal (α) and frontal (β) planes, in degrees (mean ±SD)

	Navicular/Talus		Cuboid/Calcaneus	
	Sagittal (α)	Frontal (β)	Sagittal (α)	Frontal (β)
DM+PN Control	-21.3 ± 5.1	-13.2 ± 6.0	29.8 ± 8.2	-8.3 ± 6.3
NCA, Uninvolved foot	-14.1 ± 5.9	-15.1 ± 6.4	30.5 ± 7.6	-11.0 ± 10.6
NCA, Medial Pressure Pattern	-20.0 ± 12.0	-4.2 ± 8.0*	15.1 ± 16.1*	12.0 ± 23.7*
NCA, Lateral Pressure Pattern	-16.0 ± 5.3	-24.9 ± 8.3*	40.9 ± 8.6*	-7.2 ± 10.2

Sagittal plane (α): + = plantarflexion, - = dorsiflexion; Frontal plane (β): + = inversion, - = eversion. * = $p<0.05$