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The influence of lower extremity coronal plane movements on the passive regulation of instantaneous loading mechanics during running

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A – Study Design
B – Data Collection
C – Statistical Analysis
D – Data Interpretation
E – Manuscript Preparation
F – Literature Search
G – Funds Collection

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Key words: *impact loading, kinematics, coronal plane, passive regulation.*

Abstract

Background: *The repetitive transmission of impact forces may contribute to the aetiology of overuse injuries. Therefore determining the mechanisms that regulate impact loading has potential clinical significance.This study aimed to determine the influence of lower extremity coronal plane kinematics on the regulation of impact loading during running.*

Material/Methods: *Thirty-six participants ran at 4.0 m.s-1striking the centre of a piezoelectric force platform with their dominant limb. Coronal plane angular kinematics about the hip, knee and ankle joints were measured using an eight-camera motion analysis system operating at 250 Hz. Regression analyses with instantaneous loading rate magnitude as a criterion were used to identify the coronal plane parameters associated with impact loading.*

Results: *The overall regression model yielded Adj* $R^2 = 0.37$, $p \le 0.01$. *Two biomechanical parameters were obtained as significant predictors of the instantaneous loading rate. Peak ankle eversion Adj* $R^2 = 0.22$, $p \le 0.01$ *and peak eversion angular velocity of the ankle Adj* $R^2 = 0.15$, $p \le 0.01$ *were found to be significant predictors of instantaneous loading rate.*

Conclusions: *The findings of the current investigation therefore suggest that passive joint motions in the coronal plane can regulate the magnitude of impact loading, linked to the development of chronic injuries.*

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Introduction

During the impact phase of the ground reaction force, the momentum from the decelerating limb changes rapidly as the foot makes contact with the ground surface, resulting in a transient force which is transmitted up the skeleton. During typical running velocities, these forces can reach magnitudes of up to three times body weight [1]. Whilst an optimal loading window exists for the positive development of bone and tissue strength, movements beyond this window can lead to the breakdown of body tissue and overuse injuries [2, 3, 4, 5].

The transient impact that occurs as a result of footstrike propagates through the musculoskeletal system [6, 7]. A number of attenuation mechanisms exist that may be internal or external to the musculoskeletal system [6]. The majority of analyses in this area have examined the influence of footwear on the impact loading magnitude during running. To characterize the load attenuating properties of footwear, mechanical impact testing techniques have been developed using simple material test machines [8, 9, 10]. Such testing procedures offer a consistent tool with which to characterize the in vitro mechanical properties of the shoe; however, this technique may not necessarily relate to variations in impact magnitude when evaluated in situ during running [11, 12]. Numerous studies have been conducted varying the geometry of the midsole, its hardness or a combination thereof [13, 14, 15, 16]. The shock attenuation properties of footwear and midsoles have been evaluated with force platforms, bone-mounted accelerometers, and surface mounted accelerometers.

Passive tissues and active joint movements have also been shown to moderate the magnitude of the impact load [17]. Denoth [18] proposed the concept of effective mass whereby impact magnitude is dependent on the knee angle in the sagittal plane at footstrike. This concept is supported by the results of Bobbert et al. [19]. Lafortune et al. [20], who investigated the effect of surface hardness and the initial knee angle on the impact force and tibial acceleration magnitudes in a human pendulum approach. However, despite these early propositions there still remains a paucity of research regarding the protection from transient impact loading afforded by joint alignment during running.

Coronal plane eversion of the foot has previously been hypothesized as a mechanism by which impact loading may be attenuated [21]. Preliminary evidence has related reductions in impact loading mechanics to variations in footwear construction [22]. Valgus aligned footwear has been shown to decrease peak tibial accelerations produced as a result of footstrike [23]. However, Perry and Lafortune [22] and Yingling et al. [21] documented that rear foot parameters had no significant influence on impact loading during running.

It remains unclear which of the kinematic or muscular factors may explain variations in impact magnitude. Whilst the coronal plane movement of the ankle joint has been proposed as a passive mechanism by which impact loading is regulated, the results from previous analyses in investigating this mechanism have been conflicting. Furthermore, there is currently a paucity of information regarding the influence of coronal plane movements in the more proximal hip and knee joints on the regulation of impact loading during running. The aim of the current investigation was therefore to determine the influence of lower extremity coronal plane parameters on the passive regulation of impact loading during running.

Material and methods

Participants

Thirty-six male participants who were free from musculoskeletal injury volunteered to take part in this study. The mean characteristics of the participants were: age 28.59 ± 4.15 years, height 176.72 ± 5.08 cm and body mass 77.97 ± 5.79 kg. All participants were classified as natural rearfoot strikers by exhibiting a clear first peak in their vertical ground reaction force. A statistical power analysis was conducted using G* Power Software using a moderate effect size [24] to reduce the likelihood of a type II error and determine the minimum number of participants needed for this investigation. It was found that the sample size was sufficient to provide more than 80% statistical power. The study was approved by the School of Psychology ethical committee, and all participants provided written informed consent in accordance with the guidelines outlined in the declaration of Helsinki.

Procedure

Participants ran at 4.0 ms^{-1} over a force plate (Kistler, Kistler Instruments Ltd., Alton, Hampshire, UK Model 9281CA) embedded in the floor (Altrosports 6 mm, Altro Ltd.,) of a 22 m biomechanics laboratory. The running velocity was quantified using Newtest 300 infrared timing gates (Newtest, 300 OyKoulukatu, Finland); a maximum deviation of ±5% from the set velocity was allowed. Stance time was defined as the time over which a vertical force of 20N was applied to the force platform [25]. A successful trial was defined as one within the specified velocity range, where all tracking clusters were in view of the cameras, the foot made full contact with the force plate and there was no evidence of gait modifications due to the experimental conditions. Runners completed five successful trials.

All kinematic data were captured at 250 Hz via an eight camera motion analysis system (Qualisys^{TM} Medical AB, Goteburg, Sweden). Calibration of the Qualysis^{TM} systems was performed before each data collection session. To ensure that high quality kinematic data was obtained, only calibrations which produced average residuals of less than 0.85 mm for each camera for a 750.5 mm wand length and points above 4000 in all cameras were accepted prior to data collection.

The marker set used for the study was based on the calibrated anatomical systems technique (CAST) using a 6 degrees of freedom model [26]. A static trial was conducted with the participant in the anatomical position allowing the positions of the anatomical markers to be referenced in relation to the tracking clusters, following which they were removed. Markers used for tracking remained in place throughout.

Retro-reflective markers were attached to the 1st and 5th metatarsal heads, medial and lateral malleoli, calcaneus, medial and lateral epicondyle of the femur, greater trochanter of the right leg, iliac crest, anterior superior iliac spines (ASIS) and posterior superior iliac spines (PSIS) with tracking clusters positioned on the shank and thigh (Figure 1). All markers were positioned by the first author. The hip joint centre was determined using regression equations based on the positions of the ASIS and PSIS markers on [27]. Each rigid cluster comprised four 19 mm spherical reflective markers mounted to a thin sheath of lightweight carbon fiber with length to width ratios of 2.05:1 and 1.5:1 for the femur and tibia respectively, in accordance with Cappozzo et al. [28] recommendations. Participants wore the same footwear throughout (Saucony pro grid guide 2), in sizes 6-9.

Data processing

Trials were processed in Qualisys Track Manager in order to identify anatomical and tracking markers then exported as C3D files. 3-D Kinematic parameters were quantified using Visual 3-D (C-Motion, Germantown, MD, USA) after marker data was filtered using a low pass Butterworth 4th order zero-lag filter at a cut off frequency of 12 Hz which was selected as being the frequency at which 95% of the signal power was below. Coronal plane kinematics (not normalised to standing posture) about the hip, knee and ankle joints were calculated using the euler technique via an XYZ cardan sequence of rotations (where X is sagittal; Y is coronal and Z transverse plane rotation) (29). All data were normalized to 100% of the stance phase then processed gait trials were averaged. Coronal plane measures from the hip, knee and ankle which were extracted for statistical analysis were 1) angle at footstrike, 2) range of motion from footstrike to toe-off during stance, 3) peak angle during stance, 4) angular excursion from footstrike to peak angle and 5) peak angular velocity. These variables were extracted from each of the five trials for each joint in all three planes of rotation and the data was then averaged within subjects for statistical analysis. Participant's kinematic curves were time normalized to stance and were ensemble averaged across subjects for visual purposes only. Forces are reported in bodyweights (BWs) to allow normalization of the data among participants. From the force plate data, instantaneous loading rate was quantified as the maximum increase in the vertical force between frequency intervals [30, 31].

Instantaneous loading rate was selected based on the Greenhalgh et al. [32] recommendations as a more practical and representative measure of impact loading.

Statistical analysis

Multiple regression analyses with instantaneous vertical loading rate as a criterion variable and the 3-D kinematic parameters as independent variables were carried out using a forward stepwise procedure with significance accepted at the $p \leq 0.05$ level. The independent variables were examined for co-linearity prior to entry into the regression model using a Pearson's correlation coefficient matrix and those exhibiting high co-linearity $R \geq 0.7$ were removed. All statistical procedures were conducted using SPSS 19.0 (SPSS Inc, Chicago, USA).

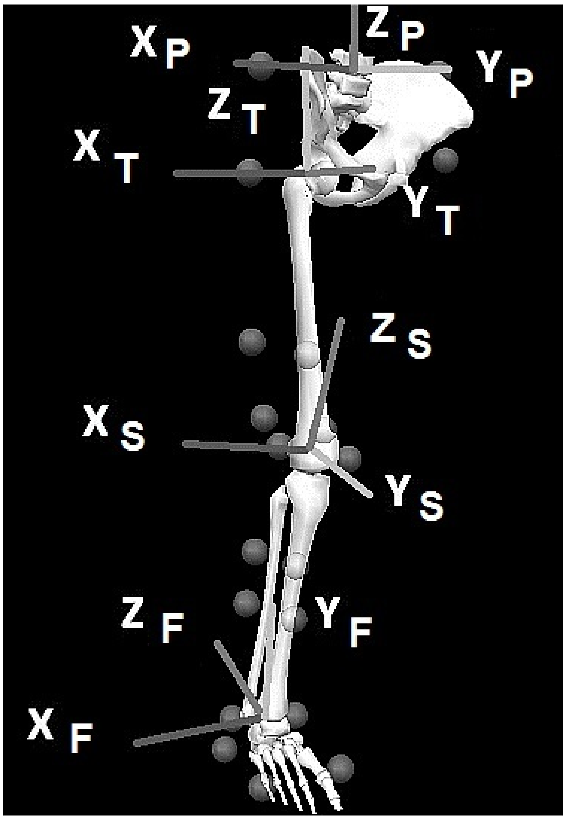

Fig. 1. Pelvic, thigh, tibial and foot segments, with segment co-ordinate system axes (P= pelvic, S= shank, T = tibia and F = foot)

Results

Figure 2 and Tables 1-2 present the mean ± standard deviation coronal plane angulations from the stance phase of running. The overall regression model yielded an R= 0.64, R^2 = 0.41 and Adj R^2 = 0.37, $p \leq 0.01$. Two biomechanical parameters were obtained as significant predictors of the instantaneous loading rate. Peak ankle eversion (B = 0.56, t = 3.96) Adj R^2 = 0.22, $p \leq 0.01$ and peak eversion angular velocityof the ankle(B = 0.42, t = 2.96) Adj R^2 = 0.15, $p \leq 0.01$ were found to be significant predictors of instantaneous loading rate.

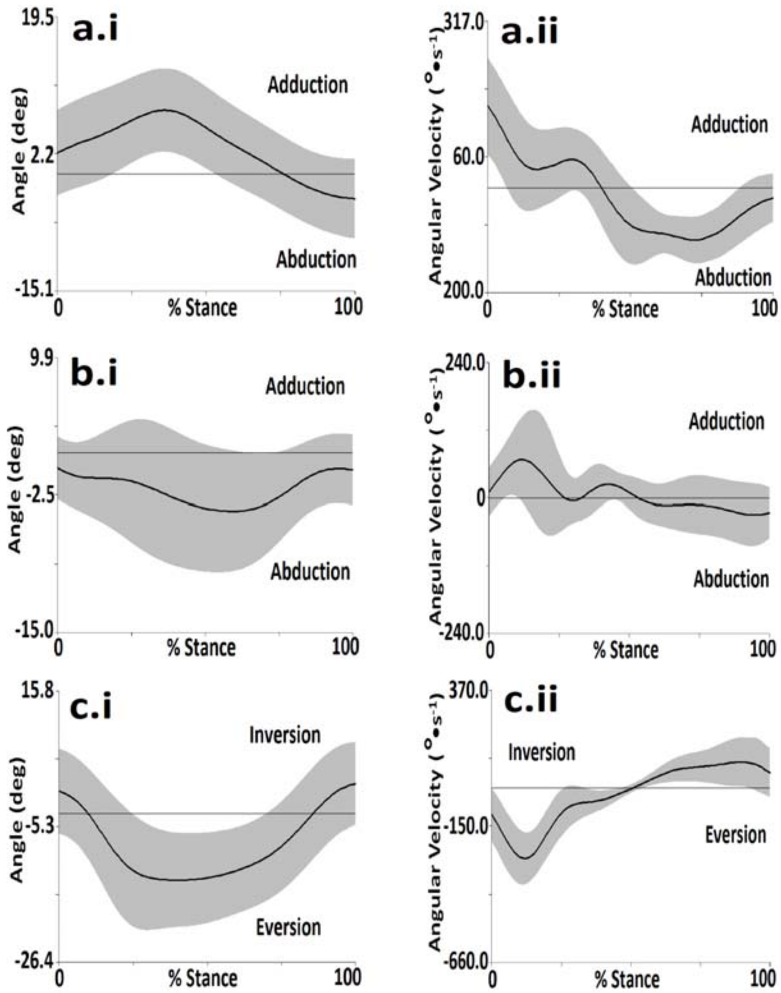

Fig. 2. Mean and standard deviation angulations about the a. hip, b. knee and c. ankle joints in the coronal plane

Table 1. Mean and standard deviation angulations of hip, knee and ankle joints in the coronal plane

Angulation	Hip	Knee	Ankle
Angle at footstrike (°)	2.95 ± 4.61	-1.41 ± 2.44	5.84 ± 4.03
Angle at toe-off (°)	-2.47 ± 5.04	-1.61 ± 2.92	0.74 ± 4.29
Range of motion (°)	6.42 ± 3.16	1.74 ± 0.95	5.67 ± 3.86
Relative range of motion (°)	5.45 ± 2.68	4.38 ± 3.50	14.29 ± 3.59
Peak angle (°)	8.40 ± 4.01	-5.80 ± 4.23	-8.45 ± 4.25

Table 2. Mean and standard deviation angular velocities of hip, knee and ankle joints in the coronal plane

Angular velocity	Hip	Knee	Ankle
Angular velocity at footstrike (°/s)	165.23 ± 76.76	62.36 ± 113.15	-142.89 ± 83.73
Angular velocity at toe-off (°/s)	-25.85 ± 57.66	-42.14 ± 38.82	116.73 ± 65.49
Peak adduction/inversion angular velocity (°/s)	22.34 ± 45.98	176.97 ± 109.06	176.15 ± 60.56
Peak abduction/eversion angular velocity (°/s)	-112.57 ± 42.42	-117.06 ± 60.83	-276.45 ± 119.80

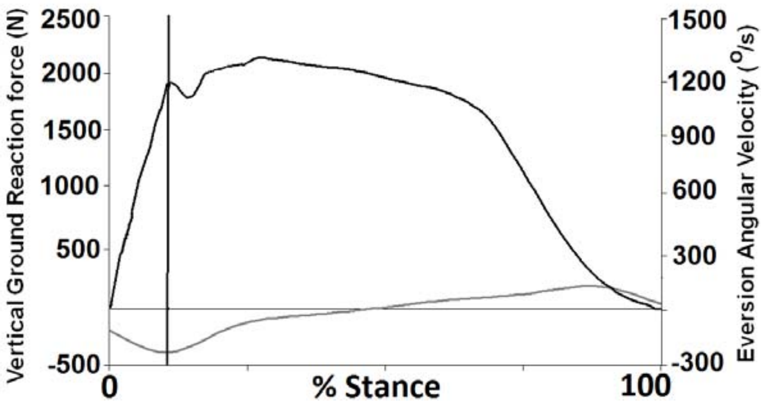

Fig. 3. Temporal synchronization of the vertical ground reaction force (black line) and ankle eversion angular velocity (grey line)

Discussion

The aim of the current investigation was to determine the influence of coronal plane kinematic parameters on instantaneous loading mechanics. This study represents the first to consider the effects of lower extremity coronal plane movements on the passive regulation of impact loading during running.

The primary observation of this study was that rear-foot eversion parameters served as significant regulators of the instantaneous loading rate. This finding opposes Yingling et al., [21] and Perry and Lafortune [22] who both noted that increases in rearfoot eversion did not significantly influence impact loading during running. There are several potential explanations for this lack of continuity between studies. Firstly, Yingling et al., [21] and Perry and Lafortune [22] utilized a two dimensional method with four markers (positioned in the middle of the heel, on the upper part of the calcaneus, on the Achilles tendon at the height of the malleoli, and 15 cm above in the middle of the gastrocnemius) for the quantification of rearfoot kinematics in the coronal plane. This is in contrast to the current investigation in which a three-dimensional technique was utilized, whereby coronal plane angulation is considered to occur about the segment co-ordinate axes of the foot segment relative to those of the tibia. Secondly, both Yingling et al., [21] and Perry and Lafortune [22] performed comparative analyses using modified footwear designed to increase rearfoot eversion in order to determine whether impact loading was subsequently reduced. This differs from the statistical approach utilized in the current study. Regression looks for weighted relationships between multiple variables and a criterion variable rather than statistical differences between conditions.

Whilst the results of the current investigation appear to confirm that coronal plane angulation about the ankle joint can significantly influence impact loading magnitude, there remains a large proportion of unexplained variance. Therefore, the remaining variance with regard to understanding the mechanisms behind the regulation of impact remain unknown. Future research may wish to consider the non-coronal regulation of impact loading in an attempt to determine where the remaining variance lies. Whilst previous analyses have been conducted examining the influence of lower extremity kinematics on the regulation of impact forces their criterion variable was impact peak of the vertical ground reaction force as opposed to the loading rate which was used in the current study. Shorten and Mientjes [33] determined that the impact peak of the vertical ground reaction force is not a reliable indicator of impact magnitude, as it does not comprise any temporal elements of the load being experienced by the lower extremities.

A number of investigations and authors have discussed the influence of both ankle plantar flexion and knee flexion to attenuate the impact load following footstrike. However, despite these propositions ankle plantar flexion occupies 80 ms and the peak of stance phase knee flexion occurs 50 ms after initial contact, whereas the high frequency transient component of the vertical ground reaction force occurs during the first 20 ms of the stance phase. Therefore, whilst it appears reasonable to suppose that these sagittal plane mechanisms may reduce the rate at which body weight is transferred to the lower extremities, it is unlikely that they provide sufficient protection on their own against the heelstrike transient. However, if the vertical ground reaction force and ankle angular velocity curves are plotted together (Figure 3), there is clear temporal synchronization between the two modalities in terms of the peak eversion angular velocity and the transient element of the stance phase. This leads to the notion that coronal plane ankle motion may act as an initial regulator of impact loading during the early stance phase, although more analyses are required using prediction modelling in order to fully corroborate this suggestion.

That rearfoot eversion serves as a significant regulator of impact loading may confound footwear manufacturers, as rearfoot eversion (in addition to high impact loading) has also been linked to the development of overuse injuries in runners, such as tibial stress syndrome, plantar fasciitis and anterior knee pain [34, 35, 36, 37]. Therefore, the concept of a general running shoe proposed by Nigg [38] that serves the needs of all runners in terms of their protection from both impact loading and rearfoot eversion appears to be invalid. As reductions in impact rearfoot eversion would place runners at a greater risk from loading related injuries, allowing increases in eversion would subsequently place runners at risk from injuries due to a lack of rearfoot stability.

Limitations

This study did not evaluate the electromyographical potentials of the lower extremity muscles. This may serve as a limitation of the current study as muscle pre-activation particularly in the period prior to footstrike has been proposed as one of the mechanism by which loading of the lower extremities during running may be attenuated [39]. It is recommended that future analyses examine this pre-activation mechanism in conjunction with 3-D kinematic analyses. In addition, the current investigation quantified only the time domain characteristics of impact loading. Shorten and Winslow [40] noted that impact loading either from a tibially mounted accelerometer or from the vertical ground reaction force can be transformed from the time into the frequency domain using a fast fourier transform function, allowing the frequency content of the signals to be examined. It has been documented that both the time and frequency characteristics of impact loading are pertinent to the development of chronic injuries in runners [30, 31, 39]. Therefore, it is further recommended that additional investigations consider the regulation of frequency domain properties in addition to the conventionally measured time domain measures.

A final limitation which future research may wish to resolve is the all-male sample utilized in the current investigation. Previous analyses have demonstrated that loading mechanics differ between genders. Heinng [41] and Stefanyshyn et al. [42] found that at matched running velocities females were associated with significantly greater loading rates than males; thus it is unlikely that the results of the current investigation can be generalized to females. Furthermore, it has previously been documented that females exhibit different coronal plane kinematics in comparison to males at all of the lower extremity joints [30, 43, 44, 45]. Therefore, it is unlikely that female runners regulate impact loading using coronal plane mechanics in the same manner as males. Future research should therefore repeat the current investigation using a female sample.

Conclusion

In conclusion this study appears to confirm the notion that rear-foot eversion can influence impact loading during running. Therefore, it may be possible to implement training / technique adaptations for runners in order to maximize their passive shock attenuation. However, future research aimed at identifying the mechanisms governing these regulation processes is needed in order to further understand their implications for runners.

Acnowledgements

University's Code of Conduct for Research can be found at the following web address:http://www.uclan.ac.uk/information/research/research_degrees/ethics_research_governance.php.

References

1. Cavanagh PR, Lafortune MA. Ground reaction forces in distance running. J Biomech. 1980;13:397-406.
2. Taunton JE, Clement DB, McNicol K. Plantar fasciitis in runners. Canadian Journal of Applied Sport Sciences. 1982;7:41-44.
3. Simon SR, Radin EL, Paul IL, Rose RM. The response of joints to impact loading, II. In vivo behavior of subchondral bone. J Biomech. 1972;5:267-72.
4. Folman Y, Wosk J, Voloshin A, Liberty S. Cyclic impacts on heel strike: a possible biomechanical factor in the etiology of degenerativedisease of the human locomotorsystem. Archives of Orthopaedic Trauma Surgery. 1986;104:363-5.
5. Collins JJ, Whittle MW. Impulsive forces during walking and their clinical implications. Clin Biomech. 1989;4:179–87.
6. Whittle MW. The generation and attenuation of transient forces beneath the foot; a review. Gait & Posture. 1999;10:264-275.
7. Sinclair J, Bottoms L, Taylor K, Greenhalgh A. Tibial shock measured during the fencing lunge, the influence of footwear. Sports Biomechanics. 2010;9:65-71.
8. Kolitzus HJ. Functional standards for playing surfaces. In: Frederick EC, editor. Sport shoes and playing surfaces. Champaign, Ill: Human Kinetics; 1984, 98-118.
9. Cook SD, Kester MA, Brunet ME. Shock absorption characteristics of running shoes. Am J Sport Med. 1985;13:248-253.
10. Krabbe B, Baumann W. Mechanical properties of running shoes – measurement and modelling. In: Herzog W, Nigg B, van den Bogert T, editors. Proceedings of the Canadian Society for Biomechanics VIIIth Biennial Conference, August 1994. University of Calgary; 1994, 20-21.
11. Nigg BM, Bahlsen A. Influence of heel flare and midsole construction on pronation, supination, and impact forces for heel-toe running. International Journal of Sport Biomechanics. 1988;4:205-219.
12. Stacoff A, Denoth J, Kaelin X, Stuessi E. Running injuries and shoe construction: some possible relationships. International Journal of Sport Biomechanics. 1988;4:342-357.
13. Bates BT, Osternig LR, Sawhill JA, James SL. An assessment of subject variability, subject-shoe interaction, and the evaluation of running shoes using ground reaction force data. J Biomech. 1983;16:181-191.
14. Nigg BM, Morlock M. The influence of lateral heel flare of running shoes on pronation and impact forces. Med Sci Sport Exer. 1987;19:294-302.
15. DeWit B, DeClerq D, Lenoir M. The effect of varying midsole hardness on impact forces and foot motion during foot contact in running. J Appl Biomech. 1995;11:395-406.
16. Kersting UG, Bruggemann GP. Midsole material-related force control during heel-toe running. Research in Sports Medicine. 2006;14:1-17.
17. Wright IC, Neptune RR, van Den Bogert AJ, Nigg BM. Passive regulation of impact forces in heel-toe running. Clin Biomech. 1998;13(7):521–531.
18. Denoth J. Load on the locomotor system and modelling. In: Nigg BM, editor. Biomechanics of running shoes. Champaign, IL: Human Kinetics; 1986, 63-116.
19. Bobbert MF, Yeadon MR, Nigg BM. Mechanical characteristics of the landing phase in heel-toe running. J Biomech. 1992;25:223-234.
20. Lafortune MA, Hening EM, Lake MJ. Dominant role of interface over knee angle for cushioning impact loading and regulating initial leg stiffness. J Biomech. 1996;29:1523-1529.
21. Yingling VR, Yack HJ, White SC. The effect of rearfootmotion on attenuation of the impulse wave at impactduring running. J Appl Biomech. 1996;12:313.
22. Perry SD, Lafortune MA. Influences of inversion/eversion of the foot upon impact loading during locomotion. Clin Biomech. 1995;10:253-257.
23. Milani TL, Schnabel G, Hennig EM. Rearfoot motion and pressure distribution patterns during running in shoes with varus and valgus wedges. J Appl Biomech. 1995;11:177-187.
24. Erdfelder E, Faul F, Buchner A. G*Power: a general power analysis program. Behav Res Meth Ins C. 1996;28:1-11.
25. Sinclair J, Edmundson CJ, Brooks D, Hobbs SJ. Evaluation of kinematic methods of identifying gait Events during running. International Journal of Sports Science and Engineering. 2011;5:188-192.
26. Cappozzo A, Catani F, Leardini A, Benedeti MG, Della CU. Position and orientation in space of bones during movement: Anatomical frame definition and determination. Clin Biomech. 1995;10:171-178.
27. Bell AL, Brand RA, Pedersen DR. Prediction of hip joint centre location from external landmarks, Human Movement Science. 1989;8:3-16.
28. Cappozzo A, Cappello A, Croce U, Pensalfini F. Surface-marker cluster design criteria for 3-D bone movement reconstruction. IEEE Transactions on Biomedical Engineering. 1997;44:1165-1174.

29. Sinclair J, Taylor PJ, Edmundson CJ, Brooks D, Hobbs SJ. Influence of the helical and six available cardan sequences on 3-D ankle joint kinematic parameters. Sports Biomechanics. 2012;11:430-437.
30. Sinclair, J., Greenhalgh, A., Edmundson, C.J., Brooks, D.,& Hobbs, S.J. (2012). Gender differences in the kinetics and kinematics of distance running: implications for footwear design, International Journal of Sports Science and Engineering, 6, 118-128.
31. Sinclair J, Greenhalgh A, Edmundson CJ, Brooks D, Hobbs SJ. The efficacy of barefoot and shod running and shoes designed to mimic barefoot running, Footwear Science. 2013;5:45-53.
32. Greenhalgh A, Sinclair J, Protheroe L, Chokalingam N. Predicting impact shock magnitude: which ground reaction force variable should we use? International Journal of Sports Science and Engineering. 2012;6:225-231.
33. Shorten M, Mientjes MIV. The ‘heel impact’ force peak during running is neither ‘heel’ nor ‘impact’ and does not quantify shoe cushioning effects. Footwear Science. 2011;3:41-58.
34. Taunton JE, Clement DB, McNicol K. Plantar fasciitis in runners. Canadian Journal of Applied Sport Sciences. 1982;7:41-44.
35. Duffey MJ, Martin DF, Cannon DW, Craven T, Messier SP. Etiologic factors associated with anterior knee pain in distance runners. Med Sci Sport Exer. 2000;2:1825-1832.
36. Willems TM, De Clercq D, Delbaere K, Vanderstraeten G, De Cock A, Witvrouw E. A prospective study to gait related risk factors for exercise-related lower leg pain. Gait & Posture. 2006;23:91-98.
37. Lee SY, Hertel J, Lee SC. Rearfoot eversion has indirect effects on plantar fascia tension by changing the amount of arch collapse. The Foot. 2010;20:64-70.
38. Nigg BM. Biomechanics of running shoes. Champaign, Ill.: Human Kinetics Publishers, Inc.; 1986.
39. Sinclair J, Taylor PJ, Edmundson CJ, Brooks D, Hobbs SJ. The influence of footwear kinetic, kinematic and electromographical parameters on the energy requirements of steady state running. Sports Science and Human Movement. 2013;80:39-49.
40. Shorten MR, Winslow DS. Spectral analysis of impact shock duringrunning. International Journal of Sports Biomechanics. 1992;8:288-304.
41. Hennig EM. Gender differences for running in athletic footwear. In: Proceedings of the 5th Symposium on Footwear Biomechanics. Zuerich 2005.
42. Stefanyshyn DJ, Stergiou P, Nigg BM, Rozitis AI, Goepfert B. Do females require different running footwear? In: Proceedings of the Sixth Symposium on Footwear Biomechanics. 2003; 91-92.
43. Ferber R, Davis IM, Williams DS. Gender differences in lower extremity mechanics during running. Clin Biomech. 2003;18:350-357.
44. Malinzak RA, Colby SM, Kirkendall DT, Yu B, Garrett WE. A comparison of knee joint motion patterns between men and women in selected athletic tasks. Clin Biomech. 2001;16:438-445.
45. Hewett TE, Myer GD, Ford KR, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. Am J Sport Med. 2005;33:492-501.