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Original Article

EFFECTS OF MEDIALLY POSTED INSOLES ON FOOT AND LOWER LIMB MECHANICS ACROSS WALKING AND RUNNING IN OVERPRONATING MEN

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ABSTRACT

Anti-pronation orthoses, like medially posted insoles (MPI), have traditionally been used to treat various of lower limb problems. Yet, we know surprisingly little about their effects on overall foot motion and lower limb mechanics across walking and running, which represent highly different loading conditions. To address this issue, multi-segment foot and lower limb mechanics was examined among 11 overpronating men with normal (NORM) and MPI insoles during walking (self-selected speed 1.70 ± 0.19 m/s vs 1.72 ± 0.20 m/s, respectively) and running (4.04 ± 0.17 m/s vs 4.10 ± 0.13 m/s, respectively). The kinematic results showed that MPI reduced the peak forefoot eversion movement in respect to both hindfoot and tibia across walking and running when compared to NORM ($p<0.05$-0.01). No differences were found in hindfoot eversion between conditions. The kinetic results showed no insole effects in walking, but during running MPI shifted center of pressure medially under the foot ($p<0.01$) leading to an increase in frontal plane moments at the hip ($p<0.05$) and knee ($p<0.05$) joints and a reduction at the ankle joint ($p<0.05$). These findings indicate that MPI primarily controlled the forefoot motion across walking and running. While kinetic response to MPI was more pronounced in running than walking, kinematic effects were essentially similar across both modes. This suggests that despite higher loads placed upon limp limb during running, there is no need to have a stiffer insoles to achieve similar reduction in the forefoot motion than in walking.

1 INTRODUCTION

Foot orthotics represent one of the most popular techniques to alter lower extremity movement. In particular, many orthotics, like medially posted insoles (MPI), are designed to reduce excessive foot pronation motion and this reduction has been thought to be a central mechanism behind beneficial management of various lower extremity injuries (Donatelli et al. 1988; Collins et al. 2008; Eng & Pierrynowski 1993; Shih et al. 2011).

Previous research examining the effects of medially posted insoles (MPI) on walking and running mechanics has typically demonstrated a small reduction in the peak hindfoot eversion (up to 1-3 degrees) (Liu et al. 2012; Stacoff et al. 2000; Nester et. al.2003; McClean et. al. 2006; Eslami et al. 2009; Nawoczenski et al. 1995). Some studies have also reported alterations in the frontal plane moments, particularly at the ankle and knee joints, when walking or running with MPI (Nester et al. 2003; Telfer et al. 2013; Nigg et al. 2003). These changes are suggested to be resulting from a shift of the center of the pressure (COP) to the direction of insole posting, which thus can affect the lever arm of the ground reaction force to the joint center (Nigg et al. 2003; Kakihana et al. 2005).

While considerable information has accumulated about the effectiveness of insoles on the lower limb mechanics, the majority of the prior studies have been limited in using a single-segment foot model, which primarily reflects the movement of the hindfoot (calcaneus) component (Cheung et al. 2011; Nester 2009). Consequently, relatively little is known for how MPI alters the overall foot motion. Furthermore, many studies have quantified foot motion by using external shoe markers, which do not fully represent the motion of the foot inside the shoe (Sinclair et al. 2014; Cheung et al. 2011).

The present knowledge of the effectiveness of MPI on the overall foot function stems from few studies, which have utilized a multi-segment foot model in quantifying the motion of the forefoot and hindfoot components separately. However, during walking the results have been controversial: while one prior study has found decreased peak forefoot pronation (Hsu et al. 2014), the others have reported alterations primarily at the hindfoot rather than forefoot motion (Bishop et al. 2015; Ferber & Benson 2011). During running, where much greater loads are placed upon the foot and lower limbs (Farley & Ferris 1998), the insole effects on multi-segment foot biomechanics have been reported by only one prior study (Sinclair et al. 2015). The findings suggested that MPI were more effective in reducing pronation movement at the forefoot than hindfoot component. However, based on the above findings the interpretation of the insole effects on the overall foot motion, particularly across both walking and running, is challenging because of differences in participants, insole characteristics and experimental protocols between studies. Knowing whether and how the effectiveness of MPI to control the overall foot motion depends on the mode of locomotion would be clinically important by providing valuable guidance for selecting an appropriate insole according to a certain need.

Therefore, the purpose of this study was to investigate the effects of MPI on walking and running mechanics in overpronating men, using a multi-segment foot model. Based on the existing knowledge, it was hypothesized that MPI would reduce peak eversion movement of both forefoot and hindfoot across walking and running when compared to normal insole (NORM). It was also expected that MPI would alter frontal plane moments by shifting the path of the COP medially under the foot during both walking and running.

2 METHODS

Subjects. Overpronating male subjects were recruited from a student population of University of Jyväskylä via advertisements placed on notice boards. An a priori sample size calculation was conducted based on data of previous multi-segment foot model studies in walking (Hsu et al. 2014) and running (Sinclair et al. 2015). Using frontal plane forefoot-hindfoot data, it was revealed that a sample size of ten subjects would be needed to detect a difference in the mean change in eversion movement with a power of 80% and at $\alpha = 0.05$. After initial telephone screening, 20 injury-free volunteers with self-estimated overpronation were invited to participate in a foot pronation assessment defined by a navicular drop test (Mueller et al. 1993). The eligibility criterion for overpronation was a navicular drop value over 10 mm. This was measured as the distance (mm) between navicular height in barefoot standing with the subtalar joint in a neutral position and in a relaxed stance. This method has been found to have intratester and intertester reliabilities ranging from 0.73 to 0.96 (Mueller et al. 1993; Sell et al. 1993). Out of 20 volunteers, eleven (height 1.76 ± 0.08 m, body mass 74 ± 8 kg) were identified as overpronators and were qualified to be subjects in this study. The mean navicular drop value of these eleven subjects was 12.0 ± 1.1 mm. Participants read and signed an information form and confirmed that they had no previous history of any musculoskeletal problems, such as a recent injury or surgery, which could affect the gait pattern of the subject. All the methods of the study were approved by the local ethics committee and were performed in accordance with the Declaration of Helsinki.

Insole preparation. Preparation of MPI was based on heated orthotic blanks (Footbalance Systems Ltd., Vantaa, Finland). These blanks were set on a molding pillow, after which the subject stepped on it. Molding pillows reacted to the pressure and heat. The physiotherapist guided the foot towards a neutral alignment. The windlass mechanism, with knees slightly flexed (20º) and the first metatarsophalangeal joint dorsiflexed, was used to get a mold of the orthotics. Subjects were instructed to use their MPI every day for approximately two weeks before biomechanical walking and running measurements to familiarize themselves with the insoles.

Biomechanical data collection. Biomechanical walking and running measurements were conducted in an indoor sports hall along a 30 m long track. A Ten-camera system (Vicon T40, Oxford Metrics, Oxford, UK) and three force platforms (BP1200600, AMTI, Watertown, MA, USA) recorded marker positions and ground reaction force (GRF) data synchronously at 300 Hz and 1500 Hz, respectively. The subjects first performed walking trials at a self-selected speed and then performed running trials at a target speed of 4.0m/s. Five successful force plate contacts in any of the three force plates were collected in each condition, which typically required three to six walking and running trials. Data were collected using the same new running shoes (Nike Pegasus 30, neutral shoe) with normal insoles of the shoes (NORM) and with MPI in random order. Two photocells positioned at the mid-part (15-20 m) of the track were used to control the speed within and between shoe conditions (± 10%). The same part of the track was used as a data capture area.

Anthropometric measurements (height, weight, leg length, and knee and ankle diameters) and placement of 28 retro-reflective markers (14 mm in diameter) were performed according to Plug-in gait and Oxford foot models (OFM) (Vicon, Oxford Metrics, Oxford, UK) (figure 1). Holes in the shoe outsole allowed placement of the markers directly on the foot according to OFM. The size of the holes was kept under 1.7cm x 2.5cm so that the integrity of the shoe would not be severely altered (Schultz & Jenkyn, 2012). To eliminate errors associated with marker replacement, the bases of markers were not removed and reattached between NORM and MPI test conditions. To enable this, the markers were taped from the base on the foot and when changing the insole conditions, the marker balls were unscrewed from the base as the shoe was taken off (figure 1). The base of the marker stayed in place while the insole was changed, and when the shoe was put back on, the markers were screwed back on the bases.

Data analysis. Vicon Plug-in gait model with OFM was used to analyze kinematic and kinetic data. Marker trajectories and GRF data were low-pass filtered using a fourth-order Butterworth filter with cutoff frequencies of 12 and 50 Hz, respectively. Five successful force plate contacts of the right leg within each test condition were selected for the analysis. Although we analyzed right side orthotic devices were placed inside both shoes. Period of ground contact was determined based on 20 N vertical force thresholds at the time of the heel contact and toe-off. Data over the stance phase were time normalized (0-100%) and averaged across five accepted walking and running trials to get individual mean curves for the analysis. Lower limb joint moments in the sagittal and frontal planes were calculated via inverse dynamics about an orthogonal axis system located in the distal segment of a joint by taking into account the magnitude of the segmental masses, and location of the mass centres, which were determined based on relative segmental masses reported by Dempster (1955) and the participant's anthropometric data. Joint moments were expressed as external moments and normalized to the body mass (Nm/kg). The path of the center of pressure (COP, derived from the force plate data) under the footwear was analyzed during the stance phase in running based on the COP position in the anterior–posterior and medial–lateral directions of the global system with respect to locations of the heel and 2nd toe markers (Forghany et al. 2014).

Statistical analysis. Kinematic parameters of interest were forefoot angles in respect to hindfoot and tibia and hindfoot angles in respect to tibia in all three planes (sagittal, frontal and transversal). In addition, sagittal and frontal plane moments across ankle, knee and hip joints and COP path in the medio-lateral direction during the stance phase were selected for the comparison. Two-way repeated measures ANOVA were used to test main and interaction effects of locomotion modes (walking, running) and insole conditions (MPI, NORM). For significant events, a Student's paired t-tests were performed to determine the effect of type of insole for walking and running separately. Statistical tests were performed with IBM SPSS software (Version 22.0, Chicago, IL, USA). P values less than 0.05 were considered significant. Symbols are used to describe statistically significant differences as follows: * = $P < 0.05$; ** = $P < 0.01$.

3 RESULTS

Walking and running speeds did not differ between normal insole and MPI (1.70 ± 0.19 m/s vs 1.72 ± 0.20 m/s and 4.04 ± 0.17 m/s vs 4.09 ± 0.13 m/s, respectively). For any of the kinematic parameters, there was no significant interaction between the insole type and locomotion mode, indicating that the responses to MPI were virtually similar during both walking and running. There were significant main effects for forefoot dorsiflexion in respect to hindfoot and tibia. Univariate analysis indicated that walking with MPI increased forefoot-hindfoot dorsiflexion angle, where both greater heel contact (-0.2 ± 3.7 vs 3.0 ± 4.8, p<0.01) and peak (6.2 ± 3.4 vs 8.8 ± 4.1, p<0.05) values were present, when compared to NORM. A trend towards increased forefoot-hindfoot dorsiflexion was also observed at heel contact during running (2.8 ± 4.7 vs 5.1 ± 5.9, p=0.073) with MPI.

In the frontal plane, there were significant main effects for the forefoot pronation in respect to hindfoot and tibia. A significant reduction in the peak forefoot pronation angle in respect to hindfoot was present with MPI during both walking (-0.6 ± 4.0 vs -4.0 ± 3.1 deg., p<0.05) and running (-1.8 ± 4.4 vs -4.7 ± 4.1 deg., p<0.01) when compared to NORM (figure 2A). In addition, reduced forefoot pronation in respect to tibia was also observed with MPI during walking (-5.5 ± 3.1 vs -7.3 ± 4.3 deg., p<0.05) and running (-19.1 ± 8.3 vs -20.7 ± 8.9 deg., p<0.05) compared to NORM (figure 2B). There were no other significant differences in the ankle and foot kinematics during walking or running (supplementary table 1).

In the kinetic parameters, there was a significant main effect for insole for all frontal plane joint moments and a significant interaction effect for the ankle joint moment. Univariate analysis indicated that the peak frontal plane moments were significantly different during running, where 28 % lower ankle (-0.18 ± 0.11 vs -0.25 ±0.11 Nm/kg, $p<0.05$), but 7 % higher knee (2.14 ± 0.69 vs 1.99 ± 0.64 Nm/kg, $p<0.05$) and 6% higher hip (2.00 ± 0.40 vs 2.13 ± 0.45 Nm/kg, $p<0.05$) moments were noted with MPI compared to NORM (figure 3). Frontal plane moments in walking as well as sagittal plane moments across walking and running showed no significant differences (Supplementary table 2). In the COP movement there was a significant interaction for insole type. It was noted that during running MPI shifted the path of the COP medially under the foot on average 5.5 mm ($p<0.05$) compared to NORM, while no such changes were present during walking (figure 4).

4 DISCUSSION

The aim of this study was to examine the effects of medially posted insoles (MPI) on walking and running kinematics and kinetics in overpronating men. It was hypothesized that MPI would reduce peak eversion of the hindfoot and forefoot during both walking and running when compared to NORM. This hypothesis was partially supported since peak forefoot eversion was reduced with MPI. However, no differences were found in the peak hindfoot eversion. For the second hypothesis, it was suggested that MPI would alter frontal plane moments by shifting the path of COP medially under the foot across walking and running. This prediction was supported in running, but not in walking.

As far as we are aware the present study was the first one to determine the effects of MPI on the multi-segment foot biomechanics across walking and running. The results showed that regardless of substantial biomechanical differences between walking and running gaits (Farley & Ferris 1998), the kinematic response to insoles was essentially similar across the modes. This knowledge may be important from a clinical point of view because it has been previously unclear whether greater external loads during running than walking affect the effectiveness of anti-pronation insoles to control the foot motion. Thus, the present findings imply that there is no need to have stiffer insoles for running to achieve similar reduction for the foot motion than in walking, at least not with the insoles that were used in this experiment.

The greatest insole effects were present in the forefoot motion, where MPI consistently reduced the peak pronation angle in respect to hindfoot (~3°) and tibia (~1.5°) during both walking and running when compared to NORM. In addition, the forefoot segment in respect to hindfoot and tibia showed a consistent tendency towards greater dorsiflexion with MPI, suggesting increased height of the foot arch compared to NORM, although the difference in running did not reach statistical significance. While these results agree well with some previous findings in walking (Hsu et al. 2014) and running (Sinclair et al. 2015), not all studies (Bishop et al. 2015; Ferber & Benson 2011) have found alterations in the forefoot motion when wearing medially posted insoles. It is possible that differences in the participants or insoles used explain these different effects.

Contrary to our expectations, there were no differences in the peak hindfoot eversion during walking or running between insole conditions. This finding may be explained by the fact that there were no controlling features like medial wedging under the heel of the MPI. It is possible that such a design is necessary for achieving a reduction in the hindfoot eversion movement as has been previously shown in several studies across walking (Nester et al.2003; Telfer 2013) and running (Stacoff et al. 2000; Eslami et al. 2009; McClean et al. 2006).

Because all the subjects participating in this study had excessive pronation, the kinematic changes with MPI in the forefoot component most likely resulted from the mechanical support of the medial longitudinal arch, which also led to alterations in the frontal plane kinetics. It was interesting, however, that although the kinematic response to MPI was virtually similar across walking and running, alterations in the kinetic parameters were gait-specific. While no differences were observed in walking, running with MPI increased the hip and knee adduction moments but decreased the ankle inversion moment and changed the path of the COP medially under the foot. Although not all studies (Bishop et al. 2015) have associated the usage of insoles to kinetic changes, there is evidence to suggest that the insole posting often causes a shift in the COP path towards the posting side (Nigg et al. 2003; Kakihana et al. 2005), thus altering the frontal plane joint moments (Nester et al. 2003; Telfer et al. 2013). Possibly the mechanical support effect and thus the extent to which insoles alter COP and frontal plane moments, depends at least partly on the amount of midsole compression (and foot depression into midsole), which becomes greater with increased external loads (Verdejo & Mills 2004). This could explain the different kinetic response to MPI between walking and running in the present study. Another possible explanation for greater kinetic response to MPI during running may be greater ankle eversion movement in running compared to walking, which may accentuate insole supporting effect on the medial longitudinal arch. It is, however, unclear what effects this medial shift of the COP and slightly increased frontal plane moments in the hip and knee joints seen in the current experiment could have in long-term on the corresponding joints.

Certain limitations of the current study should be considered when interpreting its findings. All participants were males with overpronation, and therefore, caution must be made in generalizing these results to females or people with different foot structure. In addition, it should be noted that walking and running speeds in the present study were greater (~1.7 and ~4.0 m/s, respectively) than those used in many previous investigations (Cheung et al. 2011) (~1.3 and ~3 m/s, respectively), so it may be that our results cannot be generalized to lower walking and running speeds. Finally, skin marker based approach has its limitations. Especially, the kinematic calculations are highly dependent upon marker placement and may also be influenced by soft-tissue movement artifact (Westbland et al. 2002). However, to eliminate errors associated with marker replacement, the bases of markers were not removed and reattached between test conditions. We therefore believe that our overall conclusions drawn from the data are not significantly influenced by measurement errors.

In conclusion, the present study showed that MPI primarily affected the forefoot motion by reducing the peak pronation movement across walking and running when compared to NORM. While kinetic responses (alterations in COP path and frontal plane moments) to MPI were more pronounced in running than walking, kinematic changes were essentially similar across both modes, suggesting that there is no need to have stiffer insoles for running to achieve similar reduction for the foot pronation than in walking. This may, however, depend on the insole type used.

Conflict of interest statement

The authors declare that they have no conflict of interest relating to the material presented in this article.

REFERENCES

Bishop C , Arnold JB & May T 2015. Effects of Taping and Orthoses on Foot Biomechanics in Adults with Flat Arched Feet, Medicine and Science in Sport and Exersice, In press.

Cheung, R.C., Chung T.T. & Ng G.Y. 2011, Efficacies of different external controls for excessive foot pronation: a meta-analysis. British Journal of Sports Medicine, 45, 743-751.

Collins N, Crossley K, Beller E, Darnell R, McPoil T, Vicenzino B. 2008. Foot orthoses and physiotherapy in the treatment of patellofemoral pain syndrome: randomised clinical trial. British Medical Journal. 337, 1-8.

Donatelli, R., Hurlbert, C., Conaway, D. & St.Pierre R. 1988. Biomechanical foot orthotics: A retrospective study. The Journal of Orthopaedic and Sports Physical Therapy 6, 615 - 624.

Eng JJ & Pierrynowski MR. 1993. Evaluation of soft foot orthotics in the treatment of patellofemoral pain syndrome. Physical Therapy 73: 62-68.

Eslami, M., Begon, M., Hinse, S., Sadeghi, H., Popov, P. & Allard, P. 2009. Effect of foot orthoses on magnitude and timing of rearfoot and tibial motions, ground reaction force and knee moment during running. Journal of Science and Medicine in Sport 12, 679 - 684.

Farley, C. T. & Ferris, D. P. 1998. Biomechanics of walking and running: center of mass movements to muscle action. Exercise and Sport Sciences Reviews 26, 253-285.

Ferber, R. & Benson, B., 2011. Changes in multi-segment foot biomechanics with a heat-mouldable semi-custom foot orthotic device. J Foot Ankle Res, 4, 1-8.

Forghany S., Nester C.J., Richards B, Hatton A.L., Liu A. 2014. Rollover footwear affects lower limb biomechanics during walking. Gait & Posture. 39, 205-12

Hsu, W-H., Lewis, C.L., Monaghan, G.M., Saltzman, E., Hamill, J. & Holt, K.G. 2014. Orthoses posted in both the forefoot and rearfoot reduce moments and angular impulses on lower extremity joints during walking. Journal of Biomechanics 47, 2618 - 2625.

Kakihana, W., Akai, M., Nakazawa, K., Takashima, T., Naito, K & Torii, S., 2005. Effects of laterally wedged insoles on knee and subtalar joint moments. Archives of physical medicine and rehabilitation, 86, 1465-1471

Leardini, A., Chiari L., Della Croce U. & Cappozzo A. 2005. Human movement analysis using stereophotogrammetry: Part 3. Soft tissue artifact assessment and compensation. Gait & posture 21, 212-225.

Liu, A., Nester, C.J., Jones, R.K., Lundgren, P., Lundberg, A., Arndt, A, & Wolf, P., 2012. Effect of an antipronation foot orthosis on ankle and subtalar kinematics. Medicine and science in sports and exercise, 44, 2384-2391.

MacLean, C., McClay Davis, I. & Hamill, C.L. 2006. Influence of a custom foot orthotic intervention on lower extremity dynamics in healthy runners. Clinical Biomechanics 21, 623 - 630.

McClay, I. & Manal, K. 1997. Coupling parameters in runners with normal and excessive pronation. Journal of Applied Biomechanics 13, 109 - 124.

McClay, I. & Manal, K. 1998. A comparison of three-dimensional lower extremity kinematics during running between excessive pronators and normals. Clinical Biomechanics 3, 195 - 203.

Mueller M.J., Host J.V., & Norton B.J. 1993. Navicular drop as a composite measure of excessive pronation. Journal of the American Podiatric Medical Association 83: 198–202.

Nawoczenski, D.A., Cook, T.M. & Saltzman C.L. 1995. The effect of foot orthotics on three-dimensional kinematics of the leg and rearfoot during running. Journal of Orthopaedic and Sports Physical Therapy 21, 317 - 327.

Neil, B.S., Griffiths, I.B., Dowling, G.J., Murley, G.S., Munteanu, S.E., Frattenovich Smith, M.M., Collins, N.J. & Barton, C.J. 2014. Foot posture as a risk factor for lower limb overuse injury: a systematic review and meta-analysis. Journal of Foot and Ankle Research 7, 55.

Nester, C.J., 2009. Lessons from dynamic cadaver and invasive bone pin studies: do we know how the foot really moves during gait? Journal of Foot and Ankle Research 2. 1-7.

Nester, C.J., van der Linden M.L. & Bowker, P. 2003. Effect of foot orthoses on the kinematics and kinetics of normal walking gait. Gait and Posture 17, 180 - 187.

Nigg, B.M., Stergiou, P., Cole, G., Stefanyshyn, D., Mündermann, A. & Humble, N. 2003. Effect of shoe inserts on kinematics, center of pressure, and leg joint moments during running. Medicine and Science in Sports and Exercise 2, 314 - 319.

Rubin, R. & Menz, H.B. 2005. Use of laterally wedged custom foot orthoses to reduce pain associated with medial knee osteoarthritis. Journal of the American Podiatric Medical Association 4, 347 - 352.

Schultz, R., & Jenkyn, T. 2012. Determining the maximum diameter for holes in the shoe without compromising shoe integrity when using a multi-segment foot model. Medical Engineering and Physics 34, 118 - 22.

Sell KE, Verity TM, Worrell TW, Pease BJ, Wigglesworth J. 1994. Two measurement techniques for assessing subtalar joint position: a reliability study. Journal of Orthopaedic Sports Physical Therapy. 19:162–7.

Shih, Y-F., Wen, Y-K. & Chen, W-Y. 2011. Application of wedged foot orthosis effectively reduces pain in runners with pronated foot: a randomized clinical study. Clinical Rehabilitation 10, 913 – 923.

Sinclair, J., Isherwood, J. & Taylor, P.J. 2015. The effects of orthotic intervention on multi-segment foot kinematics and plantar fascia strain in recreational runners. Journal of Applied Biomechanics, 1, 28-34.

Sinclair, J., Taylor, P. J., Hebron, J., & Chockalingam, N. 2014. Differences in multi-segment foot kinematics measured using skin and shoe mounted markers. The Foot and Ankle Online Journal. 7, 7.

Stacoff, A., Reinschmidt, C., Nigg, B.M., van den Bogert, A.J., Lundberg, A., Denoth, J. & Stussi E. 2000. Effects of foot orthoses on skeletal motion during running. Clinical Biomechanics 15, 54 - 64.

Telfer, S, Abbott, M, Steultjens, MP & Woodburn, J. 2013. Dose–response effects of customised foot orthoses on lower limb kinematics and kinetics in pronated foot type. Journal of Biomechanics 46, 1489-1495.

Verdejo, R. & Mills, N.J., 2004. Heel–shoe interactions and the durability of EVA foam running-shoe midsoles. Journal of biomechanics 37, 1379-1386.

Westblad, P, Hashimoto, T, Winson, I, Lundberg, A & Arndt, A. 2002. Differences in ankle-joint complex motion during the stance phase of walking as measured by superficial and bone-anchored markers. Foot & ankle international 23, 856-63.

FIGURE 1. Marker placement of the Oxford multi-segment foot model in the test shoe. The bases of the markers were kept attached on the foot when changing insoles by unscrewing the reflecting balls.

FIGURE 2. Peak forefoot pronation angle in respect to hindfoot (A) and tibia (B) during walking and running.

FIGURE 3. External moments in the frontal plane for the hip, knee and ankle joints during walking (A) and running (B).

FIGURE 4. The path of the Center of the Pressure (COP) under the foot during walking (A) and running (B).