PROCEEDINGS
OF THE
VIII\textsuperscript{TH}
FOOTWEAR BIOMECHANICS
SYMPOSIUM

edited by E. C. Frederick & S. W. Yang

National Yang-Ming University
Taipei, Taiwan, June 27-29 2007
Organization

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The FBG expresses its deepest appreciation to the organizations that contributed to the financial support of this Symposium. Without their generosity the Taipei meeting, our 8th Symposium as a technical group, would not have been possible.

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The FBG also wishes to express its appreciation to the ISB and to the local groups that contributed to the organization of this Symposium. We are deeply indebted to them for their exceptional efforts that kept the meeting organization on track and helped us maintain a high standard. Without their skilled support the Taipei meeting, our 8th Symposium as a technical group, would not have been possible.

National Yang-Ming University
www.ym.edu.tw/index_en.html

International Society of Biomechanics
isbweb.org/

Taiwan Society of Biomechanics
biomech.bme.ntu.edu.tw/TSB-web/index.htm

Taiwan Society of Biomechanics in Sports
fbs2007.ym.edu.tw/www.tpec.edu.tw/tsbs

Footwear Biomechanics Group (FBG)
- A Technical Group of the ISB
footwearbiomechanics.org
AWARDS

Awards for the 8th Footwear Biomechanics Symposium -Taipei
Each award will be in the amount of US$1000. The competition for these awards is open to all presenters at the conference, with the exception of invited speakers and any presenters who request that they not be considered for an award. Judging is by a committee of selected FBG members, and is based on the quality of the work presented in the submitted abstracts, the podium and poster presentations, and any other scientific criteria that the judges deem appropriate. The decision of the judges is final, and winners will be announced at the closing banquet, along with the announcement of the winner of the $25,000 Nike prize, an award judged independently from the conference presentations. The decisions of the judges are final. All winners are responsible for all taxes, currency exchange fees, etc. to which they may be liable.

Chairperson, Awards Committee:
Keith R. Williams, Ph.D. University of California, Davis, CA, USA

Awards available for this biennial symposium are:

The Adidas Applied Research Award
The Lar New Young Investigator Award
The Li Ning Basic Research Award
The Nike Performance Research Award
The RSscan Pressure Research Award

NB: Also awarded during the symposium will be the prestigious $25,000 Nike Award for Athletic Footwear Research.
Previous Footwear Biomechanics Symposia Proceedings:

Hamill, J., E. Hardin, and K. Williams (eds). Proceedings: 7th Symposium on Footwear Biomechanics, Technical Group on Footwear Biomechanics, Case Western Reserve University, Cleveland, OH, USA 2005


8th Footwear Biomechanics Symposium
Wednesday June 27, 2007
Introductory Session – NYMU Activity Center

3:00p-5:00p  Special Tutorial Lecture
THE HUMAN FOOT FROM EARLY CHILD- TO ADULTHOOD
Ewald Hennig: Universität Duisburg-Essen, Germany
Julie Steele: University of Wollongong, Australia

5:30p-6:00p  Opening Session & Business Meeting

6:00p-7:00p  Keynote Lecture - MOTION CONTROL CONCEPTS REVISITED
Gert-Peter Brüggemann, Ph.D.
Professor and Leader
Institute for Biomechanics and Orthopedics, German Sport University - Cologne, Germany

7:15p-9:00p  Opening Reception

Thursday June 28, 2007

8:00-9:00  Keynote Lecture – CAN FOOTWEAR AFFECT SPORT PERFORMANCE?
Darren J. Stefanyshyn, PhD, P.Eng.
Associate Professor
Faculty of Kinesiology, University of Calgary, Canada

9:00-9:15  THE INFLUENCE OF STANCE LEG TRACTION PROPERTIES ON KICKING
PERFORMANCE AND PERCEPTION PARAMETERS IN SOCCER
Thorsten Sterzing¹, Ewald M. Hennig²
¹Department of Human Locomotion, Chemnitz University of Technology, Chemnitz, Germany
²Biomechanics Laboratory, University of Duisburg-Essen, Essen, Germany

9:15-9:30  CUSHIONING AND PERFORMANCE IN TENNIS FOOTWEAR
Gaspar Morey Klapsing¹, Pedro Pérez Soriano², Salvador Llana Belloch³,
¹INESCOP-Spain, ²Dept. Of Physical and Sports Education, Univ. Valencia-Spain

9:30-9:45  THE INFLUENCE OF FOREFOOT SHOE ELEVATION ON VERTICAL JUMP
PERFORMANCE
Torsten Brauner¹, Thorsten Sterzing¹, Ewald M. Hennig²,
¹Department of Human Locomotion, Chemnitz University of Technology, Germany
²Biomechanics Laboratory, University Duisburg-Essen, Germany

9:45-10:00  FOREFOOT MIDSOLE BENDING STIFFNESS DURING CUTTING MOVEMENTS
Geng Luo and Darren J. Stefanyshyn
Human Performance Laboratory, Faculty of Kinesiology, University of Calgary, Canada

10:00-10:15  DESIGN AND CONSTRUCTION OF A SPRINT SHOE WITH A SELECTIVE LASER
SINTERED NYLON-12 SOLE UNIT
Daniel Toon, Neil Hopkinson and Mike Caine
Mechanical and Manufacturing Engineering, Loughborough University, UK

10:15-10:30  THE CHARACTERISTICS OF UNDER-FOOT PRESSURE DISTRIBUTION FOR AIR
RIFLE SHOOTERS
Vasiljev Radivoj¹, Jelicic Bojana², Vasiljev³ A. Irina³
¹University of Novi Sad, Faculty of Sport, Novi Sad, Serbia
²Institute of Neurology, Clinical Center, Novi Sad, Serbia
³Russian State University of Sport, Moscow, Russia

10:30-11:00  COFFEE BREAK
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<td>Foot Morphology and Fit – SongNing Zhang, Session Chair</td>
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<td>FEMALE FOOT MORPHOLOGY – IMPLICATIONS FOR LAST DESIGN</td>
<td>Inga Krauss, Stefan Grau, Marlene Mauch, Christian Maiwald, Thomas Horstmann</td>
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<td>STUDY OF THE INFLUENCE OF FITTING AND WALKING CONDITION IN FOOT DORSAL</td>
<td>José Olaso, Juan Carlos González, Sandra Alemany, Enric Medina, Amparo López, Carlos</td>
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<td>Martin, Jaime Prat and Carlos Soler</td>
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<td>11:30-11:45</td>
<td>THE INFLUENCE OF BODY POSITION AND PHYSICAL ACTIVITY ON FOOT DIMENSIONS</td>
<td>Sabrina Kunde, Thorsten Sterzing, Thomas L. Milani</td>
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<td>Department of Human Locomotion, Chemnitz University of Technology, Germany</td>
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<td>11:45-12:00</td>
<td>A 3D FINITE ELEMENT SIMULATION OF FOOT-SHOE INTERFACE</td>
<td>Jason Tak-Man Cheung¹, Bruno Bouchet¹, Ming Zhang¹, Benno M. Nigg¹</td>
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<td>¹Human Performance Laboratory, Faculty of Kinesiology, University of Calgary, Canada</td>
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<td>²Dept of Health Technology &amp; Informatics, The Hong Kong Polytechnic University, Hong</td>
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<td>12:00-12:15</td>
<td>A 3D FINITE ELEMENT ANALYSIS OF HUMAN FOOT WITH HIGH-HEELED SHOE</td>
<td>Ming Zhang, Jia Yu, Yan Zhang, Jason Tak-man Cheung, Yan Cong, Aaron Kam-lun Leung</td>
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<td>Department of Health Technology and Informatics, The Hong Kong Polytechnic University, Hong Kong</td>
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<td>1:30p-2:45p</td>
<td>Methodological Advances– Berthold Krabbe, Session Chair</td>
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<td>1:30-1:45</td>
<td>GROUND REACTION FORCES AND TRACTION IN GOLF SHOES</td>
<td>Keith R. Williams¹, Jennifer M. Neugebauer²,</td>
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<td>¹Exercise Biology Program,</td>
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<td>²Biomedical Engineering Graduate Group, University of California, Davis, USA</td>
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<td>1:45-2:00</td>
<td>USING APPROPRIATE BOUNDARY CONDITIONS IN TRACTION TEST DEVICES</td>
<td>Bob Kirk¹, Steve Haake², Christer Rolf³, Ian Noble³, Tom Mitchell³ and Matt Carré¹,</td>
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<td>¹Sports Engineering Research Group, University of Sheffield, UK</td>
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<td>²Department of Health and Wellbeing, Sheffield Hallam University, UK</td>
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<td>³Sheffield Centre of Sports Medicine, University of Sheffield. UK</td>
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<td>MODELIZATION OF THE FOOT THERMAL RESPONSE FOR A WIDE RANGE OF FOOTWEAR</td>
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<td>Department of Sport and Exercise Science, The University of Auckland, New Zealand</td>
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12:15-1:30p LUNCH

1:30p-2:45p Methodological Advances– Berthold Krabbe, Session Chair
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\(^1\) Biomechanics/Sports Medicine Lab, The University of Tennessee, USA  
\(^2\) Biomechanics Laboratory, University of Massachusetts at Amherst, USA     | 65   |
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\(^1\) Prince of Wales Medical Research Institute, Sydney, Australia  
\(^2\) University of New South Wales, Sydney, Australia  
\(^3\) La Trobe University, Melbourne, Australia  
\(^4\) University of Wollongong, Wollongong, Australia | 67   |
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\(^1\) Biomechanics Research Laboratory, University of Wollongong, Australia  
\(^2\) Prince of Wales Medical Research Institute, Australia  
\(^3\) La Trobe University, Australia | 69   |
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<td>Faculty of Health, Staffordshire University, Stoke on Trent, UK</td>
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<td>1 Sole Technology Institute, 20162 Windrow Dr, Lake Forest, CA, USA; 2 Exeter Research, Inc., 80 Haigh Road, Brentwood, NH, USA</td>
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<td>María José Rupérez¹, Carlos Monserrat¹, Mariano Alcañiz¹, Sandra Alemany², Sergio Puigcerver²</td>
<td>¹MedIClab, Universidad Politécnica de Valencia, Spain. ²IBV, Instituto de biomecánica de Valencia, Spain.</td>
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<td>Gait Study Center, Temple University School of Podiatric Medicine</td>
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<td>Elissa J. Phillips, Uwe G. Kersting</td>
<td>Department of Sport and Exercise Science, The University of Auckland, New Zealand</td>
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<td>Liu Jingmin¹, Zheng Xiuyuan¹, Cai Yuhui², Liu Hui³, Jin Jichun³</td>
<td>¹Tsinghua University, Beijing; China ²Institute of Education and Sports Beijing Normal University, Beijing, China ³Beijing Sport University, Beijing, China</td>
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<td>THE VARIABILITY OF BIOMECHANICAL PARAMETERS DURING RUNNING DIFFERENT RUNNING SHOES - A 3-DAY TESTING DESIGN</td>
<td>Jens Heidenfelder, Thomas L. Milani</td>
<td>Department of Human Locomotion – Chemnitz University of Technology, Germany</td>
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<td>Lam MH¹, Hong Y¹, Cheng EYL¹, Fong DTP²</td>
<td>¹Dept. of Sports Science and Physical Education, The Chinese University of Hong Kong, Hong Kong ²Dept. of Orthopaedics and Traumatology, The Chinese University of Hong Kong, Hong Kong</td>
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<td>8:00-9:00</td>
<td>Keynote Lecture- FORCES, MOTION AND OUTCOMES WITH FOOT ORTHOSES AND RUNNING SHOES</td>
<td>Craig B. Payne, DipPod(NZ), MPH</td>
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<td>Clinical Footwear Biomechanics – Howard Hillstrom, Session Chair</td>
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<td>A NEW AND SIMPLE MECHANICAL “SUPINATION SPRAIN SIMULATOR” FOR EVALUATING THE PROTECTIVE EFFECTS OF FOOTWEAR ON ANKLE SPRAIN, Yue-Yan Chan¹, Daniel Tik-Pui Fong¹, Yuk-Ming Tang¹, Kin-Chuen Hui², Kai-Ming Chan¹, ¹Department of Orthopaedics and Traumatology, Prince of Wales Hospital, Faculty of Medicine ² Department of Mechanical and Automation Engineering, Faculty of Engineering, The Chinese University of Hong Kong, Hong Kong</td>
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<td>9:15-9:30</td>
<td>INFLUENCE OF MOTION CONTROL SHOES ON LOWER EXTREMITY MOVEMENT AND PRESSURES BENEATH THE SHOE IN OVER-PRONATORS Sharon Dixon¹, Michelle Giles² and Nachiappan Chockalingham², ¹School of Sport and Health Sciences, University of Exeter, Exeter, EX1 2LU, UK. ² Faculty of Health Sciences, Staffordshire University, Stoke-on-Trent, ST4 3DF, UK</td>
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<td>9:30-9:45</td>
<td>THE INFLUENCE OF OBESITY ON THE PERCEPTION OF TOUCH AND VIBROTACTILE THRESHOLDS UNDER THE FOOT Ewald M. Hennig, Dennis Breuing, Biomechanics Laboratory, University Duisburg-Essen, Germany</td>
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<td>9:45-10:00</td>
<td>EFFECT OF LATERAL HEEL WEDGE INSOLE IN MANAGEMENT OF SUBJECTS WITH OSTEOARTHRITIC KNEE SaiWei Yang¹, Fen-Ling Kuo², Chao-Jung Hsieh³, Lin-Fen Hsieh⁴, ¹Institute of Biomechanical Engineering, National Yang-Ming University, Taipei, Taiwan ²Department of Rehabilitation, Taipei Municipal Wanfang Hospital, Taipei, Taiwan ³Institute of Science and Technology, National Yang-Ming University, Taipei, Taiwan ⁴Department of Rehabilitation, Shin Kong Wu Ho-Su Memorial Hospital, Taipei, Taiwan</td>
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<td>10:00-10:15</td>
<td>PROPRIOCEPTION OF ANKLE JOINT COMPLEX IN YOUNG HOCKEY PLAYERS AND RUNNERS Jing Xian Li, School of Human Kinetics, University of Ottawa, Canada</td>
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<td>10:15-10:30</td>
<td>EVALUATING THE TENDENCY FLAT FOOT IN NORMAL ADULTS USING A TWO-DIMENSIONAL COORDINATE SYSTEM Chi-Wen Lung¹, Jen-Suh Chern² and Saiwei Yang¹, ¹Institute of Biomechanical Engineering, National Yang-Ming University, Taipei, Taiwan ²Department of Occupational Therapy, Chang Gung University, Tao-Yuan, Taiwan</td>
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| 11:00-11:15| A COMPARISON OF 'NEUTRAL' RUNNING SHOES ON PLANTAR PRESSURE PATTERNS AND PERCEIVED COMFORT IN ATHLETES WITH A CAVUS FOOT TYPE: A CROSSOVER, RANDOMIZED CONTROLLED TRIAL | Caleb Wegener¹, Joshua Burns² and Stefania Penkala¹  
¹ Podiatry Program, The University of Western Sydney, Australia  
² Faculty of Medicine, The University of Sydney, Australia |                                                                 |
| 11:15-11:30| THE EFFECT OF ARCH STRUCTURE MIDSOLE ON THE HEEL DURING RUNNING               | Yaodong Gu¹, Jianshe Li¹, Su Li², Guoqing Ruan²  
¹ Human movement Lab, Ningbo University, China  
² Anta Company Limited, China |                                                                 |
| 11:30-11:45| EFFECT OF FOREFOOT ROCKER OUTSOLE IN IMPROVING AMBULATORY ABILITY FOR SUBJECTS WITH DIABETES MELLITUS - | Wei Lin¹, SaiWei Yang², Shi-Ming Lai³  
¹ Inst. of Rehabilitation Science & Assistive Technology, National Yang-Ming University, Taiwan  
² Institute of Biomedical Engineering, National Yang-Ming University, Taipei, Taiwan  
³ Department of Endocrinology, Shin Kong Wu Ho-Su Memorial Hospital, Taipei, Taiwan |                                                                 |
| 11:45-12:00| RELATIONSHIP OF PLANTAR PRESSURE PATTERNS AND LOWER LIMB KINEMATICS IN BAREFOOT RUNNING | Maiwald, C., Grau, S., Mauch, M., Krauss, I., Horstmann, T.,  
Medical University Clinic, Department of Sports Medicine, Tuebingen, Germany |                                                                 |
| 12:00-12:15| FROM HIGH SPEED PLANTAR PRESSURE MEASUREMENT TO LOWER LIMB SKELETAL MOTION   | Jempi Wilsens, Friso Hagman  
RSscan International, Belgium |                                                                 |
| 12:15-1:30p| LUNCH                                                                        |                                                                         |                                                                 |
| 1:30-1:45  | DIFFERENCES IN KNEE JOINT LOADING BETWEEN FOREFOOT AND REARFOOT STRIKE RUNNING PATTERNS | Frank I. Kleindienst¹, Sebastian Campe¹,², Eveline S. Graf¹,³, Kerstin Witte  
¹ Biomechanical Lab SEF, adidas Innovation Team, adidas AG, Scheinfeld, Germany  
² Institute for Sport Science, Otto-von-Guericke-University of Magdeburg, Germany  
³ Institute for Biomechanics, Dept. Mechanical & Processing Eng., ETH Zürich, Switzerland |                                                                 |
| 1:45-2:00  | DAY-TO-DAY VARIABILITY OF KINEMATIC VARIABLES IN RUNNING                     | Stacoff A., Bachmann Ch., List R., Ukelo T., Stüssi E., Wolf P.,  
Institute for Biomechanics, ETH Zurich, Switzerland |                                                                 |
| 2:00-2:15  | FOOTWEAR AFFECTS THE GEARING IN THE MUSCULOSKELETAL SYSTEM OF THE LOWER EXTREMITIES WHILE RUNNING | Bjoern Braunstein, Adamantios Arampatzis, Gert-Peter Brüggemann  
Institute of Biomechanics and Orthopaedics, German Sport University Cologne, Germany |
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<td>2:15-2:30</td>
<td>MOTION ANALYSIS OF TREADMILL RUNNING FOR FOOTWEAR DESIGNING</td>
<td>Seigo Nakaya and Tsuyoshi Nishiwaki</td>
<td>ASICS Corporation, Japan</td>
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<td>2 Swedish School of Sport and Health Sciences, Stockholm, Sweden</td>
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<td>4 Centre for Rehabilitation and Human Performance Research, University of Salford, UK</td>
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Keynote Lectures
INTRODUCTION

Influenced by the fitness and running boom in the last three decades, athletic footwear established from simple shoes to a high tech product. New designs, manufacturing techniques and materials were introduced into the construction of technical athletic shoes. It can be estimated that between 33-56% of runners are affected by at least one injury per year (e.g. Clement et al. 1981, Taunton et al. 2002). Pre injuries, excessive training volume, training mistakes, excessive impact forces, excessive pronation or excessive knee joint moments (adduction-abduction moments, external rotation moments) have been proposed as major reasons for the development of overuse injuries. Footwear and sport surface were assumed to influence impact forces, foot pronation and knee joint moments. Consequently, the concepts of cushioning and motion control were introduced into the development of athletic footwear design. One can observe that over the last twenty years the percentage of motion control shoes increased in the different footwear companies to counteract instability and uncontrolled motion. However, results of more recent studies challenged the proposed association between skeletal alignment, foot pronation and running injuries.

Thus the purposes of this paper are (1) to critically discuss the potential association between foot motion control and the development of running related injuries and (2) to synthesize the technical solutions developed for motion control and to critically discuss the impact of these motion control concepts to injury prevention and performance enhancement.

MOTION CONTROL AND OVERUSE INJURIES

Running related injuries have often been associated with a static or dynamic malalignment of the skeleton. Excessive foot varus or valgus positions have been speculated to create high joint loading and for repeated loading cycles overuse injuries. Additionally, excessive foot eversion and related tibia rotations have been proposed to increase the chance of overuse such as patellofemoral pain syndromes (PFPS), shin splints, Achilles tendon inflammation or even partial ruptures, plantar faciitis, and stress fractures (e.g. James et al. 1978). Thus the proper alignment of the skeleton has been proposed as being one of the most important functions of running shoes as well as shoe inserts. It has been proposed that overuse injuries due to excessive foot and leg movement and especially to excessive foot eversion, could be reduced with shoes by correcting, aligning or limiting the skeletal movement of foot and leg.

In an epidemiological study the lower extremity alignment was suggested not to be a major risk factor for running injuries bases on alignment measurements on a group of runners enrolling in a marathon training program (Wen et al. 1997). A prospective studies with 131 runners showed that foot and ankle movement did not act as a predictor for an increase in running injuries over a 6-month period (Nigg 2001). Bone pin studies demonstrated small and non systematic effects of shoes with different midsoles or inserts on the kinematics of the calcaneus, the tibia and the femur during running (Stacoff et al. 2000, 2001).

Injuries of biological structures are caused by mechanical force and stress. Skeletal kinematics is not necessary related to forces or moments applied. Therefore the relation between skeletal kinematics and overload injuries should be generally weak and non systematic. A prospective study showed the relation between excessive knee joint moments (in the frontal and transverse planes) and an increased risk to receive an overuse injury (PFPS) (Stefanyshyn et al. 2006).
MOTION CONTROL CONCEPTS

The support of the medial foot arch has often been proposed as one of the most effective correction or motion control strategies for foot eversion and pronation or/tibia rotation. The traditional technical concepts from the seventies consequently used a medial support integrated in the midsole of the shoe consisting of harder material. The “duo density” midsole construction was the technical solution in the last thirty years and received in principal no changes over more or less three decades. This technology was used more or less by all athletic footwear companies.

Some trials were performed with midsole modifications at the lateral side. Wide flares, straight soles and rounded soles were constructed to control the angular pronation velocity or to reduce the lever arm for the ground reaction forces and thus to reduce the external eversion moment (Stacoff et al. 2001). The large lateral heel flares were found neither to increase eversion velocity nor internal tibial rotation. The used shoe sole modifications did not produce the expected systematic effects.

More recently lateral material modifications (soft elastic, soft viscous) and/or geometrical modifications were applied to midsoles of running shoes. The concept was used to increase the muscle potential of ankle plantar flexor muscles, foot inversion muscles and knee extensor muscles.

CONCLUSION

Motion control concepts are used for more than 20 years in running footwear. The frequency of running related injuries did not change significantly in this period of time. Remarkably even the injury distribution and location were not affected. The relative number of knee and shank injuries did no differ when comparing data reported from the same group (Clement et al. 1981, Taunton et al. 2002). One can conclude that the purely mechanical concept of motion control by duo density midsole technology did not meet the expectations to decrease the risk of injuries. The skeleton changes its path of movement for a given task only minimally and non systematic when exposed to a mechanical intervention. The locomotor system seems to choose a strategy to keep a “minimal resistance movement path” (Wilson et al. 1996). An optimal or appropriate shoe concept would affect muscle activity and muscle force potential rather than a mechanical support.

REFERENCES

CAN FOOTWEAR AFFECT SPORTS PERFORMANCE?

Darren J. Stefanyshyn
Human Performance Laboratory, University of Calgary

INTRODUCTION
Various footwear characteristics such as traction, mass, energy return, etc. have been proposed to influence sport performance. Some of these claims are anecdotal without strong scientific evidence. Others may even be erroneous. However, there are some publications that attempt to relate specific footwear properties to performance improvements in various sports.

ENERGETICS
Sport shoes can influence the energetics of human movement. The three main aspects where sport shoes can play a role are in maximizing the energy which is returned to the athlete, minimizing the energy which is lost by the athlete and optimizing the musculoskeletal system (Nigg et al., 2000). Maximum values of energy storage in a shoe sole are on the order of 10 J (Shorten, 1993). If all of this energy is returned to the athlete, it should be enough to influence performance. However, shoe midsoles lose approximately 30 % of the energy input (Alexander and Bennett, 1989). Additionally, depending on the movement, energy return sometimes occurs at the wrong time, frequency, location and in the wrong direction which compromises the ultimate influence on improving performance (Nigg and Segesser, 1992). As a result, the actual influence that energy return has on performance is probably minimal.

Examples of the strategy to minimize energy loss include (1) reducing the mass of the shoe, (2) using appropriate midsole materials which dissipate unwanted vibrations, (3) implementing constructions which improve the stability of the ankle joint and (4) increasing the bending stiffness of shoe midsoles which reduces the energy lost at the metatarsophalangeal joint. It has been shown that even small decreases in shoe mass can have a significant influence on performance, especially during activities where the lower leg encounters large accelerations (Catlin and Dressendorfer, 1979). Increased forefoot bending stiffness has been shown to increase running, sprinting and jumping performance (Stefanyshyn and Nigg, 2000; Stefanyshyn and Fusco, 2004; Roy and Stefanyshyn, 2006).

Footwear can influence the dynamic force production of muscles by manipulating the intrinsic musculoskeletal characteristics of the force-length and force-velocity relationships. That is, footwear can have a large influence on the technique that an athlete is using, which can have a substantial influence on performance. A dramatic example of this is the introduction of the klapskate. The incorporation of a hinge in traditional speed skates shifted where speed skaters operate on the force-velocity and force-length relationships, allowing greater energy production from the quadriceps muscles (Houdijk et al., 2000). The result was a dramatic increase in performance and a virtually instantaneous shattering of current world records. Although optimizing the musculoskeletal system is often overlooked in sport shoe development, it is proposed that this concept has tremendous potential to influence performance.
OTHER FACTORS

Traction
Numerous studies have alluded to the fact that footwear traction can affect performance. It is intuitive that insufficient traction will result in an athlete slipping and potentially even falling, obviously compromising performance. However, there exists a lack of published scientific studies to support these claims and to identify minimum traction requirements necessary for sports performance.

Ankle Support
Some studies have suggested that increasing footwear ankle support to improve stability comes at a performance cost (Robinson et al., 1986; Brizuela et al., 1997). Low top basketball shoes lead to increases in jump height and faster times in an obstacle course in comparison to high top shoes. It was proposed that the reduced performance with the high top shoes was due to greater restriction of ankle joint plantarflexion.

Training
Although not directly during the sport performance, some shoes have claimed to increase performance indirectly through a training effect by incorporating specific characteristics such as increased mass or increased forefoot height in footwear worn during training. Few studies support these claims but Kraemer et al. (2000) have shown that a shoe designed to overload the plantar flexor muscles and improve the linear up and down phases of movement can improve jump height and 60-yard dash times after eight weeks of training

CONCLUSION
Footwear can definitely influence performance. Published studies have shown that footwear can have an influence on performance, both positively and negatively. In some cases, the influences can be dramatic. However, the number of studies directly relating footwear to performance is limited and opportunities exist for additional research to be performed.

REFERENCES
Nigg, B.M., Stefanyszyn, D.J. et al. (2000) Biomechanics and Biology of Human Movement, Ed B.M.
Overuse injury and pathology in the lower limb is widely considered to be, in part, due to abnormal position, motion or function of the foot. This biomechanical dysfunction is thought to overload the tissues beyond normal physiological ranges resulting in these overuse symptoms. Foot orthoses are commonly used in clinical practice for the treatment of the assumed causes of this biomechanical dysfunction of the foot that is thought to result in this overuse pathology. Foot orthoses directed at correcting this assumed dysfunction have been widely shown, in randomised controlled trials, outcome surveys and patient satisfaction surveys, to be clinically effective. Running shoes with motion control features are also commonly used to treat this assumed biomechanical dysfunction, but the effects on injury rates are not known. Despite this, running shoes with these features are widely recommended for those with these biomechanical dysfunctions.

A lot of recent literature has challenged the accepted theoretical underpinning that is most widely used for the prescription and clinical use of foot orthoses and the motion control features of running shoes. The clinical practice of foot biomechanics and the use of foot orthoses clinically to alter foot biomechanics and the associated dysfunction are seen as having a poor fit between the theory of clinical practice and the science or evidence to underpin it. A large part of clinical practice and the use of motion control running shoes is based on the foot being in an abnormal position or functioning abnormally. However several prospective studies have shown that these abnormal positions or structure of the foot are not strong prospective predictors of the development of pathology. The most commonly accepted abnormal function of the foot is that of ‘excessive pronation’, but in light of this evidence (or lack of) the concept of excessive pronation of the foot as being pathologic has been questioned. Foot orthoses used to treat this abnormal pronation have been shown, in reviews of the available literature, to be clinically effective in treating a wide range of overuse symptoms in the lower limb. No prospective data is available on the use of motion control running shoe technology actually effecting injury rates. This dissonance or paradox between excessive foot pronation is not necessarily being associated with clinical pathology in prospective studies and the clinical success of treating excessive pronation of the foot is problematic.

Due to this apparent problem, further consideration of the role of other factors that may prospectively predict injury and the determination of the clinical response to foot orthoses or the use of design features in running shoes may be necessary. In light of the abnormal position or structure of the foot related to excessive pronation not being necessarily pathologic, the concept of the forces behind these abnormal positions, structure or function are worthy of consideration as one of these factors. Intuitively, it is the forces that are responsible for actual damage to tissues and not motion and position. As excessive pronation of the foot is the most commonly assumed cause of biomechanical dysfunction, this consideration the forces that are pronating the foot and the forces need to supinate the foot is
considered one solution to the paradox. Little emphasis has been given in the literature to forces behind abnormal structure and function of the foot.

Previous research on the role of biomechanical dysfunction mentioned above has focused on the assumption that kinematics (ie abnormal position and motion) is a cause of overuse pathology and has measured the kinematic response of the foot and lower limb to foot orthoses and running shoes (ie changes in abnormal position and motion). No work has been done on the role of kinetics (ie forces) as a possible cause of overuse pathology and only minimal research has considered the kinetic response of the foot and lower limb to foot orthoses and running shoes.
Abstracts of Presentations
THE INFLUENCE OF STANCE LEG TRACTION PROPERTIES ON KICKING PERFORMANCE AND PERCEPTION PARAMETERS IN SOCCER

Thorsten Sterzing¹, Ewald M. Hennig²
¹Department of Human Locomotion, Chemnitz University of Technology, Chemnitz, Germany
²Biomechanics Laboratory, University of Duisburg-Essen, Essen, Germany

INTRODUCTION

Kicking is the most typical sport specific technique in soccer. Especially the full instep kick, the fastest kicking technique, has received attention in biomechanical research [2]. Ball velocity of kicks on goal is capable to decide games since a faster ball means less reaction time for the goal keeper. Soccer shoes have been shown to take influence on kicking velocity [3,4,5]. So far, studies on the influence of soccer shoes on kicking velocity focused on the shoe of the kicking leg. However, the shoe at the stance leg is also important for kicking performance. In kicking, due to kinetic chain theories, the transfer of momentum to the ball is based on the whole sequence from foot strike of the stance leg to the end of the collision phase of the kicking leg. Higher vertical GRFs of the stance leg were already linked to increased ball velocity during kicking [1]. Thus, traction properties of the stance leg most likely take influence on kicking success, too. Therefore, the purpose of this study was to evaluate the influence of different traction properties of soccer shoes at the stance leg on kicking velocity. Thereby, performance parameters of the full instep kick in soccer, like peak ball velocity and action time, were investigated. Action time was defined as the time period from initial foot strike of the stance leg to initial movement of the ball. Furthermore, it was examined whether soccer players are able to perceive their achieved ball velocity and the different traction properties at the stance leg during foot strike.

METHODS

23 experienced soccer players participated in this study (age: 24.6±3.1 years; height: 177.4±4.8 cm; weight: 73.8±5.4 kg). The subjects performed six maximum full instep kicks in four different shoe conditions at the stance leg which only differed in their outsole configurations (Picture 1). The basic shoe model used was the Nike Mercurial Vapor II. The regular firm ground (FFG) and soft ground (FSG) versions as well as two modified versions of the firm ground model were used. The studs of these modified versions were functionally cut to either 50% (HFG) and 0% (ZFG) of their original length. With regard to the kicking leg subjects wore the same shoe (FFG) all throughout the testing in order to achieve neutral conditions. Subjects were required to perform a standardized three-step-approach for all kicks.

![Picture 1: Shoe Conditions – Different Outsole Configurations](image)

The study was designed as a laboratory experiment. Kicking took place on artificial turf (DD-Soccer Grass HPF CROWN/DIN 18035 T 7 120 μ, 8800 dtex, density 30000/m2) towards a spanned net. Peak ball velocity was measured by a Stalker Pro radar gun (Applied Concepts Inc., Plano, TX, USA). GRFs of the stance leg at foot strike were taken by a Kistler 9281 force plate. Action time was measured from...
the beginning of foot strike of the stance leg to initial ball movement, which was controlled by a photo cell. Furthermore, subjects were required to give a perception ranking of achieved ball velocity (1 – best, 4 – worst) as well as a perception rating of stance leg traction properties (anchored perception scale: 1 – very high, 9 – very low) for the different shoe conditions.

RESULTS AND DISCUSSION

Results show that subjects achieved a higher ball velocity in the HFG and FFG shoes. In these shoe conditions subjects also appeared to have shorter action time values and higher values for maximum resultant shear force. With exception of the post-hoc tests of ZFG to FSG and of HFG to FFG all other comparisons were statistically significant according to Fisher’s PLSD (Figure 1). Perception parameters show that subjects were able to differentiate between their achieved ball velocity as well as between the different traction properties. It is remarkable, that despite actual and perceived ball velocity appeared to be comparably low in the FSG shoe, it was rated to provide almost as high traction as the best shoe in this study (FFG) (Figure 2). These findings strongly call for functionally high traction properties rather than solely mechanically high traction properties.

CONCLUSION

Only high and functional traction properties of the stance leg (FFG, HFG) resulted in higher resultant shear forces and thus shorter action time leading to higher ball velocity. High but dysfunctional (FSG) as well as low and dysfunctional traction properties (ZFG) resulted in weaker full instep kicking performance. A follow-up study should examine whether smaller differences of outsole configurations at the stance leg influence kicking performance as severe as the set of shoes in this study did.

REFERENCES


ACKNOWLEDGEMENTS

This research was supported by Nike Inc., USA.
CUSHIONING AND PERFORMANCE IN TENNIS FOOTWEAR

Gaspar Morey Klapsing1, Pedro Pérez Soriano2, Salvador Llana Belloch2
1INESCOP-Spain, 2Dept. Of Physical and Sports Education. Univ. Valencia-Spain

INTRODUCTION

There is evidence on that cushioning is perceived as comfortable, might prevent pain and overuse injuries as well as delay soreness (Mündermann et al. 2001). For running shoes the sometimes claimed energy return is negligible (Nigg et al. 1986). On the other hand cushioning is related to energy dissipation and is therefore is minimized in sprint shoes. In court sports fast reaction and stop and go motions are essential for performance. Nevertheless court sports specific footwear generally incorporates cushioning elements to damp the frequent and also high impacts occurring. Whether the cushioning might have a significant effect on the reactive performance (fast reaction) of sports footwear has received only little attention (Nigg et al. 1989).

This intervention study, done on concrete court tennis, is a pilot towards identifying the potential influence of cushioning on reactive performance parameters. It aims to gain knowledge towards an adequate tuning of sport footwear cushioning.

METHODS

Two tennis shoe models (NIKE Air Zoom Animo and NIKE Air Resolve) were tested with and without an additional cushioning insole (5mm Poron). Two tests related to energy absorption were performed: UNE-EN ISO 20344-2005 and SATRA TM142. The first one calculates the energy needed to compress the sole from 50 N to 5000 N at 10 mm/minute (Fig 1). The SATRA TM142, calculates the energy return basing on the rebound height after an impact at 4.16 J (8.5 Kg released from 50mm height) (Fig 2), which is supposed to correspond to a running average male. In addition 10 male tennis players (at least national competitive level) participated in this study. Plantar pressures were measured during different game situations on a concrete court: Baseline playing, alternating short and long balls, combined volley, lob/smash. Ground reaction forces were recorded during three tasks: Maximal squat jump after an acoustic signal, 5 consecutive squat jumps, 180° sidestep turn. All tests were done with and without wearing the additional Poron insole in random order.

Figure 1. Energy absorption test following UNE-EN ISO 20344-2005. The Poron insole led to a decrease in the slope, but the area remained nearly unaffected.

Figure 2. Energy return test following SATRA-TM142. The Poron insole led to a 30% decrease in the energy return.
RESULTS
The mechanical testing revealed no major differences in energy absorption (similar areas below the curves with and without Poron). The maximal difference was below 5% generally around 2% but in both directions. The energy return (SATRA TM142 standard), revealed that adding the poron insole led to more than 30% decrease in the energy returned.

Adding the poron insole generally resulted into lower peak plantar pressures, but in several cases also in higher median pressures.

Regarding the performance tests, no differences were found neither for the maximal squat jump, nor for the five consecutive squat jumps (P>0.3). However for the 180° sidestep turn, the Poron caused a significant delay (P=0.028) which in average was above 20 ms meaning almost 4% delay.

DISCUSSION AND CONCLUSIONS
This study compared several parameters obtained from 2 commercially available tennis shoes as they are delivered with these same shoes having an additional insole made of 5mm Poron, a typical material for shock insulation. The slow loading (UNE-EN ISO 20344-2005) revealed almost no differences among conditions in energy absorption. This test is supposed to serve for evaluating shock absorption in security footwear, but due to its slow force ramp it is probably not adequate for this purpose. The impact test at 4.16 J (SATRA TM142) is quite more specific and revealed that adding the Poron insole reduces the energy return by more than 30%, which in absolute values corresponds to only 0.3 J. When looking at the effects of the additional insole on human performance we only found a small but significant detrimental effect of increased cushioning for the 180° sidestep turn.

The low differences found, indicate that in general no big differences in performance might be expected even when cushioning is considerably increased. Although not analyzed statistically, most of the subjects perceived the additional insole as comfortable but would not wear it for competition. Some of them claimed that it felt slightly unstable.

It has to be considered that in this study adding the insole altered the geometry of the shoe. This could have influenced the results. Furthermore, the analyzed shoes already provide cushioning. Therefore it is possible, that when comparing shoes having the same geometry but differing cushioning (from low to high), differences might become clearer. Finally, even if the differences were low, in competition small differences might make the point. A more comprehensive study could possibly allow identifying sport and level specific cushioning properties. Besides comfort and performance parameters, it might be advisable to include also the influence of cushioning on stability into any further study.

REFERENCES

Acknowledgements: Study supported by: local government of Valencia (IMPIVA – IMCOCA/2006/64) and by the Spanish Ministry of Industry, Tourism and Commerce (FIT-020600-2006-90). Special thanks to: Angel Camp (Diabetic Foot Care, Hospital Virgen del Consuelo) and Norberto Portas (INESCOP).
THE INFLUENCE OF FOREFOOT SHOE ELEVATION ON VERTICAL JUMP PERFORMANCE

Torsten Brauner¹, Thorsten Sterzing¹, Ewald M. Hennig²
¹Department of Human Locomotion, Chemnitz University of Technology, Germany
²Biomechanics Laboratory, University Duisburg-Essen, Germany

INTRODUCTION
Vertical jumping ability is a key factor in many modern sports. This is the reason for many scientific articles about training methods to maximize vertical jump performance of athletes. Less literature is found dealing with the influence of shoe modifications on vertical jump performance. Some inventions of shoes or external shoe attachments decrease the heel height in relation to the forefoot [1, 4] which is supposed to improve the jumping performance of athletes. The supposedly positive effects of a forefoot lift are discussed controversially in the scientific literature. Reaper et al., for instance, found no positive influence of strength shoes on jumping performance after six weeks of plyometric practice [6]. However, Larkins and Snabb showed that athletes jumped significantly higher while wearing shoes with a negative inclination [5]. Therefore, the goal of this study was to show whether training in sport shoes with a smaller pitch compared to regular shoes has a positive effect on vertical jumping performance. A forefoot lift was integrated into regular basketball shoes, so that the basketball players were able to wear these modified shoes during regular team practice. The reduced heel height is supposed to result in a more extensive use of eccentric work thus leading to more explosive muscle activity and increased jumping height [2].

METHODS
Thirty-seven experienced basketball players (13 male, 24 female; 23,7yrs ±5,0; 180,4cm ±9,8; 72,0kg ±13,3) of five local Basketball teams participated in this study. Maximum vertical jump of the subjects was tested in the pre-measurement using a Kistler 9281 force plate. Jumping height was determined by double integration over time of the vertical ground reaction force [3]. The subjects performed five countermovement jumps starting in an upright standing position on the force plate.

The members of each team were randomly assigned to the test or control group. Basketball shoes with build-in forefoot elevation (Fig. 1, left shoe) were given to the test group, the same shoe model without the elevation was given to the control group. Both groups wore the shoes in all basketball practice sessions for six weeks (on average 26 hours of practice). After this period post-measurements were taken. To eliminate any effects of the modified shoes during the pre- and post-measurements all subjects had to wear their own shoes for the experimental jumps.

The elevations of the test group shoe were 7mm for shoes smaller than US 9 and 8mm for shoes larger than US 9. So the insert reduced the shoe pitch from 16mm originally to 9mm respectively 8mm in the test group (Fig. 1).

In addition to jumping height further biomechanical parameters like peak vertical force (PFZ), maximum force rise (FRIs), maximum force relaxation (FRel), peak velocity (PVel), and mean velocity rise (VRS) were calculated. Paired t-tests were performed to identify differences between pre- and post-measurements.

Fig. 1 Longitudinal cut of test shoe with orthotic (left) and control shoe without (right)
RESULTS AND DISCUSSION

After six weeks of practice no significant increase in jumping height was found in either group. In the control group all the other biomechanical parameters remained unchanged, too, whereas the test group showed some significant changes in relevant biomechanical parameters (Figure 2, 3, 4). These parameters were PFZ (−4%, p < .001), FRis (−10%, p < .05), and FRel (−4%, p < .05). Additionally, VRS was significantly reduced, too (−5%, p < .05).

![Fig. 2 Peak Vertical Force (PFZ)](image)
![Fig. 3 Force Rise Rate (FRis)](image)
![Fig. 4 Force Relaxation Rate (FRel)](image)

So although there was no decrease in vertical height, the test group showed significant decreases in parameters characterizing speed of force generation and force relaxation. Additionally, there was a significant increase in time of the downward phase of the countermovement jump (+5%, p < .05). The changes in these biomechanical parameters have the opposite effect of the expected outcome of this study. Possibly, the modified shoe construction may have induced a protective behavior of the basketball players in their movement pattern. All biomechanical parameter values in the post measurements in the test group point towards a loss in force explosiveness. A more cautious movement behavior to protect the body against injuries is a likely reason for this finding.

CONCLUSION

Even though a moderate change of footwear pitch of basketball shoes showed no effect on jumping height, this study showed that shoe forefoot elevation does have an influence on relevant biomechanical parameters of vertical jump performance. As the test group showed a decrease in explosive muscle force behavior it can be concluded that wearing shoes with a moderate decrease in heel height is counterproductive.

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ACKNOWLEDGEMENT

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FOREFOOT MIDSOLE BENDING STIFFNESS DURING CUTTING MOVEMENTS

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INTRODUCTION
Despite the fact that running shoes have been widely studied during the past two decades, the number of studies focusing on court shoes is limited. While running involves primarily linear heel-to-toe motions, in court sports lateral cutting maneuvers are of high frequency. It has been shown that such cutting maneuvers lead to much higher peak ground reaction forces in both medial-lateral and vertical directions compared to running (McClay et al., 1994). In running shoe studies it has been shown that increasing shoe longitudinal bending stiffness improves jumping and sprinting performance by reducing the loss of energy at the metatarsophalangeal (MTP) joint (Stefanyshyn and Nigg, 2000; Stefanushyn and Fusco, 2004). It has also been found that increasing midsole stiffness improves running economy (Roy and Stefanushyn, 2006). However, in-depth knowledge regarding the kinematics, kinetics, and joint energy of the MTP joint during cutting movements is not adequate in the literature. Furthermore, the effects of midsole bending stiffness on the MTP joint energy during cutting are not known. Thus, the purposes of the current study were: a) to investigate the MTP joint kinematics and kinetics and b) to examine the influence of midsole bending stiffness on MTP joint energy during cutting maneuvers.

METHODS
Five university level basketball players playing the position of power forward participated in the study. Midsole bending stiffness was adjusted by inserting different carbon fiber plates into the forefoot part of the testing shoes. Two of the three carbon fiber plates were cut either horizontally or longitudinally to give bending stiffness only along one direction. Four shoe conditions were examined: 1) full plate; 2) horizontal cut; 3) longitudinal cut; and 4) control (no plate). Subjects performed 10 trials of v-cut maneuvers in each shoe condition. 3D kinematic data of the forefoot, rearfoot and shank segment were collected using an eight high speed camera system (Motion Analysis Corp.) sampling at 240Hz. A Kistler force platform was used to collect the kinetic data at sampling frequency of 2400Hz. The raw data were filtered using a fourth-order low-pass Butterworth filter with cutoff frequencies of 100Hz and 10Hz for the kinetic and kinematic data respectively. Kintrak (Motion Analysis Corp.) was used to perform inverse dynamics data analysis. For the calculation of MTP joint kinetics, it was assumed that the resultant joint moment was zero until the center of pressure moved distal to the joint. A one-way repeated measures ANOVA was performed for each variable in order to compare between the four shoe conditions. The level of significance was set at α= 0.05.

RESULTS
During the v-cut movement, the range of MTP dorsiflexion was significantly lower in the full plate (14.4°) and longitudinal cut (15.1°) conditions than in the control (18.5°) and horizontal cut (19.4°) conditions. The full plate shoe reduced the MTP dorsiflexing velocity significantly (74.1°/sec lower than the horizontal cut shoe; 56.3°/sec lower than the control shoe).

There was no significant difference in peak MTP joint dorsiflexor moments across the four shoe conditions. However, there was a trend of larger MTP dorsiflexor moments with the full plate shoe (full plate: 61.0Nm; longitudinal cut: 53.6Nm; control: 57.2Nm; horizontal cut: 52.2Nm). Increasing the longitudinal midsole bending stiffness tended to reduce the energy absorbed at the MTP joint (Figure 1:...
full plate: 11.9J; longitudinal cut: 10.7J; control: 14.0J; horizontal cut: 13.3J). No statistically significant differences in energy returned at the MTP joint were found across the four shoe conditions (Figure 2), however, the full plate shoe tended to show more positive work (full plate: 2.3J; longitudinal cut: 1.8J; control: 1.9J; horizontal cut: 1.7J).

![Figure 1: Energy Absorbed at the MTP Joint](image1)

![Figure 2: Energy Returned at the MTP Joint](image2)

**DISCUSSION**

The range of dorsiflexion motion at the MTP joint during a v-cut is much lower than what has been found in running shoe studies. Shoes with higher bending stiffness can reduce the range of motion and peak angular velocity at the MTP joint during a v-cut movement. The trend that the stiffer forefoot midsole reduced negative work at the MTP joint during a v-cut is similar to the results from running shoe studies. However, compared to results from the running shoe studies, the current study showed a larger ratio of energy generated over absorbed at the MTP joint. It is believed that the larger amount of returned energy during cutting compared to running is due to the greater plantarflexion motion at the MTP joint at toe-off. This may suggest manufacturers could engineer devices enhancing energy return at the forefoot section of court shoes. The effects of the reduced energy dissipation by increasing midsole bending stiffness on cutting performance need to be quantified in future studies.

**REFERENCES**


DESIGN AND CONSTRUCTION OF A SPRINT SHOE WITH A SELECTIVE LASER SINTERED NYLON-12 SOLE UNIT

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INTRODUCTION
Currently, sprint shoe outsoles are invariably injection moulded. This process imposes constraints upon outsole design, principally because of the necessity for hardened steel tooling. Selective laser sintering (SLS) is an additive manufacturing process that uses a laser to selectively sinter compacted powder. This tool-less process permits production of complex 3D forms and enables cost effective, low-volume manufacture. It is thus suited to manufacture of products where customisation is desired.

Whilst investigating whether shoe midsole properties influence athletic performance, Stefanyshyn & Nigg (2000) discovered that the lost energy at the metatarsophalangeal joint (MPJ) was decreased by increasing the bending stiffness of the shoe’s midsole and consequently resulted in significant improvements in jump height. Different shoe bending stiffness values were achieved by adapting shoes such that carbon fibre plates could be inserted into the midsoles. In 2004, Stefanyshyn & Fusco used similar customising techniques by inserting carbon fibre plates with bending stiffness of 49, 90 and 120 N-mm\(^{-1}\) into sprint spikes. Increasing shoe bending stiffness typically resulted in improved sprint performance, however, the authors asserted that tuning of an athlete’s shoe stiffness to their individual characteristics is required if performance is to be maximised.

It is apparent from the literature that a performance benefit may be gained from customising the mechanical properties of sprint shoes. The current paper reports on the feasibility of using SLS nylon-12 to manufacture sprint spike sole units.

DESIGN
A sprint spike last, uppers and sole units were sourced. A 3D solid model of the sole unit was produced (SolidWorks, 2005) with corresponding curvatures between the superior surface of the sole unit and the inferior surface of the shoe last and upper. Consistency between the curvatures of the two mating surfaces was important to ensure good bond strength and structural integrity. The superior surface of the sole unit was simplified and re-meshed until an acceptable curvature was achieved. The re-meshing process was iterative; each CAD design was produced and manufactured (Vanguard HS, 3D Systems, USA). The sole units were then physically handled and relationships between the mating surfaces inspected to evaluate fit. Once what was believed to be an optimal fit between the shoe last and sole unit had been achieved the construction process was initiated.

CONSTRUCTION
The inferior surface of the lasted upper was roughened around the perimeter using a rotating scourer. The lasted upper and SLS nylon sole unit were then cleaned and a freshly mixed primer applied to the mating surfaces. The primer was left to dry for 30 min. Next, a single coat of PU adhesive was applied liberally to the mating surfaces of the sole units and uppers and left to dry for 30 min. The adhesive was activated in a flash oven to ensure that that the SLS nylon sole unit reached a temperature whereby it would conform under pressure. On removal from the oven the sole units were manually aligned on the lasted upper, placed on a fixture and subjected to 4 bar of pressure for 30 s to complete the bond. Finally, to ensure suitable traction in subsequent wear trials, an outsole tread was bonded to the forefoot region of the completed sprint shoe. Sprint shoes were constructed using SLS nylon-12 sole units with thicknesses of 2.0, 2.5, 3.0, 3.5, 4.0 and 4.5mm.
MECHANICAL TESTING

The shoes with SLS sole units and a selection of current commercially available sprint spikes were tested using a modified procedure based on the American Society for Testing and Materials Standard Test Method for Flexibility of Running Shoes (ASTM F-911 – 85). Each shoe was secured at the forefoot with the flex fulcrum defined as a line perpendicular to the shoe centre line at a distance of 70% of the total shoe length from the rearmost part of the heel counter. Force data were recorded throughout a period of flexion and a separate period of extension; the results are shown in Figures 1 & 2.

RESULTS

The mechanical testing showed that as the thickness of the SLS sole units was increased, the bending stiffness about the point corresponding with 70% of shoe length also increased. Additionally, sprint spikes A, B and C are benchmark values used to describe the range in stiffness performance of current commercially available sprint spikes. In comparison to the benchmark values, the bending stiffnesses, in both flexion and extension, achieved by the shoes with SLS nylon sole units appropriately spans the range. The maximum loads (mean ± S.D.) in flexion for the sprint spikes were 28±0.2N, 22±0.2N and 15±0.1N. The range in maximum load for the sprint spikes with SLS nylon sole units spanned a range from 12±0.4N to 45±1.5N. The maximum loads (mean ± S.D.) in extension for the sprint spikes were 115±1.1N, 95±0.7N and 25±0.3N. The range in maximum load for the sprint spikes with SLS nylon sole units spanned a range from 48±2.3N to 145±1.7N. Athletes provided feedback on the SLS outsoled spikes having worn them to perform bounce drop jumps and other sprint related tasks.

DISCUSSION AND CONCLUSION

SLS of nylon-12 produces mechanically robust soles units capable of being used within the manufacture of sprint shoes. The results of the mechanical testing coupled with the athlete feedback suggest that the shoes with SLS sole units were comparable to commercially available spikes. The SLS process opens up the possibility of developing personalised footwear to further explore the potential to enhance an individual’s sprint performance via tuning of the outsole’s bending stiffness to the athlete characteristics.

REFERENCES

THE CHARACTERISTICS OF UNDER-FOOT PRESSURE DISTRIBUTION FOR AIR RIFLE SHOOTERS

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INTRODUCTION

The technique of performing air rifle shooting is a complex action, assembled out of a large number of technical elements: a) posture of a shooter - that differs from the usual standing posture (Nasonova, A.A., 2005), b) aiming technique, c) breathing management and d) control of triggering (Dzagamadze T.A. et al. 1998, Korph, A.Y. 1987). When taking his position, the shooter stands under the right angle, with his left side facing the direction of the target (Dzagamadze T.A. et al. 1998, Korph, A.Y. 1987), while the impact on the body is up to 90% (Reinkemeier, H., et al. 2002). As far as the width of the posture goes, different opinions exist relating to the distance between one's feet. Certain authors believe that the ideal position of the shooter's feet is at shoulders’ width (Dzagamadze T.A. et al. 1998), others are considering that it should be narrower (Pullum, W., and Hanekrat F.T., 1973), while the rest claim that the posture could also be a bit wider (Reinkemeier, H., et al. 2002). Theoretically, the balance is attained when the total body mass is symmetrically distributed on both sides of the central line that separates the human body. This posture, found in certain athletes, recognizes equally distributed pressure on both feet (Korph, A.Y. 1987, Dzagamadze T.A. et al. 1998, Reinkemeier, H., et al. 2002). It is necessary to emphasize that top performers put more pressure on one foot than on the other, which is found to be the result of moving the body mass projection closer to the left foot (Dzagamadze T.A., et al. 1998, Korph, A.Y. 1987). When in balance, the total body mass distributes under one foot in means of 50:50 proportion, while pressure in the frontal part of the foot increases (leaning on the big toe – “toe heavy” technique) within certain top performance shooters (Reinkemeier, H., et al. 2002). The goal of our research project was to define the characteristics of the distribution of under-foot pressure within a population of air rifle shooters.

MATERIAL AND METHODS

The podometrics Footscan® platform (length 0.55m, width 0.4m and working frequency 300 Hz) of the RSscan International company was used for reaching the goal that was set at the beginning of the research. Active air rifle shooting athletes (n=12), with a minimum 3 year experience in competitive shooting, participated in the experiment. The experiment was conducted at an indoor rifle range in the city of Novi Sad (Serbia), in standard competitive conditions. The characteristics of the under-foot pressure distribution were registered using the usual tests for evaluating body stability. The vertical posture was focused on, with extended arms at shoulder height and spread out fingers, while the feet were parallel and at a shoulder width apart (Mistulova, T.E. 2005). Two versions of the test were used - with open eyes (OE) and with closed eyes (CE), in time frame of 100s and frequency of 10 Hz. Induced tests were used according to a recommendation accepted for an evaluation of psycho-physiological conditions in humans (Finaeva, E.V., and A.G. Bolonev, 2005). Taken postures were completely suitable for the shooters’ positioned legs and feet on surfaces during the shooting. Except these mentioned tests, the characteristics of under-foot pressure distribution during performance of five shots from the air rifle (AR), on a 10m distance, were also registered. The registration of parameters was conducted on a podoplatform in 20 sec time interval and a 50 Hz frequency. The testing (OE, CE, AR) was performed in sequences without pauses. The registered parameters gave an evaluation of a level of
vertical body position coordination, and, with the help of trajectory of pressure center projection, an information about increasing fatigue (Finaeva, E.V., and A.G. Bolonev, 2005) and, also, an evaluation of the direction that the under-foot pressure shifts in. The dynamics of under-foot pressure gave us an opportunity to evaluate with which leg/foot, and with which part of it – frontal (forefoot) or heal (rearfoot), does the shooter put more pressure on the surface. As a result of the conducted tests, values of under-foot pressure, given in percents, that characterize air rifle shooters, were obtained.

RESULTS
The results showed a relation between under-foot pressure in different zones expressed through percentage of body mass, received after conducting the tests (OE, CE and AR). Statistically significant difference was found between scores of under-foot pressure in particular tests. For example, in frontal zone of the left foot with (AR 23.6% and OE 17.7%), the statistically significant difference is (p>0.006), while between (CE 17.7%) and (OE 21.3%) is (p>0.029). Statistically significant difference of pressure under the right foot heel, is found when comparing (AR) and (OE), (CE) and (OE) at (p>0.003). Observing the position of the projection of central pressure, it is found that in (OE) test it is closer to the right foot, while in (CE) and (AR) to the left. The relation between the left and the right foot in (AR) is 53.7:46.3% and the difference is statistically significant, specially in the left rear-foot zone (p>0.006).

DISCUSSION AND CONCLUSION
During the performance of the shots, the pressure is more distributed on the left leg (53.7%), while the shooter uses the rearfoot of the same leg (30.1%) as a support. It is necessary to remark that, while taking their positions, shooters register and memorize, in codes, the solutions of the previously accomplished actions (Korh, A.Y. 1987). Therefore, we believe that the explanation, of various results received in tests (OE) an (CE), is found in the subconscious program that controls and regulates the shooter’s posture. This largely influences the relation between under-foot pressure distribution in zones. Under these circumstances, the process of adaptation can not be used as an explanation of found under-foot pressure, since it has a situational short-term effect and is a result of conductance of a certain memorized program. Therefore, depending on the goal, the results of the surface contact will be, in some cases, a reflection of subconscious contents, while in the others it will function on the conscious level – information will be received through the sense of sight.

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FEMALE FOOT MORPHOLOGY – IMPLICATIONS FOR LAST DESIGN

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INTRODUCTION

Knowledge of foot morphology is important for correct shoe fitting which is one of the most important buying criteria for the majority of runners (Kleindienst, 2003). The silhouette of the shoe including upper and midsole mainly depends on last contour. Therefore proper fit is achieved if shoe shape corresponds to foot shape (Janisse, 1992). Gender specific differences in foot shape are described in several papers (e. g. Wunderlich et al., 2001; Manna et al., 2001). In the mentioned studies, absolute or relative foot measures were averaged across different shoe sizes or comparisons between genders were only made for one shoe size. The use of averaged relative or absolute values seems to be questionable, since foot proportions alter with foot size (Anil et al., 1997; Krauss, 2006). Comparisons of men’s and women’s feet of the same size (e. g. EU 39) show that on average men’s feet are higher and wider than women’s feet (Krauss et al., 2005). Running shoes for women are often manufactured with a men’s last which is only down-graded for the appropriate shoe sizes for women (Frey, 2000). This procedure seems to be deficient, since gender differences exist when comparing the same foot size. The present study evaluates female foot morphology with respect to different foot types in the given foot sizes, and compares these data with the men’s lasts used to manufacture both women’s and men’s running shoes.

METHODS

424 European women between the ages of 18 and 60 years were measured using a 3-dimensional (3D) foot scanner (Pedus®, Human Solutions Inc., Germany). Data were analyzed with a customized version of Scan Worx 2.8.5. SL 1 (Human Solutions Inc., Germany) and different foot types were classified using a cluster analysis. Medial metatarsophalangeal joint (MPJ) length, lateral MPJ length, ball width, heel width, toe height and dorsum height at 50% foot length (FL) were incorporated into the analysis. Four running shoe lasts (US Women 6, 7, 8.5 and 9.5) used for the manufacturing of both men’s and women’s running shoes were measured with the Infoot 3D Foot Scanner (AIST, Japan). According to the mentioned foot measures, the same 6 variables were quantified on the lasts using Rhinoceros® software. Last and foot quantities were compared to the different foot types represented in cluster 1-3, and the mean values (grand mean, GM) in each shoe size.

RESULTS

![Figure 0: Cluster distribution, n=424.](image-url)
Three different foot types were detected: Cluster 1 includes wide and high feet with a mean MPJ length, Cluster 2 includes flat and rather narrow feet with long toes, and Cluster 3 incorporates narrow and rather flat feet with short toes. In US Women 4 to 6.5, more than 50% of women’s feet are in Cluster 1. Within size 7 to 8, all 3 clusters are present. Size 9 to 13 incorporates Clusters 2 and 3 (Figure 0).

Table 1: Grading of grand mean (Δ GM) and last (Δ L), corresponding difference (Diff)

<table>
<thead>
<tr>
<th>Variable</th>
<th>Women US 6 - 9.5</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Δ GM</td>
</tr>
<tr>
<td>Medial MPJ length [mm]</td>
<td>21.2</td>
</tr>
<tr>
<td>Lateral MPJ length [mm]</td>
<td>21.8</td>
</tr>
<tr>
<td>Ball width [mm]</td>
<td>3.1</td>
</tr>
<tr>
<td>Heel width [mm]</td>
<td>2.5</td>
</tr>
</tbody>
</table>

Table 1 shows that the grading of the lasts (Δ L) is appropriate for MPJ length measures, with differences of less than 1 mm (Δ GM, Diff). Ball and heel width show a discrepancy of 3.5 to 5.9 mm (Diff) between grading of the last and the increase of width and height measures in real feet. Figure 0 demonstrates that differences between last and grand mean increase with foot size. Agreement between foot and last measures only exists when comparing Cluster 1 with the last.

**DISCUSSION AND CONCLUSION**

Considering different foot types seems to be necessary for last design, since foot proportions are related to foot size (Anil et al., 1997). Discrepancies between lasts and feet are most obvious in very small and large sizes. Results of the present study indicate that the use of a men’s last for a women’s shoe is not appropriate, especially with regard to the heel and ball widths. The practice of using a men’s last for a women’s shoe has already been questioned by Frey (2000), and therefore a revised grading system is recommended for a better shoe fit in female running shoes. Future research should focus on the edges of the size range to get a better insight in female foot morphology, and improve lasts for the entire range of feet/shoe sizes instead of just the common foot proportions.

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**ACKNOWLEDGEMENTS**

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STUDY OF THE INFLUENCE OF FITTING AND WALKING CONDITION IN FOOT DORSAL PRESSURE
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INTRODUCTION
Footwear fitting is one of the most important aspects in footwear comfort, and also one of the most important considerations for users when purchasing new footwear (Kos and Duhovnik, 2002). Several aspects of the footwear-foot interaction influence the fitting perceived by the user. Among these aspects, pressure provided by the upper to the feet is one of the least studied.

Two references were found in the bibliography analyzing foot dorsal pressure (Jordan and Bartlett, 1995; Jordan et al., 1997). On one hand, in the study developed by Jordan and Barlett (1995), comparisons between two specific footwear types, an oxford laced shoe and a slip-on shoe, were analyzed. Dorsal pressure was collected for two different activities: bipedestation and walking. They found statistically significant differences between the footwear types and conclude that the type of shoe worn and the activity being performed can influence the magnitude of dorsal pressures. On the other, Jordan et al. (1997) performed the analysis with ten different footwear models, classified in two groups: 5 models considered as comfortable shoes and 5 as uncomfortable shoes. Pressure was measured on two specific dorsal areas, the flex line and the lace area, while subjects walked on a straight pace. Reported results included a significantly greater maximum force in the uncomfortable group, and a smaller contact area for the comfortable group. They concluded the necessity of the researchers to focus in specific regions of the shoe when measuring comfort.

The aim of this study was to analyze the pressure distribution provided by mountain boots on the dorsal side of the foot. The three influencing aspects suggested in the bibliography were considered: type of shoe worn, type of activity being developed and specific areas on the foot dorsum.

METHODS
Four subjects (aged 27 ±1 years) without foot pathologies performed the test walking following a self-selected velocity. Pressure between the upper and the foot dorsum was collected using a device with an array of 8 pressure sensors (TEKSCAN Flexiforce®) able to gather both static and dynamic pressures, connected to a computer through wireless technology. Sensors were placed on six selected relevant anatomical points of the subjects’ right foot dorsum (figure 2): the external face of the first metatarsal head (ap1), external face of the fifth metatarsal head (ap2), the third metatarsal head (ap3), the external face of the proximal phalange head (ap4), most prominent point of the heel near the Achilles insertion (ap5), most prominent point of the instep (ap6), the distal point of the talotibial articulation (ap7) and apophysis of the fifth metatarsus (ap8).

Three fitting conditions and five walking conditions were considered. To induce the fitting conditions, three pairs of commercially available boots only differing in the girth at the metatarsophalangeal joint line were used: a narrow boot (fc1), a medium width boot (fc2) and a wide boot (fc3). The walking conditions were induced by changing the relative inclination of the path respect the user walking direction: walking on flat surface (wc1), walking transversally to a 10 deg. inclined surface with the measured foot in the lower part (wc2), walking transversally to a 10 deg. inclined surface with the measured foot in the upper part (wc3) and, finally, walking upwards (wc4) and...
downwards (wc5) on a 10 deg. inclined surface. All the subjects performed at least two valid trials of three steps with each boot on each condition.

Maximum peak pressure was extracted for each measure and analysis of variance tests and Bonferroni correction multiple-comparison tests were used to determine significant differences between fitting and walking conditions on each anatomical point.

**RESULTS**

The obtained interval of maximum peak pressures was [6.03, 193.83] (kPa), which is coherent with the bibliography (figure 1). On one hand, considering walking condition influence, significant differences (p<0.05) were found in the maximum peak pressure between wc1 and the other walking conditions in five anatomical points (ap2, ap4, ap5, ap6, ap8). On the other hand, significant differences were obtained among the boots for the anatomical points situated on the metatarsophalangeal joint sides (ap1, ap2, ap4). In general, the maximum peak pressure was significantly greater (p<0.05) with the narrow boot (fc1) than with the wide boot (fc3) for different walking conditions (wc1, wc2, wc4, wc5). In addition, maximum peak pressure was significantly greater (p<0.07) with the narrow boot (fc1) than with medium width boot (fc2) when walking downwards (wc5).

![Figure 1. Mean maximum peak pressures obtained in each anatomical point for the three boots.](image1)

![Figure 2. Situation of the anatomical points](image2)

**DISCUSSION & CONCLUSIONS**

Results reported in this study support the thesis that dorsal pressure caused by footwear directly depends on the fit between foot and footwear in specific anatomical points, but also on the activity being developed. In that sense, results suggest that discrimination among different footwear designs is possible through the study of the pressures caused by the footwear upper. This study also supports the idea of a dynamical nature of the interaction between footwear and foot, emphasizing the need of dynamical tools to deeply characterize such interaction.

Relevant information for Footwear design as well as for Biomechanical research can be extracted from the analysis of the dorsal pressures of the foot provided by footwear.

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THE INFLUENCE OF BODY POSITION AND PHYSICAL ACTIVITY ON FOOT DIMENSIONS MEASURED BY A FOOT SCANNING SYSTEM

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INTRODUCTION
The influence of physical activity on foot volume was examined in several studies. It was shown that walking and running lead to an increase of foot volume (Cloughly et al., 1995; McWorther et al. 2003). McWorther et al. (2003) showed a highly significant foot volume increase of 2 % after 10 min of walking and of 3 % after 10 min of running. In contrast to this, Chalk et al. (1995) did not find significant changes in foot volume before and after intensive volleyball practice of female players. Nevertheless, it may be concluded that physical activity leads to an increase of foot volume. Specifically, with respect to footwear, McWorther et al. (2003) showed that foot volume increase was higher for those subjects wearing relatively bigger shoe sizes compared to their foot sizes. Therefore, changes of foot volume due to physical activity need to be considered in footwear design as it might effect shoe comfort, shoe fit and other functional aspects. Modifications of the functionality of footwear may occur due to different tightness of the foot shoe unit. Especially, different foot shoe unit conditions may affect foot sensitivity and change the athlete’s perception of footwear performance.

In the literature, foot volume is usually determined by volumeter devices measuring the amount of water displaced by the foot. Today, foot dimensions may be measured by optical laser scanning systems. With respect to physiological as well as to methodological questions, the goal of this investigation was to examine the influence of (a) body position and (b) running exercise on foot dimensions measured by a foot scanning system.

METHODS
The foot dimensions foot length (FL), foot breadth (FB), heel breadth (HB), ball girth circumference (BGC) and instep circumference (IC) were measured by the Infoot 3D Foot Scanning System with data accuracy of 1 mm (I-Ware Laboratory Co. Ltd, Osaka, Japan). Foot volume was calculated by use of the data processing software Rhinoceros 3.0 (McNeel, Seattle, WA, USA). For each testing condition in study (a) and (b) three foot scans were performed and averaged for statistical analysis.

In study (a) 20 subjects (male = 10, age: 23.5 ± 2.5 years, weight: 75.1 ± 6.6 kg, height: 179.6 ± 6.5 cm; female = 10, age: 21.1 ± 2.3 years, weight: 58.6 ± 4.8 kg, height: 166.4 ± 3.2 cm) were recruited. Measurements were performed after 15 min of supine rest, after 10, 20 and 30 min of upright standing and finally after another 15 min of supine rest. For statistical analysis an ANOVA was performed to identify changes in foot dimensions according to time of measurement for both gender groups.

In study (b) 23 male subjects (age: 25.8 ± 4.1 years, weight: 74.3 ± 8.5 kg, height: 179.3 ± 6.8 cm) were included. They were divided into two groups based on their sports activity level, resulting in group HA (high activity level, n = 13) and group LA (low activity level, n = 10). HA was defined by weekly exercise of more than 2 hours. Measurements were performed after 15 min of supine rest (pre test) and after 30 min of running (post test). Running intensity was set to a heart rate of 140-150 bpm and/or a perceived exertion level of 13 (a little hard) according to the Borg scale. Paired and unpaired t-tests were used to identify changes in foot volume before and after exercise as well as between HA and LA groups.

RESULTS AND DISCUSSION
The Infoot 3D Foot Scanning System in combination with the Rhinoceros 3.0 software proved to be suitable to obtain foot volume data.
For study (a) the statistical analysis showed that upright standing after supine rest results in highly significant increases in foot volume whereas turning back to supine rest later on results in a decrease of foot volume (n=20, p<0.01). Figure 1 demonstrates that, when splitting subjects into gender groups, foot volume changes are statistically significant for the female group only. For both gender groups interindividual variability is considerably high.

In study (b) results show significantly increased foot volumes after running exercise for the LA and HA group (Figure 2). Average normalized volume increase of the foot was 2.3% (n=23). No statistical differences were observed in the amount of volume increase between the HA and LA group.

<table>
<thead>
<tr>
<th></th>
<th>pre</th>
<th>post</th>
<th>increase</th>
</tr>
</thead>
<tbody>
<tr>
<td>FL</td>
<td>200.0</td>
<td>206.0</td>
<td>0.4</td>
</tr>
<tr>
<td>FB</td>
<td>100.0</td>
<td>102.5</td>
<td>0.6</td>
</tr>
<tr>
<td>HD</td>
<td>65.0</td>
<td>65.8</td>
<td>0.9</td>
</tr>
<tr>
<td>SGC</td>
<td>251.1</td>
<td>252.6</td>
<td>1.2</td>
</tr>
<tr>
<td>IC</td>
<td>248.7</td>
<td>247.1</td>
<td>0.6</td>
</tr>
</tbody>
</table>

Table 1: Foot Dimensions – Pre and Post Exercise

Table 1 summarizes the average changes of selected foot dimensions due to running exercise (n=23). However, foot dimension changes in most cases remain within measurement accuracy specifications of the scanning system.

CONCLUSION

The presented findings of an increase of foot volume due to an upright body position and exercise are in agreement with the majority of literature results. With regard to footwear, consequences of these findings need to be considered during the design process. Especially, human perception of footwear properties may be affected by a tighter foot shoe unit due to an increase of foot volume.

REFERENCES


ACKNOWLEDGEMENTS

This research was supported by Puma Inc., Germany.
A 3D FINITE ELEMENT SIMULATION OF FOOT-SHOE INTERFACE

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INTRODUCTION
The computational approach such as the finite element (FE) method provides an efficient and objective approach to quantify the biomechanical effects of a single or a combination of design factors of footwear without the prerequisite of fabricated footwear and replicating patient trials. Three-dimensional FE analysis of the foot and footwear in the literature has been restricted to the simulation of static postures or isolated instances of the stance phases of gait (Actis et al., 2006; Chen et al., 2003; Cheung and Zhang, 2005; Cheung et al., 2005). In this study, a three-dimensional (3D) FE quasi-static simulation of the foot-shoe interface was established to study the biomechanical effect of custom-molded orthotics on normal and flat-arched foot structures during the stance phases of gait.

METHODS
A shoe model was incorporated into a previously developed 3D FE foot model (Cheung et al., 2005), consisting of 28 bones, 72 ligaments and the plantar fascia embedded in a volume of bulk soft tissue (Figure 1). The assembled shoe structure consisted of an uppershoe cover, insole, midsole and outsole layers. The encapsulated soft tissue and shoe soles were defined as hyperelastic while the foot bones, ligaments, ground support and uppershoe shell were assumed as linearly elastic. Contact interactions among the major joints, foot-insole and shoe-ground interfaces were defined. Nine extrinsic musculotendon forces defined by contraction forces via axial connector elements, were estimated from normalized electromyographic (EMG) data and physiological cross-sectional area assuming a linear EMG-force relationship. The ground reaction force was applied at the inferior ground support. The inclination of the ground support relative to the shank was prescribed while the superior surfaces of the ankle joint were fixed throughout the stance phases. The biomechanical response of flat-arched foot structure, which was defined by reducing 50% of the elastic modulus of the ligaments and the plantar fascia, was compared to the normal condition.

RESULTS
Figure 2 depicts the deformation of the foot soft tissue and bones and plantar pressure distribution during simulated heelstrike, midstance and pushoff. The FE model predicted the highest plantar pressure at the second metatarsal head (0.47 MPa) during push-off and at the heel (0.21 MPa) during heelstrike. Highest
von Mises stress of foot bones was predicted at the posterior subtalar joint (30.5 MPa) followed by the ankle joint (26.4 MPa) and mid-shaft of the third metatarsal (15 MPa) during pushoff. Comparing to the normal condition, a flat-arched foot structure resulted in peak plantar pressure reduction. However, the magnitude of pressure reduction was minimal especially during heelstrike and pushoff. A flat-arched foot structure sustained a higher metatarsal and rearfoot joint von Mises stress during pushoff.

![Image](image.png)

Figure 2. Deformed mesh plot and predicted plantar pressure distribution of the foot-shoe interface during simulated heelstrike, midstance and pushoff.

**DISCUSSION**

The developed FE approach allows efficient evaluation of the parametrical effects of the shape and material properties of an assembled shoe structures on the stress distribution of the plantar foot, joints and ligaments during the stance phases of gait.

**REFERENCES**


**ACKNOWLEDGEMENTS**

This work was supported by the Croucher Foundation Fellowship of Hong Kong.
A 3D FINITE ELEMENT ANALYSIS OF HUMAN FOOT WITH HIGH-HEELED SHOE

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Email: ming.zhang@polyu.edu.hk

INTRODUCTION
Wearing high-heeled shoes increase peak pressure in the forefoot area, which may cause discomfort. Meanwhile, temporary reduction of symptoms from plantar fasciitis could be achieved via high-heeled shoes. The understanding of the effect of high-heeled shoes, especially how heel elevation contributes to foot biomechanics are still far from complete. Computational modeling based on finite element (FE) method provides opportunities for parametric study of the designs efficiently. We have developed a geometrically accurate three-dimensional FE foot model for male (Cheung, 2005). In this study, a FE model for a female foot was developed to study the effect of high-heeled support and other design parameters on the foot weight-bearing biomechanics. Experiments have been conducted for the model validations.

METHODS
The geometry of the FE model was obtained from 3D reconstruction of magnetic resonance images (MRI) from the right foot of a normal young female. Coronal MR images with 1 mm intervals in the neutral unloaded position were segmented to obtain the boundaries of the skeleton and skin surfaces using MIMICS v9. A solid model of 2 inches high-heeled support was built according to the foot size in computer-aided design software, CATIAv5.

The solid models of foot bones encapsulated with foot soft tissues and high-heeled support were then imported and assembled in the FE package ABAQUS v6.6. The FE model, as shown in figure 1, consisted of 78 ligaments, 28 bony segments, including the distal segments of the tibia and fibula, 26 foot bones and a high-heeled support.

The interactions among bones were defined as frictionless contact surfaces, which allow relative articulating movement. The articulation of joints was constrained by ligaments. The encapsulated soft tissue was defined as hyperelastic while the rest of the bony, ligamentous and foot support were idealized as homogeneous, isotropic, and linearly elastic. Ligaments and fascia were defined as tension-only truss elements. A balanced standing foot position was simulated by applying a vertical half body weight force underneath the high-heeled support and Achilles tendon forces at the insertion of the posterior calcaneus. The superior surface of the soft tissue, distal tibia, and fibula was fixed throughout the analysis to serve as the boundary conditions.

A series of controlled experiments was performed in order to validate the computational model.
Three pairs of foot supports (Figure 2) which simulated the profiles of standard commercial lady’s shoe lasts with different heel heights (0, 2, 4 inches) were custom manufactured. The profile of the support with 2-inch heel height was same as the one used in the FE model. The plantar foot pressures were evaluated for different heel height under different weight-bearing conditions for 24 female participants using F-Scan system (Tekscan Inc).

![Experiment Setup](image1)

**Figure 2. The experiment setup (a balanced standing)**

### PRELIMINARY RESULTS AND CONCLUSION

The developed female foot model provided estimation of plantar pressure and internal stress distribution of the bony and soft tissue structures. In the standing position, forefoot plantar region experienced increased contact pressure with high-heeled support. A pronounced increase in von Mises stress at the first metatarsophalangeal (MP) joint (2.37 MPa) and a von Mises stress concentration effect at the posterior talus (37.1 MPa) were predicted compared with level standing. The von Mises stress at dorsal soft tissue round MP joint also increased, as shown in figure 3. The stress on plantar fascia between metatarsal and calcaneus decreased while plantar fascia within MP joint increased significantly compared with level standing. The work is on-going to validate the model and parametrically evaluate the shoe designs.

![Stress Distributions](image2)

**Figure 3. FE predicated von Mises stress distributions on foot: (a) level standing; (b) with high-heeled support; (c) bulk soft tissue under 2-inch high-heeled support**

### REFERENCES


### ACKNOWLEDGEMENTS

This work was supported by research grant council of Hong Kong (Project No: PolyU 5249/04E, PolyU5317/05E).
GROUND REACTION FORCES AND TRACTION IN GOLF SHOES

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INTRODUCTION

A powerful and accurate golf shot depends on a complex series of muscle activations, movements and forces occurring, and necessarily includes a need for a stable connection between the shoe and surface, allowing the feet to exert forces on the ground without slipping. Studies of ground reaction forces (GRF) during the golf swing show how shear and vertical forces change during the golf swing (Hume et al. 2005, Williams & Cavanagh 1982, Williams & Sih, 1998, 2004, see Figure 1). When center of pressure (CP) information is combined with shear force data, the direction and magnitude of forces tending to cause the feet to slip can be identified (Williams & Sih, 1998, 2004, see Figure 2). GRF information was used to identify conditions where slip was most likely to occur during the golf swing, and to design a traction test for golf shoes.

Figure 1. Example vertical and shear forces during the golf swing for left and right feet. A slip coefficient, derived from the ratio of shear to vertical force, is also shown.

METHODS

A Kistler force platform measured horizontal (Fx) and vertical (Fz) GRF’s while a 250N vertical load was applied to the center of the foot and a pendulum mechanism applied a horizontal force to shoes. The horizontal force pulled backward on the heel of the right shoe or forward on the toe of the left shoes, approximating the direction of shear forces in the middle of the downswing. The pendulum had 250N in weights on its distal end, and was dropped from a 45° angle to catch on a cable/spring system attached to a last inside the shoe. The rate of rise of the horizontal force approximated loading conditions during the downswing. The shoe rested on a surface of grass (Fescue, left and right shoes tested) or artificial grass-like surface (right shoes only tested) mounted on a box that was attached to the top the force platform.

Eleven golf shoes from a variety of manufacturers were tested for traction, including a shoe with 8mm metal spikes (SP) and a flat bottom surface (FL) to establish a range of maximum and minimum slip coefficients. Four of the shoes had removable plastic “soft” spikes (RS1-RS4), and five had molded outsoles (MO1-MO5). Each shoe was tested ten times and the slip coefficient (Cs=Fy/Fz, see Figure 3)

Figure 2. Example CP and shear force patterns.
averaged across trials. Five tests were done on a given piece of grass, with a different location on each grass sample used for each test.

**RESULTS**

Table 1 shows the absolute (Cs) and relative (Cs(rel)) values for each shoe in each test condition.

\[
Cs(rel)(i) = [Cs(i)-Cs(FL)]/[Cs(SP)-Cs(FL)]
\]

The correlation between slip coefficients for tests of left and right shoes (omitting the flat shoe) was significant for a one-tailed \( p=0.05 \) level of significance (\( r = 0.64 \)). The flat shoe scores were omitted because its very low slip coefficient inappropriately inflated the correlation to 0.97. The correlation of the slip coefficients for right shoes on grass compared to the artificial surface was very low (\( r=0.15 \)). Left shoe slip coefficients for eight of the shoes were also compared with a set of similar tests done three months earlier on the same shoes, with the correlation between the test dates (\( r=0.83 \)) indicating good test-to-test reliability.

<table>
<thead>
<tr>
<th>Shoe</th>
<th>Left Shoe Turf</th>
<th>Right Shoe Turf</th>
<th>Right Shoe Artificial Surface</th>
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<tr>
<td></td>
<td>Cs</td>
<td>Cs(rel)</td>
<td>Cs</td>
</tr>
<tr>
<td>SP</td>
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<td>SP</td>
</tr>
<tr>
<td>RS1</td>
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<tr>
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<td>RS2</td>
</tr>
<tr>
<td>RS3</td>
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<td>RS3</td>
</tr>
<tr>
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<tr>
<td>FL</td>
<td>0.83</td>
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</tr>
</tbody>
</table>

**DISCUSSION**

All of the golf shoes tested provided good traction compared to a flat soled shoe, and several of the removable spike shoes and one of the molded outsole shoes gave slip coefficients similar to that for a metal-spiked shoe. Generally, the shoes with removable spikes did better on the tests on grass than did the molded outsole shoes, with the exception of MO1 that was comparable to the removable spike shoes. The molded outsole shoes showed slightly better slip coefficients on the artificial surface, though slip coefficients showed less variability among shoes on that surface. The poor correlation between slip coefficients on grass and the artificial surface indicates that traction testing on an artificial surface may not be a good indicator of traction on grass.

**REFERENCES**


INTRODUCTION

Previous authors have developed traction test devices with the aim of replicating the boundary conditions at the studded shoe-surface interface during movements in which the athlete is at risk of suffering injury or losing performance (Barry and Milburn, 2000; Cawley et al., 2003). While subject tests can provide loading and movement patterns which are more representative of key movements during competition, mechanical test devices are repeatable, thus allowing the influence of different shoes and surfaces to be more clearly understood. However, replicating characteristic player loading conditions and movements is vital in order to note meaningful trends. For example, Nigg (1990) showed that a particular surface provided the highest traction forces when tested at a low normal force but gave the lowest traction when a higher, more representative, normal force was applied. Attention should be drawn to the loading conditions at the time within a movement when the player is at the highest risk of injury or loss in performance. The parameter obtained from mechanical tests which is used describe traction must also be carefully selected in order to be related to injury risk or performance for each particular movement. This study contains analysis of a forefoot push-off (performance) movement in which the player seeks sufficient traction in order to avoid slipping relative to the surface.

METHOD

Five subjects carried out a variety of football (soccer) specific movements 10 each times on third-generation artificial turf mounted on a piezoelectric force-platform (Kistler 9281CA). This was synchronized to a Phantom (v4.2) high-speed camera, operating at 1000 Hz, used to capture the velocity and orientation of the shoe at impact and also the orientation of the shoe with respect to the surface throughout contact with the ground. A forefoot push-off from jogging to full sprint was used to represent a performance movement, where the player seeks sufficient traction so that the foot does not slip relative to the surface. A player’s movement pattern is known to depend on the mechanical properties of the surface (Denoth et al., 1985) and also can be influenced by the laboratory environment. Therefore, high-speed video observations of Academy footballers at a Premiership club were carried out on both an artificial and a natural training pitch. The players carried out the forefoot push-off movement as part of a regular training session thus providing minimum disruption to the players and increased the likelihood that the movements captured were close to those they would carry out in a game scenario.

RESULTS

Fig. 1 shows typical ground reaction force curves for the forefoot push-off together with a plot of the horizontal-vertical force ratio during the subject’s contact with the surface. The time to the peak horizontal force occurred on average at 69% of the total contact time (S.E. 0.36%). The normal force at this time was 1.80 BW (S.E. 0.03 BW). The peak horizontal-vertical force occurred on average 97% through contact (S.E. 0.55%) when the normal force was 0.10 BW (S.E. 0.02 BW). The timings to these peak values were highly consistent between repeated trials and subjects. Fig. 1c. shows the variation in the angle between the heel segment of the shoe and the ground. This angle was approximately 40° at the time of the peak horizontal force and nearly 90° at the time of the peak horizontal-vertical force.
DISCUSSION

When carrying out a mechanical test which replicates a forefoot push-off movement the boundary conditions applied should be similar to those suffered by a player at the time in the movement when they are at the highest risk of losing performance. It is known that the force required to cause failure of a surface increases with normal force (Nigg, 1990). Following this argument the time of the highest value of horizontal-vertical force could be believed to be the most crucial moment in the movement. However, due to the fact that the normal force can influence traction in a non-linear manner (Cawley, et al., 2003), together with the fact that the peak value of horizontal-vertical force occurs at the end of the movement when the vertical force is very low (0.1 BW), it can be deduced that the time of the peak horizontal force is more crucial. At this time it would be recommended to use a normal force of approximately 1.8 BW and align the shoe heel segment at 40° to the surface.

The (active) horizontal force exerted by the player was positive throughout impact, reaching a single maximum and decreasing. Analysis of the motion of the shoe relative to the surface during contact showed that the shoe scarcely moved in the anterior-posterior direction after the initial impact. If the player were to slip relative to the surface, this would be perceived after a small anterior posterior movement. Previous authors have applied a velocity to the shoe and measured the peak traction force which can often occur at displacements of 100 mm or more. It is recommended that more relevant traction quantities can be obtained at the time when the shoe first starts to move, for example the value of the initial stiffness. It would also be recommended that force-controlled loading would be more representative of the loading experienced at the shoe-surface interface under the action of the player.

Similar methodology can be applied to mechanical tests which simulate movements associated with high injury risk, such as a 180° turn. In such cases the timings to the highest knee and ankle moments must also be considering requiring the use of inverse dynamics.

REFERENCES

MODELIZATION OF THE FOOT THERMAL RESPONSE FOR A WIDE RANGE OF FOOTWEAR PROPERTIES AND CLIMATIC CONDITIONS

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INTRODUCTION
Thermal comfort is a feature increasingly demanded by users for outdoor footwear due to the characteristics of the environment and the activities performed. However, the design or selection of the optimal footwear for a specific activity and climate is still an unresolved problem. In this paper, it is presented the development and validation of a numeric thermal model that permits the prediction of the temperature and humidity of the shod foot for a wide range footwear properties and climatic conditions.

REVIEW AND THEORY
In the bibliography, there are only two specific studies where a numerical model of the foot (Lotens, 1989) and toes (Xu et al., 2005) are described. These models calculate the foot temperature as a function of the heat supply from the body and the heat loss to the environment. The Lotens’ model (1989) considers the following specifications: i) The foot is divided into two compartments: the skin with a control of the thermoregulatory blood flow and the core, thermally passive except by a small nutritional blood flow; ii) The function for the blood flow control is a linear combination of the central and mean skin body temperatures and the foot skin temperature; iii) The efficiency in the blood heat transport is 60% due to the counter-current heat exchange. These models have been validated for cold conditions; however, they do not include the evaporative heat loss and the sweating transmission, resulting inappropriate for high temperatures where most of the heat is lost due to the evaporation. In this sense, there is not any specific model for the prediction of the production of foot moisture, existing in the bibliography diverse models for general body areas (Fiala, 2001).

METHODS
The numeric thermal model of the foot has been based on the thermoregulation of the foot and in the heat and mass transfer equations consisting of three modules: 1) A model of the complete body which provides the information of the organism thermal state. It is based on the validated analytic models used in the standards for the evaluation of thermal risks in hot and cold environments (ISO 7933, 2004; ISO 11079, 1999); 2) A model of heat transfer, developed from the work carried out by Lotens (1989). It has been introduced the heat loss due to the evaporation of the sweating among other improvements; 3) A model of mass transfer, developed from the equations of Fiala (2001) for the production of sweating and the equations of mass balance between the foot and the environment, simplified by applying an efficiency coefficient of evaporation.

The model parameters have been fitted for the heat and mass transfer modules, applying genetic algorithms. The validation has been done by comparing the experimental data with the results of the simulation in two ways: calculating the correlation between both type of data at fixed time intervals and comparing qualitatively the whole temporal function.

Experiments with users have been carried out for the model fitting and validation. Five healthy men used to outdoor activities participated in the trials. Temperature and humidity were measured during the experiment between the first and second toes and on the longitudinal arch. Skin temperature on the foot plant and instep were also recorded. Tests were carried out outdoor in two localizations, in cold-dry and...
hot-humid conditions. The test was divided into three phases of activity and rest (Kurz et al., 2002). Twelve boots were manufactured for tests in cold-dry conditions and ten for hot-humid environment according to an experimental design in which diverse materials were used. Two boots from the market were included in the experiment. The global thermal isolation and water vapour resistance where calculated with a foot manikin test, these parameters were used as input for the model.

RESULTS
The linear fitting for the measured and experimental temperature in the final phases of the test varied from 57% to 68% (R² in %). The correlation for the measured and calculated humidity presents lower values with a R² from 30% to 62% for the final phases. When comparing the temporal curves for the foot temperature, the calculated curves follow the experimental ones with a standard deviation that falls within the experimental sample. The error in the mean temperature calculated is lower than 1ºC in most of the time interval (figure 1. a and b). In the case of the vapour pressure, the approximation calculated by the model is good for the cold-dry climate (figure 1. c); on the other hand, in the hot-humid condition the model is not capable to predict appropriately the high increase of humidity produced at the beginning of the activity phase (figure 1. d).

![Graphs of measured vs calculated foot temperature and vapour pressure](image)

**DISCUSSION AND CONCLUSIONS**
Comparing with the studies presented in the bibliography (Kuklane et al., 2000; Lotens, 1989), the model developed presents better results in the prediction of the temperature in cold conditions. On the other hand, this is the first foot model validated for hot environments and capable to predict the humidity in the foot/footwear microclimate. The lack of fitting in the predictions could be due to diverse reasons, standing out the variability in the thermo-physiologic response of subjects and the complexity of the mass transfer mechanisms.

REFERENCES
DEVELOPMENT OF A MULTI-DIMENSIONAL FORCE SENSOR FOR FIELD TESTING IN SNOWBOARDING

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INTRODUCTION

The lower body has been reported to be involved in 33 – 43% of all snowboarding injuries (Bladin et al., 2004). Of particular concern is the over-representation of Lateral Process of the Talus (LPT) fractures in snowboarders. In order to prevent trauma it is important to first understand the underlying mechanisms. Approximately 40% of all snowboarding injuries are sustained while performing aerial maneuvers. At present the loads applied to the lower body with jump landings have not been quantified. Such an analysis would benefit prevention of lower limb injuries and has been supported within the literature (Funk et al., 2003).

This work aimed at developing a testing procedure to allow the collection of accurate data in a real snowboarding environment. Specialized portable measuring equipment was designed and validated to allow for this approach to data collection. The finalized procedure and equipment was intended for use in future injury prevention studies.

METHODS

Force Sensor Development

A snowboard mounted force platform (SFP) was designed to fit beneath the bindings of any standard snowboard. Six unidirectional force transducers were fixed between two aluminium plates in an arrangement inspired by the Stewart Platform. The accuracy of the SFP was assessed under static and dynamic loading conditions. A Bertec force platform and an Instron machine were used as dynamic and static reference measures, respectively. The plate was tested under static (up to 5 kN) and dynamic conditions (up to 1.6 kN). Three trials were recorded for each loading condition. Error calculations were made for all force and moment components.

Collection of Preliminary On-Snow Data

On-snow kinematic and kinetic data were collected from two experienced snowboarders performing a series of jumps. A four camera SIMI motion system (Unterschleissheim, Germany) was used to collect video data at 120 Hz. A set of 23 black markers were fixed bilaterally onto white tights worn by the participants. The SFP was mounted beneath the lead foot and data were recorded by a data logger (biovision) at 960 Hz. Only the landing phase of jumping was analysed. Ankle joint kinematic data were calculated based on boot and leg markers. Ankle joint range of motion was calculated for all joint axes. Mean values of each measure were calculated for each participant individually. Inverse dynamics calculations were carried out using SIMI motion software.

RESULTS

Force Sensor Evaluation

The mean absolute error of the calculated vertical static load was 12 N (SD = 10.2). The maximum error seen for incremental loading was 37.3 N. The response to loading was found to be linear with a correlation between the applied and calculated load of r = 0.9997, p < 0.001. In Table 1 dynamic loading results are listed.

Table 1 RMS error results for all GRF output components of the SFP under dynamic loading conditions

<table>
<thead>
<tr>
<th>RMS Error (N or Nm)</th>
<th>Force Component</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Fx</td>
</tr>
<tr>
<td>mean</td>
<td>13.4</td>
</tr>
<tr>
<td>SD</td>
<td>2.3</td>
</tr>
<tr>
<td>max</td>
<td>15.9</td>
</tr>
</tbody>
</table>
The magnitude of error differed between the measurement axes of the snowboard force plate. During dynamic testing the Fx, Fy, Fz and Mx components matched very closely in both magnitude and timing.

**On Snow Testing**

Riders landed with their front foot in an abducted position. Following touch-down, both underwent further abduction of the ankle joint with subject two exhibiting a greater range of motion. Average abduction ROM values were M = 6° (SD = 2.9) and M = 13° (SD = 1.4). Both participants landed with inverted ankle joints. However, subject one decreased inversion while subject two increased inversion. The average peak ankle joint compressive load was -2776 N (+/- 845.9) and -2342 N (+/- 724.2) for subjects one and two respectively (Fig. 1). Expressed as a proportion of body weight these mean compressive loads were 3.77 BW and 3.51 BW. The greatest loading in the shear directions was observed along the x axis of the ankle joint, with peaks of approximately 1.68 BW (+/- 0.29) and 1.88 BW (+/- 0.93) respectively. Mean peak y axis loads of -1.33 BW (+/- 0.57) and -1.06 BW (+/- 0.10) were calculated for subject one and two, respectively.

**DISCUSSION**

Expressed as a percentage of applied load, the dynamic measurement errors of 5.4%, 5.8% and 2.2% of applied load for the x, y and z force components respectively which compares well to others (Bally & Taverney, 1996). In summary, the dynamic load response of the SFP was satisfactory.

Cadaver experiments revealed fractures for nine out of ten ankle specimens at 10° ankle inversion when coupled with forced dorsiflexion and axial loads of 2200 to 8900 N (Boon et al., 2001). Calculated loads from the present study lie within this loading range and, therefore, indicate a high injury risk.

**CONCLUSIONS**

The technique proposed and evaluated will allow further understanding of injury mechanisms and the evaluation of interventions in snowboarding. Currently there is no published research of this kind. These data provide a basis to be built upon by future biomechanics research.

**REFERENCES**


NEW RUNNING SHOE SOLE STIFFNESS PREDICTION METHOD  
BY USING AN EIGENVIBRATION ANALYSIS

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ASICS Corporation

INTRODUCTION
In order to produce the smooth guidance racing shoe sole needs enough stiffness or rather stiffness distribution under the bending and torsional deformations. Exactly speaking, the stiffness must be evaluated under the condition corresponding to the practical usage environment. However the environment, which means boundary and loading conditions cannot be identified. The establishment of the sole stiffness evaluation method considered the environment is an important key to the designing process of racing shoes for high speed running. In this paper, the sole stiffness evaluation method based on the eigenvibration analysis is proposed for the effective designing of racing shoes.

SOLE DEFORMATION
Figure 1 shows the sole deformation angle changes calculated by the motion capture system. Horizontal axis is a contact time from heel contact to toe off. In this measurement, sole is divided into 3 portions, A1, A2 and A3, which are defined by each 3 reflective markers. The relationships between A1 and A2 and between A2 and A3 are denoted by \( \phi \) and \( \theta \), respectively. Here subscripts \( b \) and \( t \) indicate the bending and torsion, respectively. \( \phi_b \) corresponds to the forefoot bending angle. Figure 1(b) shows that forefoot torsional angle, \( \phi_t \) can be ignored. Therefore these 3 deformation angles, \( \phi_b \), \( \theta_b \) and \( \theta_t \) are important parameters for the designing racing shoes. It is evident that these parameters depend on sole stiffness distribution.

![Diagram](image)

(a) Changes in bending angles  
(b) Changes in Torsional angles

Fig.1 Sole deformation angle changes during contact phase.

SOLE STIFFNESS EVALUATION METHOD
Sole stiffness evaluated by the mechanical testing methods such as 3-point bending test and torsional test is not accurate for the prediction of the above deformation angles. The main reason is that the deformation mode in the mechanical evaluation is quite different with that under the running. In this study a new sole stiffness prediction method based on the finite element method is proposed. An eigenvibration analysis is widely used in various structural designing fields. The main purpose is to
avoid the resonance. I found that the eigenvibration is equivalent with stiffness corresponding to the deformation mode under the constant weight.

Figure 2 shows the analytical method and results for the prediction of $\phi b$. At first the foot pressure distribution is measured at the maximum value of $\phi b$ shown as a white circle in Fig. 1(a). The result is shown in Fig. 2(a). Fig. 2(b) shows the numerical model with mass density distribution, which is equivalent with Fig. 2(a). Figure 2(c) is an eigenvibration mode corresponding to a forefoot bending. This eigenfrequency means the sole stiffness under the forefoot bending.

![Foot scan result, mass distribution, and eigenvibration mode](image)

(a) Foot scan result  (b) Mass distribution  (c) Eigenvibration mode (1st bending)

**Fig. 2** Sole stiffness prediction method

In order to check the validity of the above prediction method, deformations of 4 typed shoes with different sole stiffness distributions were measured and compared with eigenfrequencies. Figure 3 shows the test shoe soles. These bold lines denote the groove, whose cross-section are 5mm × 5mm constant. Figure 4 shows the eigenfrequencies with eigenvibration modes obtained from 4 soles plotted against maximum deformation angles. Judging from Fig. 4, it was concluded that the eigenfrequencies calculated were effective for the prediction of sole deformation angles during contact phase.

![Test shoe soles with different grooves](image)

Fig. 3 Test shoe soles with different grooves

![Graphs of foot pressure](image)

(a) Midfoot bending  (a) Midfoot torsion  (a) Forefoot bending

**Fig. 4** The relationship between maximum deformation angles and eigenfrequencies.
NEW DESIGNS IN CHILDREN’S FOOTWEAR BASED ON A FOOT TYPE CLASSIFICATION

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INTRODUCTION

Several studies postulate that ill-fitting footwear can be the cause of foot problems and pathologies (Rao, 1992; Sachithanandam, 1995). A child’s foot does not complete growth until during or after adolescence, and is therefore at particular risk of developing problems through external factors. If footwear is to support the foot in its physiological function, the shape and dimension of the shoe’s internal space must correspond to the human foot (Hawes, 1994). To create well-fitting shoes that fit a wide range of children, it is necessary to know about their different foot shapes. However, the human foot shows biological variations in its shape, structure and function. Moreover, the anthropometrics of children’s feet differs from that of adults, and children’s shoes therefore need special standards in design. Few studies have investigated structural variations of the children’s foot regarding single foot measurements (e.g. Kouchi, 1998). Much of the clinical research has focused on the medial longitudinal arch (plantar measurements) due its vital role in proper foot function (Forriol, 1990). Unfortunately, current foot systems are based on a simple 2 dimensional categorisation regarding the foot length and forefoot width. Even though 3 dimensional (3D) measuring devices have become more available in recent years (Mochimaru & Kouchi, 2000), there is still a lack of information about 3D data of children’s feet and the use of statistical multivariate methods to classify similar feet into groups. Therefore, the purpose of this study was to detect different foot types in children and thereby provide a basis for their implementation in the design of children’s shoes.

METHODS

The feet of 2881 children (1446 \(\delta\), 1435 \(\varphi\)), aged 2 to 14, were measured throughout Germany in the years 2003-2004. Foot anthropometry was taken using a 3 dimensional foot scanner (Pedus\(\textsuperscript{\circledast}\), Human Solutions, Germany) with the children standing in a bipedal upright position. The scans were analyzed using ScanWorX 2.8.5SL software to determine 15 different (anatomical) foot measures. Analysis of correlation was used to detect five widely independent and uncorrelated measures in the forefoot, mid- and hindfoot: ball-of-foot length, ball width, ball angle, heel width, and instep height. A cluster analysis (Ward’s method) was applied to classify the feet into different types based on the z-values of these foot measures. The resulting clusters exhibit high internal (within-cluster) homogeneity and high external (between-cluster) heterogeneity (Hair, 1998).

RESULTS AND DISCUSSION

Three foot types were identified, characterized by their distinctive cluster centers as follows: Cluster (1) (31\%) represents a foot type with a wide heel width, and a high instep height, as well as a small ball-of-foot length and ball angle, which means a rather squared forefoot shape; the feet of cluster (2) (29\%) are characterized by a short ball-of-foot length and a large ball angle; whereas the feet of cluster (3) (40\%) show a slender shape with small values in ball and heel width.

The distribution of the three foot types with regard to the foot length of the children (Fig. 1) show a high proportion of cluster 1 in the smaller sizes (size 25-29), whereas cluster 3 is disproportionately high in the larger sizes (size 36-41); an equal mixture of all types is found between sizes 30-35. This necessitates a division into three sectors (size groups) to meet all requirements of the different
proportions of children’s feet (Fig. 1). For each sector, separate cluster analyses were re-calculated resulting in three independent foot types, respectively.

![Graph showing distribution of different foot types across shoe sizes](image)

Figure 1: Distribution of the different foot types with regard to age

**CONCLUSION**

The outcome can be directly applied to create specific designs of children’s shoes for different morphological foot types based on 3D measurements. This foot typing system allows for the multifaceted differences in the shape of children’s feet, and paves the way for an unimpaired development of children's feet. However, to allow for an optimal linkage between manufacturing and fitting the shoes, a foot measuring device in the shoe shops could be used which includes foot type data to insure that children receive the best fitting shoes possible. Furthermore, to continue improving the fit of children’s shoes, dynamical 3D scanners should be developed and applied to assure that shoes follows foot function.

In conclusion, further research should start with a prospective study focusing on the evaluation of this new shoe system, as well as on further factors influencing the development of the child's foot in its morphology and function.

**REFERENCES**


GROWTH CHANGES IN THE FOOT SHAPE OF JAPANESE

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INTRODUCTION
A 3-year longitudinal study was conducted to investigate the development of running motion and foot morphology. In this study, growth changes in 3D foot shape are analyzed using foot scan data from 110 Japanese children

SUBJECTS
A total of 115 children living in and around Tokyo participated in this study, and were divided into 4 age groups. Age at the start of study was 3±0.25, 5±0.25, 7±0.25, and 9±0.25 years for each group. The 4 groups are hereinafter referred to as age groups 3, 5, 7, and 9. An age range of 3 to 11 years was covered by the end of study. One-hundred and ten children participated in all measurements.

METHODS
Eight measurements, including height and weight, were taken using traditional methods. Right and left feet were scanned using an Infoot system (I-ware Laboratory Co., Ltd.) Development of the plantar arch was examined by observing these foot scans. A homologous model consisting of 295 data points was created for each of the right foot scans for 89 children (Figure 2). Models were normalized for foot length. Morphological distance between two feet was defined as the sum of Euclidean distances between corresponding data points of homologous models. Distance matrices were analyzed for each age group by multidimensional scaling (MDS) using Human Body Statistica (Digital Human Technology Inc.) (Mochimaru and Kouchi, 2000).

RESULTS
Figure 1 shows the percentage of flat sole at each age. Soles of 73% of children were flat at age 3. The percentage of flat sole markedly decreased from age group 3 to 5, but the changes in age groups 7 and 9 were minimal. Figure 1 suggests that the right foot develops somewhat earlier than the left foot.

A 4-dimensional solution was obtained for the MDS analysis conducted using 267 (3 × 89) models. RSQ was 0.95-0.97, that is, 4 MDS scales explain 95%-97% of the information contained in the distance matrix. Significance of between-age differences in MDS scores was tested by ANOVA for each age group separately (Table 1). Between-age differences were significant in 3 MDS scales (Dims-2, -3 and -4) in age group 3, but only 1 MDS scale showed significant between-age differences in age groups 5, 7 and 9. Figure 2 shows the comparison of virtual shapes at both ends of each MDS scale for age group 3 calculated by the method of Mochimaru and Kouchi (2000). During growth, the foot becomes less pronated, relatively narrower, and develops a relatively lower dorsal arch, a relatively higher sphyiron and sphyiron fibulare, and a more protruding heel. Shape changes in age groups 5, 7, 9 are such that the foot becomes relatively narrower and develops a relatively lower dorsal arch (Figure 3). It is interesting to note that the big toe in age group 7 becomes more valgus during the 2 years from age 5 (Figure 3, middle).

DISCUSSION AND CONCLUSION
Development of the plantar arch is very conspicuous, but when the 3D foot shape is considered, its variation is not very important. The most drastic shape change occurs in age group 3. This change may
be partly due to the short legs in small children. Feet must be 17-18 cm apart for scanning, which is a relatively large distance for small children with short legs. Shape changes in other age groups are similar to the allometric shape difference among adults. As the walking motion becomes more similar to adults (Sutherland et al., 1980), foot morphology also becomes more similar to adults.

REFERENCES

ACKNOWLEDGEMENTS
This study was supported by Nike Inc.

Table 1. Between-age difference of MDS scores

<table>
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<tr>
<th>Cohort</th>
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<th>Dim-2</th>
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<td>p&lt;0.001</td>
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<td>ns</td>
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<td>ns</td>
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</tr>
<tr>
<td>7-year</td>
<td>ns</td>
<td>ns</td>
<td>ns</td>
<td>p&lt;0.001</td>
</tr>
<tr>
<td>9-year</td>
<td>ns</td>
<td>ns</td>
<td>p=0.021</td>
<td>ns</td>
</tr>
</tbody>
</table>

ns: not significant

Figure 1. Decrease in flat foot with age.

Figure 2. Virtual shapes at mean +3 S.D. (black) and mean –3S.D. (gray) on each MDS scale.

*: inter-age difference was significant at the 5% level,
**: inter-age difference was significant at the 1% level.

Figure 3. Virtual shapes at mean +3 S.D. (black) and mean –3S.D. (gray) on each MDS scale.

*: inter-age difference was significant at the 5% level,
**: inter-age difference was significant at the 1% level.
LONGITUDINAL ANALYSIS OF KINEMATIC DEVELOPMENT FOR CHILDREN RUNNING

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INTRODUCTION

A 3-year longitudinal study was conducted to investigate the development of running motion and foot morphology. Cross-sectional analysis based on age-group statistics has been used in both cross-sectional (Sutherland, 1980) and longitudinal studies (Kimura, 2005). We proposed a method to analyze kinematic development patterns based on longitudinal data without age-group statistics.

SUBJECTS

A total of 115 children living in and around Tokyo participated in this study, and were divided into 4 cohorts. Age at the start of study was 3±0.25, 5±0.25, 7±0.25 and 9±0.25 years in each cohort. Data for 12 subjects (3 from each cohort) were used for kinematic analysis.

METHODS

Whole body movements were measured by a motion capture system (1st year: MAC FALCON 60 Hz; 2nd-3rd year: Vicon MX 200 Hz) using 30 surface makers (Figure 1). Subjects ran barefoot through a 10 [m] path. Running speed was controlled as subject height × 2 [m/sec] using a pacemaker. The ground reaction force was measured simultaneously with the motion. Five complete trials were measured for each subject. Angular displacement, joint moments and joint powers for flexion-extension of the ankle, knee and hip were calculated using DIFF software (developed and distributed by the Japan Clinical Gait Analysis Society). Inter-trial dissimilarity in kinematics was defined using flexion-extension angular displacements of 3 joints (right ankle, knee and hip) during the stance phase. Angular displacements were normalized against stance duration and 100 temporal samples were calculated for each trial. Kinematics can be described as the trajectory in 3-D angular space (Figure 2). Inter-trial dissimilarity was defined as the sum of Euclidean distances between 2 trajectories at each of 100 temporal samples. Inter-trial dissimilarities were also calculated for joint moment and joint power of the 3 joints using the same method.

RESULTS AND DISCUSSION

A distribution map based on inter-trial dissimilarities in kinematics was obtained by multidimensional scaling (Figure 3). The first axis contrasts the large and small flexions of knee and ankle (difference in amplitude of flexion angles), and the second axis contrasts the earlier and later flexions (difference in temporal flexion pattern). The developmental pattern of each child is described by 2 vectors in the distribution map: one is from the first to the second year, and the other is from the second to the third year. Individual variation in the developmental pattern was analyzed using 4 components of the 2 vectors by the cluster analysis. Twelve subjects were classified into 3 groups (Figure 4). In group A, the range of the flexion-extension angle of the knee and the ankle was larger in the first year, and earlier flexion of the knee and the ankle was observed the next year. On the other hand, in group B, earlier flexion was observed in the first year, and flexion angle was larger the next year. Group C exhibited a unique development process: the range of flexion angle increased in the first year but decreased in the second year. The developmental processes of the joint moment and power were...
analyzed in the same manner as with kinematic analysis. A large positive hip power was observed in children with mature and powerful running.

Five subjects from the 3-year cohort were divided into 2 groups based on development of the non-contact area of the sole obtained from 3-D foot morphological analysis, and members of these groups were compared with the groups based on kinematic/kinetic development. The relationship between development of the non-contact area of the sole and kinematics was unclear.

REFERENCES

ACKNOWLEDGEMENT
This study was supported by Nike Inc.
FOOT AND ARCH STRUCTURE CHARACTERISTICS ACROSS AGE GROUPS

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INTRODUCTION
Abnormal foot structures and mechanics are often associated with risks of lower extremity injuries. Clinical measurements such as arch index are often used in clinical and research settings to classify foot types. Arch and foot structures vary widely between people (Razeghi et al. 2002) and can be related to gender, age and side dominance. Information about foot and arch structures is available in the literature (Cavanagh et al. 1987; Williams et al. 2000), however, knowledge about foot structural changes related to age, gender and body side is still rare. Therefore, the purpose of this research project is to document foot structural changes across different age, gender and body side groups using a large sample size.

METHODS
A total of 300 subjects were recruited in the southern region in US. The subjects were between 18 and 70 years of age and grouped into eight age groups that generally match the national gender and age distribution of the 2000 US Census (http://www.census.gov/). Fifty percent of the subjects were men and the remaining 50% were women. All subjects were in good general health, mobile and agile enough to execute the study protocol. Clinical measurements of both feet including full and truncated foot lengths, foot widths at metatarsal heads, midfoot and heel, dorsum heights in sitting and bilateral standing, and navicular heights in sitting and standing were taken first with some of the measurements conducted on an AHIMS device (Arch Height Index Measurement System, JAK Tool and Model, LLC). Plantar pressure distributions of each foot in static and quasi-dynamic postures were measured using a plantar pressure mat system (Tekscan, Boston, MA) in a randomized order.

Arch index (AI) was determined based on dorsum height and the truncated foot length measurements (Williams et al. 2000). An arch index based on navicular height was also computed in a similar fashion. Arch stiffness (AS) was computed (Zifchock et al. 2006) and navicular drop (ND) was calculated as the difference between the navicular heights in its neutral and relaxed positions. Three one-way ANOVAs were conducted on selected clinical variables with respect to the age, gender and side groups and post hoc comparisons were performed using a pair-wise t-test in the ANOVAs across the age groups (p < 0.05).

RESULTS
The results showed a gender difference in most examined variables. While there were significant gender differences in dorsum heights (sitting & standing), truncated foot length and AI (Figure 1) were not significantly different between gender groups. Navicular drop and navicular AI also did not show significant gender differences.

The statistical results exhibited age differences in the metatarsal width, navicular neutral, AS and navicular drop. The post hoc comparisons showed a significant difference for metatarsal width between the 20-24 and 60-64 groups. For AS measurements, the 35-44 age group was significantly different from all other groups. Finally, the ND of the 20-24 group was significantly smaller than the ND of the 60-64 age group. The comparison between the left and right sides indicated that only the midfoot width was statistically significant.
DISCUSSION AND CONCLUSIONS
This study provided a complete database of common clinical measurements of foot and arch structures across a wide age range (18 – 70 years). It was interesting to note that the two arch indices did not demonstrate any significant changes although the dorsum height and truncated foot length were different between the males and females. The arch index reported here (0.32 – 0.34) are slightly higher than the arch index reported elsewhere (Williams et al. 2000). However, we did not find any significant age changes suggesting that the medial longitudinal arch structure does not seem to experience significant structural changes as one gets older. In addition, the clinical measurements on the left and right feet do not generally vary from each other. However, the results on the body side should be interpreted with cautions since we did not take side dominance into account.

ACKNOWLEDGEMENTS
Supported by Schering-Plough HealthCare Products Inc.

REFERENCES
WHAT ARE THE EFFECTS OF A HIGH HEEL-COLLAR AND A TREAD SOLE SHOE ON STOPPING OVER DRY AND SLIPPERY SURFACES IN OLDER PEOPLE?
Jasmine C. Menan1, 2, Hylton B. Menz3, Bridget J. Munro4, Julie R. Steele4 and Stephen R. Lord1, 2
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INTRODUCTION
Slips and trips contribute to about a half of the falls experienced by older people. Walking around in ill-fitting or inappropriate footwear (Sherrington and Menz, 2003), or without shoes (Menz et al., 2006) may predispose older adults to falls both in and outside of the home, especially when walking on rough or slippery surfaces. Despite the number of studies on the effects of footwear on balance and gait, it remains unclear as to which specific shoe features can be classified as safe or unsafe for older people when conducting daily living activities. A systematic approach whereby individual features of a standard shoe pertaining, for instance, to heel collar height or slip-resistance of the sole are modified one at the time would assist in this process. Unexpected stopping presents a postural stability challenge and has been used to assess dynamic balance in shoes of varying midsole hardness (Perry et al., 2007). The aim of this study was to determine the effects of high heel-collar and tread sole shoes on the ability of older people to stop rapidly on dry and slippery surfaces. It was hypothesized that a high heel-collar might reduce the time to achieve whole-body stability by enhancing leg propriocception, whereas a tread sole shoe might assist in reducing the distance required to stop due to its frictional properties, particularly in the case of a slippery surface.

METHODS
Fourteen community-dwelling older people (mean age: 79.1±3.9 yr; 8 females) walked at a self-selected speed over a dry and then a wet linoleum surface in three randomized footwear conditions: a standard shoe, a high heel-collar shoe and a tread sole shoe. While wearing a whole-body safety harness, subjects were required to stop as quickly as possible in 3 out of 8 walking trials in each condition, at the sound of a buzzer, which was randomly activated at right heel strike by depressing a pressure sensor placed in the shoe. Three-dimensional positions of infrared markers affixed to the shoe heel and toe-box, back of the pelvis, and back and front of the head were collected at 200 Hz using two CODA scanner units. A complete stop was assumed when the pelvis marker’s sagittal velocity decreased to less than 0.1 m.s⁻¹. Stopping distance, reaction time (between stopping stimulus and final heel contact), stabilization time (between final heel contact and complete stop) and total stopping time (reaction time + stabilization time) were calculated and normalized to the mean walking velocity of the walk-through trials. Step width at complete stop was also determined. For each condition, the stopping trial with the fastest reaction time was selected for further statistical analysis. Repeated measures ANOVA with simple contrasts were performed to identify any significant within-subject effects of surface or shoe sole hardness (p<0.05).

RESULTS
As anticipated the subjects walked significantly slower on the wet surface compared to the dry surface (F_{1,13}=6.693, p=0.023). There was a significant surface effect on stopping distance (F_{1,13}=9.523, p=0.009), and a trend for a surface effect on total stopping time (p<0.1). A significant surface x footwear interaction suggested that subjects adopted a longer total stopping time in the high heel-collar than in the standard shoe condition on the dry surface but that this effect was reversed on the wet surface
Trends for a surface x footwear interaction on stabilization time and step width between the high-collared shoe and the standard shoe were also noted ($p<0.1$). This indicates that stabilization time was greater in the high-heel collar shoe than the standard shoe on the dry surface but smaller on the wet, and that step width was greater on the wet surface in the high heel-collar shoe compared to the standard shoe. There was also a trend for a greater number of steps taken in the tread sole shoe condition compared to the standard shoe condition ($p<0.1$).

**DISCUSSION**

Despite adopting a slower walking velocity when walking over the slippery surface, the older subjects required more time and a greater distance to terminate gait and took longer to regain their whole-body stability after the stop. Although not significant, there was a reduction in total stopping time observed when subjects walked on the wet surface in the high heel-collar condition. This may be due to a decrease in stabilization time, possibly enabled by the subjects adopting a greater step width in this condition. Contrary to our hypothesis, stopping was not facilitated in the tread sole shoe on either surface, and subjects did indeed take more steps in this shoe condition compared to the standard shoe.

Previous research performed on a “slipping machine” mimicking heel strike has shown that the addition of a tread sole to an Oxford type shoe did not enable a safe dynamic coefficient of friction to be reached during ambulation when exposed to wet/slippery conditions (Menz et al., 2001). Thus, the various shoe conditions may have posed a similar postural threat to the older individuals. In addition, awareness of a potentially slippery surface has been shown to lead to significant proactive gait adjustments in young adults, such as a flatter foot-floor angle and reduced braking impulse (Marigold and Patla., 2002). Thus prior knowledge of the slippery surface combined with a high-level of stopping expectation and the freedom of walking at a self-selected speed may have reduced the challenging aspect of the task, and obscured differences between the shoe conditions. Furthermore, the modified shoes may have been too similar to the standard shoe to affect older people’s balance, although the modifications represent features found in regularly-worn, commercially available shoes, and therefore warrant investigation.

**REFERENCES**

SOFT TISSUE THICKNESS, PLANTAR SENSATION AND FOOT PAIN: WHAT IS THE EFFECT OF OBESITY ON THE OLDER FOOT?

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INTRODUCTION
The prevalence of overweight and obesity is rising at an alarming rate worldwide, such that a high percentage of the population is now classified as obese. The elderly are no exception with 15% of men and 28% of women aged over 60 years considered to be obese (Perissinotto et al., 2002). As the proportion of elderly people is rapidly increasing and the negative health consequences of obesity are numerous and well documented, it is imperative that effective interventions aimed at combating the obesity epidemic in the elderly are implemented. One strategy to reduce the effects of obesity, and potentially improving quality of life in the elderly, is the provision of safe and comfortable shoes to enable older people to comfortably participate in physical activity. However, before functional footwear can be designed for this population, information pertaining to their foot structure and function must be derived. Furthermore, although overweight and obesity has been shown to affect the structure and function of the feet of children (Mickle et al. 2006) and adults (Hills et al. 2001), few studies have investigated the elderly obese foot. Therefore, the purpose of this study was to determine the effects of obesity on the soft tissue thickness, plantar sensation and foot pain in older adults.

METHODS
Three-hundred and twelve older men and women aged between 60-90 years (154 women; age = 71.2 ± 7.7 yr; height = 1.66 ± 0.1 m; BMI = 28.5 ± 5.4 kg.m⁻²) were randomly recruited, using the Australian Electoral roll, from the Sydney and Illawarra statistical regions of New South Wales, Australia, to participate in the study. A portable SonoSite® 180PLUS ultrasound system (10-5 MHz, maximum depth 7 cm; SonoSite, Washington, USA) was used to measure the thickness of the plantar soft tissue at the 1st and 5th metatarsals, midfoot and heel. The average of three measurements was taken at each site. Plantar sensation was measured using 5 Semmes-Weinstein Monofilaments at the heel, midfoot, 1st and 5th metatarsals, hallux and lateral malleolus. The finest monofilament that was correctly identified 3 times was recorded for each foot. Each subject was also asked, as part of a comprehensive questionnaire, whether they suffered from foot pain and recorded as a dichotomous “yes” or “no” variable. From the larger sample, 103 individuals (33%) were identified as obese (Body Mass Index (BMI) > 30; World Health Organization (WHO), 1998). Eighty-one individuals (26%) had a BMI less than 25 and were therefore classified as normal weight for comparison to the obese adults. Independent t-tests and χ² tests were used to determine any differences in plantar soft tissue thickness, sensation and foot pain between the obese and normal weight participants (p < 0.05).

RESULTS
The obese adults displayed significantly thicker soft tissue at the heel, midfoot and 5th metatarsal head compared to their normal weight counterparts (see Figure 1). Although the soft-tissue thickness at the 1st metatarsal did not differ between the two groups, the overweight and obese older adults displayed a significantly higher fat pad-to-soft tissue ratio (0.25 ± 0.09) than the normal weight participants (0.22; t = -3.4, p = 0.001). The obese individuals were also more insensate at the heel (χ² = 27.74; p = < 0.001) and midfoot (χ² = 10.58; p = 0.005) regions of the foot. Furthermore, 63.4% of the obese older

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adults suffered from foot pain, which was significantly higher than the 41.3% of normal weight adults who reported foot pain ($\chi^2 = 8.16; p = 0.005$).

Figure 1: Soft tissue thickness at the heel, midfoot, 1st metatarsal (1MTH) and 5th metatarsal (5MTH) of the obese and normal weight participants. * denotes a significant difference between the two groups ($p < 0.05$).

**DISCUSSION**

Despite having more soft tissue on the plantar surface of the foot, the obese older adults were more insensate and reported suffering a higher incidence of foot pain than their normal weight counterparts. Of concern, 45% of the obese participants were unable to detect the 5.07 monofilament on their heel, indicating that these individuals had lost protective plantar sensation, placing them at an increased risk of sustaining an injury and, in those who also suffered diabetes, a higher risk of plantar ulcers. However, only 24% of the obese participants who were not able to detect the 5.07 monofilament on their heel had been diagnosed with diabetes. Therefore, the presence of diabetes in this population did not fully explain the high incidence of peripheral neuropathy in older obese adults. It is speculated that peripheral neuropathy in obese adults might be associated with increased mechanical loading of their feet caused by bearing additional mass over time.

Almost two-thirds of the obese participants reported suffering from foot pain. Foot pain has been shown to impair balance and functional ability in older people (Menz & Lord, 2001) and may deter older individuals from being active. With the number of people aged over 65 increasing around the world, the incidence of elderly obese individuals is likely to increase. Therefore, footwear choices for this population are paramount to reducing their foot pain and allowing them to remain active. Footwear interventions that can protect the older obese foot and provide better quality of life for older people are recommended, so they can become more physically active, which may in turn reduce the incidence of obesity.

**REFERENCES**


POTENTIAL USE OF CENTER OF PRESSURE IN SHOE MATCHING – A PILOT STUDY

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INTRODUCTION
Evidence suggests that self-perceived comfort in footwear may be associated with lower injury rate in running (Mundermann et al., 2001). After an early finding of lower injury rate in barefoot running compared to shod running (Robbins & Hanna, 1987), various studies have investigated the biomechanics of barefoot running (De Wit et al., 1999; Kurz & Stergiou, 2003), which may be viewed as the natural running form in healthy individuals. It has been hypothesized that the large variance in barefoot running is positively correlated to overuse injury reduction by spreading impact forces across a larger amount of tissues and decreases injury incidence (Kurz & Stergiou, 2003).

The aim of this study was to explore whether the center of pressure (COP) pattern could be used to match a footwear model to an individual. The matched footwear model would allow the individual to run in a pattern that closely resembled the individual’s barefoot running pattern. Assuming that subjective comfort was positively associated with objective measurement, it was hypothesized that the matched footwear model would be in agreement with the individual’s preferred model.

METHODS
One male subject (age = 21 yrs, height = 1.67 m, mass = 78.5 kg) who was free from recent lower-extremity injury or pain participated in the study. Informed consent was obtained prior to data collection. The subject ran at a self-selected speed across a force platform (Advanced Medical Technologies Inc., Watertown, MA) in the laboratory with three different running shoes and while barefoot. The three different shoes were classified according to their function: stability (Spira Genesis), cushioning (Spira Volare), and light-weight cushioning (Spira Clarion). The order of footwear conditions were randomly assigned and 10 trials were collected for each condition. Contact with the force platform was made with the right foot and kinetic data were recorded at 2400 Hz. The subject was allowed time to get accustomed with proper foot placement on the force plate and was instructed to maintain a consistent pace throughout all trials and shoe conditions. The subject was not informed on the function of the shoes and was asked to select the model that he preferred most after the experiment based on subjective ‘feeling’.

In order to locate the foot position on the force platform, kinematic data were collected simultaneously with the kinetic data using an eight-camera motion analysis system (Vicon, Los Angeles, CA) operating at 240 Hz. Spherical reflective markers were placed on the right heel, second metatarsal head and lateral malleolus of the subject to allow identification for a period during which the whole foot was in contact with the force platform. The heel marker position during this period was used as the reference point in each trial. Force data were filtered using a cubic spline with a degree of smoothing calculated based on a cut-off frequency of 50 Hz. Data were discarded if the vertical force was below 50 N to remove noisy data in the beginning and at the end. Center of pressure was calculated from the filtered force data and adjusted to the heel marker position. The COP data were time normalized and 101 data points were used per trial. An average COP pattern was calculated from the 10 trials for each footwear condition. The root mean square (RMS) differences in average COP time histories between barefoot and each shoe model were calculated in both medio-lateral and anteroposterior directions.
RESULTS
Figure 1 compares the average COP patterns of the four running conditions. The RMS differences between barefoot and each shoe model are shown in Table 1. The subject chose the cushioning shoe as the model that he preferred most during running.

![Figure 1. Comparison of COP patterns during running under four different footwear conditions.](image)

**Table 1. Root mean square differences in COP between barefoot and three shoe models**

<table>
<thead>
<tr>
<th>RMS difference</th>
<th>stability model</th>
<th>cushioning model</th>
<th>light-weight model</th>
</tr>
</thead>
<tbody>
<tr>
<td>medial lateral (mm)</td>
<td>6.21</td>
<td>4.76</td>
<td>9.53</td>
</tr>
<tr>
<td>anteroposterior (mm)</td>
<td>20.77</td>
<td>19.72</td>
<td>26.46</td>
</tr>
<tr>
<td>average (mm)</td>
<td>13.49</td>
<td>12.24</td>
<td>17.99</td>
</tr>
</tbody>
</table>

DISCUSSION
The COP pattern during running in the cushioning shoe best resembles that of barefoot running. This is in agreement with the subject’s preferred shoe model. Such findings may reinforce the hypothesis that subjective comfort may be positively associated with objective measurement. If barefoot running is regarded as the natural running pattern, COP pattern may potentially be used to characterize running gait and to match an appropriate shoe model to an individual.

CONCLUSIONS
The footwear which allowed for a running pattern close to barefoot running, as characterized by COP pattern, matched with individual preference chosen upon subjective comfort.

REFERENCES

Acknowledgements: The authors would like to acknowledge Spira Footwear for their financial support.
THE GAIT CHARACTERISTICS FOR SUBJECTS WITH DIFFERENT SEVERITY OF KNEE OSTEOARTHRITIS

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INTRODUCTION

Knee osteoarthritis (OA) is major problem in older people. The common complaints from knee OA subjects are pain and stiff knee in initial walking, decrease of range of motion, reversion of screw home mechanism, slower walking speed, shorter stride length, shorter stance phase, longer double-leg support, and asymmetric gait pattern (Messier, et al.,1992; Stauffer, et al.,1977; McGibbon , 2002; Zahrani and Bakheit, 2002; Chen, et al.,2003). Due to excessive knee varus torque the compressive force acts on medial site of knee joint is 2.5 times higher than the lateral condyle (Sharma, et al, 1998; Kerrigan, et al.,2002). The vertical ground reaction at time of loading response and push off are smaller and delayed (Kaufman,2001;Chen,2003), but with faster loading rate after initial contact (Radin ,et al.,1991). The purpose of this study was to quantitatively analyze the kinetic pattern of different stages of OA knee in order to better understanding the characteristics of OA knee gait.

MATERIALS AND METHODS

Forty knee subjects (64.15 ± 8.75 yrs) and twenty age matched normal subjects (63.20 ± 2.68 yrs) were recruited from the outpatient clinic of Department of Rehabilitation, Shin Kong Wu Ho-Su Memorial Hospital. The OA subjects were classified by Kellgren-Lawrence scale into grade II (44 knees) and grade III (13 knees). Lower limbs anthropometric measurements, Lequesne quaternary, visual-analog scale, kinetic parameters of plantar pressure and ground reaction force measurements using force platform (AMTI, USA), and pressure mat (Footscan, Rsscan, Belgium) were statistically analyzed via two-way ANOVA to discriminate the gait characters between OA grades.

RESULTS AND DISCUSSION

The walking speed of OA subjects was slower than the normal but no significant. The normal subjects showed higher peak plantar pressure than the OA subjects, significance at T1 (Hallux), M3 (Metatarsal 3), lateral heel regions. The grade 3OA subjects had larger peak plantar pressure than grade 2 and profoundly at M3, M4, and MH ( Fig 1). The contact time at each mask region revealed that the normal subjects were significantly smaller than OA subjects (Fig 3). In term of contact % of stance phase, there is no significant difference between different grades except M5 region (Fig 4). In general, OA subjects showed larger impulse than normal one; grade 3 subjects were significantly higher at lateral forefoot and medial heel. The normal subject also showed significant larger impulse at M1, M3 but smaller at M4,5.

CONCLUSION

The severe the OA condition the less shock absorption capacity at the rear foot, slower walking speed, and delay loading response during a gait cycle. The varus moment on the OA knee results in varus foot compensation and poor weight transferring between limbs. A well cushion heel, lateral wedge insole, and rocker shoes can be manufactured to dissipate the ground reaction shock at the time of initial contact and to realign the varus knee as well as reduce the forefoot pressure and to help the push off propelling

REFERENCES

Disability and rehabilitation

![Figure 1: Peak plantar pressure at different regions](image1)
![Figure 2: Impulse at different regions](image2)

T1: Hallux, M1: metatarsal head 1, MH: medial heel

![Figure 3: contact time in ms](image3)
![Figure 4: Contact time in % of stance phase](image4)

ACKNOWLEDGEMENT

This study was supported by the National Science Council of R.O.C through grants NSC95-2622-B010-001. NSC94-2622-B010-004. and Shin-Kong Wu Ho-Su Memorial Hospital grant 95E-008
PEAK KNEE ADDUCTION MOMENT VALUES FOR FIVE OFF-THE-SHELF WALKING SHOES

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INTRODUCTION
Knee osteoarthritis (OA) occurs more frequently in the medial compartment than in the lateral compartment. The peak knee adduction moment (PKAM) can be used to predict the loads in the medial compartment (Schippelein et al. 1991), and can therefore be used to assess the effectiveness of interventions, such as shoe modifications.

Studies by Crenshaw et al. (2000), Kerrigan et al. (2002), and Fisher et al. (2002, 2004) have demonstrated reductions in the PKAM with the use of shoe modifications such as lateral wedges and variable midsole stiffness. But would interventions be further enhanced by starting with a particular shoe model? The purpose of this study was to evaluate off-the-shelf walking shoes, before any modifications were made, in order to establish a baseline of PKAM values.

METHODS
Five off-the-shelf walking shoes were tested: A, B, C, D, E. Each shoe model was a neutral (cushioning) shoe with no specific motion control features. Eighteen healthy women were recruited for this study (avg. ht. 63 in., avg. wt. 139 lbs., avg. age 38 years, shoe size US7). Subjects walked at 1.6m/s while kinetic (1200 Hz, Kistler, Amherst, NY, U.S.A.) and 3D kinematic (240 Hz Motion Analysis Corporation, Santa Rosa, CA, U.S.A.) data were collected for five trials in each shoe. Retrorereflective markers were placed on predetermined bony landmarks on the lateral aspect of the right leg. Inverse dynamic calculations were used to calculate the peak knee adduction moments.

RESULTS AND DISCUSSION
Maximum impact and propulsion ground reaction force (GRF) peaks are presented for each shoe (Table 1). C had a significantly lower impact peak than A, B, D, and E. There were no differences in the propulsion peak among the five shoes.

<table>
<thead>
<tr>
<th>Table 1: Maximum impact and propulsion vertical GRF peaks (avg and SD); * indicates statistical significance (p &lt; 0.10)</th>
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<tbody>
<tr>
<td><strong>Shoes</strong></td>
</tr>
<tr>
<td>Impact (BW)</td>
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<tr>
<td>Propulsion (BW)</td>
</tr>
</tbody>
</table>

The PKAM curve has two peaks, and this study reports on both peaks (Figure 1). The 1st PKAM values ranged from 4.25-4.43 (%BW*Ht) (Figure 2). B had a significantly lower 1st PKAM than A (p=0.047) and E (p=0.003). These values were slightly higher than those reported previously by Kerrigan et al. (2002), Crenshaw et al. (2000), and Fisher et al. (2004), but the higher values reported here could have been influenced by the faster walking speed used in this study. The 2nd PKAM values ranged from 2.87-3.02 (%BW*Ht) (Figure 1). E had a significantly higher 2nd PKAM than A (p=0.039), B (p=0.0007), and C (p=0.095).
The results from this study demonstrated differences in both peaks of the PKAM curve, suggesting that any interventions added to shoes should be full-length in order to further reduce the load in the medial compartment.

CONCLUSIONS
This study presented the range of PKAM values when five off-the-shelf walking shoes were tested on healthy subjects. There was a 4% difference in the 1st PKAM, and a 5% difference in the 2nd PKAM, between the highest (E) and lowest (B) values. It is possible that the shoe chosen for use with a knee OA intervention (orthotics, wedge, etc.) might help play a role in reducing the load on the medial compartment of the knee. The vertical GRF values indicated that four of the five shoes in this study had similar cushioning. Further research into specific shoe design characteristics, including a broader range of cushioning, and their effect on PKAMs would be helpful.

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ACKNOWLEDGEMENTS
The authors would like to thank Michael Amos in the NSRL for his work on this project.
HOUSEHOLD SHOE WEARING HABITS OF OLDER ADULTS:
ARE THEY ASSOCIATED WITH FALLS RISK?

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INTRODUCTION
Falls are the leading cause of injury, disability, hospitalization and death for older people. Over half of these falls occur while in the home (Rubenstein et al. 1988) and unsafe footwear is often cited as a major falls risk factor. However, it is unknown whether particular shoe-wearing habits of older people are in-fact placing individuals at a higher risk of falling or whether other factors, such as foot pain, influence the shoe wearing habits of older people which, in turn, might place these people at risk of a fall. Therefore, the purpose of this study was to determine what footwear older people commonly wear inside their home and whether these shoe wearing habits are associated with their falls risk.

METHODS
Three-hundred and twelve older men and women aged between 60-90 years (154 women; age = 71.2 ± 7.7 yr; height = 1.66 ± 0.1 m; BMI = 28.5 ± 5.4 kg.m⁻²) were randomly recruited, using the Australian Electoral roll, from the Sydney and Illawarra statistical regions of New South Wales, Australia, to participate in the study. Each subject was asked, as part of a comprehensive questionnaire, about their household shoe-wearing habits, including the type of footwear, if any, they most frequently wore while at home. Subjects were also asked whether they suffered from foot pain and this was recorded as a dichotomous “yes” or “no” variable. Each subject had their falls risk assessed using the short version of the Physiological Profile Assessment (PPA; Lord et al., 2003), which included five tests to assess vision, proprioception, quadriceps muscle strength, reaction time and body sway. A two-way ANOVA design was then conducted to determine whether there were any significant (p < 0.05) main effects of shoe-wearing habits or gender on falls risk score whereas Chi-square tests were used to determine any differences in foot pain reported by the participants who wore the different shoe types.

RESULTS
Fifty-nine of the participants (19%) stated that they did not regularly wear shoes in and around the home. Of those who wore shoes in the home, the most frequently worn footwear were slippers (74 respondents), followed by a shoe with a fastener (athletic or non-athletic), which 70 of the participants chose as their regular indoor shoe. The falls risk scores and foot pain reported by the participants who chose one of these three categories (barefoot, slippers, fastened shoes; 65% of all participants) were then analysed to determine the effects of the most common household footwear types on these variables. Interestingly, the falls risk score was found to be significantly higher in those participants who wore slippers in the home compared to those participants who wore no shoes or a fastened shoe (F₂,197 = 4.037, p = 0.02; see Figure 1). However, there was no main effect of gender (F₁,197 = 1.899, p = 0.17) and no significant gender x shoe interaction (F₂,197 = 1.424, p = 0.24).

The incidence of foot pain was significantly different between those participants who wore slippers and those who either wore no shoes or a fastened shoe in the home. That is, 60% of slipper-wearing participants stated that they suffered from foot pain compared to only 37% of those who did not wear shoes (χ² = 7.17; p < 0.03).
DISCUSSION
The results of the present study imply that older individuals who wear slippers around the home have a higher falls risk score than those individuals who either wear a fastened shoe or wear no shoes in the home. Although it could be speculated that slippers pose a safety risk based on these findings, it is unknown whether habitually wearing slippers around the home is contributing to their higher risk of falling by affecting components of the PPA, such as balance, strength or proprioception, or whether there are other confounding factors to be considered. For example, it was noted that those individuals who wore slippers in the home tended to suffer from a higher incidence of foot pain than those who did not. It is hypothesized that older people who suffer from foot pain prefer to wear slippers in the home as they are comfortable and easy to don, due to their lack of structure. Foot pain has been found to impair balance, gait and functional ability in older men and women (Menz & Lord, 2001), which may be placing these individuals at an increased risk of falling. Therefore, it is imperative that safe and comfortable footwear for older people, particularly those who suffer from foot pain, are developed.

REFERENCES

Figure 1: The mean falls risk scores for the participants who did not wear shoes in the home, wore slippers or a fastened shoe. * denotes a significant difference between the two groups (p < 0.05).
IN SHOE MOTION OF THE CHILD’S FOOT WHEN WALKING

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The University of Sydney, Australia

INTRODUCTION
Because a skeletally immature and developing child is particularly susceptible to the influences of footwear, it is important to understand the likely influence of a shoe design. Although the common aims of controlling rearfoot motion have been questioned for adults, they have not yet been investigated for children. Furthermore, a compensatory increase in rearfoot motion may be caused by the limitation to the midfoot motion of a shoe, with varying responses for pronated feet, at least in adults (Attwells and Smith, 2001). Some of the challenges in footwear research are concerned with knowing what motion is occurring in the foot within the shoe. Previous concerns that surface markers may provide misleading information about foot motion can be overcome by the use of markers attached to the heel while the foot is in the shoe. The aims of the current study were to compare motion of the heel (rearfoot) within the shoe via a rearfoot wand (Tong and Smith, 2003) to the shoe counter during walking, in children.

METHODS
Healthy active children aged between 6 to 10 years were recruited for this study. Foot shape and type i.e. pronated/supinated were scored according to the Foot Posture Index (FPI; Redmond et al, 2006).

In-shoe motion of the rearfoot segment was determined using a brass wand device in which three retro-reflective markers were attached to a single rigid shaft mounted onto a flat stirrup. The stirrup provided a large area of contact around the heel and was attached firmly. The wand extended posteriorly through a 10 mm hole in the rear of the shoe counter. For each study subject it was established from video recordings that the wand shaft did not contact the shoe during walking or running gait. Markers were also attached to the shoe, and on the shank, at the knee and at the ankle to calculate segment motion with six degrees of freedom and joint centre locations for the knee and ankle joint complex.

A 10 camera motion analysis system (EVaRT5.0, MAC) with a sampling rate of 200Hz was used to calculate the marker trajectories in 3D. An embedded segment axis system was used to determine joint motion. The 3D coordinate system for the Shoe Counter and Wand segments was determined from the global coordinate system. A static reference trial was used to determine all three segment axes and for which the long axis of the foot was aligned with the global x axis.

Participants wore well-fitted leather school shoes. Five walking trials were recorded at a self-selected pace. Data were time normalized to the stance phase. Comparisons were made between the 3D motion of the shoe counter relative to the leg, and the rearfoot wand relative to the leg.

RESULTS
Preliminary results are presented for sagittal and frontal plane motion. In Table 1 are the total ranges of motion, to date, for four children. All four had FPI scores for left and right feet between +1 to +6. While mean sagittal plane motion was within 1° for the shoe counter and the rearfoot wand, mean frontal plane motion was 3° less for the shoe counter compared to the rearfoot wand. Time series graphs for one representative subject are presented in Figure 1. The stance phase pattern for sagittal motion shows that although dorsiflexion occurs to the same extent for the shoe and for the rearfoot while in the shoe, a lesser range of rearfoot sagittal motion occurs in the direction of plantarflexion. For this subject,
most of this reduction in the plantarflexion motion occurred prior to midstance. In the frontal plane, the lesser motion of the shoe counter compared to the rearfoot-in-shoe occurs in the direction of inversion. Except for a short interval of time around the time of foot flat (approx. 22% stance), the position of the shoe counter is not co-incident with that of the rearfoot-in shoe; it instead remains relatively everted. Variability (SD) between subjects (Table 1) was less for the shoe counter compared to the rearfoot wand in both sagittal and frontal planes. The between trial variability (95% confidence interval) for the representative subject (Fig 1) shows that in the sagittal plane, except for the time of peak plantarflexion (around the time of push-off), this greater variability of the shoe counter occurs throughout stance phase and also just prior to heel contact and after toe-off. In the frontal plane, the graph suggests that most of the variability occurs at the time of foot flat and continues until after toe-off.

Table 1. Sagittal and frontal motion (°) for Shoe Counter and Rearfoot Wand: mean and SD of 4 subjects and 5 trials.

<table>
<thead>
<tr>
<th></th>
<th>Shoe Counter</th>
<th>Rearfoot Wand</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sagittal Plane</td>
<td>mean 25.8</td>
<td>25.6</td>
</tr>
<tr>
<td></td>
<td>SD 3.1</td>
<td>6.7</td>
</tr>
<tr>
<td>Frontal Plane</td>
<td>mean 14.7</td>
<td>17.1</td>
</tr>
<tr>
<td></td>
<td>SD 6.9</td>
<td>9.1</td>
</tr>
</tbody>
</table>

Figure 1. Sagittal and frontal Plane Motion of the Shoe Counter and Rearfoot Wand for one study participant: mean of five trials and 95% confidence intervals. Heel contact is 0% and toe-off is at 100%.

**DISCUSSION**

These data indicate that several assumptions about the co-incidence between shoe and rearfoot motion are incorrect. A leather school shoe appears to limit rearfoot inversion more than is actually the case. The functional significances of any relative limitation to rearfoot plantarflexion and also the greater motion variability of the rearfoot compared to the shoe are not yet clear. The implications of all of these preliminary data will become more evident as the current analyses are completed, and will be elucidated by the complete analyses of the whole foot across the breadth of the walking running balance activities while barefoot and in different types of shoes.

**REFERENCES**


DOES CALF STRETCHING REDUCE FOREFOOT PLANTAR PRESSURE?

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INTRODUCTION

Calf muscle flexibility is imperative to allow the ankle joint (AKJ) to achieve “normal” range of motion and therefore “normal” foot function (Wernick and Volpe 1996). It is accepted that a tight calf muscle complex (also known as an ankle joint equinus) increases forefoot plantar pressure (Armstrong et al. 1999, Maluf et al. 2004, Willrich et al. 2005). A tight calf muscle complex (CMC) translates mechanically into an early heel lift (Thomson et al. 2002) and therefore, with each step there will be less time spent on the hindfoot and more time spent on the forefoot. This increased pressure on the forefoot can lead to numerous pathological conditions such as metatarsalgia, capsulitis, corns, callous, stress fractures and ulcerations (Thomson et al 2002). There appears to be a growing trend to increase the length of the tendon Achilles (TA) surgically in an attempt to reduce the forefoot pressure (Armstrong et al 1999, Maluf et al 2004, Willrich et al 2005). Surgical procedures are invasive, expensive and not without risk. A simple safe and proven alternative would therefore be of great value. Such a method does already exist and is practiced by clinicians but there appears to be a lack of evidence to support the more conservative approach of stretching the CMC. This pilot study will investigate the affects of stretching the CMC and changes to forefoot plantar pressure. The aim of this investigation is to; therefore, scientifically evaluate the claim that “calf stretching reduces forefoot plantar pressure”.

METHODOLOGY

Thirteen runners (4 male and 9 female) with an average age of 34.43 ±6.93 years, height of 170.65± 8.44 cms, weight of 67.24 ±8.36 kg and shoe size of 7.43 ±1.59 participated in the study. Ethical approval was sought and granted by the University ethics committee and all participants gave an informed consent. The subjects had no known musculoskeletal pathology and had ≤ 5° of ankle joint dorsiflexion. During the initial assessment the calf muscle flexibility of both legs was assessed by the same investigator using both the goniometer and the flexeramp. After recording the anthropometric details and the information on preferred footwear, foot posture index (FPI) score was recorded. The FPI is an objective, validated, classification, assessment tool for foot type (Redmond et al 2006) and was used to rule out any contributory biomechanical insufficiencies. The subjects were allocated with a pair of trainers and in-shoe pressure distribution pattern was recorded using F-Scan (Tekscan Inc., USA) system. Subjects were advised to use the same pair of shoes for future data collection sessions. After collecting the pressure data, each subject was advised to stretch in the morning and in the evening, seven days a week for 8 weeks. The stretch was carried out on the flexeramp and both the right and left leg was stretched together. Since each subject had a similar range of motion at the start of the study, each subject will start on either the 5° or 8° angle. If the ramp angle is set too high the subject will compensate by arching their back and/or flexing their knees and/or leaning forward which could create new pathologies. Once the subject can stand up straight with normal posture and if there was only a very slight pull in their CMC the subject was advised to increase the ramp angle. Diary sheets were issued to each subject to record the time, length and height of each stretch. The subjects were re-assessed at 4 weeks and 8 weeks. Once the temporal parameters were extracted from the pressure data, statistical tests
were conducted to check the normality, and then repeated measures ANOVA was carried out as appropriate.

RESULTS AND DISCUSSION

This pilot study indicated that calf stretching for 8 weeks increases the flexibility/ROM within the calf muscle complex and ankle joint ROM. It also significantly reduced the forefoot plantar pressure by allowing the foot function to naturally improve.

Further studies with larger samples will substantiate the reported results and lead to newer footwear designs. Furthermore, the results will influence the clinical intervention in subjects with a reduced ankle joint range of motion.

REFERENCES


REARFOOT MOTION DURING RUNNING RELATIVE TO
ACTIVE AND PASSIVE JOINT LIMITS

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University of Massachusetts at Amherst

INTRODUCTION
Abnormal frontal plane rearfoot motion has been implicated as a predisposing factor of repetitive lower extremity injuries (Williams et al, 2001). Classically, biomechanical studies have focused on a joint’s peak range of motion (ROM) during an activity (Hamill et al., 1992). However, researchers have not evaluated this peak motion relative to the ROM available or how closely a joint has come to reaching its end range. As a joint nears its end range, increased stress and strain will be placed on the surrounding bone and soft tissue, which may increase its risk of injury. Therefore, the purpose of this study is to analyze the magnitude of rearfoot motion during running relative to the available active and passive joint ROM.

METHODS
Four subjects with no history of lower extremity injury over the last year were studied. Rigid lower leg and foot segments were defined using retroreflective markers placed on the first and fifth metatarsal heads, medial and lateral malleoli and medial and lateral knee joint line. Additionally, two rigid tracking plates with markers were secured on the lateral aspect of the lower leg and the posterior aspect of the calcaneus. A standing calibration was used to define each subject’s zero position for joint ROM. Individual markers were removed and segments were tracked via the tracking plates.

Kinematic data were collected at 200Hz with a calibrated 8-camera Qualysis 3-D motion analysis system (Qualysis, Inc., Gothenburg, Sweden). Data were then exported to Visual 3D (C-Motion, Rockville, Md) where joint angles were calculated after being filtered with a Butterworth low pass filter with a cut off frequency of 8Hz. Joint angles were calculated using an XYZ Cardan rotation sequence.

Active rearfoot ROM boundaries were recorded with the subject standing and performing open chain rearfoot inversion, eversion, abduction and adduction. Passive ROM boundaries were recorded by having a licensed physical therapist manually invert, evert, abduct and adduct the subject’s rearfoot, with the subject side-lying on a treatment table. The maximum eversion and inversion value from these four movements was used to define the joint’s end range. Rearfoot ROM was then recorded with the subject running barefoot on a treadmill at 6.0 mph.

RESULTS AND DISCUSSION
Differences in active and passive ROM were seen in all four subjects. Additionally, each subject had a varying amount of rearfoot motion while running barefoot at 6.0 mph. More importantly though, subject 1 surpassed her active eversion ROM boundary by 3.3 degrees, and came within 3.1 degrees of her passive eversion ROM boundary. Subject 2 on the other hand was 4.8 degrees from her active eversion ROM boundary and 12.7 degrees from her passive eversion ROM boundary (Figure 1). Using this method of analysis, it could be hypothesized that subject 1 may be more prone to injuries because she functioned closer to her passive end range placing more stress on the surrounding tissue.

Given that this method relies on looking at rearfoot motion relative to its boundaries, a consistent and reliable protocol must be developed in order to validate these conclusions. All joints should have more passive than active ROM, however it can see that both subject three and four demonstrated more
active than passive eversion ROM. This difference is most likely attributed to differing degrees of ankle dorsiflexion when measuring eversion, however could also be a result of subject resistance and the amount of force applied by the examiner. Future studies should evaluate the inter-rater reliability of finding the end-range of passive ROM, as well as the reliability of these measurements across days.

![Graphs showing rearfoot motion](image)

**Figure 1**: Frontal plane rearfoot motion from one footfall while running at 6.0 mph. Active and passive ROM boundaries for inversion and eversion are represented by the dashed lines. Heel strike is indicated by the HS and toe-off by the TO.

**CONCLUSION**

Examining rearfoot motion relative to its boundaries may provide useful information in identifying individuals with increased potential for repetitive overuse injuries. To validate this method of analysis, a consistent and reliable ROM protocol must be developed.

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BIOMECHANICAL ASSESSMENT OF A NEW ICE HOCKEY SKATE BLADE DESIGN

Peter Federolf, Aurel Coza, Robert Mills, Benno Nigg
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INTRODUCTION

Ice hockey is a fast and dynamic sport. Speed and agility are key characteristics of the participating athletes. Speed and agility of an ice hockey player depend on the athlete’s conditioning and on the skating equipment. In 2006, a new skate blade design was presented by the company CT Edge. The differences to conventional blades are additional blade angles milled into the side of the blade. This gives the CT blades (i) a wider contact area at the bottom of the blade in straight gliding and (ii) advantageous edge angles, which may provide more resistance when the blade is inclined. Players using this new blade have reported that straight gliding was possible with less physical effort, that turning and cutting corners at high speeds are easier, and that the new blades allow better control when stopping. However, scientific evidence supporting these claims is missing.

The purpose of this study was to determine differences in the performance characteristics of the new blade compared to regular skate blades. Specifically the following four issues have been investigated: (a) frictional properties, (b) performance in agility tests, (c) subjective assessment by high level ice hockey players and (d) changes in muscle activation patterns in comparison to activities with conventional blades.

METHODS

The frictional properties of the skate blades were characterized by determining the coefficient of friction (COF) from the deceleration of a sled equipped with four skate blades. Three different skate blade designs, with blade angles of 4° (CT4) 6° (CT6), and 8° (CT8), were measured and compared to regular skate blades.

All experimental measurements were done with players of the University of Calgary’s ice hockey team (age 22-24, weight 90 ± 8kg, height 183 ± 5cm). The acceleration and speed tests consisted of 45m straight skating during which the acceleration time (measured between 0m and 10m) and the maximal speed (averaged between 30m and 40m) were measured. For the glide turn (agility test 12 skaters cut four sharp 180° turns using two blade conditions, the CT8 and the players’ regular skate blades.

All subjects filled out a questionnaire after the tests in which the blade performance was assessed using a visual analog scale. The scale was divided into the sections “standard (STD) blade preferred”, “neutral”, “CT8 blade preferred” (STD—0—CT8). In the analysis the distance of the subjects’ marks from the neutral line were measured.

Muscle activation patterns were measured using bipolar surface electromyographic electrodes (Ag/AgCl, 2400 Hz) on the bellies of biceps femoris, flexor digitorum, gracilis, gastrocnemius lateralis, gluteus medius, peroneus longus, vastus lateralis, and vastus medialis. Four specific situations were selected for further analysis: an acceleration step, a 0.6s phase of a glide turn, one crossover step, and 0.6s of stopping. The raw EMG data was high-pass filtered at 10Hz and smoothed by calculating the root mean square RMS over 30ms for each data point. This process yields a smoothed envelope of the raw EMG signal.

All tests were carried out in the Calgary Olympic Oval between November 2006 and February 2007. For the statistical analysis Student’s t-tests were used. The level of statistical significance was set at $\alpha \leq 0.05$. 
RESULTS
The friction measurements confirmed that compared to regular skate blades the new blade design reduced the blades’ coefficient of friction, COF, significantly (12.6% for CT4, 20.6% for CT6 and 23.6% for CT8).
The results for the acceleration measurements showed an average improvement of 0.9% (statistically not significant). The results for the maximal speed showed for the CT8 blade significant improvements of 1.3% (p=0.048). The results for the glide turns (agility) showed an average improvement of 1.4% (p=0.0058). The experimental results showed for all variables substantial individual differences.
The EMG measurements showed large individual differences in the muscle activation patterns. Some of the subjects showed different muscle activation patterns when using different blades, but these differences were not consistent between all subjects.
The subjects’ average feedback concerning the blades showed a clear preference of the CT8 blades. The mean results for the tasks “glide turns”, “crossover turns”, changing direction”, “stopping”, “glidability”, and “overall performance” were 69%, 53%, 68%, 52%, 68%, and 74%, respectively. All subjects stated that given the choice, they would buy CT blades.

DISCUSSION
The coefficients of friction, COF, of the CT blades were in the range reported by Kobayashi (1973), de Koning et al. (1992) for speed skating blades. The coefficients of friction, COF, of the standard ice hockey blades were slightly higher. The results of this study suggest that the paradigm of “the thinner the blade the lower the friction coefficient” does not hold true for blade designs, which may be related to the changes of design, or the fact that the contact area was increased. Further investigations of blade properties such as blade width, rocker radius, hollow radius, or hollow shape might lead to the development of blade designs with even less friction. In the acceleration and maximal speed tests the subjects could on average benefit from the new blade design, but not in the same extend as friction was reduced. This was expected, because in most situations in ice hockey the players skate on inclined skates (e.g. in the push-off phase of a step or in any turning maneuver). The results of the glide turn test indicate that most ice hockey players also can improve their performance in situations in which the skate blades are inclined. The subject feedback substantiated the results of the agility tests. In summary, the results of this study indicate that changes in blade design have an effect on performance in skating, changing various important aspects such as COF, speed and agility.

REFERENCES

ACKNOWLEDGEMENTS
The support of the Olympic Oval, the University of Calgary’s hockey team and Jamie Wilson is acknowledged. The study was financially supported by CT Edge.
VARIATION IN FOOT SHAPE UNDER VARIOUS LOADING CONDITIONS
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INTRODUCTION
Previous research has quantified differences in foot shape between genders (Wunderlich & Cavanagh, 2001), and changes in foot shape throughout various degrees of loading up to 100% of body weight (Houston et al., 2006). The purpose of this study was to compare changes in foot shape with increasing loads of up to 150% of body weight, and to assess if feet of different sizes change in similar proportions to the increasing loads above 100% body weight.

METHODS
Twenty-four (24) subjects participated in this study. Subjects were chosen based on self-reported shoe size (women’s size 7: n=8, mean weight = 144.9 lbs; men’s size 9: n=8, mean weight 179.0 lbs; men’s size 11: n=8, mean weight = 200.5 lbs).

Body weight and foot lengths were collected. Subjects were then asked to stand barefoot on a foot image platform, which enabled digital images of each foot to be taken from medial and inferior views. While on the foot image platform, subjects were able to touch a rail for balance purposes, but were instructed not to load any weight on the rail (Figure 1).

Static digital images of each foot were taken under different loading conditions: 0%, 100% and 150% of body weight. For the 0% body weight trial, subjects were asked to lightly touch the platform with each foot. For the 150% body weight trial, 50% of the subject’s body weight was added to the subject using a weight vest (see Figure 1).

Digital images were analyzed and scaled on a computer using a CAD based system. Foot lengths and foot widths were measured from the inferior view, and arch height was measured from the medial view. Foot length was defined as the distance between the farthest point back on the heel to the farthest point forward on the toes. Foot width was defined as the distance between the widest point on the medial and lateral sides of the foot. Arch height was defined as the distance between the bottom plane of the foot to the top of the foot at 50% of the foot length (Williams & McClay, 2000). Changes in these variables between loading conditions were analyzed.

The inter- and intra-rater reliability for the foot image platform was high (foot length inter-rater ICC R=0.97, intra-rater ICC R=0.98; foot width inter-rater ICC R=0.99, intra-rater ICC R=0.98; arch height inter-rater ICC R=0.98, intra-rater ICC R=0.90).

RESULTS
Foot length and foot width were significantly different across sizes and across loads (P < 0.05; Tables 1 and 2). Arch height was significantly different between the men’s size 9 and men’s size 11, and women’s size 7 and men’s size 11 at all 3 loads, but not significantly different between the women’s size 7 and the men’s size 9 at all 3 loads (Table 3).

Within the women’s size 7 and the men’s size 9 subjects, foot length increased significantly with increased loads (P < 0.05: Table 1). For both arch height and foot width, significant differences were seen between the 0% and 100% body weight loads as well as the 0% and 150% body weight loads, but arch height and foot width between the 100% and 150% body weight loads were not significantly different.

Within the men’s size 11 subjects, foot width and arch height were significantly different with increased loads (P < 0.05; Table 1). Foot length was significantly different between the 0% and 100% body weight loads as well as the 0% and 150% body weight loads, but foot length between the 100% and 150% body weight loads was not significantly different.
Table 1. Mean (+std) foot lengths of women’s size 7, men’s size 9 and men’s size 11 feet under increasing loads.

<table>
<thead>
<tr>
<th>Foot Length</th>
<th>0% BW (mm)</th>
<th>100% BW (mm)</th>
<th>150% BW (mm)</th>
<th>% Change 0-100% BW</th>
<th>% Change 0-150% BW</th>
<th>% Change 100-150% BW</th>
</tr>
</thead>
<tbody>
<tr>
<td>Women’s 7</td>
<td>231.2±2.7*</td>
<td>235.4±2.1**</td>
<td>236.7±2.0**</td>
<td>1.8%</td>
<td>2.4%</td>
<td>0.6%</td>
</tr>
<tr>
<td>Men’s 9</td>
<td>254.4±6.2*</td>
<td>259.9±6.5**</td>
<td>260.3±6.5**</td>
<td>1.8%</td>
<td>2.3%</td>
<td>0.1%</td>
</tr>
<tr>
<td>Men’s 11</td>
<td>267.4±6.2*</td>
<td>273.0±6.3**</td>
<td>273.2±4.6^</td>
<td>2.1%</td>
<td>2.2%</td>
<td>0.1%</td>
</tr>
</tbody>
</table>

* All significantly different between sizes (P < 0.05)
# All significantly different within women’s 7 and men’s 9 (P < 0.05)
^ Significantly different within men’s size 11 from 0% body weight load (P < 0.05)

Table 2. Mean (+std) foot widths of women’s size 7, men’s size 9 and men’s size 11 feet under increasing loads.

<table>
<thead>
<tr>
<th>Foot Width</th>
<th>0% BW (mm)</th>
<th>100% BW (mm)</th>
<th>150% BW (mm)</th>
<th>% Change 0-100% BW</th>
<th>% Change 0-150% BW</th>
<th>% Change 100-150% BW</th>
</tr>
</thead>
<tbody>
<tr>
<td>Women’s 7</td>
<td>89.9±3.4*</td>
<td>92.4±3.6**</td>
<td>92.0±3.1**</td>
<td>2.8%</td>
<td>2.4%</td>
<td>-0.4%</td>
</tr>
<tr>
<td>Men’s 9</td>
<td>97.4±5.0*</td>
<td>101.4±4.9**</td>
<td>101.7±5.4**</td>
<td>4.0%</td>
<td>4.3%</td>
<td>0.3%</td>
</tr>
<tr>
<td>Men’s 11</td>
<td>105.2±4.0^</td>
<td>108.1±3.9^</td>
<td>108.7±4.0^</td>
<td>2.8%</td>
<td>3.3%</td>
<td>0.5%</td>
</tr>
</tbody>
</table>

* All significantly different between sizes (P < 0.05)
# Significantly different within sizes women’s 7 and men’s 9 from 0% body weight load (P < 0.05)
^ All significantly different within men’s size 11 (P < 0.05)

Table 3. Mean (+std) arch heights of women’s size 7, men’s size 9 and men’s size 11 feet under increasing loads.

<table>
<thead>
<tr>
<th>Arch Height</th>
<th>0% BW (mm)</th>
<th>100% BW (mm)</th>
<th>150% BW (mm)</th>
<th>% Change 0-100% BW</th>
<th>% Change 0-150% BW</th>
<th>% Change 100-150% BW</th>
</tr>
</thead>
<tbody>
<tr>
<td>Women’s 7</td>
<td>67.1±3.9@</td>
<td>64.0±3.6@@</td>
<td>64.4±3.7@@</td>
<td>-4.7%</td>
<td>-4.1%</td>
<td>0.6%</td>
</tr>
<tr>
<td>Men’s 9</td>
<td>71.0±4.1@</td>
<td>66.7±3.7@@</td>
<td>66.2±3.2@@</td>
<td>-6.0%</td>
<td>-6.7%</td>
<td>-0.7%</td>
</tr>
<tr>
<td>Men’s 11</td>
<td>76.7±5.9@</td>
<td>73.0±5.3@@</td>
<td>71.9±5.4^</td>
<td>-4.8%</td>
<td>-6.2%</td>
<td>-1.5%</td>
</tr>
</tbody>
</table>

@ Significantly different between groups from men’s size 11 (P < 0.05).
# Significantly different within sizes women’s 7 and men’s 9 from 0% body weight load (P < 0.05)
^ All significantly different within men’s size 11 (P < 0.05)

DISCUSSION

Although the measured variables were different across sizes, the percent changes within each size over different loads were similar. This indicates that feet change shape in similar proportions regardless of foot size.

The results of this study were similar to Houston et al. (2006), which showed a 1.5% increase in foot length, 4.3% increase in ball width in subjects bearing 100% of body weight. Arch height decreased to a lesser degree (Houston et al., 2006 showed a -9.3% decrease in arch height with subjects bearing 100% body load).

The biggest changes in foot shape were seen between 0% and either 100% or 150% body weight loading. The changes between 100% and 150% body weight loading were less than 1% for most variables across all sizes. Loading the foot with an extra 50% of body weight may not be enough to elicit further changes in foot shape.

REFERENCES


REPORTING SHEAR GROUND REACTION FORCES DURING SKATEBOARD LANDINGS

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INTRODUCTION
Skateboarding is a popular activity that had over 11 million participants in the US alone during 2005 (Anon, 2006). Despite its popularity, little is known about the biomechanics of this growing sport. To rectify this we aimed to analyze one of the more popular maneuvers practiced by skateboarders, the rail slide, and, in the process, examine ways to analyze and report shear force data.

In most human locomotion movements (i.e. walking and running) the subject’s body is oriented parallel to the direction of forward motion. In board sports however, such as skateboarding and snowboarding, the subject’s body is oriented perpendicular to the direction of forward motion. Therefore, we propose a new naming convention when describing the axes of board sport shear ground reaction force (GRF) movements.

The terms nose ground reaction forces (NGRF) and tail ground reaction forces (TGRF) will be used to describe the movements of the system’s center of mass towards the nose or the tail of the skateboard. The terms front-side ground reaction forces (FGRF) and back-side ground reaction forces (BGRF) will be used to describe the movement of the system’s center of mass towards the front toe-side edge of the board (i.e. an anterior body movement) or back heel-side edge of the board (i.e. a posterior body movement). The proposed axis system can be seen in Figure 1.

In addition, due to the uniqueness of each landing, subjects rarely landed on top of their skateboards in plane with the orthogonal Fx and Fy axes of the force plate. Therefore, we were curious about how correcting for these off-angled landings with a cosine correction coefficient might affect shear force results acting upon the system of subject and skateboard.

METHODOLOGY
GRF and moment data were collected on 12 healthy top-amateur and professional male skateboarders (BM = 70.2 ± 9.1 kg) as they slid down a sloping handrail on their skateboards, eventually leaving the rail and landed on a force plate (AMTI model BP12001200) located at the base of the rail. A successful landing trial was defined when subjects landed on top of their skateboards with all four wheels completely hitting the force plate before eventually rolling off the plate without falling. Each subject performed only 3 trials due to the violent nature of these landings. All subjects wore the same model of skate shoes. However, each subject used his own skateboard for reasons of safety. GRF and moment data were collected for 8 seconds at 1000 Hz for each landing. A 4th order low pass Butterworth filter was applied to the data with a cutoff frequency of 100 Hz.

Center of pressure (COP) coordinates (x,y) were calculated using force and moment data from the force plate and the manufacturers specified vertical offset (Zoff) of the plate (see equations below). By plotting the coordinates against each other we could track the direction of motion of the COP across the
force plate. A best fit line was calculated for a portion of the data when the entire skateboard was still on the plate and the system was relatively stable. Usually this portion of data occurred towards the end of the trial just before the skateboard’s front wheels rolled off the force plate. The slope of the best fit line was calculated and an angle of motion of the COP was calculated against the original Y-axis of the force plate. The cosine of this angle was then multiplied to the original Fx and Fy GRF data to calculate the new N/TGRF and F/BGRF data.

$$\text{COP}(x) = \left[ \frac{M_y + (Z_{off} \times F_x)}{F_z} \right] \times (-1)$$

$$\text{COP}(y) = \left[ \frac{M_x - (Z_{off} \times F_y)}{F_z} \right]$$

**RESULTS**

The F/BGRF and the N/TGRF results were quite variable both within subjects and between subjects. The only regular feature of the F/BGRF curve we can report is that the sinusoidal shape of the curve seemed characteristic of an oscillation: an expected result given the nature of stabilizing body movements. The average offset angle between the skateboard and the force plate was found to be 11.6º ± 5.5º, though offset angles across all trials and subjects ranged between 0.8º and 29.2º. Mean BGRF forces during the landings were found to be 944.6 ± 145.3 N, while mean FGRF forces were found to be 549.6 ± 210.1 N. N/TGRF results were found to be relatively minimal due to the fact the wheels on the skateboard were free to roll upon landing. Mean TGRF forces during the landings were found to be 489.1 ± 258.3 N, while mean NGRF forces were found to be 51.1 ± 31.0 N.

**DISCUSSION**

Shear forces in the landing trials were found to be quite variable both within and between subjects. This seemed largely due to two factors: the instability of the skaters on their boards shortly after landing and slight differences in orientation of the board relative to the force plate. Upon first landing, the subject may not be centered on their board in the transverse (front-back) and frontal (nose-tail) planes. The instability of these landings, both front to back and nose to tail, required the subject to make corrective movements to stabilize their position on the board. These stabilizing movements, which were different for each subject and each landing, created, in part, the observed variability we measured.

By examining the data from the trial with the greatest offset angle (29.2º), we found that differences in the skateboard’s orientation upon landing can affect shear force variability within a subject by as much as 13%. This suggests that the magnitude of off-angle-landing errors may indeed be significant. Our enthusiasm for this conclusion is somewhat tempered by the fact that the COP coordinates take some 20 to 30 ms to find a consistent orientation and that this may undermine the estimation of an accurate offset angle. Nevertheless, such off-angle landings seem common in other sports as well as skateboarding and snowboarding and our data suggest that this may lead to significant errors in the reporting of shear forces. We propose the use of similar cosine error correction procedures and reporting conventions that more accurately convey the body’s orientation to the force plate’s orthogonal planes. Compared with the conventional approaches these modifications should enhance our collective understanding of shear forces and the effects they have on the body.

**REFERENCES**

FOOTWEAR VIRTUAL SIMULATOR FOR THE ANALYSIS OF COMFORT AND FUNCTIONALITY IN NEW DESIGNS OF SHOES.

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¹MediClab, Universidad Politécnica de Valencia, Spain.
²IBV, Instituto de biomecánica de Valencia, Spain.

ABSTRACT
We present in this paper a multidisciplinary project that uses computer graphic techniques (Virtual Reality and Simulation) to create a new design and evaluation methodology of footwear virtual prototypes without the need to construct the real prototype. This permits to reduce the time and money required to launch new products into the market. There are not available methodologies currently to perform this kind of virtual evaluation with the footwear.

INTRODUCTION
At the market level, consumers demand more and more comfort and functionality in the footwear. As a consequence, these aspects are taken into account by the footwear companies in the design of a new shoe collection. For the functional evaluation of the footwear, there are two kinds of tests: subjective tests and objective tests. Subjective tests are based on the user perceptions with feet that are standard both in form and size (Shackel et al., 1969; Corlett, 1981). Objective tests are based on the measure of biomechanical variables during the use of the shoes in real or simulated conditions (Alcántara et al., 2001; González et al., 2001). Objective tests provide useful information about the functionality of a shoe but they are costly and time-consuming. In this context, the use of the virtual reality and the simulation appear as the alternative procedures. In this paper, we describe SIMUCAL, a new application to perform comfort tests in a virtual way. SIMUCAL is a computer application developed by the research group MediClab in collaboration with the IBV.

Figure 1.- Animation generated by the Simulator that allows a functional evaluation: frame(0) (left), (30) (middle) and (50) (right).

Figure 2.- Animation generated by the Simulator that permits a aesthetic evaluation: frame(0) (left), (15) (middle) and (30) (right).
METHODS
The proposed software has two separated tools: the Aligner and the Simulator. The Aligner aligns a virtual shoe with a real foot that has been scanned. It also performs an initial adjustment of the virtual shoe and the scanned foot. The Simulator takes the output of the Aligner as its input. The Simulator generates an animation that shows the pressures generated by the shoe-foot interaction in a complete step. The Simulator also generates (using Shaders and BumpMapping) a photorealistic visualization of the shoe behavior in the complete step. In order to simulate the behavior of the virtual shoe in the Simulator, we developed a new deformable model based on Finite Element Models (FEM). This deformable model simulates the real mechanical behavior of the shoe upper and computes the force intensities in the contact zones between the foot and the shoe.

RESULTS
Results are shown in Figure 1&2. Figure 1 shows a part of an animation during a functional study in which the distribution of forces at the contact zones is shown. Figure 2 shows a part of an aesthetic evaluation where the deformation of the design could be studied.

CONCLUSIONS
As a result of this work, we have obtained an innovative simulator prototype that enables footwear designers to perform virtual evaluations (in a virtual environment) of the functionality and aesthetic of their designs. The developed application offers the subsequent advantages with respect to other R+D projects and existing commercial software:
1. The final evaluation of the prototype is not only static (Hwang et al., 2005; Kos and Donovnik, 2002; Witana et al., 2004) but also dynamic because it can be performed during a complete step.
2. The scanned foot is a real foot without a standard form.
3. For the virtual evaluation of the shoe-foot fit, it has been developed a deformable model that simulates the physical features of the shoe upper. This is a novel approach since the majority of the biomechanical research related to the foot has been focused on either measuring the plantar pressure distribution (Onwuanyi, 2000; Holewski et al., 1989; Verdejo and Mills, 2004; Chen et al., 2003) or a static analysis of the differences between the standard form and the scanned foot.

REFERENCES
A SIMPLIFIED FOOTWEAR SOLE STABILITY EVALUATION METHOD ON TRAIL RUNNING

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ASICS Corporation

INTRODUCTION
Nowadays, trail running, which is a running on rough roads in forest and mountain is getting popular. Footwears’ requirement properties depend on surface conditions. In general, running shoes are designed based on normal running (NR) motion on the flat road surface. However, trail running face to many different conditions such as uphill, downhill and so on. Judging from each condition on trail running, it can be conformed severe pronation on sloped downhill running (SDHR) as shown in fig.1. This indicates that sole stability is required trail running shoes to reduce running injuries. In this study, we carried out motion analyses on SDHR and the sole stability evaluation method based on acceleration history is proposed. In order to check the validity of the proposed method, the results are compared with that obtained from motion capture system.

MOTION ANALYSIS
For measurement of stability on SDHR, motion capture system (VICON-MX, oxford) was used at a sampling frequency of 200 Hz. Reflective markers are attached to the foot and shank illustrated in fig.2. Inversion-eversion (β) and adduction-abduction (γ) of calcaneus relative to tibia were calculated (Grood et al., 1983). Fig.3 shows the example of β and γ histories, it is conformed that the maximum β and γ (|β| and |γ|) emerge during 30-40 % phase of stance. On flat road surface, |β| generally occurs during 15-20 % phase of stance (Nishiwaki et al., 1999). Namely, this indicates that |β| appearance time on SDHR delay depends on surfaces.
EVALUATION METHOD

A lot of methods for the purpose of and measurements by using motion capture system have been reported (Michelson et al., 2002). However, these are many restrictions such as weather and location for outside measurements due to measure various movements directly. Therefore, it is required a simpler evaluation instead of the conventional methods. In this study, we focused on acceleration history during SDHR. In order to obtain the accelerations, the accelerometer (Kyowa Electronic Instruments, Co.) was attached 100mm apart from the tibial plateau on the shank as shown in fig.4. 3 types shoes with different sole stiffness and upper were used. The histories of acceleration outputs are measured for two subjects at a sampling frequency of 5kHz. The typical result of acceleration history is also shown in Fig.3.

Figure 3 shows the comparison of acceleration history with \(\beta\) and \(\gamma\) on SDHR. Judging from this fig.3, it is conformed that \(|\beta_{\text{max}}|\) and \(|\gamma_{\text{max}}|\) is affected by the hatched area in acceleration history. In the stability evaluation, parameter \(V\) was defined as the following Eqn1.

\[
V = \int_{A}^{B} \text{Accel}(t) dt
\]  
(Eqn.1)

\(V\) is equivalent to hatched area. Relationships between \(|\beta_{\text{max}}|\) and \(V\), and between \(|\gamma_{\text{max}}|\) and \(V\) checked in fig.5. It is confirmed that \(|\beta_{\text{max}}|\) and \(|\gamma_{\text{max}}|\) are increasing with \(V\) increases. These tendencies indicate that \(V\) obtained from an accelerometer is effective for the evaluation of sole stability in SDHR.

Fig.5 Relationship between \(|\beta_{\text{max}}|, |\gamma_{\text{max}}|\) and \(V\)

REFERENCES

James D. Michelson, Andrew J. Hamel, et al., Kinematic behavior of the ankle following malleolar fracture repair in a high-fidelity cadaver model, J. Bone and Joint Surgery. 84: 2029-2038
The purpose of this study is to examine the immediate effects of footwear design on postural sway. This study is part of a larger investigation examining Birkenstock® footwear technologies. Data was collected on 20 different subjects wearing five different shoe designs. The goal was to examine the effects of overall shoe design as well as shoe upper design. Three Birkenstock shoes were used to evaluate upper shoe design effects: the Arizona Sandal (AP), the Boston Clog (BO), and the London Shoe (LO). Note that all three models have identical footbed technology with pronounced arch support. Three shoes from different manufacturers were investigated to compare overall shoe design effects: the New Balance walking shoe (NB573), the Land’s End Leather Comfort Casual Oxford (LE), and the Birkenstock LO.

**METHODS**

Center of pressure (COP) data (Figure 1) was collected on each subject in 5 different shoe conditions. Each subject was asked to stand in their comfortable stance on a Kistler™ force plate, sampled at 120 Hz, for a total of 1 minute with only the last 40 seconds of data analyzed to eliminate transient effects (Carroll, 1993). The subject’s feet were traced onto a sheet of paper overlaying the force plate to ensure repeatability in subject’s foot position across all trials. A total of 3 trials were collected for each shoe condition. The best-fit elliptical and circular areas of COP sway were calculated. The deviation from an assumed ideal COP position was estimated by calculating the midpoint between the left and right feet at the posterior third of the foot length. Statistical analysis included a 2-way mixed effect Analysis of Variance and the Bonferroni-Dunn post hoc tests with significance at P < 0.05.

**RESULTS AND DISCUSSION**

In the comparison of overall shoe design effects, no significant differences were observed (see Table 1). Upper shoe design effects (see Table 2) yielded significantly reduced mean circular and elliptical sway areas. In specific, circular sway area was significantly lower in the LO compared with the BO. In this same comparison, the deviation from the assumed ideal COP was statistically smaller for the clog (BO) than the sandal (AP).

**CONCLUSIONS**

In the overall shoe design comparison, there was no statistically significant difference in postural sway across the footwear evaluated. However, in the upper shoe design comparison, postural sway was affected. When wearing a sandal (AP) with minimal upper support or a clog with only toe box support, postural sway was increased compared to the shoe (LO) which included rearfoot support.
REFERENCES

Table 1: Shoe Design Description and Parameters

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Shoe</th>
<th>Mean</th>
<th>SD</th>
<th>p-value</th>
<th>Post-Hoc</th>
</tr>
</thead>
<tbody>
<tr>
<td>Circular Area (mm²)</td>
<td>LO</td>
<td>359.98</td>
<td>184.96</td>
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<td>352.99</td>
<td>187.43</td>
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<td>LE</td>
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<td>184.96</td>
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<td>Elliptical Area (mm²)</td>
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<td>82.82</td>
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<td>NB</td>
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<td>LE</td>
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<td>82.82</td>
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<td>Deviation (mm)</td>
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<td>LE</td>
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<td>5.72</td>
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Table 2: Shoe Upper Design Description and Parameters

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<thead>
<tr>
<th>Parameter</th>
<th>Shoe</th>
<th>Mean</th>
<th>SD</th>
<th>p-value</th>
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<tr>
<td>Circular Area (mm²)</td>
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<td>428.47</td>
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<td></td>
<td>BO</td>
<td>459.38</td>
<td>181.56</td>
<td></td>
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</tr>
<tr>
<td></td>
<td>LO</td>
<td>301.59</td>
<td>181.56</td>
<td></td>
<td>b</td>
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<td>Elliptical Area (mm²)</td>
<td>AP</td>
<td>179.47</td>
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<td>BO</td>
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<td></td>
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<td>LO</td>
<td>139.36</td>
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<td>Deviation (mm)</td>
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<td>BO</td>
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<td>LO</td>
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THE EFFECT OF SHOE SOLE MODIFICATIONS ON LOWER EXTREMITY LOADING IN CHANGE OF DIRECTION TASKS ON ARTIFICIAL TURF

Elissa J. Phillips, Uwe G. Kersting
Department of Sport and Exercise Science, The University of Auckland, New Zealand

INTRODUCTION
The rapid increase in usage of artificial turf surfaces at recreational and professional levels has lead to the development of various turf products. However, there is little understanding of the interactions between athletes and sport surfaces. Previous research has focused on mechanical impact or rotational friction testing as a means of assessing artificial surfaces. Translating such results directly into recommendations may be misleading (Nigg & Segesser, 1988). Subject testing has the advantage of evaluating shoe-surface combinations using actual loading as created during these movements (Dixon et al., 1999). Only very few studies exist where joint loading using athletes was assessed. Examinations of knee loading in the literature is subject to great variability due to individual variation and limitations of experimental protocols used (Shorten et al., 2003).

The aim of this study was to examine shoe sole affects on lower extremity mechanics during several change of direction tasks on artificial turf focusing on forces and moments acting at the knee joint.

METHODS
Fifteen experienced subjects were asked to perform repeated trials of three tasks in the laboratory: side CUT, TURN and STOP. Athletes were tested on medium pile artificial turf with sand infill. Tasks were initiated by a symbol randomly displayed on a screen at one end of the lab after the subject ran through a set of timing gates at full effort, 2m before reaching the force plates. A turf shoe (Nike Total90 Shift) was used in three modifications: Normal; no alterations, Custom; several pimples were removed, creating an arrow shaped pattern, Short; all pimples where shaved down to approximately half their original length. Mechanical properties of the shoe sole were assessed by translational and rotational friction experiments on the turf surface. Athletes’ perception of traction and comfort were assessed by ratings on visual analogue scales following each shoe condition.

Retro-reflective markers were fixed to lower limb landmarks to record three-dimensional lower limb movements by using an 8-camera VICON motion analysis system (Oxford Metrics Ltd., Oxford, UK) capturing at 250Hz. A customized model was created using Vicon BodyBuilder and three-dimensional kinematics and kinetics at the ankle and knee joints of the front leg were obtained.

Waveforms of loading parameters over stance were analysed with a generalised linear model using regression equations and multivariate analysis of coefficients of variation (MANCOVA) (Nelder & Wedderburn, 1972) in SAS (9.1, SAS Institute Inc, 2003). For TD and maximum/minimum key values, a separate repeated measures Analysis of Variance (ANOVA) was performed for each movement using NCSS (NCSS 2001). A value of p <0.05 was chosen to established significant differences and interactions of independent variables.

![Figure 1: Example of knee varus-valgus torques for three shoe conditions in one subject.](image-url)
RESULTS
The waveform analysis did not reveal any significant overall effects of shoe sole modification on any of the joint force or torque curves at the knee. However, in 11 subjects significant interactions of shoe and movement and/or component of the respective parameter were shown. Five subjects revealed substantial changes in peak valgus loading which reached 80% in one case indicating that some subjects experience dramatic changes in critical joint loading components (Fig. 1). In Table 1 all of the key/TD values are listed which demonstrated significant shoe effects. These represent exclusively changes to ankle joint kinematics and anterior posterior ground reaction force (GRF).

Table 1: Min/Max Values and TD Key Events with Significant Shoe Effect For All Tasks: Mean (sd) F ratio, p value, group difference and subject interaction p value.

<table>
<thead>
<tr>
<th>Task</th>
<th>Parameter</th>
<th>Cust</th>
<th>Norm</th>
<th>Short</th>
<th>Fratio (2,28)</th>
<th>p value</th>
<th>diff</th>
<th>subject int</th>
</tr>
</thead>
<tbody>
<tr>
<td>CUT</td>
<td>Ank VV TD</td>
<td>11.4 (9.1)</td>
<td>9.7 (8.6)</td>
<td>8.9 (9.9)</td>
<td>3.71</td>
<td>0.037*</td>
<td>s-c</td>
<td>0.03138*</td>
</tr>
<tr>
<td></td>
<td>Ank IETD</td>
<td>-4.1 (5.2)</td>
<td>-2.0 (4.7)</td>
<td>-1.4 (8.1)</td>
<td>4.73</td>
<td>0.017*</td>
<td>c-n,s</td>
<td>0.00573*</td>
</tr>
<tr>
<td></td>
<td>AnkAng IE</td>
<td>-16.6 (10.3)</td>
<td>-13.5 (5.6)</td>
<td>-13.6 (5.4)</td>
<td>5.52</td>
<td>0.009*</td>
<td>c-n</td>
<td>0.00738*</td>
</tr>
<tr>
<td>TURN</td>
<td>AnkAng VV</td>
<td>-16.8 (13.8)</td>
<td>-18.7 (14.4)</td>
<td>-18.7 (14.4)</td>
<td>3.92</td>
<td>0.032*</td>
<td>c-s,n</td>
<td>No</td>
</tr>
<tr>
<td></td>
<td>GRF AP</td>
<td>-19.0 (5.7)</td>
<td>-19.6 (5.7)</td>
<td>-18.5 (5.6)</td>
<td>3.77</td>
<td>0.036*</td>
<td>s-n</td>
<td>No</td>
</tr>
</tbody>
</table>

DISCUSSION
Results indicate that some subjects adapt substantially to changes in shoe sole condition while others keep their knee joint kinetics virtually unchanged. Subjects may be split into two groups such as responders vs. non-responders as suggested for running by Reinisch et al. (1991). At the same time significant shoe effects for the ankle joint kinematics were observed. It could be speculated that individuals attempt to alter initial conditions in response to changes in shoe sole properties but dependent on factors which are still to be determined these changes lead to joint loading effects or not.

CONCLUSIONS
Based on these results it is obvious that athletes modify kinematics and kinetics between shoe-surface combinations. The individuality within this interaction makes it difficult, if not impossible, to provide recommendations as to the ideal footwear for a given surface. Rather than concluding that subject testing has limited use, research to increase the understanding of subjects’ adjustments is required.

REFERENCES
STUDY OF EVALUATING METHODS ON JOGGING SHOES
Liu Jingmin\textsuperscript{1}, Zheng Xiuyuan\textsuperscript{1}, Cai Yuhui\textsuperscript{2}, Liu Hui\textsuperscript{3}, Jin Jichun\textsuperscript{3}
\textsuperscript{1}Tsinghua University, Beijing; PRC; \textsuperscript{2}Institute of Education and Sports Beijing Normal University, Beijing, PRC; \textsuperscript{3}Beijing Sport University, Beijing, PRC

INTRODUCTION
How to choose a pair of good jogging shoes is very important for the runners. Good shoes can absorb impact forces much and keep comfortable, even can improve the runner’ performance. Therefore, the methods of testing and evaluating a pair of jogging shoes were our study. The aim of this study was to investigate the effect of six different running shoes on different biomechanics variables during walking and running. Particularly, if these variables about the plantar pressures and the comfortableness could be grouped differently was studied. Furthermore, we tried to find the best representative variable in all variables to evaluate running shoes.

METHODS
This research measured six different brands of jogging shoes and gym shoes (similar to the barefoot condition). Number the shoes as: A(Do-win marathon shoes), B(Do-win jogging shoes), C(Li-ning jogging shoes), D(Reebok jogging shoes), E(Anta jogging shoes), F(Double-star marathon shoes), G(gymnastics shoes). Twenty-five well trained Chinese runners were selected for the study, including 15 Male runners (aged 21.2±2.3 yrs, height 179.1±3.8cm, weight 66.8±6.0 kg) and 10 female runners (aged 20.0±1.3 yrs, height 167.6±4.2 cm, weight 53.9±3.7 kg). Subjects were allowed a 5-10 min warm up to be familiar with treadmill running. Every athlete walked or ran in each of the seven different brand shoes on treadmill at three different constant running speeds (1.5m/s, 3m/s, 5m/s). Ground reaction force and pressure distribution data under the plantar heel surface of the feet were collected with a PEDAR pressure insole system with 99 sensors each produced by Germany Novel Company. Five trials were performed for each condition and mean values were calculated for ground reaction force(GRF), heel maximal force (MF), peak pressure(PP), mean pressure(MP), maximal touched area(TA), pressure-time integral(PTI). A repeated measures ANOVA was used to test for significant differences (p<0.05).

RESULTS
Maximal ground reaction forces of the whole foot and the heel were selected as the index of the impact forces absorption measure and evaluation. All running shoes had absorbed the ground reaction force compared with the gymnastics shoes (similar to the bare foot) (p<0.05). Six brand shoes ranked simply by the mean of whole maximal ground reaction forces under three speeds were as follows: walking(1.5m/s): C<A<B<E<D<F; jogging(3m/s): B<A<E<C<D<F; running(5m/s): A<E<B<C<D<F. Ranked simply by the mean of heel maximal ground reaction forces were as follows: walking(1.5m/s):B<C<A<E<F<D; jogging(3m/s): A<B<C<E<F<D; running(5m/s): A<B<C<E<D<F. The results showed that A could absorb the ground reaction forces better while running.

The perceived comfort level related well with the peak pressure measurements on the plantar surface. Although results were not statistically significant, according to the peak pressure measurements seven brand shoes ranked simply by the comfortableness under three speeds were as follows: walking(1.5m/s): D>A>E>C>B>F>G; jogging(3m/s): B>A>D>C>E>F>G; running(5m/s): B>D>C>E>A>F>G. (See figure 1). According to the mean of whole average pressure, ranked simply by the comfortableness under three speeds were as follows: walking(1.5m/s): D>E>A>C>F>B>G; jogging(3m/s): B>E>A>C>D>F>G; running(5m/s): B>C>D>A>E>F>G. The results showed that B had the lowest pressure and felt comfortable most while running, in the other hand, F and G felt comfortable least.
The maximal touched area of the shoes with feet was decided by the difference of each insole surface. Suppose the touched area was more, the perceived comfort level was higher. Seven brand shoes ranked simply by the mean of whole maximal touched area under three speeds were as follows: walking(1.5m/s): B>C>D>F>A>E>G; jogging(3m/s): B>D>E>C>A>F>G; running(5m/s): B>D>A>E>C>F>G. The results showed that B was the most comfortable shoe under all three speeds.

The runner’s work cost can be measured by pressure-time integral better. If this value was too much the cumulative pressure will be harmful to feet. Therefore, the shoes with the least pressure-time integral in this study will be the best. Seven running shoes ranked simply by the mean of whole pressure-time integral under three speeds were as follows: walking(1.5m/s): A<D<E<C<F<B<G; jogging(3m/s): A<B<C<D<E<F<G; running(5m/s): A<B<C<D<E<F<G (See figure 2). The results showed that A was the least pressure-time integral shoe under all three speeds. A had a better protective role to runners. G was the worst one. The result was the same as the questionnaire to the feelings of subjects. They will not feel much tired very soon when wearing A than the others.

In order to evaluate the different running shoes wholly, the cumulative scores were calculated. The original score was the same as the rank of the six indexes which were ground reaction force, heel ground reaction force, peak pressure, mean pressure, maximal touched area and pressure-time integral values. The separate total scores were: A-41, B-44, C-59, D-64, E-68, F-102. Therefore, the final evaluation to six running shoes under the impact force absorption and comfortableness by the rank from good to bad was: A, B, C, D, E, F. A ranked first and it was the most adaptive to long running.

![Figure 1. Peak pressure of seven shoes under three speeds](image1.png)

![Figure 2. Pressure-time integral of seven shoes](image2.png)

**CONCLUSION**

The methods of testing and evaluating a pair of jogging shoes can used the six indexes which were ground reaction force, heel ground reaction force, peak pressure, mean pressure, maximal touched area and pressure-time integral. They can be grouped three. The impact force absorption can be evaluated by the whole and heel plantar maximum forces. The comfortableness can be evaluated by peak pressure, mean pressure and maximal touched area. The economic work can be evaluated by pressure-time integral which has also the same result with the whole evaluation by six indexes. Therefore, the plantar pressure-time integral can represent for all six indexes in this study. The plantar pressure-time integral may be one of the best indexes to evaluate a pair of running shoes.

**REFERENCES**


**ACKNOWLEDGEMENTS**

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THE VARIABILITY OF BIOMECHANICAL PARAMETERS DURING RUNNING IN DIFFERENT RUNNING SHOES - A 3-DAY TESTING DESIGN

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INTRODUCTION
Footwear development is characterized by several test procedures that involve mechanical, subjective and biomechanical studies. Whereas mechanical testing is shown to be reliable and highly objective, the results of subjective and biomechanical studies are characterized by the variability that is produced by the individuality of the subjects [1]. Since most studies are performed on several following days with a large number of shoes and subjects, it has to be proved if the intraindividual and day to day variability of parameters during running allows the repeated measures design. The purpose of this study was to analyze the intraindividual variability of biomechanical parameters over 5 trials in different running shoes on three different days. Furthermore, the day to day variability of 9 subjects as a group was analyzed to verify the significance of this test design for the description of biomechanical properties of running shoes.

METHODS
Nine male subjects with mean age of 25 years (SD 2.1), height of 175 cm (SD 5.2) and body mass of 71 kg (SD 6.4) were recruited. The subjects were well-trained runners with good experience in laboratory running measurements. The experimental setup consisted of a force platform (KISTLER 9287BA; 60x90cm) that was integrated into a 13-m indoor running surface. A miniature accelerometer was attached to the medial aspect of the tibia at mid location between malleolus and the tibial plateau. A light weight electrogoniometer was fixed to the shoe heel counter with its axis of rotation at the approximate height of the subtalar joint [2]. Vertical force rising rate (FRR) and peak vertical impact force (PVF) were calculated. Peak tibial acceleration (PTA), the range of total pronation (TPR) and the maximum pronation velocity (MPV) were calculated. Running speed was set at 3.5 m/s (±0.1). Data of each runner were collected for a total of 5 successful trials. Before starting the measurements, each subject performed several running trials to adapt to the lab running conditions. The methodological protocol involved three different running shoes. Two pairs were identical models (A1, C1), the third pair (B1) was a different brand with differences in constructional and mechanical properties. This was assured by mechanical impact testing as described in [3]. Since the individual preferred shoe brand may influence the individual running procedure, subjective bias was eliminated by blinding all shoes. The testing procedure was repeated for all subjects under the same conditions and the same time on day 1, 3 and 5 of the same week. For each subject, a 5-trial mean value was calculated for each parameter, for each shoe and every day. In order to test the reliability between days over all subjects, group mean values were calculated for each parameter, each shoe and every day. Furthermore, intraindividual variability within five running trials was quantified by the coefficient of variation (COV) for all conditions (parameters, shoes, days). Over all subjects a mean COV was calculated for all conditions (parameters, shoes, days). Multivariate analyses were used to perform statistical evaluation.

RESULTS & DISCUSSION
The intraindividual variability within five trials ranges from 0.5% to 34.5% between subjects overall. A wide range of intraindividual variability between subjects could be observed for all investigated parameters in all shoes and every day.
Nevertheless, between the biomechanical parameters there are differences in intraindividual variability (Tab. 1). The lowest variability was found for PVF, whereas the highest intraindividual variability could be observed for PTA and MPV. This result could be evaluated for all three days and all shoes. The statistical analysis revealed no significant differences between the day to day intraindividual variability (COVs) of the biomechanical parameters and between shoes. Referring to the results of the biomechanical group parameters, no significant differences could be revealed between the three days (Fig. 1). This result was observed for all evaluated parameters and all shoes. This result indicates the reliability of the test procedure. Furthermore, over three days highly significant differences could be found when comparing the shoes. The multivariate analysis revealed significant differences for FRR, MPV (p<0.01) and PTA (p<0.05) between A1-B1 and C1-B1 over three days. Therefore, independently from day to day and individual variability, reliable results for the description of biomechanical properties of running shoes could be evaluated. This result is confirmed by at least significant correlations (p<0.05) between the identical shoes (A1 and C1) for all biomechanical parameters. Although the running shoes (A/C versus B) are considerably different in mechanical properties, no significant differences for PVF and TPR could be determined between A1-B1 and C1-B1.

<table>
<thead>
<tr>
<th>day parameters</th>
<th>A1</th>
<th>C1</th>
<th>B1</th>
<th>MEAN</th>
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<tr>
<td></td>
<td>Ø COV</td>
<td>SD</td>
<td>Ø COV</td>
<td>SD</td>
</tr>
<tr>
<td>1 PVF</td>
<td>7.75</td>
<td>4.09</td>
<td>6.61</td>
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</tr>
<tr>
<td>1 TPR</td>
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<td>4.89</td>
<td>9.91</td>
<td>7.48</td>
</tr>
<tr>
<td>1 MPV</td>
<td>15.99</td>
<td>6.15</td>
<td>18.01</td>
<td>9.46</td>
</tr>
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</table>

Tab. 1: Intraindividual variability over 5-trials with nine subjects (Ø COV), one day

Fig. 1: Vertical force rising rate over 3-days for three shoes

CONCLUSION

The study demonstrates that the test design produces reliable results for the description of biomechanical properties of running shoes independently from day to day and individual variability. In aspect of the distinct correlation of biomechanical parameters between identical, but blinded shoes it may be concluded that the running style was unchanged and not influenced by an unconsciously adaptation due to a preferred brand. Further studies may investigate the significance of biomechanical parameters that are mainly influenced by running style versus parameters that are influenced by shoe properties.

REFERENCES


ACKNOWLEDGEMENTS

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FOOTWEAR FITTING: RELATING OBJECTIVE AND SUBJECTIVE MEASUREMENTS
A PILOT STUDY

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2Department of Orthopaedics and Traumatology, Faculty of Medicine;
The Chinese University of Hong Kong, Hong Kong, China;

INTRODUCTION
Fitting in footwear is not yet generally standardized. It is a kind of complex which involves different shapes and ought to be more than just length and width. Studies concerning subjective fitting were seldom reported. The purpose of this study is to relate two footwear fitting measurements: subjective from questionnaire and objective from pressure sensors.

METHODOLOGY
Five physically healthy male (age = 21.8 ± 1.9 yr, mass = 70.6 ± 7.7 kg, height = 1.79 ± 0.02 m) were recruited as subjects. Their foot sizes were limited to be within the range of 26-27.5 cm. All subjects were injury free at the moment of study and no injury history to be result in abnormal gait.
Subjects were asked to evaluate the tightness or looseness of the running shoes by filling the fit questionnaire which was designed by using a visual analogue scale (VAS)1 with 150mm. There were totally 12 questions in the questionnaire which include: Q1) overall fit Q2) free space in front of toe Q3) fit at the side of forefoot Q4) upper forefoot fit Q5) overall forefoot fit Q6) height of dorsal arch Q7) height of plantar arch Q8) fit at the side of mid-foot Q9) overall mid-foot fit Q10) free space at the back of rear-foot Q11) fit at the side of the rear-foot Q12) overall rear-foot fit. Subjects were required to stand for 30 seconds and then fill in the questionnaire. Subjects were first assess the fit of the control shoes and followed by randomized trial of the five testing shoes.
Twelve flexible sensors were attached on the right foot of the subject on the following location: P1) tip of toe 1 P2) metatarsal tibiale P3) tip of toe 2 P4) toe 1 Joint P5) toe 5 joint P6) metatarsal fibulare P7) instep height P8) navicular P9) tuberosity of 5th metatarsalis P10) pternion P11) medial calcaneou P12) later calcaneous. Pressure data were collected for each testing and control shoe while subjects were asked to stand steadily for 5 seconds.
A set of fit data (Q1-Q12) with 12 questions were collected for each testing shoe of each subject. The corresponding pressure values (P1-P12) from 12 sensors were exported from the mode during 5 seconds standing period. Correlation between fit data and pressure data were calculated by Pearson Correlation. Pressure data were used to predict each of the fit data by linear stepwise regression.

RESULTS
Pearson correlation showed that a few fit and pressure data were fairly correlated. Correlation values and their corresponding significant levels were indicated in table 1 (only significant values were shown). Eight prediction models were constructed and were shown in table 2.
Table 1 – Correlation between fit and pressure data

<table>
<thead>
<tr>
<th></th>
<th>P1</th>
<th>P3</th>
<th>P4</th>
<th>P5</th>
<th>P6</th>
<th>P7</th>
<th>P9</th>
<th>P10</th>
<th>P11</th>
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<td>0.022</td>
<td>-0.023</td>
<td>-0.154</td>
<td>-0.311</td>
<td>0.261</td>
<td>-0.138</td>
<td>-0.428(*)</td>
<td>-0.136</td>
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</tr>
<tr>
<td>Q2</td>
<td>0.078</td>
<td>0.151</td>
<td>0.144</td>
<td>-0.315</td>
<td>-0.071</td>
<td>0.328</td>
<td>-0.483(*)</td>
<td>-0.157</td>
<td>-0.315</td>
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<tr>
<td>Q3</td>
<td>-0.071</td>
<td>0.122</td>
<td>-0.048</td>
<td>-0.237</td>
<td>0.149</td>
<td>-0.366(*)</td>
<td>-0.108</td>
<td>-0.144</td>
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</tr>
<tr>
<td>Q4</td>
<td>-0.057</td>
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<td>-0.271</td>
<td>-0.420(*)</td>
<td>0.311</td>
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<td>0.248</td>
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</tr>
<tr>
<td>Q5</td>
<td>-0.018</td>
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<td>-0.034</td>
<td>-0.393(*)</td>
<td>0.187</td>
<td>-0.162</td>
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<td>Q6</td>
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<td>-0.232</td>
<td>0.158</td>
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<td>-0.077</td>
<td>-.434(*)</td>
<td>-0.202</td>
</tr>
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</table>

** Correlation is significant at the 0.01 level (2-tailed)  
* Correlation is significant at the 0.05 level (2-tailed).

Table 2 – Prediction models using 12 pressure values

<table>
<thead>
<tr>
<th>Predic t value</th>
<th>Prediction equation</th>
<th>R²</th>
<th>SEE</th>
<th>Predict value</th>
<th>Prediction equation</th>
<th>R²</th>
<th>SEE</th>
</tr>
</thead>
<tbody>
<tr>
<td>Q1</td>
<td>-2.819 - 24.860(P9) - 3.687(P4)</td>
<td>0.302</td>
<td>14.70</td>
<td>Q7</td>
<td>-51.598 + 7.497(P4)</td>
<td>0.273</td>
<td>21.39</td>
</tr>
<tr>
<td>Q2</td>
<td>-8.550 - 35.198(P9)</td>
<td>0.233</td>
<td>25.36</td>
<td>Q8</td>
<td>-16.086 - 6.562(P7) + 0.889(P6)</td>
<td>0.395</td>
<td>13.27</td>
</tr>
<tr>
<td>Q3</td>
<td>-48.974 + 1.420(P6)</td>
<td>0.147</td>
<td>22.62</td>
<td>Q10</td>
<td>15.482 - 1.885(P10) - 1.815(P5)</td>
<td>0.445</td>
<td>13.90</td>
</tr>
<tr>
<td>Q6</td>
<td>-9.554 - 4.378(P7)</td>
<td>0.134</td>
<td>13.41</td>
<td>Q12</td>
<td>12.756 - 1.847(P10) - 1.287(P5)</td>
<td>0.323</td>
<td>15.64</td>
</tr>
</tbody>
</table>

(Unit: Pressure in psi)

DISCUSSION

The result showed that 16 out of 144 pairs of parameters were significantly correlated. P2, P8 and P12 had no correlation with any questionnaire data which indicated that the bony landmarks: metatarsal tibiale, navicular and lateral calcaneou did not represent the subjective perceptions on footwear fitting. This may due to the particular foot and shoe shape such that there was relatively less space near these bony landmarks. Question 8 had the highest correlation (0.524) with pressure sensor 7. It was reasonable because both parameters evaluated mid-foot. Q2 and P1, Q8 and P6, Q10 and P10, Q12 and P10 were also significantly correlated with 0.474, 0.421, 0.408 and 0.434 respectively. It indicated that question items were significantly correlated with the pressure sensors located in the corresponding fore-foot, mid-foot and rear-foot regions.

In each regression model, only 1 or 2 pressure parameters were used to predict question data. It matched with the correlation results that pressure sensor in specific region was only related to specific question. The explained variance (R²) was around 0.15-0.45. This was an acceptable value because the numbers of subjects were limited. A new method to predict subjective fitting measurements by objective pressure sensors was presented.

REFERENCE

A NEW AND SIMPLE MECHANICAL “SUPINATION SPRAIN SIMULATOR” FOR EVALUATING THE PROTECTIVE EFFECTS OF FOOTWEAR ON ANKLE SPRAIN

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²Department of Mechanical and Automation Engineering, Faculty of Engineering;
The Chinese University of Hong Kong, Hong Kong, China.

INTRODUCTION
Inversion platforms were commonly used to introduce simulated ankle sprains in evaluating the protective effects of brace, tape and footwear¹. Since most ankle sprains are supination sprains, it is essential to initiate ankle supination instead of just inversion to better simulate ankle sprain kinematics. Similar devices were used²,³, however, before the spraining motion, the feet were already plantarflexed instead of at anatomical positions. This study presents a simple free-fall mechanical “supination sprain simulator” which allows a starting anatomical position and different combination of inversion and plantarflexion for better simulating a supination ankle sprain motion.

METHODS
The “supination sprain simulator” has a platform and a rotating disc (radius = 15.5 cm) on top (Fig 1a). The front-medial corner of the platform is cut to allow putting two feet closely in a supination setting (Fig 1c). The rotation axis is at the medial edge. Trapdoors are at the lateral edge to support the platform, which are controlled by a solenoid shutter. When the shutter opens, the platform falls from the lateral edge to introduce a sudden ankle motion (maximum = 30 degrees). The right foot of the subject is fixed on the rotating disc as shown in Figure 1a by straps (not shown). In this position, the device provides an inversion motion as the foot axis is in parallel with the rotation axis (Fig 1b).

![Diagram](image)

Figure 1 – (a) “Supination sprain simulator”; (b) inversion, (c) supination, and (d) plantarflexion settings
When the disc rotates clockwise, or relatively when the platform rotates anti-clockwise, the foot axis starts to make an angle with the rotating axis, and thus introducing a supination motion (Fig 1c). When the platform further rotates to 90 degrees, the device provides a plantarflexion motion (Fig 1d). One subject performed five trials of the three motions captured at 120Hz by a motion analysis system (MotionAnalysis, USA). Ankle joint angles were evaluated in plantarflexion/dorsiflexion and inversion/eversion planes from 0.5 second before until 1 second after the start of the motion.

RESULTS

Ankle joint angle profiles are shown in Fig 2. Two local peaks were found at the first 0.5 seconds in plantarflexion and inversion sprain. The plantarflexion and inversion sprains are not pure single-plane motions but accompanied by motion in each of another plane. The supination sprain shows similar range of plantarflexion (26.0 deg) and inversion (13.5 deg) to the other two sprains.

![Ankle Joint Angles in Three Simulated Spraining Motions](image)

**Figure 2** – Ankle joint angles in the three simulated spraining motions

DISCUSSION

This study presents a simple free-fall mechanical “supination sprain simulator” for studying ankle sprain motions. This study shows that ankle kinematics is different in plantarflexion, inversion and supination sprains. Considering the anatomy and rotating axis of the human ankle joint, a supination sprain should be employed in studying the functions of footwear to protect the foot against sprains.

REFERENCES

INFLUENCE OF MOTION CONTROL SHOES ON LOWER EXTREMITY MOVEMENT AND
PRESSURES BENEATH THE SHOE IN OVER-PRONATORS

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INTRODUCTION

Motion control shoes are typically selected by individuals believed to experience a greater than average amount of rearfoot pronation (eversion). These shoes aim to reduce peak eversion during the initial stance phase by providing medial support to the foot through dual-density midsole materials and medial (varus) wedging. Whilst the introduction of medial wedges to footwear has been demonstrated to reduce peak rearfoot eversion (Perry and Lafortune, 1995), a recent study of a current commercially available motion control running shoe demonstrated no difference in peak eversion compared with a cushioning shoe (Butler et al., 2007). Conflicting results in these studies are likely contributed to by the use of different shoe models and subject selection criteria. Since motion control shoes are directed at individuals with high rearfoot eversion during running, the present study selected subjects with a pronated foot type. This selection was based on the recently developed foot posture index (Redmond et al., 2006) and the measurement of barefoot pressure data during running. Quantification of the influence of different shoe types on lower extremity biomechanics typically involves the collection of kinematic data. It has recently been suggested that pressure data collected from beneath the shoe may be a useful tool for monitoring footwear effects, providing a measure of the medial-lateral foot balance during running (Dixon and McNally, 2005). The purpose of the present study was to investigate the influence of motion control running shoes on lower extremity movement and pressure distribution for individuals identified as over-pronators. It was hypothesized that motion control footwear would reduce peak rearfoot eversion and result in a reduced loading on the medial aspect of the shoe relative to the lateral aspect during early stance.

METHODS

Ten recreational athletes believed to be over-pronators volunteered to participate in the study. Each subject underwent an initial podiatry assessment and foot type was classified using the foot posture index (8 criteria format: Redmond et al., 2006). The foot posture index classified all ten subjects as pronators, with scores ranging from +8 to +13. All subjects were then assessed dynamically using plantar pressure data. Five barefoot pressure time-histories were collected for each foot while running at a self-selected running pace (±5%) over an RSscan pressure plate (500 Hz). These pressure data were used to determine the medial-lateral balance of pressures across the foot during early stance (RSscan D3D®). Within the software the foot was divided into anatomical areas of medial and lateral heel (HM, HL) and metatarsal areas (M1, M2, M3, M4, M5). Heel balance was defined as [HM-HL] ÷ [average force] * 100. Foot balance was defined as [(M1+M2+HM)-(M3+M4+M5+HL)] ÷ [average force] * 100. Subjects with heel balance values greater than 20 during the initial 10% of stance or foot balance values greater than 20 during the period from 25% to 35% stance were assumed to pronate more than average. Based on the combined podiatry assessment, foot posture index and pressure distribution data, all 10 subjects were classified as over-pronators.

Each of the 10 subjects performed 10 running trials under each of two footwear conditions – motion control shoe and cushioning shoe, using the same running pace as selected for the initial pressure.
assessment. Three-dimensional kinematic data were collected at 120 Hz using a Peak realtime system (Peak Technologies, USA). For each running trial, peak angles were determined for rearfoot inversion-eversion and tibial internal rotation. In addition, the balance of pressure on the lateral to medial sides of the shoe was determined using the same procedures as outlined for the barefoot pressure assessment. For heel balance and foot balance, the minimum value during early stance was used to indicate differences in lateral concentration of pressure at initial ground contact and the maximum to indicate the peak concentration on the medial side of the shoe during early stance. Study variables were compared for the two test conditions using paired t-tests (p<0.05).

RESULTS AND DISCUSSION
Wearing motion control shoes compared with a cushioning shoe model resulted in significant changes in kinematic and pressure variables (Table 1). The kinematic data revealed no significant difference in peak eversion between the two footwear conditions (p>0.05). However, there was a significant reduction in internal tibial rotation when wearing motion control shoes compared with a cushioning shoe (p<0.05). These results are consistent with those of Butler et al. (2006), who observed no change in rearfoot eversion, but a reduction in tibial internal rotation when a motion control shoe was compared with a cushioning shoe. The peak lateral heel balance was significantly higher in magnitude for the motion control shoe, indicating a more lateral concentration of pressure at initial ground contact for this footwear condition (p<0.05). The medial heel balance was also significantly higher for the motion control shoe, indicating a greater peak medial concentration of pressure during stance (p<0.05). Foot balance data were not significantly different for the two footwear conditions (p>0.05). The hypotheses that peak eversion would be lower and that pressure data would reveal a more lateral (less medial) concentration of pressure for the motion control shoe were both rejected.

Table 1 – Group mean (±SD) kinematic variables and pressure data for cushioning shoe and motion control shoe (* p<0.05)

<table>
<thead>
<tr>
<th></th>
<th>Peak rearfoot angle (°)</th>
<th>Peak tibial angle (°)</th>
<th>Lateral heel balance</th>
<th>Medial heel balance</th>
<th>Lateral foot balance</th>
<th>Medial foot balance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cushioning</td>
<td>-9.1 ± 2.8</td>
<td>-10.1 ± 3.1</td>
<td>-16.1 ± 13.6</td>
<td>25.1 ± 19.9</td>
<td>-18.9 ± 17.4</td>
<td>30.4 ± 18.8</td>
</tr>
<tr>
<td>Motion Control</td>
<td>-9.7 ± 2.0</td>
<td>-7.9 ± 4.0*</td>
<td>-24.3 ± 9.3*</td>
<td>47.4 ± 11.5*</td>
<td>-24.7 ± 9.2</td>
<td>24.7 ± 13.3</td>
</tr>
</tbody>
</table>

CONCLUSIONS
Overall, the data of the present study do not support the suggestion that a motion control shoe reduces peak rearfoot eversion for over-pronators during running compared with a cushioning shoe. In contrast, kinematic data indicate no difference in rearfoot eversion and pressure data suggest a greater amount of rearfoot movement when wearing the motion control shoe model.

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THE INFLUENCE OF OBESITY ON THE PERCEPTION OF TOUCH AND VIBROTACTILE
THRESHOLDS UNDER THE FOOT

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INTRODUCTION

The foot as a sensory organ has received considerable attention in recent years. Studies by Perry et al. (2000) demonstrated the importance of the mechanoreceptors under the foot for balance control during gait. A reduction of plantar sensation will cause less stability to maintain balance during standing and walking. A study, investigating threshold sensitivities at 30 different anatomical locations, has shown that touch and vibrotactile sensitivity varies considerably across the plantar and dorsal surfaces of the foot (Hennig et al., 2004). As compared to the men women demonstrated a better threshold sensitivity, especially for the vibrotactile stimuli. In general, touch thresholds were reduced under the heel, above the Achilles tendon, medial and lateral malleoli and above the articulatio talocruralis. The most sensitive skin locations were found under the medial arch and the distal phalanges. In contrast to the low touch sensitivity under the heel the vibrotactile sensitivity at this site was surprisingly good. Obese patients have an increased fat pad thickness under their feet (Nass et al., 1999). More fatty tissue underneath the foot may result in an attenuation of external mechanical stimuli on the foot. Reduced sensation could be one of several factors, causing postural instability, as it is often observed in obese patients. This study investigated the influence of overweight on the sensitivity in perceiving mechanical stimuli.

METHODS

38 male and 40 female subjects participated in this study. From the 38 men 16 subjects with a mean Body Mass Index of 34.7 were compared to the control group of 22 subjects with a BMI of 24.7. Similarly, 21 women with a BMI of 36.0 were compared to 19 women with a BMI of 24.0. The chosen anatomical locations (right foot) for the measurement of the sensitivities were the heel (P1), the medial arch (P2), the 3rd distal phalanx (P9), and the dorsum of the hallux (D8). Additionally, sensitivities were also determined at the middle distal phalanx (Finger) at the inside of the right hand. Touch and vibrotactile threshold sensitivities were determined by using “Semmes Weinstein” monofilaments and a “Horwell Neurothesiometer”. The order of measuring the anatomical sites was randomized between subjects. An infrared lamp was used to maintain constant foot temperature. Touch threshold detection was realized with the “Semmes Weinstein Monofilaments”. Including zero force stimuli, threshold recognition was accepted for an anatomical site if four out of five right answers were given (Jeng et al., 2000). A ramp generator was used for a slow and linear amplitude increase of the 50 Hz vibration stimulus from the “Horwell Neurothesiometer”. As soon as the subjects detected a vibration sensation at the measuring site they triggered a push button to record the current amplitude value.

RESULTS AND DISCUSSION

In figures 1 and 2 the results of the perception threshold values from both groups are shown. Because no gender effect was detected the data from the men and women were pooled. Therefore, the results from the obese group is based on 37 subjects, whereas the control group represents the means from 41 subjects. The ANOVA showed statistically significant differences (p<0.01) between the measuring sites as well as between the experimental groups. From the post hoc t-tests the obese group showed for all anatomical sites (except P9) highly significantly increased touch threshold values (p<0.01). For P9 a
significant higher threshold value was found for the obese group (p<0.05). Similarly, the vibrotactile threshold values for the obese group were significantly higher at all anatomical sites (p<0.05).

Figure 1: Touch threshold values (in Semmes Weinstein numbers) at the 6 anatomical locations

Figure 2: Vibration threshold values (in units, representing vibration amplitude) at the 6 anatomical locations

Touch and vibration detection threshold values were higher at all measuring locations in the group of the obese subjects. Therefore, the increase in fat pad thickness for the overweight group may explain the reduced sensitivity in recognizing external mechanical stimuli. Between the anatomical sites a similar sensitivity distribution is present as published previously (Hennig et al., 2004)

CONCLUSIONS
The perception of touch and vibration is reduced in obese subjects. A higher mechanical input is needed for this group to detect external stimuli. This lowered sensitivity in perception may be a contributing factor for less balance control, as it is frequently observed in obese patients.

REFERENCES
EFFECT OF LATERAL HEEL WEDGE INSOLE IN MANAGEMENT OF SUBJECTS WITH OSTEOARTHRITIC KNEE

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2Department of Rehabilitation, Taipei Municipal Wanfang Hospital, Taipei, Taiwan
3Institute of Science and Technology, National Yang-Ming University, Taipei, Taiwan
4Department of Rehabilitation, Shin Kong Wu Ho-Su Memorial Hospital, Taipei, Taiwan

INTRODUCTION
Knee osteoarthritis (OA) is one of major problems in the elderly. Due to overuse and abnormal loading on the knee cartilage surface, it results in knee pain, deformity, and motion disability. The progressive uneven distributed reaction force on knee in walking produces an excessive varus knee moment which induces the medial compartment pain (Stauffer, et al 1997). The typical biomechanical characteristics are a delay and smaller first peak vertical ground reaction force (Yang, et al 2006), weak and delayed response of quadriceps (Birmingham, et al 2001), poor proprioception and postural control, poor static and dynamic balance, large sway of center of pressure trajectory (Kul-Panza and Berker, 2006), as well as having difficulties in functional activities, such as walking, squatting and, rising from a chair (Lo, 2005). To manage the problem, functional knee brace and heel wedge insole is often prescribed. The survey study showed the functional knee brace would significant decrease pain and corrected the femur-tibia angle (Otis et al 1996) and the heel valgus wedge insole could reduce pain and realigned the femur-tibia angle as well as the talar tilt angle (Toda and Segal, 2002). However, controversial result from the gait analysis showed no significant difference on knee varus moment reduction, postural control and static stability with and w/o either brace or wedge insole (Hewett et al 1998; Kakihana et al 2005).

The purpose of this study was to find out if the lateral heel wedge insole would effectively reduce knee pain and improve the ambulatory ability for different severances of OA knee subjects.

MATERIALS AND METHODS
Thirty subjects with medial unicompartmental knee osteoarthritis were recruited from the outpatient clinic of Department of Rehabilitation, Shin Kong Wu Ho-Su Memorial Hospital. The subjects were classified by Kellgren-Lawrence scale and divided into mild group- grade II (4 males, 12 females, Ages 59.7± 10.7 Kgs , weight 62.9±11.6 Kgs, BMI 25.2±3.6, affected sideR:12,L:6 ) and severe group-grade III and IV (7 males, 7 females, ages:65.7±9.6 weight 69.4±10.2 Kgs, BMI 27.5±4, affected side R:8, L6). All participants were asked to wear a pair of base-line shoes, then the custom made 10° lateral heel wedge insoles. The evaluation tasks included a) X-ray evaluation before and after each conditions, b) functional evaluation using VAS (Visual Analogue Scale and WOMAC (Western Ontario and McMaster University Osteoarthritis Index), c) gait analysis on 10m level walking. The kinematic and kinetic data such as range of joint motion, knee moment, peak planter pressure, and center of pressure loci, pressure-time integral and etc. collected from gait analysis (Vicon, MX 13 system, UK), force platform (AMTI, USA), and pressure mat (Footscan, Rsscan, Belgium) were analyzed via two-way ANOVA. The significance of alpha lever was set to 0.05.

RESULTS AND DISCUSSION
The X-ray evaluation before and worn the wedge insole showed the talar tilt angles were significant smaller than wearing the baseline shoes (p=0.044). The WOMAC and VAS evaluation showed that the total score was reduced significantly with the lateral wedge insoles (p=0.005 and p=0.009, respectively). In gait analysis, the walking time, speed and step length increased significantly. With the lateral wedge insoles, the double/ single support time ratio was reduced. The sever group with the insole had fast walking speed and less double limbs support time (Table 1).

The vertical ground reaction forces were significant increased at the phase of loading response when wearing insole. In midstance, the ground reaction forces were increased significantly at the affected side. This might interpret the stronger knee stability when wearing the wedge insole. It also showed significant decrease the knee varus moment, similar results were presented by Kakihana [3] and Kerrigan [4] concluded that the knee OA subjects after wearing 6 to 10 degrees of lateral heel wedge insole for 6 months would help to decrease the knee varus moment at the early stance phase.

CONCLUSION
In gait analysis, the kinematic and kinetic data showed the subjects with various degree of unicompartmont knee OA would increased the dynamic stability and immediate pain relief when wearing the lateral heel wedge insoles. The efficacy was profounded on the mild OA group, the knee varus moment, WOMAC score were reduced singificantly, the ambulatory ability was significantly improved, and the user acceptance was much higher than earing the knee brace.
Table 1: The significant variables for the efficacy of lateral heel wedge insoles

<table>
<thead>
<tr>
<th>Variables</th>
<th>Baseline shoes</th>
<th>Insole</th>
<th>P value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Talar tilt angle from X ray (degree)</td>
<td>6.39±3.64</td>
<td>5.78±3.90</td>
<td>0.044</td>
</tr>
<tr>
<td>Visual analog scale (VAS)</td>
<td>3.3±2.4</td>
<td>1.1±1.8</td>
<td>0.009</td>
</tr>
<tr>
<td>Walking Speed (m/sec)</td>
<td>0.96±0.17</td>
<td>1.01±0.22</td>
<td>0.047</td>
</tr>
<tr>
<td>Stride Length (m)</td>
<td>1.11±0.07</td>
<td>1.18±0.08</td>
<td>0.050</td>
</tr>
<tr>
<td>Step Length (m)</td>
<td>0.54±0.09</td>
<td>0.60±0.07</td>
<td>0.045</td>
</tr>
<tr>
<td>Ankle maximum internal rotation moment (%)</td>
<td>47.87</td>
<td>52.59</td>
<td>0.028</td>
</tr>
<tr>
<td>Ankle maximum abduction moment (%)</td>
<td>45.02</td>
<td>29.90</td>
<td>0.005</td>
</tr>
<tr>
<td>Vertical ground reaction force (N)-Initial contact</td>
<td>56.96±16.45</td>
<td>72.54±19.81</td>
<td>0.004</td>
</tr>
<tr>
<td>Vertical ground reaction force (N)-Loading response</td>
<td>103.93±6.90</td>
<td>106.99±6.99</td>
<td>0.008</td>
</tr>
<tr>
<td>Anterior/Posterior shear force (N)-Midstance</td>
<td>11.29±3.33</td>
<td>13.06±3.44</td>
<td>0.002</td>
</tr>
<tr>
<td>Onset time of medial/lateral shear force (%)</td>
<td>6.13±1.21</td>
<td>8.87±1.93</td>
<td>0.002</td>
</tr>
<tr>
<td>F8-Medial/Lateral shear force (N)-Midstance</td>
<td>5.08±1.47</td>
<td>5.41±1.50</td>
<td>0.009</td>
</tr>
<tr>
<td>V1-Velocity of CoP (m/sec)-Loading response</td>
<td>0.88±0.24</td>
<td>1.10±0.38</td>
<td>0.043</td>
</tr>
<tr>
<td>V3-Velocity of CoP (m/sec)-Terminal stance</td>
<td>0.46±0.22</td>
<td>0.81±0.46</td>
<td>0.014</td>
</tr>
<tr>
<td>A3-Acceleration of CoP (m/sec²)-Terminal stance</td>
<td>4.32±2.11</td>
<td>9.22±4.73</td>
<td>0.036</td>
</tr>
<tr>
<td>Contact time of medial side of foot (%)</td>
<td>91.0</td>
<td>85.8</td>
<td>0.009</td>
</tr>
<tr>
<td>Contact time of lateral side of foot (%)</td>
<td>94.4</td>
<td>88.0</td>
<td>0.001</td>
</tr>
<tr>
<td>Peak plantar pressure of medial heel (N/cm²)</td>
<td>13.4</td>
<td>9.9</td>
<td>0.000</td>
</tr>
<tr>
<td>Peak plantar pressure of lateral heel (N/cm²)</td>
<td>14.0</td>
<td>10.8</td>
<td>0.000</td>
</tr>
<tr>
<td>Peak plantar pressure of medial forefoot (N/cm²)</td>
<td>16.8</td>
<td>14.8</td>
<td>0.000</td>
</tr>
<tr>
<td>Peak plantar pressure of lateral forefoot (N/cm²)</td>
<td>15.7</td>
<td>13.6</td>
<td>0.012</td>
</tr>
<tr>
<td>Pressure-time integral of medial heel (N/cm² sec)</td>
<td>3.77</td>
<td>2.77</td>
<td>0.001</td>
</tr>
<tr>
<td>Pressure-time integral of lateral forefoot (N/cm² sec)</td>
<td>5.69</td>
<td>4.19</td>
<td>0.002</td>
</tr>
<tr>
<td>CoP maximum medial/lateral displacement (mm)</td>
<td>108.59±34.40</td>
<td>90.54±30.16</td>
<td>0.045</td>
</tr>
<tr>
<td>Ground reaction force of sit-to-stand (N)</td>
<td>264.72</td>
<td>173.32</td>
<td>0.022</td>
</tr>
<tr>
<td>CoP anterior/posterior displacement in single-leg standing (mm)</td>
<td>40.1±9.3</td>
<td>52.5±14.5</td>
<td>0.001</td>
</tr>
<tr>
<td>Sound side leading-standing time of tandem gait(s)</td>
<td>22.3±8.4</td>
<td>27.8±11.1</td>
<td>0.001</td>
</tr>
<tr>
<td>Affect side leading-standing time of tandem gait(s)</td>
<td>22.1±8.4</td>
<td>28.3±0.8</td>
<td>0.001</td>
</tr>
</tbody>
</table>

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ACKNOWLEDGEMENT

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PROPRIOCEPTION OF ANKLE JOINT COMPLEX IN YOUNG HOCKEY PLAYERS AND RUNNERS

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INTRODUCTION

Proprioception is the afferent information that contributes to conscious sensation (muscle sense), total posture (postural equilibrium), and segmental posture (joint stability). It is mediated by proprioceptors located in the skin, muscles, tendons, ligaments, and joint capsules (Lehert et al, 1997). Specifically, research found that declined ankle proprioception results from ankle injuries such as sprain, while aging reduces postural stability (Fu, et al., 2005; Pai, et al., 1997). Considering its importance in postural stability and movement control, it would therefore be beneficial to understand the effects of exercise on proprioception function so as to promote injury prevention in athletes and contribute to the rehabilitation of patients with injuries and foot and ankle diseases. However, it is still being debated upon whether or not exercise training can improve proprioception. In particular, long-term Tai Chi practice was found beneficial to the proprioception of the lower limb (Xu, et al. 2004). A recently published work which studied the influence of a five-month professional dance training without concurrent additional coordinative training found that such training did not lead to improvements in ankle joint position sense or improved measures of balance (Schmitt et al., 2005). These studies indicate that exercise may benefit proprioception. However, the effect may be influenced by the type and modality of the exercise.

Hockey involves a very high demand to postural control capacity. Currently, the design of the hockey boot only allows the ankle joint complex to move in a very small range. It was hypothesized that playing hokey has high a demand for movement control in the foot and ankle because of its multi-directional movement and very small supporting base in body. This high demand when wearing hockey boots might result in training effects on the proprioception of the ankle joint complex. Therefore, the objective of the study is to examine the proprioception of the ankle joint complex in hockey players, in comparison with runners. The findings of the study will add to the understanding of the effects of exercise on the proprioception of the ankle joint complex.

METHODS

The running group (R group) was composed of 12 university runners (6 males, 6 females) with an average body weight (BW) of 70.06±11.64 kg, body height (BH) of 171.0±8.36 cm, and body mass index (BMI) of 23.95±4.00. All of them had a regular running exercise habit, running for more than three times each week for more than five years. Meanwhile, the hockey group (H group) was composed of 13 male university hockey players with an average BW of 82.54±7.57 kg, BH of 180.50±6.97 cm, and BMI of 25.30±2.00. They also played more than three times each week for more than five years. The age of the participants ranged from 19 to 21, and all of them were free from foot and ankle injuries. A custom-made device was used to measure ankle kinesthesia during plantarflexion-dorsiflexion and inversion-eversion at 0.4 degree/sec rotation velocity of the ankle joint. The instrumentation and data collection procedures were adopted from the work of Xu and co-workers (2004), although slight changes were instituted. The custom-made device used is a box with a movable platform that rotates on a single axis in two directions. With the foot resting on the platform, plantar-dorsiflexion and inversion-eversion of ankle movements can occur. This platform is moved by an electric motor that rotates the foot on an axis at the rate of 0.4°/sec. Movement can be stopped any time with the use of a hand-held switch. During testing, the subject’s foot was rested on the platform of the device. Using a thigh cuff and a scale of the device, the extremity can be adjusted to make the foot in contact with the platform so that only 50% of the limb weight was borne by the foot. For data collection, each subject was seated on an adjustable chair, and his or her foot was
placed on the platform with the axis of the apparatus coinciding with the plantar-dorsiflexion/inversion-eversion axis of the ankle joint. The hip, knee, and ankle were positioned at 90°. Each test movement began with the foot placed on the horizontal platform, with the starting position at 0°. The subjects were instructed to concentrate on their foot and to press the hand-switch when they could sense motion and identify the direction of the movement. The researcher recorded the rotation angles of the platform and the direction of movements as passive motion sense. At least three randomized trials were conducted in each movement of each foot. Independent t-test was used to examine any significant differences in BW, BH, and BMI between the two groups. Paired t-test was used to examine any significant differences in the measurements of dorsiflexion, plantarflexion, inversion, and eversion between the dominant and nondominant foot of the group. If there was no significant difference in the data from the dominant and nondominant ankle joint complex, the data from both ankle joint complexes were merged together for the analysis. Since the participants in the running group consisted of males and females, independent t-test was used to examine any possible gender difference in the measurements of each movement between the male and female participants. Since no significant difference was found in the measurements between the male and female runners, the data from both genders were pooled together. Independent t-test was used to examine the significant differences in the measurements of kinesthesia in dorsiflexion, plantarflexion, inversion, and eversion between the two groups. The significance level was set at 0.05.

RESULTS AND CONCLUSION
Statistical analysis showed that there were significant differences in BH and BW between the two groups. However, there was no significant difference in BMI between them. In addition, paired t-test did not reveal any significant differences in the kinesthesia measured in dorsiflexion, plantarflexion, inversion, and eversion between the dominant and nondominant ankle joint complexes. Therefore, the data from both ankle joint complexes were pooled together. Further independent t-test analysis of the merged data between the two groups showed that there were significant differences in the kinesthesia measured in dorsiflexion, plantarflexion, inversion, and eversion (Table 1). As compared to the R group, the H group showed significantly better measurement scores of kinesthesia in plantarflexion, dorsiflexion, inversion, and eversion. Furthermore, the results indicate that hockey exercise results in good training effects for ankle proprioception.

Table 1. Kinesthesia measured during dorsiflexion, plantarflexion, inversion, and eversion in the ankle joint complex in the participants of the running group and hockey group (Mean ± SD)

<table>
<thead>
<tr>
<th></th>
<th>Running group (n=12)</th>
<th>Hockey group (n=13)</th>
<th>P value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dorsiflexion (°)</td>
<td>2.53±0.88</td>
<td>1.92 ± 0.65</td>
<td>0.009</td>
</tr>
<tr>
<td>Plantarflexion (°)</td>
<td>2.61±1.12</td>
<td>1.88±0.63</td>
<td>0.006</td>
</tr>
<tr>
<td>Inversion (°)</td>
<td>4.33±2.04</td>
<td>2.79±0.94</td>
<td>0.001</td>
</tr>
<tr>
<td>Eversion (°)</td>
<td>4.76±2.56</td>
<td>3.17±1.12</td>
<td>0.007</td>
</tr>
</tbody>
</table>

REFERENCES
EVALUATING THE TENDENCY FLAT FOOT IN NORMAL ADULTS USING A TWO-DIMENSIONAL COORDINATE SYSTEM

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²Department of Occupational Therapy, Chang Gung University, Tao-Yuan, Taiwan

INTRODUCTION
The height of the four points of the navicular bone (Figure 1) in the foot has been recognized as an important indicator of flat foot. But the methods published for evaluation of the height of the navicular bone was not sensitive enough. The purposes of the present study were to establish a method for the evaluation of the height of the navicular bone, using a two-dimensional coordinate system, and to compare the height of navicular bone between normal males and females. It was hypothesized that the height of the navicular bone in females will be lower than that in males.

METHODS
The bone structure of 28 feet in 14 normal males and 68 flat feet of 34 females were measured. A two-dimensional coordinate system was used to measure the height of the navicular bone of those feet based on weight-bearing lateral radiographies of the feet [1]. The line connecting the lowest point of the first metatarsal head and the lowest point (O) of the calcaneus is defined as the x-axis, and the line that is perpendicular to the x-axis and intersects the x-axis at point O is defined as the y-axis. The y-coordinate of the four points (Figure 1) of the navicular bone was defined by the coordinate system. The average of the y-coordinate of the four points of the navicular was used as the index of the height of the navicular bone. The differences of the index of the height of the navicular bone between groups was examined by independent t-test.

RESULTS AND DISCUSSION
The index of the height of the navicular bone in females was found to be lower than that in males significantly (P<0.05). However, no change was observed in the y-coordinates of the padding of calcaneus and metatarsal. This result suggested that the two-dimensional coordinate system in this study was able to recognize the difference in the height of the navicular between males and females. Therefore, this method might be valuable in identifying flat feet. In another words, the sinking of the navicular might be the primary cause of symptom in subjects with flat foot.

CONCLUSIONS
The coordinate system defined in this study is useful to identify the bone structure characteristics in normal adults. The height of navicular in females was lower than that in males.
ACKNOWLEDGEMENTS

The authors would like to thank Miss Cha-Chi Lin and Miss Yen-Hsin Chen for their assistance. This study was supported by the National Science Council of the Republic of China through grants NSC92-2218-E010-007 and NSC93-2218-E010-003.

Fig. 1. (a) Radiograph, (b) diagram demonstrating the two-dimensional coordinate system, and (c) angle on lateral radiographs.

(a) Lateral radiograph shows male in the right foot (L) of a 34-year-old man. Each measurement point in the two-dimensional coordinate system is plotted.

(b) (c) Diagram comparing with the mean values at each point and angle in the feet. Sinking of the navicular is the primary symptom in the foot shown in female. **Thick circles**, female (mean); **thin circles**, male (mean).

References

A COMPARISON OF ‘NEUTRAL’ RUNNING SHOES ON PLANTAR PRESSURE PATTERNS AND PERCEIVED COMFORT IN ATHLETES WITH A CAVUS FOOT TYPE: A CROSSOVER, RANDOMIZED CONTROLLED TRIAL

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²Faculty of Medicine, The University of Sydney, Australia.

INTRODUCTION
Athletes with a cavus foot structure have a higher rate of injury compared to athletes with other foot types (Cowan et al., 1993; Burns et al., 2005b). This increased risk of injury is thought to be associated with the elevated plantar pressure seen in people with a cavus foot (Burns et al., 2005a; Sneyers et al., 1995). Athletes with a cavus foot type are routinely recommended ‘neutral’ running shoes by their healthcare provider for the reduction of plantar pressure and to prevent and treat lower limb running injuries (Manoli and Graham, 2005). However, there is little empirical evidence to support the use of ‘neutral’ running shoes to reduce plantar pressures in this patient population. The purpose of this study was to investigate the effect of two commonly recommended ‘neutral’ running shoes on in-shoe plantar pressure distribution and footwear comfort during overground running.

METHODS
A community sample of 22 runners [16 males & 6 females; mean age 30.6 years (SD, 11.1); body mass 70.8 kg (SD, 10.7); height 1.75m (SD, 0.1)] with a cavus foot type defined by the Foot Posture Index (FPI) (Redmond et al., 2006) volunteered to participate in this experimental, randomised, single-blind, cross-over trial. All participants were regular runners with a minimum average weekly running mileage of 20km [mean 55.4 km/wk (SD, 32.9)]. Two commonly recommended ‘neutral’ running shoes were selected for testing following a Delphi interview with a group of eight sports podiatrists. The footwear conditions evaluated were: Asics Nimbus VI (Asics Oceania Pty Ltd, Sydney, Australia), and Brooks Glycerin 3 (Texas Peak Pty Ltd, Melbourne, Australia) versus a control condition (Dunlop Volley, Pacific Dunlop Ltd, Melbourne, Australia). All footwear was purchased to avoid a conflict of interest. In-shoe plantar pressures (peak pressure and pressure-time integrals) were collected for the whole foot, rearfoot, midfoot and forefoot using the Novel Pedar-X® system (Novel gmbh, Munich, Germany) at 100Hz during over-ground running at a consistent self-selected velocity on a flat horizontal concrete surface. Running velocity during all tests was monitored and controlled to within 5% of the velocity of the initial test condition. Footwear comfort was assessed using a validated 150 mm visual analogue scale (Mundermann et al., 2002). The manufacturer and model of all footwear conditions was blinded to participants using an adhered surgical bootie. A one-way, repeated measures analysis of variance was used to test significance of all pressure and comfort variables between footwear conditions.

RESULTS
Compared to the control footwear condition, the two ‘neutral’ running shoes significantly reduced peak pressure for all areas of the foot (Table 1). The Brooks Glycerin 3 was more effective at reducing peak pressure beneath the whole foot and forefoot while the Asics Nimbus VI most effectively reduced rearfoot peak pressure. Compared to the control, both ‘neutral’ running shoes significantly reduced the pressure-time integral for the whole foot and forefoot (Table 2). Only the Asics Nimbus VI reduced the rearfoot pressure-time integral compared to the control. The Brooks Glycerin 3 most effectively reduced pressure-time integral beneath the whole foot and forefoot. The Asics Nimbus VI was perceived as the
most comfortable footwear condition (P < 0.01), although both neutral running shoes were significantly more comfortable than the control condition (P < 0.001).

Table 1: Peak pressures (kPa) between shoe conditions.

<table>
<thead>
<tr>
<th>Foot Region</th>
<th>Control (N = 22)</th>
<th>Asics Nimbus (N = 22)</th>
<th>Brooks Glycerin (N = 22)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Whole foot</td>
<td>513.4 (78.9)</td>
<td>399.4 (88.6)*</td>
<td>361.2 (82.2) *†</td>
</tr>
<tr>
<td>Rearfoot</td>
<td>358.1(173.8)</td>
<td>240.9 (91.9)*</td>
<td>264.4 (90.5) *</td>
</tr>
<tr>
<td>Midfoot</td>
<td>168.6 (68.1)</td>
<td>126.3 (31.0)*</td>
<td>131.4 (34.4) *</td>
</tr>
<tr>
<td>Forefoot</td>
<td>464.2 (106.4)</td>
<td>386.1 (100.0)*</td>
<td>340.8 (89.4) *†</td>
</tr>
</tbody>
</table>

Values expressed as mean (standard deviation)  
* Indicates significant difference compared to Control (P<0.05)  
† Indicates significant difference between Asics Nimbus and Brooks Glycerin (P<0.01)

Table 2: Pressure-time integrals (kPa. s) between shoes.

<table>
<thead>
<tr>
<th>Foot Region</th>
<th>Control (N = 22)</th>
<th>Asics Nimbus (N = 22)</th>
<th>Brooks Glycerin (N = 22)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Whole foot</td>
<td>69.9 (12.4)</td>
<td>55.6 (12.2)*</td>
<td>51.7 (9.7)*†</td>
</tr>
<tr>
<td>Rearfoot</td>
<td>19.8 (10.9)</td>
<td>17.2 (6.9)*</td>
<td>18.81 (7.6)</td>
</tr>
<tr>
<td>Midfoot</td>
<td>15.3 (7.7)</td>
<td>14.4 (3.9)</td>
<td>14.8 (4.4)</td>
</tr>
<tr>
<td>Forefoot</td>
<td>63.9 (13.2)</td>
<td>50.3 (12.3)*</td>
<td>46.0 (9.6)*†</td>
</tr>
</tbody>
</table>

Values expressed as mean (standard deviation)  
* Indicates significant difference compared to Control (P<0.05)  
† Indicates significant difference between Asics Nimbus and Brooks Glycerin (P<0.01)

CONCLUSIONS

Both types of ‘neutral’ running shoes were effective at reducing plantar pressure variables compared to the control footwear condition. These results suggest ‘neutral’ running shoes are an effective strategy to reduce plantar pressures for athletic patients with a cavus foot type. However, due to regional differences in pressure reduction between the two types of ‘neutral’ running shoes, footwear recommendation should shift from being categorical in nature to be based on the location of pain or location of abnormally high plantar pressure. Further research is warranted to examine other ‘neutral’ running shoes and to investigate whether reductions in plantar pressures can reduce the high injury rates in athletes with a cavus foot structure.

ACKNOWLEDGMENT

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REFERENCES


The effect of arch structure midsole on the heel during running

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INTRODUCTION

Millions of people are involved in running and jogging activities everyday, which made the running shoes’ protect property become important. Previous studies have found that shoe design can alter the plantar pressures in specific regions of the foot, so extensive research has been conducted on the effects of constructional changes in the midsole of running shoes (Gillespie & Dickey, 2003; Lafortune, et al. 1996; Wright, et al. 1998). Finite element analysis (FEA) allowed a systematic analysis of the effects of insole properties and design on reduction of heel plantar pressure (Goske et al., 2006). This approach permits the examination of a large number of footwear designs without the burden of high volume experimentation. The purposes of this study were to evaluate the functional arch structure in midsole through FEA and to test its effectiveness through students running plantar pressure.

METHODS

The geometry of the FE model was obtained from 3D reconstruction of CT images from the left foot of a normal male subject of age 23, height 172 cm, and weight 65 kg. Coronal CT images were taken with intervals on 2 mm in the neutral unloaded position. The images were segmented using MIMICS 8.0 (Materialise, Belgium) to obtain the boundaries of skeleton surface. The Young’s modulus and Poisson’s ratio for the bony structures were assigned as 7300 Mpa and 0.3, respectively. This value was selected by weighing cortical and trabecular elasticity values. The midsole consider as non-linear material, using ogden strain energy function to represent its stress-strain relationship.

\[ U = \frac{2\mu}{\alpha^2} (\lambda_1^\alpha + \lambda_2^\alpha + \lambda_3^\alpha) \]  

Where \( \lambda_1, \lambda_3 \) are the deviatoric principal stretches and \( \alpha \) and \( \mu \) are material-specific parameters. Arch structure of the midsole and foot bones (Fig.1) were assembled in the Solidworks 2005 (Solidworks Corporation, Massachusetts). Loads and constraints only consider the heel strike condition, in which appeared the highest vertical force in heel region.

Ten male students of PE-Majoring (age 22.8 ± 1.0 years, body weight 61.2 ± 3.3 kg, height 174.0 ± 1.8) participate in this study. Plantar pressure distribution data was collected with in-shoe pressure measurement system (Novel pedar system, Germany). The system consists of pressure sensing insoles connected to a box which attaches around the subjects waist and transmits information to the Pedar software via Bluetooth wireless communication.

During the testing, subjects were asked to run on the treadmill at the speed of 10 km/hr with arch and regular structure running shoes (Fig.2). 20 sequential steps at stable running condition were analysed as follow: getting peak pressure in the heel region of every step, then average the 20 peak pressure values. All subjects’ data were averaged and the pared t-test was used to compare the mean peak pressure between the two shod conditions.
RESULTS
The model is able to predict the internal stress of foot bones under two different midsole structure shoes. The highest VonMise stress of Calcaneus (Fig.3) was found smaller in the arch structure than regular solid structure, which is 3.63Mpa and 3.96Mpa, respectively. However, the arch structure suffered high stress as the model predicted.

![Fig.3 The VonMises stress distribution in the calcaneus bone at two different structure](image)

Table 1 showed the mean peak pressure exerted on the heel region in the 20 sequential running steps. Significant differences were indicated on the both foot when wearing running shoes with the arch and control midsoles.

<table>
<thead>
<tr>
<th>N</th>
<th>Mean(Kpa)</th>
<th>Std. Deviation</th>
<th>P-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>(left foot) ARCH &amp; CONTROL</td>
<td>10</td>
<td>151.50</td>
<td>36.21</td>
</tr>
<tr>
<td>(right foot) ARCH &amp; CONTROL</td>
<td>10</td>
<td>167.65</td>
<td>34.44</td>
</tr>
</tbody>
</table>

DISCUSSION
From the FE predictions, it is clear that arch design in midsole could reduce the foot bones internal stress. However, the midsole structure suffered high stress that made this sole material must has high durability. The developed FE model can be refined to simulate more realistically the actual situations by incorporating the ligamentous and soft tissue structures, which will be the future development. Within the limitations of the materials and geometries include, the results of FEA study provide some practical guidelines for the manufacture of devices designed to guarantee the structure’s function. The running test data were in good agreement with the FE result, which indicated that the arch structure could reduce the peak pressure in the heel region.

REFERENCE
INTRODUCTION

Rocker shoes have often been prescribed as therapeutic footwear to the diabetic patient since the researches suggest that it can biomechanically restore the normal gait function and relieve the critical plantar pressure for the DM patient. Studies of plantar pressure analysis showed that the forefoot rocker, located at 55 to 60% of stick length, resulted in the best pressure reduction on metatarsal heads; in case of 65%, it would have good toe pressure reduction (Schie 2000). It was also found that for a given rocker axis position, the larger the shoe height as well as rock angle the greater forefoot pressure relief; once the axis moved from 55% to 70% of the foot length, the load-bearing on the shoes will move from the heel to the medial region (Jun 2005). However, studies using gait analysis found out that with rocker shoes, resulted in significant increase of hip extension and knee flexion during midstance (Myers et al. 2005); and increased the joint powers at terminal stance. (Bogart 2005).

Conclusion from previous studies the rocker shoes may redistribute the plantar pressure but increase the joint power and range of motion that results in dynamic instability at time of push off. The purpose of this study was to investigate how the different design of rocker sole would alter the gait pattern of the DM subjects.

MATERIALS AND METHODS

Eight subjects with diabetes mellitus (2 males and 6 females, age 58.25 ± 7.25 years, height 1.59 ± 0.1 m, and weight 66.77 ± 1.56 kg) were recruited from the outpatient clinic of Department of Endocrinology, Shin Kong Wu Ho-Su Memorial Hospital. A pair of off-shelf sandal was used as the baseline shoes. A 4.5cm height of EVA platform (Shore A 63°) was added to the baseline shoes with a fixed heel negative rocker and three locations of forefoot rockers, 55%, 60% and 65% of shoe length. All participants were asked to bear feet and wear three types of forefoot rocker shoes randomly, and then performed the gait analysis on a 10m long walkway. The kinematic and kinetic data such as range of joint motion, knee moment, peak plantar pressure, and center of pressure loci, pressure-time integral and etc. collected from gait analysis (Vicon, MX 13 system 120 HZ, 8 cameras, UK), force platform (AMTI, USA), Pedar insole sensor ( novel, Germany), and pressure mat (0.5 m Footscan, Rsscan, Belgium) were analyzed via two-way ANOVA. The significance of alpha lever was set to 0.05.

RESULTS AND DISCUSSION

The results showed that the maximum force and peak pressure were decreased at the heel and lateral forefoot regions in comparison with and without shoes but no significant difference between various rocker positions (Fig.1). In spatial temporal parameters, stride time, stride length, and walking speed were increased. When walking, the range of motion at the pelvis was increased in posterior tilting from loading response to mid stance, but the anterior tilting was decreased at the phase of mid stance to pre-swing. It may imply that heel-to-toe rocker shoes make pelvis less stable while walking. The flexion angle was reduced on hip but increased on knee and ankle joint (dorsiflexion) especial in terminal stance phase. The time of peak ground reaction forces on-sets in loading response phase was delayed in fore-aft and med-lat directions as shown on Figure 2 and 3.

CONCLUSIONS

The rocker outsole shoes may benefit the plantar pressure relief over the forefoot and does improve the walking speed and step length, but it also increases the knee flexion moment and ankle dorsiflexion angle at time or just before the forefoot
push off. This may result in dynamic instability and to be unstable. The rocker position ranges from 55 to 65% of shoe lengths show no significance in kinetics and kinematics.

Fig.1. Maximum force (left) and peak pressure (right), the data show the subtraction of rocker shoes to baseline shoes.

Fig.2. Peak Ground reaction force onset time in fore-aft direction(left Fig corresponding force diagram)

Fig.3. Peak Ground reaction force onset time in lat-med direction (left Fig corresponding force diagram)

REFERENCES
Jun, I. S., Joo. C., Park. S. B.,Oh. S. G., Lee. S. J. (2005), The relationship between peak plantar pressure and comfort using the rocker shoes, Proceeding 7th Symposium on Footwear Biomechanics, Cleveland, Ohio, USA

ACKOWLEAGEMENT
This study was supported by the National Science Council of R.O.C through grants NSC95-2622-B010-001.
RELATIONSHIP OF PLANTAR PRESSURE PATTERNS AND LOWER LIMB KINEMATICS IN BAREFOOT RUNNING

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¹ Medical University Clinic, Department of Sports Medicine, Tuebingen, Germany

INTRODUCTION

Relationships between plantar pressures and kinematic variables have been analyzed in several studies. While variables which entirely quantify the path of the center of pressure (COP) lack any relation to rearfoot motion (Cornwall & McPoil 2003), researchers have been able to model and predict calcaneal eversion based on local COP data (Hagman 2002). It has also been suggested that plantar load transitions in the heel are related to tibia internal rotation velocities (Robinson & Lake 2005). However, those approaches are restricted to the rearfoot/tibia complex in terms of the analyzed kinematic data. It remains unclear whether roll-over patterns based on total foot plantar pressures are related to hip, knee and ankle kinematics. Hence, the aim of this study was to compute barefoot roll-over patterns based on cluster analysis and associate the pressure patterns to lower limb 3D kinematic variables.

METHODS

89 recreational runners were included in the study [55 males (179.6 ± 7.3 cm; 74.1 ± 8.1 kg), 34 females (167.5 ± 1.2 cm; 58.9 ± 7.1 kg)]. An EMED-X pressure platform (Novel GmbH, Munich, Germany) was utilized to capture plantar pressure data (4 sensors/cm², 100Hz) and a 6-camera Vicon system (Vicon 460, ViconPeak, Oxford, UK) recorded lower limb motions during the stance phase of barefoot running. Measurements were taken at a pre-specified running speed of 3.3 m/s (± 5%). The plantar pressure data was automatically divided into 9 anatomical sub-areas (see Fig. 1), and relative force-time integrals (FTI) were calculated for each sub-area. Kinematic variables were processed using subjects’ static calibration trial as a reference and measurement error was quantified by calculating the root mean square error (RMSE, Tab. 2) (Bland & Altman 1996). Roll-over patterns were created using cluster analysis and utilizing subjects’ average heel (M1+M2), lateral forefoot (M6), medial forefoot (M8) and hallux (M9) sub-area FTI data. A 5-cluster solution was chosen obtained through a combined approach of hierarchical and optimizing cluster analysis techniques. Subsequently, differences between the clusters with regard to the kinematic variables were analyzed using oneway ANOVAs with post-hoc Tukey tests and statistical significance was established at $\alpha = 0.05$.

RESULTS

Results of the cluster analysis are given in Tab. 1. Anthropometrical variables including gender, weight, height, BMI and running speed (both absolute and normalized to leg length) did not differ between clusters. Tab. 3 provides an overview over the kinematic variables within the clusters. Initial hip adduction range of motion and initial hip adduction velocity differed significantly between clusters ($iHADrom: \Delta=3.5°$, $p=0.04$; $iHADvel: \Delta=83°/s$, $p=0.003$). No differences were found for knee kinematics between clusters. Frontal plane motion analysis for the ankle joint revealed differences with respect to time of maximum eversion ($AEV_{max}: \Delta=4.4\%$, $p=0.01$), initial eversion velocity ($iAEV_{vel}: \Delta=50\%$, $p=0.01$) and time of initial eversion velocity ($iAEV_{vel}: \Delta=6.8\%$, $p=0.005$). Differences between clusters were also found with regard to time of peak tibial internal rotation ($TIR_{max}: \Delta=10.2\%$, $p=0.04$).
Cluster 1 2 3 4 5
M1M2 FTI [%] 26.4 20.3 16.2 21.3 12.1
M6 FTI [%] 16.0 22.0 16.6 13.8 17.5
M8 FTI [%] 14.8 13.1 14.9 21.6 22.4
M9 FTI [%] 7.6 6.7 12.8 6.3 6.5

Tab. 1: Cluster centers (means) of the 5-cluster solution.

<table>
<thead>
<tr>
<th></th>
<th>Cluster 1</th>
<th>Cluster 2</th>
<th>Cluster 3</th>
<th>Cluster 4</th>
<th>Cluster 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>iHADrom [*]</td>
<td>9.8* (3.2)</td>
<td>8.1 (2.9)</td>
<td>7.9 (2.4)</td>
<td>8.2 (2.8)</td>
<td>6.4* (3.1)</td>
</tr>
<tr>
<td>iHADvel [°/s]</td>
<td>181* (13)</td>
<td>147 (12)</td>
<td>138 (10)</td>
<td>146 (13)</td>
<td>98* (15)</td>
</tr>
<tr>
<td>AEVtmax [% ROP]</td>
<td>41.6* (3.4)</td>
<td>38.3 2.9)</td>
<td>37.2* (3.8)</td>
<td>38.7 (5.7)</td>
<td>39.4 (5.2)</td>
</tr>
<tr>
<td>AEVtvel [% ROP]</td>
<td>14.9* (6.3)</td>
<td>12.5 (6.7)</td>
<td>8.2* (4.3)</td>
<td>10.9 (7.3)</td>
<td>7.8* (6.7)</td>
</tr>
<tr>
<td>AEVvel [°/s]</td>
<td>141* (35)</td>
<td>145 (53)</td>
<td>190* (68)</td>
<td>141* (35)</td>
<td>174 (69)</td>
</tr>
<tr>
<td>TIRtmax [% ROP]</td>
<td>47.2* (12.9)</td>
<td>45.2 (12.2)</td>
<td>37.0* (11.8)</td>
<td>42.6 (9.7)</td>
<td>39.3 (9.9)</td>
</tr>
</tbody>
</table>

Tab. 3: Kinematic variables within clusters (standard deviations); * and + denote significant differences between respective clusters.

DISCUSSION

The presented 5-cluster approach was found to be a valid solution regarding both the pressure differences between the clusters, and the interpretability of the resulting pressure patterns. An important aspect for cluster validity in the present study is the fact that running speeds did not differ between clusters and are therefore not the cause of loading differences between rear- and forefoot dominated patterns. Moreover, medially and laterally dominated pressure patterns (clusters 2 and 4) were not associated with different characteristics of frontal plane kinematics. Instead, the most prominent differences with respect to ankle and hip kinematics were found between cluster 1 and 3/5, which mainly differ with respect to heel and hallux/forefoot loading. Based on our data, only little evidence was found that plantar pressure patterns - established by calculating relative loads of rear and forefoot sub-areas - are closely related to lower extremity kinematic variables. Although differences between clusters were statistically significant, they have to be considered with care regarding the given measurement error. The results suggest that plantar pressure patterns are complex input/output signals of the human locomotor system and do not solely represent mechanical outcomes of lower limb movement (Nigg 2001).

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FROM HIGH SPEED PLANTAIR PRESSURE MEASUREMENT TO LOWER LIMB SKELETAL MOTION

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RSscan International

INTRODUCTION
At present, more and more gait labs are equipped with a motion analysis system, force plate, EMG and a pressure plate system which helps them in their efforts to understand and change the gait patterns of their patients.

In clinical surroundings, it might not be necessary to have such a complete analysis set-up. Time and costs restrictions on clinical analysis could limit the use of motion analysis and EMG systems. Predominantly, the need for a highly technical operator, set-up time, processing time, and the need for a designated working area are the restrictions for applying these methods. However, information of lower limb motion has been related to sustained injuries (in MacClay, 2000). Therefore, motion analysis cannot be discarded that easily when studying an individual’s gait.

In the present preliminary study, a midway is suggested between a full gait analysis and the sole use of a pressure plate system. By simulating lower limb motion from high speed plantar pressure measurements, the restrictions mentioned above are circumvented. In previous research by Hagman (2002, 2005), Lake et al (2005), and Dixon (2004), mechanical and statistical modelling were used to simulate or predict lower limb motion from pressure measurements.

Using the results from these studies and adding a number of pressure variables, it was the goal of this study to develop and run a forward dynamic lower limb model.

METHODS
Simultaneously measured plantar pressure and motion data of human gait were kindly made available by Dr. Mark Lake. The motion data was registered by a Qualisis system running at 1000Hz, and the pressure from a footscan® pressure system running at 500Hz (details in Lake et al. (2005)), see Figure 1.

![Figure 1: An example motion data used for the dynamical lower limb model.](image)

Subsequently, a forward dynamical lower limb model was developed within the AnyBody environment (http://www.anybodytech.com/). The model consisted out of all...
26 common foot bones and the tibia. Some joints structures were taken to be fixed, such as the joints between the navicular bone and the cuneiform bones.

Other joints were controlled rotationally and are dependant on pressure variables obtained from the pressure plate system: center of pressure path, foot balance, inversion/eversion, and local pressures.

Furthermore, the controls were not the same during heel impact, midstance, and propulsion. In Figure 2, an example of the lower limb model is depicted.

**Figure 2**: The way the lower limb model is driven. On the left the pressure image indicating the pressure variables. In the middle the model driven by the pressure variables. On the right the model in comparison with motion data.

**RESULTS**

At present, a comparison between simulated and actual kinematics has not yet been performed using a mathematical criterion. In the construction phase of the lower limb model, the combined kinematic and pressure data were used to develop these controls. The gross motion of the model is therefore within the boundaries of “normal” gait.

**DISCUSSION**

Still the necessary research has to be devoted in developing the “right” controls that drive the model. We expect that most of the present controls need fine tuning, meaning more sophisticated relationships between pressure and motion.

We believe that the applicability of this technique in clinical settings will be one everybody looks forward to. Therefore, it is our commitment to further extend the possibilities of this model and to validate it with real kinematic data.

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DIFFERENCES IN KNEE JOINT LOADING BETWEEN FOREFOOT AND REARFOOT STRIKE RUNNING PATTERNS

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INTRODUCTION

While 95% of distance runners are rearfoot strikers (RFS), only 5% have been analyzed to land with a flat foot (midfoot striker) or on the ball of their foot (forefoot striker-FFS) (Kleindienst 2003). However, some sports physicians as well as coaches propagate, particularly for recreational runners, the FFS as “natural running” and promise - beside faster running speed - a reduction of joint loading which leads to a prevention/reduction of overuse injuries (Marquardt 2002).

The reported yearly incidence of running related injuries ranges between 37 and 56% (Clement et al. 1981, Van Mechelen, 1992). The knee has been shown to be a common site of injury for runners with Patellofemoral Pain Syndrome (PFPS) being the most common of the injuries to this joint. High abduction and external rotation moments in the knee joint are related to overuse injuries like PFPS (Stefanyshyn et al., 2001).

Therefore the main goal of the present study was to determine the influence of the different running strike patterns on knee joint loading. Moreover “conventional” dynamic variables such as impact forces and loading rates as well as foot eversion and eversion velocity, which have been proposed as major reasons for the development of running injuries, were analyzed (Clement et al., 1981, Cook et al., 1990, Van Mechelen, 1992, Novacheck, 1998, Hreljac et al., 2000).

PROCEDURE

On the study participated 19 male subjects (age: Ø 33 years; body height: Ø 177cm; body mass Ø 72kg) who were serious long distance runners with an average weekly running volume of 55km. All of them were practiced RFS. Williams et al. (2000) could figure out, that runners are able to quickly alter their gait pattern from a RFS to a FFS that is mechanically similar to that of a practiced FFS. This finding represents the precondition to collect data for both locomotion patterns – RFS as well as converted FFS – measured on one subject. All subjects were wearing the same shoe model during the study (adidas® Supernova).

Kinematic data were collected using a 6-camera 3-dimensional Vicon System (200Hz). Reflective markers were placed on the pelvis, upper leg, lower leg, rearfoot and forefoot (3 per segment). Kinetic data were collected using a Kistler force plate (1000Hz). Subjects ran across the force plate in the middle of a 25m runway at a controlled velocity of 4.0±0.2ms⁻¹ regarding both RFS and FFS. Kinematic and kinetic data were collected for 5 valid trials for each subject and condition. A lower body model, which was described earlier (Michel et al., 2004) was used to determine joint centers and angles between segments. Three-dimensional knee joint moments were calculated during the stance phase using an inverse dynamics approach.

Selected values were determined from each curve and averaged for each condition and subject. Significant (sig.) differences between the conditions were detected using paired-samples T test and GLM with repeated measures (p≤0.05).

RESULTS

The analysis of the knee extension moment shows a sig. lower moment for FFS compared to RFS (Ø13%) and it occurs during the midstance phase. The knee joint moments in the frontal plane (Fig. 1) reveal no sig. differences for the max. abduction moment. However, during landing FFS indicate a higher knee abduction moment, which can be explained by the specific strike pattern. The max. knee external rotation moment (Fig. 2), which occurs at the end of midstance phase, is sig. higher in FFS (Ø33%). Moreover, during landing a sig. higher knee internal rotation moment for FFS can be observed (Ø22%). This pattern also could be caused by the initial foot strike.

Looking at the vertical ground reaction forces, the initial force peak only can be detected for the RFS. Regarding the active peak the FFS reveals sig. higher forces than RFS. The vertical loading rate is higher during
DISCUSSION AND CONCLUSION

The analyzed data of the present study are comparable to those from Williams et al. (2000) and Stackhouse et al. (2004). The results clearly demonstrate that the strike index effects kinetics and kinematics of lower extremities – particularly during landing and midstance phase.

It was shown, that increased knee abduction and knee external rotation moments are directly linked to the incidence of PFPS (Stefanyshyn et al., 2001). Consequently, FFS seems to be more risky regarding the development of PFPS than RFS. Additionally, some of the analyzed “conventional” dynamic variables, which are associated with the development of running injuries, lead the same deduction. However, there are also dynamic variables, which can not confirm a lower risk of running related injuries performing RFS such as max. eversion angle and max. vertical loading rate.

Within a medical anamnesis of 471 runners Kleindienst (2003) could not detect differences between RFS and FFS concerning running related injuries. The same is valid for the incidence of foot deformities. However, the location of foot deformities depends on the strike index and the related forces and loads respectively. Based on the findings of the present study it is to conclude that FFS does not lead necessarily to a lower risk regarding the development of running related injuries. It is likely, that the location of the injury/complaints can be influence by the strike index. However, the injury mechanisms are not completely understood. Moreover, it is questionable, whether the measured differences between RFS and FFS represent a clinically relevant reduction in the means of these variables. Therefore it is necessary to conduct prospective epidemiological laboratory and field studies in order to investigate the influence of strike index on the incidence of sport specific injuries and complaints.

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DAY-TO-DAY VARIABILITY OF KINEMATIC VARIABLES IN RUNNING

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INTRODUCTION

Instrumented gait analysis improvements over the last two decades are among others: (i) reduced projection errors (by using 3D vs. 2D); (ii) improved accuracy of segment detection (by cluster analysis and individual joint axis definitions), (iii) quantified skin marker artefacts (by using bone-pins, MRI, CT, stereophotogrammetry, video fluoroscopy), and (iv) improved test designs (by measuring the same test subject twice). Such repeated measurements are typical for prospective studies (Hamill et al., 2006) with a time lag of days or months (Ferber et al., 2001) and accepting the assumption that skin movement artefacts and marker placements of the two measurements are similar such that the registered significant difference between the measurements can be associated with the observed treatment.

However, before conclusions can be drawn with respect to a successful treatment, the day-to-day variability of a control group must be known to justify that an observed difference in patients is not only significant but also relevant. Generally, day-to day variability of kinematic variables in gait analyses are mainly been reported for walking (Kadaba et. al., 1989), less in running. Thus, the goal of this study was to establish the day-to-day variability of defined kinematic variables of the lower extremity of healthy subjects during barefoot and shod running.

METHOD

Eighteen healthy runners (8 females, 10 males; 20-50 km/week; aged 23-55 years; weight 54-85 kg; height 159-191 cm) participated in the study. Data caption was at least one week apart (mean 15 weeks, range 1-54 weeks) with the same protocol (self-selected speed, runway of 30 m length, 5-10 valid trials per subject per condition). Data capture was performed with 12 cameras at 100 Hz (Vicon MX, Oxford Metrics, UK) positioned to cover all three planes optimally. The marker set consisted of 47 markers at the lower extremities. Rotation centers/axes at the ankle, knee and hip were defined functionally for each test subject. Dorsiflexion, adduction and eversion of the ankle (A) were described relative to the tibia; flexion, adduction and internal tibial rotation at the knee (K) relative to the femur.

The reliability analysis was based on the range of motion of decomposed joint rotations during the first 50% of stance phase. Here the differences between corresponding mean ranges of motion at the ankle and knee in three plains are presented. The ninth decile of these differences was defined to propose a threshold value for future distinction between day-to-day variability for patient measurements. For comparisons with the literature, other reliability measures were calculated: Intraclass correlation coefficients (ICC3,k) and the coefficient of multiple correlation between two measurements (CMC).

RESULTS AND DISCUSSION

Mean differences between the two measurements are presented (Table 1,2; Figure 1); most were found to be less than 1° (exceptionally up to 6°). When applying the 9th decile the proposed threshold values were found to be 2° to 4°. ICC3,k parameters higher than 0.8 are considered to be very well repeatable (Benell et al., 1999) which was the case for all comparisons except for the right ankle abduction (0.790). Generally, the ICC was found highest in the sagittal plane, closely followed by the frontal and the transverse plane. CMC results were generally also over 0.8 (with three exceptions) and higher than in previous studies (Queen et al. 2006), perhaps due to the fact that in the present study the 12 cameras (rather than the often reported 6) were optimized for all three planes.
Table 1: Mean differences in [°] between the 1st and 2nd measurement of the range of motion of defined parameters at the ankle and knee. Abbreviations: A = ankle; K = knee.

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<td>barefoot</td>
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<td>1.5</td>
<td>1.0</td>
<td>0.9</td>
<td>1.2</td>
<td>1.0</td>
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<tr>
<td>shod</td>
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<td>1.4</td>
<td>0.9</td>
<td>1.1</td>
<td>0.8</td>
<td>1.5</td>
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Figure 1: Frequency distribution of the differences between the 1st and 2nd measurement of the range of motion of defined parameters at the ankle during running with shoes. Abbreviations: A = ankle; Grey = left, black = right.

Table 2: Results of the ICC3k and CMC. The threshold refers to the 9th decile of the frequency distribution presented in Figure 1 rounded up to the next full digit.

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<td>0.963</td>
<td>0.892</td>
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<td>0.834</td>
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CONCLUSION

Repeatability between two measurements of healthy subjects was generally good, in most cases equal or better than previously reported (Ferber et al., 2002; Queen et al. 2006). Barefoot and shod was about equivalent, thus, from the reliability point of view, there is no reason to eliminate one of the two in future studies. Parameters in the sagittal plane showed the best reliability, but the other two planes showed good results as well, which encourages the use of test parameters also in the frontal and transverse plane. The proposed 9th decile may be used as a threshold value for the interpretation of patient results (i.e. with and without patellofemoral pain).

ACKNOWLEDGEMENTS

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FOOTWEAR AFFECTS THE GEARING IN THE MUSCULOSKELETAL SYSTEM OF THE LOWER EXTREMITIES WHILE RUNNING

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INTRODUCTION
For a long period of time, in sport shoe research it was believed that the contribution of sport shoes is to support, control and protect the human musculoskeletal system (Nigg 2001, Bates 1989). However, new approaches tried to find solutions in sport shoe construction which are able to mimic natural running (i.e. barefoot running). From a mechanical point of view the gearing in the musculoskeletal system can effect several functional parameters related to the evolutionary history of mammals. For example small mammals show lower effective mechanical advantages (EMA) during running than larger mammals (Biewener 1989, 1990) leading to a decrease in force production per active muscle volume (Roberts et al. 1998). Therefore an excessive artificial influence on the gear ratios of the lower extremity due to footwear may induce discrepancies to the natural design of the human locomotor system and thus increases the loading of the locomotor system. However, up to now there is no study in the literature examine the influence of footwear on the gearing of the musculoskeletal system of the lower extremities while running. Therefore the purpose of this experimental study was to investigate whether runners adjust their gearing to different footwear and/or surface to get a beneficial effect on running mechanics.

METHOD
Fourteen competitive male runners participated in the study. All subjects (28.2 ± 4.5 yrs, 182 ± 6cm, 79.9 ± 6.5kg) were free of injuries for two years. 3D kinetics and kinematics were captured using a Kistler® force plate (Winterthur; 1250Hz; 60x90cm) and 12 infrared cameras (250Hz; Vicon624®). The anatomical landmarks (right leg) medial malleolus; lateral malleolus; medial femoral condyle; lateral femoral condyle; trochanter major were armed by retroflective markers (Ø 9mm). Subjects performed several velocity controlled (4.0 ± 0.2m/s) running trials with six different conditions (barefoot, conventional running Shoe 1-5) in a randomized order. The moment arm of the ground reaction force (GRF) acting about the ankle (Rankle) and the knee (Rknee) joint, the moment arm of the tendons (rpateflas, rachilles) the ratio of the moment arm of the GRF to the moment arm of the tendon (gear ratio), as well as the joint angles (αankle, αknee) and moments (Mankle, Mknee) were calculated as average values for the five intervals throughout the ground contact phase (Phase1: 5-20%; Phase2: 20-40%; Phase3: 40-60%; Phase4: 60-80%; Phase5: 80-95%). ANOVA for repeated measures was used for statistical analysis of 5 valid running trials per condition.

RESULTS
The five different shoe types showed no significant (p<0.05) differences concerning all examined parameters while running. At the ankle joint barefoot running showed compared to shod running a higher (p<0.05) gear ratio in the 1st and 2nd and a lower (p<0.05) in the 5th phase of ground contact. The moment arm of the GRF (Rankle) acting about the ankle joint showed an analogous behaviour. The differences (p<0.05) for rachilles do not exceed 3mm, while the variation in Rankle is about 35mm. Thus, the observed higher gear ratio is mainly due to the longer Rankle. The longer Rankle also leads to a higher (p<0.05) ankle joint moment (Mankle) in the 1st and 2nd phase of stance.
At the knee joint the absolute gear ratio is higher (p<0.05) in the 1\textsuperscript{st} and lower (p<0.05) in the 3\textsuperscript{rd} and 5\textsuperscript{th} phase for the barefoot compared to the shod condition. The moment arm of the GRF (R\textsubscript{knee}) behaves similar in the three mentioned phases. The difference for r\textsubscript{patella} is smaller than 1mm during the entire stancephase, which indicate that R\textsubscript{knee} is again the reason for the higher gear ratio at the knee in these phases. The absolute knee joint moment (M\textsubscript{knee}) is lower (p<0.05) in the 3\textsuperscript{rd} phase), due to a shorter (p<0.05) moment arm of the GRF (R\textsubscript{knee}) as a consequence of a more (p<0.05) extended knee (\alpha\textsubscript{knee}). The vertical GRF showed no differences in this 3\textsuperscript{rd} phase.

**DISCUSSION**

By analyzing running mechanics with different external conditions, the results supports the expectation, that environmental conditions affect the gearing at the ankle and knee joint while running. During the 1\textsuperscript{st} and 2\textsuperscript{nd} phase of stance, where plantar flexor load is moderate, shod running seems to be more efficient compared to barefoot running, and showed a lower gear ratio and therefore an increased mechanical advantage for the plantar flexors to generate propulsive force during locomotion. The behaviour of the gear ratio is contrary in the knee versus the ankle joint. In the 3\textsuperscript{rd} phase, shod running showed a higher gear ratio, M\textsubscript{knee} and a longer R\textsubscript{knee}, leading to a higher load of the structure and from a mechanical point of view to a lower safety-factor for the corresponding structures. During the 5\textsuperscript{th} phase the higher gear ratio at the ankle joint for the shod condition could/should also be considered as an advantage for the leg extensors, because the concentric working muscles are able to operate at a lower velocity and therefore a more effective force generation due to the force-velocity-relationship (Carrier 1994). This study does not allow to conclude if the differences are due to the surface (grass vs. tartan) or to footwear.

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*Clinical Journal of Sports Medicine*, 11, 2-9

INTRODUCTION

Running is a popular way to keep healthy. Recently, they run not only on a jogging course but also on the treadmill in a fitness club or their houses. Most people wear normal running shoes when they run on the treadmill. Several researchers reported different points on the running motion between overground and treadmill running. However, there was little study that could be directly useful for designing treadmill shoes. The purpose in this study is to get the basic findings to design the specific shoes for treadmill running based on the motion analyses of overground and treadmill running.

METHODS

A healthy man with little experience on treadmill running participated in this study. After the warm-up, the subject ran on two different running surfaces: a sport surface and a treadmill. Overground running was done on the all weather surface at 3.86±0.01 and 8.13±0.19 min/km. Treadmill running was done on the treadmill (Takei Co.) that have a running area of 0.5 m width × 1.6 m length at 4 and 8 min/km. The subject wore the typical running shoes (GT2110, ASICS Co.) on both conditions. To measure lower extremity motions, the reflective markers were placed on the subject’s shank and heel as shown in Fig. 1 and these marker coordinates were measured by Vicon-MX (Oxford Metrics Ltd.) operating at 200Hz. Three dimensional rotation angles of calcaneus relative to tibia; planter-dorsiflexion (α), inversion-eversion (β) and adduction-abduction (γ) were calculated using by these coordinates (Grood, 1983). Planter pressure distribution data were collected using F-Scan system (Tec-Scan Inc.). The sum values of plantar pressure (Planter force) in forefoot and rearfoot were independently calculated. The motions from side and rear angle were captured by high-speed video cameras (Nac Inc.).

RESULTS AND DISCUSSION

The typical time series of plantar forces in forefoot and rearfoot under the overground and treadmill running are shown in Fig. 2(a). In both 8min/km and 4min/km speed, plantar forces showed the same

(a) Time series plantar forces (b) Maximum values of plantar forces

Fig. 2 Typical time series of plantar forces (a) and maximal values of plantars force of forefoot and rearfoot on overground and treadmill at 8 min./km and 4 min/km (b)
tendency. Maximal values of plantar forces in forefoot and rearfoot are shown in Fig. 2(b). There was no difference of the maximum plantar force obtained from rearfoot with 8 min/km. But with 4 min/km, the maximum plantar force on treadmill running was smaller than that on overground running. In forefoot, the maximal plantar force on treadmill running is smaller than that on overground. This is derived from that runner don’t need to kick in the forward direction on treadmill running.

The typical time series of $\beta$ and $\gamma$ angles under both the conditions are shown in Fig. 3. The maximum values of rearfoot abduction angle ($\gamma_{\text{max}}$) under treadmill running and overground running were observed at nearly same time. On the other hand, the maximum value of rearfoot eversion angle ($\beta_{\text{max}}$) on treadmill running delays. The distance in both shoe edges, $L_t$ on treadmill is shorter than $L_g$, as shown in Fig. 4. This result suggests the subject stepped linearly. Because the subject automatically feels a danger to step on the edge of running belt. Therefore, their legs were inclining and the angle between his leg and ground is smaller than that on overground running at the heel contact.

**CONCLUDING REMARKS**

Some of the differences between treadmill and overground running were confirmed for the designing process of treadmill running shoes.

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DIFFERENCES IN ABDUCTION / ADDUCTION RANGE OF MOTION IN INTRINSIC FOOT JOINTS DURING WALKING AND RUNNING

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INTRODUCTION
Designers of footwear, especially for athletic shoes, have for some decades been aware of both rearfoot and forefoot pronation/supination patterns and the need to address these in the level of control provided by the shoe midsoles and lasts. The level of motion control of footwear is debatable as excessive control can prevent required natural motion at a specific joint, thus increasing stresses to ligaments, tendons or bones at more proximal levels. Furthermore, it is important to identify how much motion naturally occurs at different joints in the foot during either running or walking, so as to design footwear that can accommodate specifically for variable motion at varying anatomic locations. There are difficulties in accurately representing segment motion in the foot with standard externally applied skin marker sets1, 2. This paper presents abduction and adduction ranges of motion of the intrinsic joints of the foot measured with intracortical pins during walking and slow running. The aims were to identify the magnitude and variability of abduction and adduction so as to assist designers of athletic footwear in the construction of soles and lasts that accommodate for this motion at the correct anatomical positions and also to indicate differences in abduction/abduction between walking and running.

METHODS
Four healthy male volunteers participated in the running study (mean age: 42 years, weight: 87 kg and height: 192 cm) and six in the walking study (mean age: 38, weight 85kg and height 181 cm). The walking study included the same four subjects as the running plus two additional. The study was approved by the local ethical committee of Huddinge University Hospital, Sweden. Subjects walked and ran at self selected speed and their preferred cadence. Running consistency was maintained during all subsequent trials by the cadence set with a metronome.

Ground reaction force (Kistler, Winterthur, Switzerland) and tibia kinematic data were compared during running and walking with and without intracortical pins to control whether each subject’s gait with the pins was representative of their normal style. Kinematic data was recorded using a 10 camera ProReflex optoelectric system (Qualisys, Göteborg, Sweden). Selfdrilling, 1.6 mm diameter pins (Synthes, Bettlach, Switzerland) were inserted by an experienced orthopedic surgeon under local anaesthetic (Xylocain and Marcain, AstraZeneca, Södertälje, Sweden) into nine separate bones (tibia, fibula, calcaneus, talus, navicular, cuboid, medial cuneiform and metatarsals I and V: figure 1) with fluoroscopy guidance. Each array was equipped with three arms with reflective markers attached. After pin removal, the insertion sites were cleaned and covered with new sterile dressings. Subjects were provided with antibiotic and pain relief medication (AstraZeneca, Södertälje, Sweden).

Figure 1. Computed tomography image of a subject’s foot with the inserted intracortical pins. The calcaneus pin is obscured.
Local, technical coordinate systems for each bone were aligned with the global system in the neutral standing reference trials. Individual bone movement relative to the reference trial was determined for each motion analysis frame collected and the joint rotations were then calculated using a helical axis approach. Each segment’s rotation was projected onto the reference frame of the proximal bone under consideration of the helical axis orientation.

RESULTS
All joints except for the talar-tibial joint exhibited greater mean abduction/adduction range of motion during walking than running. In running a more even distribution of abduction/adduction was seen across all joints with the maximum occurring between metatarsal V and the cuboid (9.6° ± 2.4°) and an equal amount (mean 8.7°) in the talar-tibial and navicular-talar joints. In running all other joints except for the fibula-tibia had ranges of motion of between 4° and 6°. During walking the most dominant joint was the navicular-talar joint with 16.3° (6.5°) while the metatarsal V–cuboid joint had a similar value to running with 9.8° (2.1°). The greatest difference between walking and running was seen at the navicular-talar joint with 7.6°.

Table 1. Total range of motion (°) in abduction /adduction at 9 joints in the foot, means (SD). Differences in means between walking and slow running are presented in the bottom row. No statistics were performed to compare walking and running data due to the small subject number.

<table>
<thead>
<tr>
<th></th>
<th>Tal - tibia</th>
<th>Calc - tal</th>
<th>Nav - tal</th>
<th>Cub - calc</th>
<th>Cub - nav</th>
<th>Cun - nav</th>
<th>MtI - cun</th>
<th>MtV - cub</th>
<th>Fib - tib</th>
</tr>
</thead>
<tbody>
<tr>
<td>Running</td>
<td>8.7 (3.9)</td>
<td>5.9 (2.0)</td>
<td>8.7 (1.4)</td>
<td>6.9 (3.3)</td>
<td>5.6 (1.8)</td>
<td>4.1 (1.1)</td>
<td>4.3 (1.4)</td>
<td>9.6 (2.4)</td>
<td>1.6 (0.3)</td>
</tr>
<tr>
<td>Walking</td>
<td>7.8 (2.7)</td>
<td>7.5 (2.0)</td>
<td>16.3 (6.5)</td>
<td>8.1 (2.0)</td>
<td>8.9 (4.3)</td>
<td>6.2 (4.2)</td>
<td>6.1 (1.1)</td>
<td>9.8 (2.1)</td>
<td>3.5 (1.2)</td>
</tr>
<tr>
<td>Δ</td>
<td>-0.9</td>
<td>1.6</td>
<td>7.6</td>
<td>1.2</td>
<td>3.3</td>
<td>2.1</td>
<td>1.8</td>
<td>0.2</td>
<td>1.9</td>
</tr>
</tbody>
</table>

DISCUSSION & CONCLUSIONS
This study presented valuable in vivo information concerning the abduction and adduction ranges of motion occurring during both walking and running. An inherent limitation in the interpretation of the precise values in this study is the possibility of cross-talk between motion in different planes. This phenomenon resulted from the placement of technical coordinate systems at the various joints relative to the initial neutral standing position. These coordinate systems no not necessarily represent the anatomical axes of rotation. It can be assumed that the resulting error is greatest in eversion/inversion, however, a certain error must also be expected in the abduction adduction data. Different patterns in the distribution of this motion through the foot were found in the two gait modes and greater ranges of motion were generally measured during walking. These greater values during walking can be due to greater muscular and external force constraints during running restricting motion about individual joints. Longer contact times during walking may also play a role. Abduction/adduction was shown to occur predominantly at the navicular-talar and metatarsal V–cuboid joints. This indicates that a certain flexibility of footwear in horizontal abduction/adduction should be incorporated in shoes at these locations to permit optimal kinematics of the foot.

REFERENCES
INFLUENCE OF JOINT ANGLE, MUSCULAR PRE-ACTIVATION AND IMPACT INTERFACE ON SHOCK TRANSMISSION THROUGH THE KNEE

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INTRODUCTION

Different cadaver studies showed a shock attenuating capacity of intra- and periarticular tissues in axial impact loading (Radin and Paul, 1971; Chu et al., 1986). This shock reductions were mainly explained by deformation. In vivo reductions in shock transmission in walking or running were primarily explained by variations in leg stiffness due to modified joint angles or muscle activity (Bobbert et al., 1991; Lafortune et al., 1991). In controlled situations (human pendulum) softer shock interfaces led to lower impact loads (Lafortune et al., 1996), systematically this is not the case in running (Nigg et al. 1995). The interdependence of joint angles and muscle activation in locomotor activities make it difficult to analyze determining factors of shock transmission during locomotion. The purpose of this study was to analyze the role of muscle pre-activation and knee angle on shock transmission under controlled situations.

METHODS

Four male (35-47 years, 70-88kg) healthy subjects volunteered for this study. As illustrated in figure 1 they were positioned supine allowing to take up three different goniometric controlled knee angles (0°, 20°, 40°). The forefoot was strapped on the metal plate of an impacter device. The hip angle (40°) and ankle angle (90°) maintained constant. In each angle condition three muscle activation levels were defined in respect to prior performed maximal voluntary contractions (0%, 30%, 60% MVC). Using EMG (Biovision®) the activation was measured at both gastrocnemii (GM, GL), both vasti (VM, VL) and the semitendinosus (ST). In each of the nine different angle-activation conditions ten impacts were initiated under the subjects’ heel by means of the pneumatic driven impacter.

Figure 1: Schematic drawing of the subjects’ positioning in the three knee angle conditions. The impacter device (arrow) initiated impacts under the heel.

The shock interface was a 1cm ethylene vinyl acetate (EVA) pad as used for midsole material in running shoes. For the 40° knee angle conditions an additional interface condition with a 1cm soft foam was defined. The force was measured using a one dimensional transducer (Kistler®). Tibial and femoral shocks were measured using three dimensional accelerometers (Kistler®, m<0.0025kg). This abstract
focuses on the longitudinal components only. The sensors were attached to Apex pins (diameter 3.0mm, length 60mm) inserted under local anesthetic ca. 1.5cm into the right tibia and femur bone. The location of the insertions were medial and approximately 5-7cm below (tibia) and 2–3cm above (femur) the knee joint space. After attaching the sensors their axes were aligned with the anatomical segment axes using a correction algorithm based upon reflective markers recorded by a movement analysis system (ProReflex®). The shock transmission through the knee joint was calculated by the ratio RAT of the acceleration maxima at the tibia (ACCtib) and femur (ACCfem) (RAT=ACCtib/ACCfem·100). In the 0° knee angle condition the linear displacement of tibia and femur was estimated by double integrating the acceleration time history and the knee compression COM by the difference of tibial and femoral displacement. The sampling rate of all analog data was 1000Hz. An ANOVA (p<0.05) was carried out to identify significant differences of the mentioned parameters between knee angle or activation conditions.

RESULTS AND DISCUSSION

The average peak accelerations at the tibia varied between 2 and 4g and at the femur between 1.5 and 2.5g for the hard interface condition. With increasing muscle activation levels the reaction force increased and the tibia acceleration decreased significantly under all knee angles which was also true for the femur at 0° and 20°. In all MVC levels ACCtib increased significantly with increasing angles while ACCfem showed no systematic trend in all subjects. The average shock transmission over all trials was 59.7%. In the 0° and 20° knee angle condition three of the four subject showed a significant increase in RAT of about 10% with increasing muscle activation levels while one subject showed no substantial change. In the 40° condition increasing muscle activation had a smaller effect on RAT compared to 0° and 20°. Under the 30% and 60% activity level the softer interface led to lower impact forces and segmental peak accelerations than the harder EVA foam. The transmission RAT was not affected by interface changes. The effect of the harder foam on the increase of the impact forces was smaller than the effect of the increasing muscle force with increasing activity levels. The effect of muscle activity and therefore muscle force on the reduction of the segmental acceleration peaks was greater than the interface effect.

In all knee angle conditions the relative axial movement of tibia and femur (COM) was significantly decreasing with increasing muscle activation. Apparently the muscle forces pre-load the joint’s peri- and intraarticular structures. Considering the facts that RAT was increasing while COM was decreasing with higher muscle activity and that cadaver studies showed a shock attenuating effect of passive structures it could be assumed that the deformation of peri- and intraarticular tissue has also an effect on shock transmission in vivo. This effect could be substantially influenced by muscle force.

REFERENCES

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VARIABILITY IN SOFT TISSUE COMPARTMENT MECHANICAL PROPERTIES

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INTRODUCTION

The impact force in heel-toe running results from the rapid deceleration of the foot and leg following first ground contact (Bobbart et al., 1992) and acts as an input signal into the body that initiates vibrations of the soft tissue compartments of the leg. These vibrations are heavily damped and the concept of muscle tuning suggests the body reacts to a given input signal with goal of minimizing these vibrations (Nigg and Wakeling, 2001). Supporting evidence for this concept has shown that a muscle response does occur (Wakeling et al., 2003; Boyer and Nigg, 2004; Boyer and Nigg, 2006a) with an increase in the damping of the soft tissue compartment when the frequency of the input signal approaches the natural frequency of the soft tissue compartment (Wakeling et al., 2002; Boyer and Nigg, 2007). These muscle tuning responses to different input signals were subject-specific and may be a result of differences in the natural frequencies of the subjects’ soft tissue compartments. However, this inter-subject variability has not been assessed.

The natural frequencies of a soft tissue compartment are influenced by the muscle properties, adipose and connective tissue and the coupling between these components (Wakeling and Nigg, 2001). The effect of changes in the mechanical properties of the muscle on the total soft tissue compartment properties can be substantial, e.g. increases up to 50 Hz during isometric and isotonic leg contractions of varying intensities (Wakeling and Nigg, 2001). While changes in the frequency properties of the soft tissue compartments have been shown for increases in muscle activity in isolated experiments, the effect of increases in muscle activity for increased running speed have not yet been shown. Therefore, the purpose of this investigation was to assess the inter-subject and running speed related variation in soft tissue compartment mechanical properties for a seemingly homogeneous group of male recreational athletes. If small variations exist then it may be possible to develop interventions to modify the input signal characteristics thus minimize the chances of increased muscle activity in a locomotion task with the purpose of minimizing soft tissue vibrations.

METHODS

6 male recreational athletes (age 26 ± 4.2 years) participated in this study. Subjects ran at 3 and 4.5 m/s in five shoe conditions while segment kinematics and soft tissue accelerations were measured. The input accelerations were determined by double differentiation of the vertical position of the upper calcaneal marker (Boyer and Nigg, 2006b). Six trials per condition were collected resulting in 60 trials per subject. The mechanical properties of the soft tissue compartments were determined from a frequency response function (FRF) determined for each subject and speed. A peak in the FRF indicates a resonance frequency in the vibrating system. It was calculated from the auto power spectrum and cross power spectrum of the shoe acceleration (input) and the soft tissue compartment accelerations from the stance phase of running. For each subject the soft tissue compartment resonance frequencies were determined, for each speed, from the mean FRF of the five shoe conditions.

RESULTS

Two dominant peaks in the FRF for each subject/speed were typically found. The inter-subject variability in the resonance frequencies of the triceps surae soft tissue compartment was less than 5 Hz for jogging and 10 Hz for running for this relatively homogeneous population (Table 2). For the slow
jogging speed the groups mean resonance frequencies were located 9.8 and 15.0 Hz, for the faster running speed these frequencies shifted to 15.1 and 21.1 Hz (Figure 1).

<table>
<thead>
<tr>
<th>Subject</th>
<th>Speed 1 (3 m/s)</th>
<th>Speed 2 (4.5 m/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>f1 (Hz)</td>
<td>f2 (Hz)</td>
</tr>
<tr>
<td>E</td>
<td>12</td>
<td>18</td>
</tr>
<tr>
<td>H</td>
<td>9</td>
<td>15</td>
</tr>
<tr>
<td>F</td>
<td>7</td>
<td>13</td>
</tr>
<tr>
<td>J</td>
<td>9</td>
<td>16</td>
</tr>
<tr>
<td>K</td>
<td>7</td>
<td>13</td>
</tr>
<tr>
<td>L</td>
<td>8</td>
<td>13</td>
</tr>
<tr>
<td>Mean±SE</td>
<td>9.8±1.1</td>
<td>15.0±0.9</td>
</tr>
</tbody>
</table>

Table 1: Resonance frequencies of the triceps surae soft tissue compartment for running: 3 and 4.5 m/s.

DISCUSSION AND CONCLUSIONS

Vibrations of the soft tissue compartments may affect comfort, performance, increase stress on the connective tissues and increase the total peripheral resistance to blood flow (Mester et al., 2006). Muscle adaptations to minimize these vibrations have previously been shown. The frequency content of the input signal depends on the shoe/surface properties, landing kinematics and running speed. Since the resonance frequencies of the soft tissue compartments depend on the muscle activations levels and the surrounding tissue, requirements for muscle tuning adaptations are highly subject specific.

As expected, there was a significant increase in the resonance frequencies with increasing running speed (and muscle activity). However, this increase was small (3-6 Hz) in comparison to the 50 Hz increases found for isolated experiments (Wakeling and Nigg, 2001). This analysis has also shown that inter-subject variations in the resonance frequencies also exist for a homogeneous population. However, these variations are smaller than the intra-subject variations that resulted from increases in running speed. As changes in the input signal frequency depend on both speed and shoe properties it is highly likely that a muscle tuning response will occur for all subjects when changing their running speed during a run. However, for this sub-population a range of critical frequencies for both jogging and running exists where a muscle tuning reaction is most likely to occur. In contrast to previous shoe or surface developments these results suggest that a higher frequency input (faster deceleration of the leg) would avoid the “critical” muscle tuning region and therefore may be more beneficial.

REFERENCES


LOADING CHARACTERISTICS OF HIGH IMPACT MOVEMENTS IN BASKETBALL

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INTRODUCTION
Prolonged basketball competition and training seasons lead to tremendous amount of injuries. It was reported that 69% of professional basketball players reported some form of injuries with an estimation of 7.6% of missed game time in a 7-year retrospective study (Henry et al. 1982). Lower extremity joints experienced most significant injuries with the ankle joint accounting for the highest rate of injury at 18.2% and knee injuries causing the most games missed at 66%. These results are similar to the knee injury rate found in NCAA men’s and women’s soccer and basketball players over a period of four seasons (Arendt and Dick 1995). However, limited information on biomechanical characteristics of basketball movements can be found in the biomechanical literature (McClay et al. 1994; McClay et al. 1994). To the knowledge of the authors, no documentation of comprehensive 3D biomechanical tests can be found in the literature. Therefore, the objectives of the study were to examine ground reaction force characteristics of cutting, jumping and landing movements and to provide a high-impact loading profile in basketball.

METHODS
Six female (age: 20:00 ± 1.56 years, mass: 80.94 ±10.62 kg, height: 1.84 ± 0.11 m) and six male (age: 20:00 ± 2.00 years, mass: 84.06 ±10.26 kg, height: 1.87 ± 0.11 m) NCAA-I A basketball players participated in the study. The subjects performed five trials of following movements in each pair of two basketball shoes (A & B, adidas) in a random order: single-leg jump, two-leg jump, landing from jump-stop, landing from a two-leg vertical jump with pivot, V cut, and lateral cut. These movements were identified based upon the results of a previous study examining the frequency and intensity of major basketball movements during three regular season NCAA men’s basketball games. 3D kinematics by a 6-camera Vicon system (120 Hz), ground reaction forces (GRF) (600 Hz, AMTI), and EMG signals were recorded simultaneously during the testing session. Only the ground reaction force data are reported here.

Visual3D (C-Motion, Inc.) and customized software were used to compute 3D kinematic and joint kinetic variables. The kinematic and GRF data were smoothed using a 4th order zero-lag Butterworth low-pass digital filter at cutoff frequencies of 8 Hz and 20 Hz, respectively. A 2 × 2 (gender × shoe) repeated measures ANOVA was used to evaluate selected mediolateral (ML: X), anteroposterior (AP: Y) and vertical (Z) GRF variables (SPSS, 12.0) with post hoc comparisons and an alpha level of 0.05.

RESULTS
The results showed that the female players had smaller peak ML GRF than the males in the V cut (Figure 1A). Although not significant, the male players showed greater peak ML GRF and greater peak vertical GRF compared to their female counterparts in the lateral cut. Shoe A showed smaller peak AP GRF (Figure 1A) and impulse, but greater peak vertical loading to the body than Shoe B in the jump stop (Figure 1B).
DISCUSSION
Both cutting movements imposed average ML loading of 1.1 - 1.5 and vertical loading of 2.2 – 3.1 body weight (BW) for male and female players respectively. The male players had average peaks of 1.1 and 0.7 BW for the ML GRF and 3.7 and 2.2 BW for the vertical GRF in the 1-leg and 2-leg jump, respectively. The jump stop movement posed greater AP loading to the body compared to the landing & pivot movement with averages of 1.1 and 0.75 BW for males and 0.77 and 0.41 BW for females, respectively. The average peak vertical GRFs were 4.0 and 4.8 BW for these two landing movements, respectively, with females experiencing smaller peaks (2.8 and 3.3 BW).

CONCLUSIONS
This study documented a complete profile of ground reaction forces of major basketball movements with high impact loading for both female and male college players. The results demonstrated that female players experienced smaller lateral loading in the V cut movement while exhibiting the similar trend in the vertical loading in the two landing movements. Finally, the major shoe differences were only seen in the jump stop.

ACKNOWLEDGEMENTS
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REFERENCES


EFFECTS OF SHOE AND TRACK STIFFNESS ON THE BIOMECHANICS OF SHOE AND FOOT DURING HEEL STRIKE

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INTRODUCTION
Shoe and track stiffness are two important factors influencing the foot biomechanical behavior and performance of athletes (Chuang et al., 2005). It has been mentioned that increasing performance depends on efficiently transforming chemical energy to mechanical energy by the musculoskeletal system (Nigg et al., 1992). Three strategies, optimizing the musculoskeletal system, maximizing the energy returned and minimizing the energy loss or absorption, were considered as the major factors in the increase of performance in physical activity (Nigg et al., 1992). Furthermore, a pair of suitable shoes can reduce the incidence of injury during sports and training activities. Therefore, the purpose of this study is to investigate the biomechanics of energy exchanges in the shoe outsole and foot bone during heel strike phase to determine the correlations with respect to the stiffness of shoe as well as track.

MATERIALS AND METHODS
In this research, a 3D finite element (FE) model of a left foot including bones, cartilages, soft tissues, footwear including flat insole and outsole were developed, respectively (Figure 1). Besides, the FE model of track was also constructed using hexahedral elements. The stiffness of shoe (Ks=78, 157, 235 and 314 kN/m) and track (Kt=500, 1000, 2000, 4000 kN/m) were chosen to simulate the different shoe and track conditions. The kinematic gait data of a normal male subject was used to set the boundary conditions of the FE analyses. Outsole strain energy, calcaneal strain energy, softtissue strain energy and ground reaction force were calculated in this study.

Figure 1. Finite element model of the (a) foot soft tissue, (b) bone and cartilage and (c) simplified footwear.
RESULTS

Results of outsole, calcaneal, softtissue strain energy and ground reaction force during heel strike are shown in Figure 2. The energy stored in the deformed elastic component of the outsole decreased with shoe stiffness increasing or track stiffness decreasing (Figure 2a). However, calcaneal, softtissue strain energy and ground reaction force increased with shoe or track stiffness increasing (Figure 2a,b,c).

Figure 2. Results of (a) outsole strain energy (J), (b) calcaneal strain energy (J), (c) softtissue strain energy(J) and (d)ground reaction force(N)

DISCUSSION

According to the literature, energy recovery from the shoe depends on the applied load and the material composition of the shoe. It is suggested that running performance may be improved if the amount of energy dissipated by the shoe were decreased. More significant gains in energy storage and recovery could be achieved by decreasing the elastic stiffness of shoe cushioning materials. In the present study, the shoe with lower stiffness can store more strain energy than that with higher stiffness. Furthermore, the track with higher stiffness can result in more energy stored in the outsole. However, it was found that the outsole strain energy to be more strongly influenced by the changes in stiffness of track than shoe. For wearing a shoe with lower stiffness will decrease calcaneal, softtissue strain energy as well as ground reaction force and thus likely reduce the risk of foot injury.

REFERENCES


EXPERIMENTAL INVESTIGATIONS ON SHOCK WAVE IMPACT PROPAGATION THROUGH THE HUMAN LOWER LIMB.

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INTRODUCTION

This study focuses on experimental investigations on shock wave impact propagation through lower limbs. Ground reaction forces (Harrison et al., 1986; Dixon et al., 2003) and/or accelerometers (Laafortune et al., 1995; Mercer et al., 2003; Ogon et al., 1999) are used for determining the effect of impacts on the human body in different locomotion tasks like walking or running. On the one hand, the response of musculoskeletal system may vary depending on the mechanical midsole characteristics (Boyer and Nigg, 2004). On the other hand, subjects could actively decrease the amount of soft tissue acceleration amplitudes by muscle preactivations (Nigg et al., 2003) and, eccentric muscular activity. To our knowledge, no experiments were designed for investigating the body response when subjects can not damp the impact in modifying the kinematics pattern of the task. Therefore, the aim of this work is to associate both analyze devices (force plate and accelerometers) in order to measure the effect of the mechanical midsole characteristics on the shock wave propagation after impacts of the lower limb in two tasks.

METHODS

Six subjects participate in the experiment and realize 3 trials per condition. Two conditions consist in stepping down a stair case (18 cm). In the first, they are asked to normally perform the task (normal condition). In the second, they have to maintain their landing lower limb straight (constrained condition) while they strike the force plate. A third task is a squat-jump to measure their performance (height in meter). 6 shoes are used with values of the mechanical midsole stiffness representing the extreme values which can be obtained in an usual industrial process production (from 45 Shore C up to 70 Shore C). The cushioning value is determined with a device that impacts shoes according to the ASTM F1614 American norm. Analog data (AMTI force plate and Entran accelerometers placed at the ankle, knee, and hip) are sampled at 2000 Hz and recorded with a Labview program. Data are stored for further analysis. A customized Matlab program was built to determine peaks and time to peak of both vertical forces (: immediately after the impact and maximum peak) and vertical accelerations (: immediately after the impact at the ankle and the 3 maximum peaks at the 3 joints). Dynamic data were normalized to body weight. Conjointly, a numerical camera (digitalized at 100 Hz) records kinematics in the sagittal plan. Then, Simi Motion software enables to obtain angular range of motions in order to control the repeatability of displacements in the three tasks.

RESULTS

Results show that both ankle and knee ranges of motion are repetitive according to the condition observed. No significant differences are shown in squat jump performances whatever the shoe stiffness. Peak of vertical forces and accelerations in the constrained condition are significantly higher than in the normal condition. Indeed, peaks of the vertical force are multiplied by two and the increase of the three acceleration peaks is about 0.4g, 2g and, 4g at the ankle, knee and, hip, respectively. The mechanical midsole property influences the propagation of the shock wave as greater peaks in both forces and
accelerations are observed for higher stiffness. The constrained condition enhances the influence of midsole stiffness on forces and accelerations peaks.

**DISCUSSION**

These experimental investigations argue the effect of the muscle activities for decreasing the impact propagation in the lower limb. That is consistent with the muscle tuning for decreasing the shock wave propagation during running (Nigg et al., 2003). The relationship between mechanical midsole characteristics and shock wave propagation towards the limb will be further discussed.

**CONCLUSION**

Results of the present study show that the stepping down task while maintaining the landing leg straight may be useful to study the shock wave impact propagation through the lower limb. Moreover, our results indicate that the type of muscular activity helps in reducing the shock wave impact propagation.

**REFERENCES**


