

A comparison of the free moment pattern between normal and hyper-pronated aligned feet in female subjects during the stance phase of gait

Yazdani F.¹, Razeghi M.^{1*}, Ebrahimi S.¹

ABSTRACT

Background: Excessive range of adductory free moment of the ground reaction force may potentially increase the risk of lower extremity injuries by applying a higher torsional load transmitted to the proximal parts.

Objective: It was hypothesized that the free moment pattern might be different between hyper-pronated and normal feet subjects. Moreover, a correlation would exist between peak adduction free moment and peak ankle-foot complex abduction at the stance phase of walking.

Material and Methods: In this cross sectional study, thirty female participants were divided into two groups of asymptomatic hyper-pronated and normal feet. Kinetic and kinematic data were collected using a single force plate and a six-camera motion analysis system during three successful free speed walking trials. Ensemble average curves were extracted from the time normalized individual trials of the stance phase for both free moment and peak ankle-foot complex abduction parameters.

Results: Significant differences in peak adductory free moment, peak ankle-foot complex eversion and peak ankle-foot complex abduction were found between normal and hyper-pronated groups (4.90 ± 0.97 Vs. 5.94 ± 0.88 , $P < 0.01$), (3.30 ± 0.95 Vs. 6.28 ± 1.47 , $P < 0.01$) and (4.52 ± 1.16 Vs. 8.23 ± 2.52 , $P < 0.01$) respectively. A significant positive correlation was found between the peak adduction free moment and peak ankle-foot complex abduction in both groups, which was more strongly positive in hyper-pronated group ($r = 0.745$, $p < 0.01$ for normal group and $r = 0.900$, $p < 0.01$ for hyper-pronated group).

Conclusion: As a good measure of torque which is transmitted to the lower extremity, may free moment be a useful biomechanical indicator for both clinical and research purposes.

Citation: Yazdani F, Razeghi M, Ebrahimi S. A comparison of the free moment pattern between normal and hyper-pronated aligned feet in female subjects during the stance phase of gait. *J Biomed Phys Eng*. 2020;10(1):93-102. doi: 10.31661/jbpe.v0i0.639.

Keywords

Gait Analysis; Flatfoot; Kinematics; Kinetics

Introduction

The unique and specific structure of foot creates a dynamic link between ground and the human body, rendering it perfect for bipedal locomotion [1] and ideal for adjusting to different walking conditions [2]. Different mechanisms have been implemented for such a dynamic body-environment interaction.

The subtalar joint (STJ) complex permits the foot to act both as a flex-

¹PhD, Rehabilitation Sciences Research Centre, Department of Physical Therapy, School of Rehabilitation Sciences, Shiraz University of Medical Sciences, Shiraz, Iran

*Corresponding author: M. Razeghi
Department of Physiotherapy, Shiraz School of Rehabilitation Sciences, Abiverdi 1 Avenue, Chamran Boulevard, Shiraz, Iran
E-mail: razeghm@sums.ac.ir

Received: 3 September 2016
Accepted: 25 September 2016

ible mechanism (by pronation) and a rigid structure (by supination) during propulsion, enabling the foot to adapt to uneven surfaces as well as to facilitate the transmission of forces [3]. Due to its oblique axis, STJ causes a foot pronation (combination of eversion, abduction and dorsiflexion) directly after initial contact with the ground during normal walking [4, 5]. This maneuver is suggested to be an effective mechanism for shock absorption and foot accommodation loading response phase of gait to different terrains [6]. Any factor affecting foot's normal function may disturb this mechanism [7].

Hyper-pronation or flat foot is a medical condition in which the normal function of the subtalar joint is disrupted, and as a result, the medial longitudinal arch becomes lower than normal [8, 9]. This is a highly prevalent malalignment of the foot [10, 11] that could potentially interfere with the load transmission from the foot to more proximal segments [12]. As a consequence, Hyper-pronation could result in overloading the whole mechanical body chain, especially during weight bearing activities [12-16]. Over the time, abnormal load distribution is known to lead to an increase in the strain on soft tissues which can become symptomatic [12].

In flat-foot condition, changes in the joint kinematics [17-19], as well as the electromyographic activity of muscles have been reported [20]. However, less is known about the consequences of these alterations on the closed kinetic chain. From a clinical perspective, alteration in kinetic chain parameters could increase the risk of soft tissue injury. While plenty of studies have focused on the mechanical consequences of hyper-pronation on the movement chain [7, 17, 18, 21-25], less attention has been given to the possible effect of this malalignment on the moments and forces acting on the whole lower extremity. Some authors have studied the effects of foot pronation on the ground reaction forces (GRFs) or center of pressure (COP) [26-28].

Moreover, the effects of different orthotic devices on GRFs have been compared previously [27, 29, 30]. In these studies, no strong correlation was found between indicators of rearfoot pronation and GRF parameters. Also, some of the results about the correction effects of the orthotic devices were contrary to the accepted concept of "aligning the skeleton" with inserts and orthotics.

One component of ground reaction which has been less studied in the literature is the torque on the feet about the vertical axis of the center of pressure [31] called "free moment (FM)". Although it was reported that changes in the rearfoot pronation by specific footwear modifications in runners could potentially induce the significant changes in parameters of the FM [32], its potential differences in the true pronated population has not been investigated.

A research study showed that in runners with a remarkable history of tibial stress fracture most of the FM parameters were significantly increased [31]. It seems that, as a measure of the torsional torque about a vertical axis applied on lower extremity during the stance phase, FM is worthy of further investigation in relation to excessive pronation which is a malalignment with rotatory nature effects.

Adductory FM acts to counter the toe out and abductory FM withstands the toe in of the foot during the stance phase of the gait. Based on the fact that abduction is a component of pronation, a relationship could be expected between values for FM variables and the ankle-foot complex abduction. In the present study, the maximum adductory component of free moment was considered (ADD FM). This is a component of the free moment which acts to resist toeing out in the initial phase of stance (for details see the Method section). We believe that ADD FM in the flat foot subjects will be different from that of non-pronated subjects. This novel evidence adds to our understanding about the impact of functional flat foot on kinetic chain of lower extremity

from a biomechanical perspective and may introduce the useful indicator of the rotatory torque which is applied on the lower extremity during stance phase for research and clinical purposes.

Material and Methods

Participants

In this cross sectional study, thirty asymptomatic female subjects (aged 18-30 years), 15 with neutrally aligned and 15 with functional flat-foot type, were selected after a complete lower extremity clinical examination. A convenience sampling method was used.

The inclusion criteria for participants were having normal range of motion of the hip, knee, ankle and metatarsophalangeal joints (based on Goniometric Assessments), normal (grade five) strength in major lower extremity muscles (as manifested by manual muscle testing performed by the same examiner), being self-ambulatory and having either bilateral flat-foot (for the patient group, n=15) or natural alignment (for the control group, n=15). The subjects with functional flat-foot were determined by measuring the resting calcaneal stance position (RCSP) in the frontal plane during weight bearing. High intra- and inter-rater reliability was previously reported for this measurement [33]. RCSP between 2° of inversion and 2° of eversion represented a neutrally aligned foot, and a RCSP of more than or equal to 4° of eversion represented a flat-foot subject [34]. Moreover, feiss line test was used to define the subjects with flexible flat-foot [35], which is performed during rest and weight bearing situation [36]. Pronated feet subjects was labeled as flexible if the navicular bone was positioned under the line in both weigh bearing and non-weigh bearing conditions [37]. High intraday-, intra- and inter-tester reliability was reported for this measurement [38].

The exclusion criteria were functional or structural orthopedic maladies that would pre-

vent normal stance phase of walkingsuch as a limb length discrepancy greater than 1 cm, excessive knee hyper-extension, abnormal knee varus or valgus and chronic pain due to structural or functional problems in the lower extremity bones, ligaments or menisci, neurological ailments affecting the gait such as neuropathy or other sensory disturbances, or any past history of orthopedic lower limb surgery. The study had a cross-sectional design and hyper-pronated subjects were matched to each normal subject based on age, body mass and height.

All objective measurements of the study were performed by the same experimenter for all subjects to avoid any possible inter-examiner discrepancy.

Prior to participation, all subjects were informed about the nature of the study and signed the informed consent form, approved by the Human Ethics Committee of Shiraz University of Medical Sciences.

Measure and Procedures

Ground Reaction Force [GRF] data were collected using a single force plate (Kistler Instrument AG, Winterthur, Switzerland) sampling at 240 Hz. Kinematic data were collected using a six-camera motion analysis system (ProReflex, Qualisys AB, Göteborg, Sweden) sampling at 120 Hz.

To measure the anthropometric data and subsequently build a 6-degree of freedom model, retroreflective calibration markers with a diameter of 19 mm were placed on the following anatomical points: dominant lower extremity's medial and lateral femoral condyles, medial and lateral malleoli, the first and fifth metatarsal heads, the fifth metatarsal base and the center of calcaneus (Figure 1). One set of cluster markers containing 4 tracking markers secured on a polyform material was placed on the lateral distal one the third of the shank to track the movement of the desired segments [39,40]. The subjects stood on the force plate and assumed a normal posture for a few sec-



Figure 1: Markers placement definition

onds to capture a static trial for the sake of model building.

Following multiple practice trials, the subjects performed three barefoot walking trials at a self-selected speed, and we recorded them. Only the trials in which the subject's dominant foot landed on the force plate without any disturbance to their gait were considered for further analysis. To determine the dominant foot of the subjects, they were asked to kick a ball, whichever foot they kick was considered as dominant foot.

Data were synchronously recorded with QTM software (ProReflex, Qualisys AB, Göteborg, Sweden). All subsequent analyses were performed offline in Visual 3D software (Cmotion Inc., Rockville, MD).

Data Analysis

Raw data were filtered using a fourth order low-pass Butterworth filter with a cut off frequency of 6 Hz for kinematic data and 15 Hz for kinetic data [41]. The moment, M_z , which acts about a vertical axis at the center of force platform has two components. One component, the FM, is the torsional torque at the contact point. Depending on the direction, positive FM (ADD FM) counters toeing out

and negative FM (ABD FM) acts to withstand toeing in during stance (Figure 2). The second component is the moment of shear force which acts through the COP. Holden and Cavanagh (1991) provided detailed explanation of the relationship between the two components and M_z . FM was derived from equations defining the moment and force components from force platform output according to the manufacturer's instructions (Kistler Instrument AG, Winterthur, Switzerland). Prior to estimating FM, force plate channels were baseline set zero.

$$FM = M_z - (F_y \cdot a_x) + (F_x \cdot a_y)$$

$$a_x = -M_y / F_z \quad \text{and} \quad a_y = M_x / F_z$$

where M_z is the moment about the vertical axis, "a" the x- coordinate of force application point (COP), " F_y " the ground reaction force in

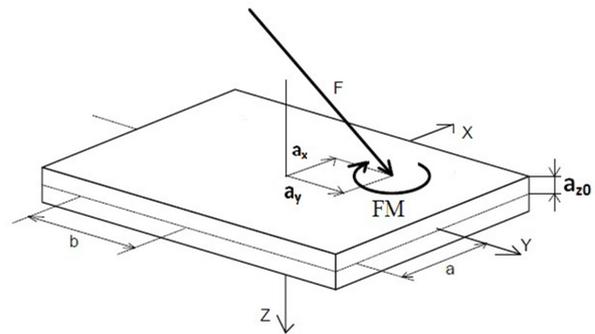


Figure 2: Representation of FM (with kind permission of the publisher, Kistler Instrument AG, Winterthur, Switzerland)

y- direction, " a_y " the y- coordinate of force application point (COP), " F_x " the ground reaction force in x-direction, M_y , the plate moment about top plate surface about y- axis, " F_z " the ground reaction force in z- direction and M_x , the plate moment about top plate surface about x- axis [31]. According to the force plate coordinate system, positive y- axis was in the direction of progression, positive x- axis was to the left when facing the direction of progression and positive z- axis was vertically downward.

FM was normalized to the body weight

and height to reduce the effect of these factors among subjects, so the resultant FM was dimensionless. Peak ADD FM was the maximum positive value of FM during stance. Moreover, peak ankle-foot complex abduction during the stance phase was determined for each trial. Each variable was averaged over three trials per subject. Ensemble average curves were extracted from time normalized individual trials of the stance phase for both FM and peak ankle-foot complex abduction parameters to aid the interpretation of any pattern differences of these parameters among groups.

Statistical Analysis

The Kolmogorov-Smirnov test was applied to test the normality of data distribution. Independent *t*-tests were used to test the significant differences among groups. Since we were only interested in whether the value of peak ADD FM, peak ankle-foot complex abduction and peak ankle-foot complex eversion would be greater than normal in the hyper-pronated group, one-tailed tests were used. Pearson correlation test was performed to evaluate the correlation between peak ADD FM and peak ankle-foot complex abduction in the normal and

hyper-pronated groups. All statistical analyses were performed using SPSS 22.0 (SPSS, Chicago, USA). The level of significance for all the tests was set to 0.05.

Results

No statistical difference was found between demographic characteristics of the subjects (Table 1). The ensemble average curves demonstrated that ADD FM variables (magnitude and duration) and the ankle-foot complex abduction variables (magnitude and duration) were both greater in hyper-pronated feet group than the normal one (Figures 3 and 4). Significant differences in peak ADD FM, peak ankle-foot complex abduction and peak ankle-foot complex eversion were found among normal

Table 1: The participants' demographic characteristics (mean \pm SD).

Variable	Normal fee group (n=15)	Hyper-pronated feet group(n=15)	*P Value
Age (year)	21.20 \pm 2.04	21.90 \pm 1.66	0.423
Height (cm)	162 \pm 5.6	159 \pm 6.3	0.426
Body mass (kg)	56.20 \pm 6.49	56.90 \pm 6.08	1.00

*Significant at $p < 0.05$

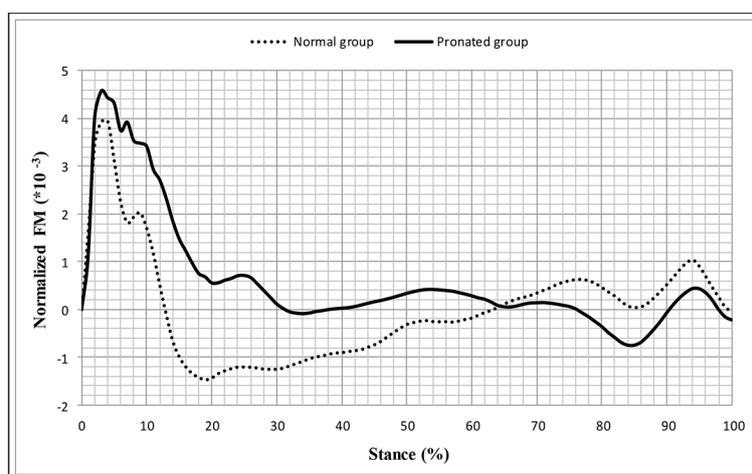


Figure 3: Ensemble average normalized free moment during stance phase in females with pronated feet and normal feet posture. Positive values indicate adductory free moment and negative values indicate abductory free moment.

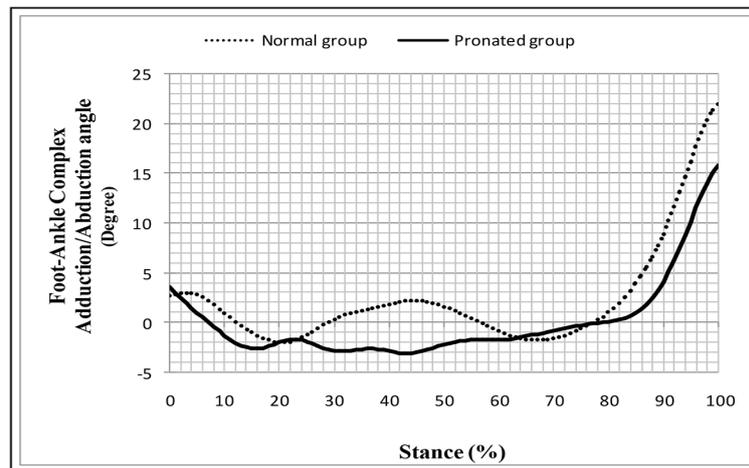


Figure 4: Ensemble average normalized foot-ankle complex abduction/adduction angle during the stance phase in females with pronated feet and normal feet posture. Positive values indicate movement toward adduction, and negative values indicate movement toward abduction.

and hyper-pronated groups (4.90 ± 0.97 vs. 5.94 ± 0.88 , $P < 0.01$), (4.52 ± 1.16 vs. 8.23 ± 2.52 , $P < 0.01$) and (3.30 ± 0.95 vs. 6.28 ± 1.47 , $P < 0.01$) respectively. Moreover, a positive correlation was found between peak ADD FM and peak ankle-foot complex abduction in normal ($r = 0.745$, $p < 0.01$) and hyper-pronated ($r = 0.900$, $p < 0.01$) groups. However, the positive correlation which was found between peak ADD FM and peak ankle-foot complex abduction in the normal group was weaker than that of hyper-pronated group.

Discussion

To the best of our knowledge, the present study is the first research which provides information on the characteristics of FM pattern in normal and hyper-pronated feet of female subjects during walking. The primary purpose of this study was to test the hypothesis that the peak ADD FM is different between hyper-pronated and normal feet subjects. Since no clear evidence on the measurements of FM can be found on the true pronated feet population in the literature, this study attempts to determine the pattern of FM in both hyper-pronated and normal feet alignment groups.

Based on the ensemble average curves, our results demonstrated that FM acts in positive direction for about 30% of support phase in the pronated group and after a short time of changing direction, it continues to act in a positive direction again till 75% of stance phase while the ankle-foot complex is in abduction position. However, in the normal group, FM resists foot abduction till 13% of stance phase, and then it acts in a negative direction to resist foot adduction (Figure 3).

The results of the present study demonstrated that the increased ankle-foot complex abduction variables (magnitude and duration) in the hyper-pronated group (Figure 4) could parallelly lead to increased ADD FM variables (magnitude and duration). The trend of pattern changes found in this study is consistent with the report of Holden and Cavanagh (1991), who showed FM vs. time pattern for three different running shoes designed to induce different rear foot postures [32]. In this study, peak FM and net angular impulse of FM both increased with increases in pronation. However, it should be considered that inducing subtalar pronation in the normal subjects emphasizes the immediate effects on the normal alignment

and not necessarily the prolonged adaptive effects.

In another study which was performed on runners, investigators found significantly greater FM parameters in subjects with tibial stress fracture history [31].

Due to different methodologies used by previously mentioned studies, the values for FM in the present study are not comparable with them [31,32]. In sum, they were conducted during running and based on the literature, speed is an important factor which can affect the movement mechanics [42] and consequently torque transmission to the lower limb [31].

We evaluated the differences in peak ankle-foot complex eversion among groups to assure that the ankle-foot complex misalignment would present during the dynamic task of walking in the hyper-pronated group. We investigated the differences in peak ADD FM and peak ankle-foot complex abduction between hyper-pronated feet female subjects and those with normal feet alignment. In agreement with our results, Levinger et al. observed the significantly greater peak forefoot abduction in the stance phase of the flat-arched foot subjects in comparison with normal foot subjects [24].

Peak ADD FM may reflect the highest internally rotating torque experienced by the entire lower extremity while subtalar pronation is happening. Therefore, the internally rotating moment may be accompanied by previously suggested kinematic changes of the lower extremity due to excessive subtalar pronation like increased internal rotation of the tibia [43] and simultaneously the internal rotation of the femur [43,44]. Some studies suggested that coupling exists between the forefoot and rear foot movement [45,46] and between rear foot eversion and lower limb transverse plane movements [47,48].

Greater ADD FM found in the pronated feet group and higher correlation between ADD FM and peak ankle-foot complex abduction

in this group suggest that higher magnitude of ADD FM may be associated with excessive subtalar pronation. In other words, as pronation increases during early stance, ADD FM would increase to resist the tendency of ankle-foot complex to be abducted.

It is important to note the limitations in this study which might affect the generalizability of the findings. First, although no universally accepted measure of foot type classification exists [49], it seems that the gold standard method to classify the foot type involves dynamic classification assessments [35], but in the present study, the static foot type classification method was used to categorize the foot posture. Although no gender-based difference in FM pattern was reported in the literature, another limitation of this study is that it was done on female subjects.

From a clinical point of view, the importance of the FM as an indicator of the rotatory torque which is applied on the lower extremity should be considered. Excessive transverse torques acting on lower extremity are suggested to be the risk factors for several hyper-pronation related injuries, especially in runners [50, 51]. In a clinical setup by means of a force plate, the assessment of this torque could be easily available.

The results of the present study support the suggested possible mechanism of the injury in flat foot condition, so further studies may provide insights into the role of different treatment methods for foot pronation correction on balancing the FM values.

Conclusion

In conclusion, this study has found that peak ADD FM is significantly higher in females with subtalar hyper-pronation compared to a control group with normal foot alignment. Based on ensemble average curves, it seems that the positive direction part of the FM of ground reaction is more prolonged in duration during the stance phase of walking in hyper-pronated feet group.

In terms of biomechanics, identifying lower extremity kinetic changes due to excessive subtalar pronation is important for clinical practice and research. FM can be used as a biomechanical variable, allowing for development of more effective treatment strategies and efficient prevention.

Acknowledgment

The present study was financially supported by Shiraz University of Medical Sciences. We thank Dr. Nasrin Shokrpour at Research Consultation Center of Shiraz University of Medical Sciences for editorial assistance.

Conflict of Interest

Authors declare no conflicts of interests

References

1. Wang WJ, Crompton RH. Analysis of the human and ape foot during bipedal standing with implications for the evolution of the foot. *J Biomech.* 2004;**37**:1831-6. doi.org/10.1016/j.jbiomech.2004.02.036. PubMed PMID: 15519591.
2. Shariatmadari M, English R, Rothwell G. RETRACTED: A review of developments into therapeutic running footwear research. *Procedia Engineering.* 2010;**2**:2811-6. doi.org/10.1016/j.proeng.2010.04.071.
3. Close JR, Inman VT, Poor PM, Todd FN. The function of the subtalar joint. *Clin Orthop Relat Res.* 1967;**50**:159-79. doi.org/10.1097/00003086-196701000-00016. PubMed PMID: 6029014.
4. Karatsolis K, Nikolopoulos CS, Papadopoulos ES, Vagenas G, Terzis E, Athanasopoulos S. Eversion and inversion muscle group peak torque in hyperpronated and normal individuals. *Foot (Edinb).* 2009;**19**:29-35. doi.org/10.1016/j.foot.2008.06.006. PubMed PMID: 20307446.
5. Pohl MB, Messenger N, Buckley JG. Forefoot, rearfoot and shank coupling: effect of variations in speed and mode of gait. *Gait Posture.* 2007;**25**:295-302. doi.org/10.1016/j.gaitpost.2006.04.012. PubMed PMID: 16759862.
6. Duval K, Lam T, Sanderson D. The mechanical relationship between the rearfoot, pelvis and low-back. *Gait Posture.* 2010;**32**:637-40. doi.org/10.1016/j.gaitpost.2010.09.007. PubMed PMID: 20889344.
7. DeLeo AT, Dierks TA, Ferber R, Davis IS. Lower extremity joint coupling during running: a current up-

date. *Clin Biomech (Bristol, Avon).* 2004;**19**:983-91. doi.org/10.1016/j.clinbiomech.2004.07.005. PubMed PMID: 15531047.

8. Donatelli RA, Hurlburt C, Conaway D, St Pierre R. Biomechanical foot orthotics: a retrospective study. *J Orthop Sports Phys Ther.* 1988;**10**:205-12. doi.org/10.2519/jospt.1988.10.6.205. PubMed PMID: 18796957.
9. Haight HJ, Dahm DL, Smith J, Krause DA. Measuring standing hindfoot alignment: reliability of goniometric and visual measurements. *Arch Phys Med Rehabil.* 2005;**86**:571-5. doi.org/10.1016/j.apmr.2004.05.014. PubMed PMID: 15759246.
10. Dunn JE, Link CL, Felson DT, Crincoli MG, Keyser JJ, McKinlay JB. Prevalence of foot and ankle conditions in a multiethnic community sample of older adults. *Am J Epidemiol.* 2004;**159**:491-8. doi.org/10.1093/aje/kwh071. PubMed PMID: 14977645.
11. Shibuya N, Jupiter DC, Ciliberti LJ, VanBuren V, La Fontaine J. Characteristics of adult flatfoot in the United States. *J Foot Ankle Surg.* 2010;**49**:363-8. doi.org/10.1053/j.jfas.2010.04.001. PubMed PMID: 20537928.
12. Tiberio D. Pathomechanics of structural foot deformities. *Phys Ther.* 1988;**68**:1840-9. doi.org/10.1093/ptj/68.12.1840. PubMed PMID: 3194451.
13. Cornwall MW, McPoil TG. Footwear and foot orthotic effectiveness research: a new approach. *J Orthop Sports Phys Ther.* 1995;**21**:337-44. doi.org/10.2519/jospt.1995.21.6.337. PubMed PMID: 7655477.
14. Klingman RE, Liaos SM, Hardin KM. The effect of subtalar joint posting on patellar glide position in subjects with excessive rearfoot pronation. *J Orthop Sports Phys Ther.* 1997;**25**:185-91. doi.org/10.2519/jospt.1997.25.3.185. PubMed PMID: 9048324.
15. Stergiou N, Bates BT, James SL. Asynchrony between subtalar and knee joint function during running. *Med Sci Sports Exerc.* 1999;**31**:1645-55. doi.org/10.1097/00005768-199911000-00023. PubMed PMID: 10589870.
16. Tiberio D. The effect of excessive subtalar joint pronation on patellofemoral mechanics: a theoretical model. *J Orthop Sports Phys Ther.* 1987;**9**:160-5. doi.org/10.2519/jospt.1987.9.4.160. PubMed PMID: 18797010.
17. Cobb SC, Tis LL, Johnson JT, Wang YT, Geil MD, McCarty FA. The effect of low-mobile foot posture on multi-segment medial foot model gait

- kinematics. *Gait Posture*. 2009;**30**:334-9. doi.org/10.1016/j.gaitpost.2009.06.005. PubMed PMID: 19615908.
18. Houck JR, Tome JM, Nawoczenski DA. Subtalar neutral position as an offset for a kinematic model of the foot during walking. *Gait Posture*. 2008;**28**:29-37. doi.org/10.1016/j.gaitpost.2007.09.008. PubMed PMID: 17988870.
 19. Hunt AE, Smith RM. Mechanics and control of the flat versus normal foot during the stance phase of walking. *Clin Biomech (Bristol, Avon)*. 2004;**19**:391-7. doi.org/10.1016/j.clinbiomech.2003.12.010. PubMed PMID: 15109760.
 20. Murley GS, Landorf KB, Menz HB, Bird AR. Effect of foot posture, foot orthoses and footwear on lower limb muscle activity during walking and running: a systematic review. *Gait Posture*. 2009;**29**:172-87. doi.org/10.1016/j.gaitpost.2008.08.015. PubMed PMID: 18922696.
 21. Barton CJ, Levinger P, Crossley KM, Webster KE, Menz HB. Relationships between the Foot Posture Index and foot kinematics during gait in individuals with and without patellofemoral pain syndrome. *J Foot Ankle Res*. 2011;**4**:10. doi.org/10.1186/1757-1146-4-10. PubMed PMID: 21401957. PubMed PMCID: 3064639.
 22. Chuter VH. Relationships between foot type and dynamic rearfoot frontal plane motion. *Journal of Foot and Ankle Research*. 2010;**3**:9. doi.org/10.1186/1757-1146-3-9.
 23. Hunt AE, Fahey AJ, Smith RM. Static measures of calcaneal deviation and arch angle as predictors of rearfoot motion during walking. *Aust J Physiother*. 2000;**46**:9-16. doi.org/10.1016/S0004-9514(14)60309-3. PubMed PMID: 11676785.
 24. Levinger P, Murley GS, Barton CJ, Cotchett MP, McSweeney SR, Menz HB. A comparison of foot kinematics in people with normal- and flat-arched feet using the Oxford Foot Model. *Gait Posture*. 2010;**32**:519-23. doi.org/10.1016/j.gaitpost.2010.07.013. PubMed PMID: 20696579.
 25. Reischl SF, Powers CM, Rao S, Perry J. Relationship between foot pronation and rotation of the tibia and femur during walking. *Foot Ankle Int*. 1999;**20**:513-20. doi.org/10.1177/107110079902000809. PubMed PMID: 10473063.
 26. Bates B, Osternig L, Mason B, James S. Lower extremity function during the support phase of running. *Biomechanics VI-b*. 1978:30-9.
 27. Cavanagh P, Clarke T, Williams K, Kalenak A, editors. An evaluation of the effect of orthotics on force distribution and rearfoot movement during running. American Orthopaedics Society Sports Medicine Meeting; Lake Placid, New York; 1978.
 28. Clarke T, Lafortune M, Williams K, Cavanagh P, editors. The relationship between center of pressure location and rear-foot movement in distance running. *Medicine And Science In Sports And Exercise*; West Camden: Williams & Wilkins; 1980.
 29. Hamill J, Bates B, White C, editors. Evaluation of foot orthotic appliances using ground reaction force data. *Locomotion II. Proceedings of the Second Biannual Conference of the Canadian Society for Biomechanics*; Biomech, Kingston, Ontario: Canadian Society for Biomechanics; 1982.
 30. Miller MD, Hinkin DT, Wisnowski JW. The efficacy of orthotics for anterior knee pain in military trainees. A preliminary report. *Am J Knee Surg*. 1997;**10**:10-3. PubMed PMID: 9051172.
 31. Milner CE, Davis IS, Hamill J. Free moment as a predictor of tibial stress fracture in distance runners. *J Biomech*. 2006;**39**:2819-25. doi.org/10.1016/j.jbiomech.2005.09.022. PubMed PMID: 16289078.
 32. Holden JP, Cavanagh PR. The free moment of ground reaction in distance running and its changes with pronation. *J Biomech*. 1991;**24**:887-97. doi.org/10.1016/0021-9290(91)90167-L. PubMed PMID: 1744147.
 33. Smith-Oricchio K, Harris BA. Interrater reliability of subtalar neutral, calcaneal inversion and eversion. *J Orthop Sports Phys Ther*. 1990;**12**:10-5. doi.org/10.2519/jospt.1990.12.1.10. PubMed PMID: 18796888.
 34. Song J, Hillstrom HJ, Secord D, Levitt J. Foot type biomechanics. comparison of planus and rectus foot types. *J Am Podiatr Med Assoc*. 1996;**86**:16-23. doi.org/10.7547/87507315-86-1-16. PubMed PMID: 8808320.
 35. Razeghi M, Batt ME. Foot type classification: a critical review of current methods. *Gait Posture*. 2002;**15**:282-91. doi.org/10.1016/S0966-6362(01)00151-5. PubMed PMID: 11983503.
 36. Feiss HO. A simple method of estimating the common variations and deformities of the foot. *The American Journal of the Medical Sciences*. 1909;**138**:213-30. doi.org/10.1097/0000441-190908000-00009.
 37. Hannigan-Downs K, Harter R, Smith G. Radiographic validation and reliability of selected clinical measures of pronation. *J Ath Tr*. 2000;**35**:12-30.
 38. Spornly-Nees S, Dasberg B, Nielsen RO, Bosen MI, Langberg H. The navicular position test-a reliable measure of the navicular bone position during rest and loading. *Int J Sports Phys*

- Ther.* 2011;**6**:199-205. PubMed PMID: 21904698. PubMed PMCID: 3163999.
39. Cappozzo A, Catani F, Croce UD, Leardini A. Position and orientation in space of bones during movement: anatomical frame definition and determination. *Clin Biomech (Bristol, Avon)*. 1995;**10**:171-8. doi.org/10.1016/0268-0033(95)91394-T. PubMed PMID: 11415549.
40. Wilken JM, Rodriguez KM, Brawner M, Darter BJ. Reliability and Minimal Detectable Change values for gait kinematics and kinetics in healthy adults. *Gait Posture*. 2012;**35**:301-7. doi.org/10.1016/j.gaitpost.2011.09.105. PubMed PMID: 22041096.
41. Khan WS, Jones RK, Nokes L, Johnson DS. The relationship of the angle of immobilisation of the knee to the force applied to the extensor mechanism when partially weight-bearing. A gait-analysis study in normal volunteers. *J Bone Joint Surg Br*. 2007;**89**:911-4. doi.org/10.1302/0301-620X.89B7.18452. PubMed PMID: 17673584.
42. Nilsson J, Thorstensson A, Halbertsma J. Changes in leg movements and muscle activity with speed of locomotion and mode of progression in humans. *Acta Physiol Scand*. 1985;**123**:457-75. doi.org/10.1111/j.1748-1716.1985.tb07612.x. PubMed PMID: 3993402.
43. Tateuchi H, Wada O, Ichihashi N. Effects of calcaneal eversion on three-dimensional kinematics of the hip, pelvis and thorax in unilateral weight bearing. *Hum Mov Sci*. 2011;**30**:566-73. doi.org/10.1016/j.humov.2010.11.011. PubMed PMID: 21459469.
44. Lafortune MA, Cavanagh PR, Sommer 3rd HJ, Kalenak A. Foot inversion-eversion and knee kinematics during walking. *J Orthop Res*. 1994;**12**:412-20. doi.org/10.1002/jor.1100120314. PubMed PMID: 8207595.
45. Chang R, Van Emmerik R, Hamill J. Quantifying rearfoot-forefoot coordination in human walking. *J Biomech*. 2008;**41**:3101-5. doi.org/10.1016/j.jbiomech.2008.07.024. PubMed PMID: 18778823.
46. Pohl MB, Buckley JG. Changes in foot and shank coupling due to alterations in foot strike pattern during running. *Clin Biomech (Bristol, Avon)*. 2008;**23**:334-41. doi.org/10.1016/j.clinbiomech.2007.09.016. PubMed PMID: 18006125.
47. Eslami M, Begon M, Farahpour N, Allard P. Forefoot-rearfoot coupling patterns and tibial internal rotation during stance phase of barefoot versus shod running. *Clin Biomech (Bristol, Avon)*. 2007;**22**:74-80. doi.org/10.1016/j.clinbiomech.2006.08.002. PubMed PMID: 17049700.
48. Souza TR, Pinto RZ, Trede RG, Kirkwood RN, Fonseca ST. Temporal couplings between rearfoot-shank complex and hip joint during walking. *Clin Biomech (Bristol, Avon)*. 2010;**25**:745-8. doi.org/10.1016/j.clinbiomech.2010.04.012. PubMed PMID: 20621756.
49. Buldt AK, Murley GS, Butterworth P, Levinger P, Menz HB, Landorf KB. The relationship between foot posture and lower limb kinematics during walking: A systematic review. *Gait Posture*. 2013;**38**:363-72. doi.org/10.1016/j.gaitpost.2013.01.010. PubMed PMID: 23391750.
50. Baumann W. On mechanical loads on the human body during sports activities. *Biomechanics VII-B*. 1981:79-87.
51. Brubaker CE, James SL. Injuries to runners. *J Sports Med*. 1974;**2**:189-98. doi.org/10.1177/036354657400200402. PubMed PMID: 4468337.