Impact Accelerations of Barefoot and Shod Running

DOI 10.1055/s-0035-1569344
Int J Sports Med 2016; 37: 364–368

Publishing House and Copyright:
© 2015 by
Georg Thieme Verlag KG
Rüdigerstraße 14
70469 Stuttgart
ISSN 0172-4622

Impact Accelerations of Barefoot and Shod Running

Authors M. Thompson[1], J. Seegmiller[2], C. P. McGowan[3]

Affiliations
[1] Fort Lewis College, Exercise Science, Durango, United States
[2] WWAMI Medical Education, University of Idaho, Moscow, United States
[3] WWAMI Medical Education Program, University of Idaho, Moscow, United States

Key words
- biomechanics
- locomotion
- heel strike
- running
- footstrike
- impact

Abstract

During the ground contact phase of running, the body's mass is rapidly decelerated resulting in forces that propagate through the musculoskeletal system. The repetitive attenuation of these impact forces is thought to contribute to overuse injuries. Modern running shoes are designed to reduce impact forces, with the goal to minimize running related overuse injuries. Additionally, the fore/mid foot strike pattern that is adopted by most individuals when running barefoot may reduce impact force transmission. The aim of the present study was to compare the effects of the barefoot running form (fore/mid foot strike & decreased stride length) and running shoes on running kinetics and impact accelerations. 10 healthy, physically active, heel strike runners ran in 3 conditions: shod, barefoot and barefoot while heel striking, during which 3-dimensional motion analysis, ground reaction force and accelerometer data were collected. Shod running was associated with increased ground reaction force and impact peak magnitudes, but decreased impact accelerations, suggesting that the midsole of running shoes helps to attenuate impact forces. Barefoot running exhibited a similar decrease in impact accelerations, as well as decreased impact peak magnitude, which appears to be due to a decrease in stride length and/or a more plantarflexed position at ground contact.

Introduction

Running is one of the most popular recreational activities, but runners are also one of the most common groups to incur overuse injuries [6,19]. Given that an estimated 20–80% of runners are injured annually, a great deal of research has focused on running and running related injuries. A number of interventions have been proposed to reduce running related injuries [14,34], the most common of which is running shoes. Recently, barefoot running form has been suggested as a potential mechanism to reduce running injuries [18]. Barefoot running has been associated with kinetic and kinematic changes, specifically, decreased stride length and a more plantarflexed position at ground contact, which may have implications for injury prevention [20].

During the ground contact phase of running, segments of the lower extremity and trunk are decelerated at different rates [4]. Segment deceleration depends on effective mass (M_{eff}), which is the portion of the total body mass needed to accurately model the impact as a point mass stopping abruptly at ground contact [11]. As segments decelerate forces are transmitted through the musculoskeletal system. These forces are progressively reduced as they travel to the head, by passive structures such as the ground, shoe midsole, and soft tissues of the lower extremity [20,24,32]. Forces can also be actively decreased by eccentric activation of the muscles crossing the hip, knee, and ankle joints [24,32]. It has been suggested that the repetitive attenuation of impact forces may contribute to overuse injuries, though this claim has yet to be supported by conclusive evidence [10].

Impact loading on the body during running can be assessed by measuring ground reaction forces (GRF) and accelerations caused by impact [22]. Body segment acceleration is dependent on the magnitude of the GRF and the damping effects of the body's passive and active shock absorbers [10]. A primary objective of modern running shoes is to reduce impact force transmission, with the goal of minimizing running-related overuse injuries that may stem from the repetitive application of these forces. However, previous studies have reported that cushioned running shoes may or may not reduce impact forces or

accepted after revision
November 21, 2015

Bibliography
DOI http://dx.doi.org/10.1055/s-0035-1569344
Published online: February 2, 2016
Int J Sports Med 2016; 37: 364–368 © Georg Thieme Verlag KG Stuttgart · New York
ISSN 0172-4622

Correspondence
Dr. Melissa Thompson
Fort Lewis College
Exercise Science
1000 Rim Dr.,
81301 Durango
United States
Tel.: +1/970/247 7580
Fax: +1/970/247 7109
mathompson@fortlewis.edu

injury rates [23,27]. Running shoes have been shown to reduce tibia accelerations [20]. Yet contacting the ground with the heel, which typically occurs when running in cushioned running shoes, is associated with higher rates of loading and impact peak forces [17]. It has also been suggested, although it has yet to be substantiated, that running shoes may limit proprioceptive feedback and hence lead to an increase in running related overuse injuries [28].
Most runners contact the ground with their heel when running in traditional running shoes that have an elevated heel [3]. In heel strike runners, a transient impact force is generated when the heel contacts the ground. This impact transient typically has a high rate and magnitude of loading and is thought to contribute to the high incidence of running related injuries [21,26]. The fore/mid foot strike pattern that is adopted by most individuals when running barefoot [17] may have important implications for attenuating impact forces. It has been suggested that the greater ankle compliance during fore/mid foot running decreases the effective mass of the body that collides with the ground, resulting in reduced impact peaks and loading rates [17]. Differences in GRF's between barefoot and shod running have been the focus of several studies. However, to the best of our knowledge, how wearing shoes and a fore/mid foot strike position independently affect impact accelerations has yet to be evaluated.
Therefore, the aim of the present study was to independently evaluate the effects of barefoot running form (fore/mid foot strike and reduced stride length) and running shoes on running kinematics, kinetics and impact accelerations. We hypothesized that 1) both running shoes and barefoot running form would reduce impact accelerations to the tibia and 2) heel strike running, whether barefoot or shod, would result in higher impact peak magnitudes.

Methods

Ten healthy, physically active heel strike runners [5 men and 5 women, age: 26±7.3 yrs; height: 1.74±0.09 m; mass: 65.6±10.2 kg] participated in this study. This study was conducted in accordance to ethical guidelines and international standards [13] and approved by the University of Idaho's Institutional Review Board. Subjects provided written informed consent prior to participation.
All subjects ran over-ground in 3 conditions: shod (SHOD), barefoot (BF) and barefoot while heel striking (BFHS). Heel strike runners were chosen so that the effect of changing to fore/mid foot strike when running barefoot could be evaluated. The BFHS condition was used to control for footstrike and isolate the effect of running shoes. Participants ran at a self-selected velocity, and were instructed to run at a pace they could maintain for a 30-min run. Foot strike was determined via foot strike angle (FSA) obtained through motion capture data. A FSA of 0° was defined as flat foot, FSA < 0° was defined as heel striking and FSA > 0º was defined as fore/mid foot striking [1]. Prior to each condition subjects ran for 5–10 min in order to familiarize to the condition.
3-dimensional motion analysis and GRF data were collected as subjects ran over a 15-m runway with a force plate (AMTI, Waterton, MA) embedded at 10 m. 10 strides from each subject were used to calculate averages for each condition. Trials in which velocity differed by > 5 % or stride length differed by > 3 % were excluded from analysis. 16 reflective markers were placed bilaterally over the anterior and posterior superior iliac spines, mid-thigh, femoral epicondyle, mid-shank, lateral malleolus, second metatarsal head and calcaneus according to the Modified Helen Hayes Marker set [15]. For the shod running, heel and toe markers were placed on the shoes overlying the anatomical landmarks. Height, weight, leg length and widths of the ankles and knees were measured for anthropometric scaling. 3-dimensional marker positions were captured at 250 Hz via a Vicon MX motion analysis system (Vicon, Oxford Metrics Ltd., UK) and filtered using a Woltring filtering routine with a predicted mean square error of 4 mm^2. The 3 orthogonal components of the GRF data were recorded at 1 000 Hz from the force plate in synchrony with the motion capture data. GRF data were low-pass filtered at 30 Hz using a second-order Butterworth filter before being downsampled and combined with the motion capture data. Joint kinematics and kinetics were computed via Vicon Plug-In Gait.
Impact peak magnitude was measured as the first observable peak in the vertical GRF. If the impact peak was absent, no value was recorded. Loading rates were calculated as the change in force divided by change in time between 20 and 80 % of the period from ground contact to impact peak [21].
Impact accelerations were measured from accelerometers placed on the lateral surface of the distal lower leg and the lateral surface of the forehead (▸ Fig. 1). Lightweight biaxial accelerometers (Freescale Semiconductor, Austin, TX; model: MMA3202KEG) were mounted to a small piece of balsa wood with epoxy resin. Each accelerometer had a minimum 50-g range and 20 mV/g sensitivity. Combined mass of the accelerometer, balsa wood and epoxy was less than 3 g. The mounted accelerometers were secured as firmly as possible to the leg with coban wrap and to the head with an elastic band. One axis of the accelerometer was oriented with the longitudinal axis of the

Fig. 1 Experimental set-up showing accelerometer placement and orientation of accelerometer axes.

tibia and the second axis was oriented with the direction of travel (▶ **Fig. 1**). This attachment method has been shown to appropriately and reliably measure impact accelerations [29, 30]. Accelerometer data were collected at 1 000 Hz via a Biometrics DataLOG MWX8 data acquisition device (Biometrics Ltd., Ladysmith, VA) simultaneously with motion capture and GRF data. Resultant accelerations were calculated from the 2 accelerometer axes, as this provides a better estimate of shock than a single axis [16]. Peak resultant accelerations were measured for each analyzed stride and averaged across trials and subjects for each running condition.

Statistical differences in the kinetic and kinematic parameters were determined using repeated-measures ANOVA in SPSS (IBM, Armonk, NY). When a significant effect was identified, a post hoc Bonferroni pairwise comparison was performed to determine which conditions were significantly different. Statistical significance was defined as $P<.05$.

Table 1 Select kinematic and kinetic parameters.

	BF	BFHS	SHOD
Stride Length (m)	**2.13 (0.15)** [c]	**2.18 (0.17)** [c]	**2.25 (0.19)** [a,b]
Velocity (m/s)	2.97 (0.19)	3.10 (0.25)	3.09 (0.25)
Impact Peak (BW)	**1.58 (0.21)** [c]	1.81 (0.25)	**1.91 (0.21)** [a]
Loading Rate (BW/s)	135.7 (38.2)	160.8 (33.6)	148.4 (48.9)
Peak vertical GRF (BW)	2.29 (0.26)	**2.23 (0.19)** [c]	**2.31 (0.23)** [b]
Peak anterior-posterior GRF (BW)	0.37 (0.08)	0.35 (0.06)	0.37 (0.06)
Peak medio-lateral GRF (BW)	0.08 (0.04)	0.07 (0.03)	0.08 (0.04)

Data are mean (standard deviation). Significant differences are indicated in bold. [a] indicates a significant difference to BF. [b] indicates a significant difference to BFHS. [c] indicates a significant difference to SHOD. $P<0.05$. BW = body weight

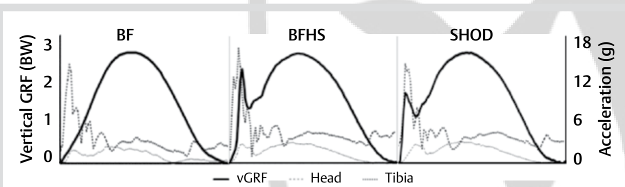

Fig. 2 Typical vertical GRF in body weights (BW) and resultant head and tibia acceleration profiles for the 3 running conditions.

Results

The SHOD condition exhibited a significantly greater stride length than the BF ($P=0.038$) and BFHS conditions ($P=0.018$) (▶ **Table 1**). There were no significant differences in running velocity between conditions (▶ **Table 1**).

There were statistically significant differences in peak resultant tibia acceleration between the SHOD and BFHS conditions ($P=0.005$) and BF and BFHS ($P=0.03$) (▶ **Fig. 2**). Peak resultant accelerations on the tibia were 11.32 ± 1.48, 13.55 ± 1.51 and 11.27 ± 1.73 g for the BF, BFHS and SHOD conditions, respectively. Peak resultant accelerations at the head were 2.44 ± 0.71, 2.73 ± 0.97 and 2.46 ± 0.85 g for the BF, BFHS and SHOD conditions, respectively. There were no significant differences for peak resultant head accelerations between conditions.

In general there was little difference in lower extremity kinematics between the 3 conditions (▶ **Table 2**). However, there were significant differences in sagittal plane ankle angle at ground contact between the BF and SHOD, and BF and BFHS conditions ($P<0.001$). In the BF condition, individuals contacted the ground in a more plantarflexed position; whereas, in the BFHS and SHOD conditions, individuals contacted the ground in a dorsiflexed position. There was also significant difference in peak sagittal plane hip angle between the SHOD and BFHS conditions ($P=0.040$) and BF and BFHS conditions ($P=0.022$).

There were significant differences in terms of impact peak and vertical GRF magnitudes between the 3 conditions (▶ **Table 1**). Specifically, there were significant differences in impact peak magnitude between the BF and SHOD conditions ($P=0.004$) and BF and BFHS conditions ($P=0.005$) (▶ **Table 1**). Impact peaks were present on 67 % of BF trials, 96 % of BFHS trials and 79 % of SHOD trials. There was a statistically significant difference in peak vertical GRF between BFHS and SHOD conditions ($P=0.034$) (▶ **Table 1**). There were no significant differences in loading rate, peak horizontal GRF, peak medio-lateral GRF or joint moments between any conditions.

Table 2 Lower extremity joint angles at ground contact and peak values.

		BF	BFHS	SHOD
Ankle Dorsiflexion (°)	At Contact	**−12.1 (7.0)** [b,c]	**7.4 (3.1)** [a]	**8.9 (5.6)** [a]
	Peak	30.0 (7.7)	30.4 (7.3)	29.0 (6.0)
Ankle Adduction (°)	At Contact	2.06 (6.1)	−4.2 (7.6)	−0.5 (7.2)
	Peak	7.3 (4.4)	11.4 (7.1)	9.6 (6.8)
Ankle Internal Rotation (°)	At Contact	−7.7 (10.5)	−13.2 (9.2)	−3.28 (14.8)
	Peak	2.2 (10.7)	3.1 (9.5)	5.2 (10.4)
Knee Flexion (°)	At Contact	8.8 (5.4)	3.2 (9.1)	6.3 (7.0)
	Peak	37.5 (6.3)	34.4 (4.3)	31.6 (6.1)
Knee Varus (°)	At Contact	5.9 (6.6)	6.0 (9.5)	3.4 (5.7)
	Peak	15.4 (9.9)	22.2 (16.1)	19.5 (8.7)
Knee Internal Rotation (°)	At Contact	−23.1 (17.6)	−27.6 (17.1)	−30.8 (13.0)
	Peak	1.6 (7.3)	4.1 (11.5)	4.9 (12.8)
Hip Flexion (°)	At Contact	36.4 (11.9)	35.2 (11.2)	36.8 (12.1)
	Peak	**36.8 (12.0)** [b]	**42.4 (10.4)** [a,c]	**38.3 (12.6)** [b]
Hip Adduction (°)	At Contact	4.6 (6.2)	5.2 (6.6)	5.6 (5.4)
	Peak	12.7 (8.0)	10.7 (8.5)	11.7 (5.0)
Hip Internal Rotation (°)	At Contact	21.9 (18.1)	24.6 (17.2)	25.4 (16.8)
	Peak	29.4 (14.9)	33.2 (13.9)	32.4 (10.5)

Data are mean (standard deviation). Significant differences are indicated in bold. [a] indicates a significant difference to BF. [b] indicates a significant difference to BFHS. [c] indicates a significant difference to SHOD. $P<0.05$. BW = body weight

Discussion

The goal of this study was to compare the effects of barefoot running form and running shoes on running kinetics and impact accelerations. The results supported our hypotheses that 1) both running shoes and barefoot running form reduced tibia impact accelerations and 2) heel strike running, whether barefoot or shod, resulted in higher impact peak magnitudes. The results of this study were consistent with previous studies that show that running shoes decrease tibia impact accelerations [4, 11]. However, contrary to McNair and Marshall, [4], our results show that running barefoot reduced impact acceleration magnitudes to the level seen with running shoes. This difference could be due to the amount of plantarflexion at ground contact, as our BFHS condition was associated with greater impact accelerations than the BF or shod conditions.

In the present study, GRF impact peak magnitudes were similar in the shod and BFHS condition; however, the shod condition was associated with reduced tibia impact accelerations. GRFs are a measure of the force applied to the ground by the body and are frequently used as a proxy for forces transmitted to the skeletal system. However, the foot's plantar surface is the only structure that receives these loads. The combination of GRF and accelerometer data used in the present study allowed for evaluation of impact force transmission. The results presented here suggests that the midsole of running shoes helps to dampen impact so the full GRF does not reach the tibia. However, in the BFHS condition more of the impact force was transmitted to the musculoskeletal system as shown by increased tibia accelerations.

The impact force created during ground contact in running is caused by inertial changes in the lower leg and results in an impact acceleration that is transmitted caudally through the musculoskeletal system [8, 12, 25]. As the impact force is transferred through the body it is partially attenuated, causing peak accelerations to occur at successively later times in each body segment [6]. Segment acceleration depends on forces applied to a segment and is influenced by joint stiffness, segment geometry, deformation, mass and moment of inertia [6]. Resulting segment accelerations equate to a portion of the body's mass, M_{eff}, stopping suddenly during impact [26, 27]. M_{eff} provides an important link between vertical impact forces and tibial acceleration, as the vertical impact force is the product of M_{eff} and tibial acceleration [28]. Changes in knee [6, 26, 28] and ankle angle [14] have been shown to alter M_{eff} and vertical impact forces.

Both the SHOD and BFHS conditions exhibited higher impact peak forces than the BF condition. This difference can be explained in part by differences in M_{eff} between conditions. In heel strike running M_{eff} at initial contact consists of the foot and lower leg and equates to 5–6.8 % of body mass [14, 26]. Alternatively, in fore/mid foot strike running M_{eff} consists of the forefoot, a portion of the rearfoot and the lower leg, and equates to 1.7 % of body mass [14]. While M_{eff} and impact peak magnitude were similar between the SHOD and BFHS conditions, the SHOD condition exhibited significantly lower tibial accelerations, suggesting the shoe midsole dampens force transmitted to the tibia. Consistent with previous studies, we have shown reduced impact peak magnitudes in the BF condition [29]. Our subjects also exhibited decreased tibia impact accelerations when running barefoot. These kinetic changes can likely be explained by a reduction in stride length and/or plantarflexed position at ground contact. Stride length is important to consider when evaluating impact attenuation, as stride length reduction has been shown to decrease peak impact accelerations [9, 30]. Our results indicate that individuals ran with a significantly greater stride length when shod. While longer stride lengths are typically associated with greater impact accelerations, the shod condition saw a reduction in impact accelerations as compared to the BFHS condition. The reduced impact acceleration, despite an increase in stride length, further supports the notion that the running shoe midsole helps to dampen impact forces. We have also shown that individuals ran at a similar stride length in both the BF and BFHS conditions, yet the BFHS condition resulted in greater impact accelerations. This would suggest that the plantarflexed position at ground contact helps to reduce impact accelerations. It has been proposed that contacting the ground on the fore/mid foot allows runners to absorb impact through compression of the medial arch of the foot, eccentric contraction of the triceps surae, and stretching of the Achilles tendon [14].

We observed no difference in resultant head accelerations between conditions, which is consistent with previous studies that have shown little effect of gait changes on head accelerations. Significant differences in the magnitude of head accelerations have been reported with changes in stride length, but the magnitude of these accelerations is considerably less than what was observed at the tibia [9, 30]. These findings indicate that, despite the magnitude of impact accelerations experienced at the lower extremity, active and passive structures reduce shock before it reaches the head. It has been proposed that several anatomical structures have evolved to attenuate shock so that vision remains stable and the brain does not experience great shock [31].

Certain limitations should be considered when interpreting the findings of our study. First, subjects wore their personal running shoes rather than standardized footwear. Previous studies show varied results for impact peak magnitude and loading rate with different shoes [32]. Second, it should be recognized that soft tissue movement can distort accelerometer signals, though we made every attempt to minimize soft tissue movement by firmly securing and using lightweight accelerometers. Additionally, changing limb orientation will influence accelerometer data; we therefore calculated resultant accelerations to better estimate lower extremity shock [24]. Lastly, it is important to note that there is considerable individual variation in kinetic and kinematic changes associated with different running conditions.

5. Conclusion

In conclusion, we have shown that both BF and shod running result in reduced impact accelerations. While shod running was associated with increased GRF impact peak magnitude, it appears that the midsole of running shoes helps attenuate impact forces, thus decreasing the amount of force transmitted through the musculoskeletal system. Barefoot running exhibited a similar decrease in impact accelerations, as well as decreased impact peak magnitude, which appeared to be due to a decrease in stride length and/or a more plantarflexed position at ground contact. Evaluating both GRFs and impact accelerations provides valuable information about the transmission of impact forces to the musculoskeletal system.

Conflict of interest: The authors have no conflict of interest to declare.

References

1 *Altman AR*, *Davis IS*. A kinematic method for footstrike pattern detection in barefoot and shod runners. Gait Posture 2012; 35: 298–300
2 *Altman AR*, *Davis IS*. Barefoot running: biomechanics and implications for running injuries. Curr Sports Med Rep 2012; 11: 244–250
3 *Bertelsen ML*, *Jensen JF*, *Nielsen MH*, *Nielsen RO*, *Rasmussen S*. Footstrike patterns among novice runners wearing a conventional, neutral running shoe. Gait Posture 2013; 38: 354–356
4 *Bobbert MF*, *Schamhardt HC*, *Nigg BN*. Calculation of vertical ground reaction force estimates during running from positional data. J Biomech 1991; 24: 1095–1105
5 *Bramble DM*, *Lieberman DD*. Endurance running and the evolution of Homo. Nature 2004; 432: 345–352
6 *Caspersen CJ*, *Powell KE*, *Koplan JP*, *Shirley RW*, *Campbell CC*, *Sikes RK*. The incidence of injuries and hazards in recreational and fitness runners. Med Sci Sports Exerc 1984; 16: 113–114
7 *Chi KJ*, *Schmitt D*. Impacts and kinematic adjustments during an exhaustive run. J Biomech 2005; 38: 1387–1395
8 *Denoth J*. Load on the locomotor system and modelling. In *Nigg* (ed.). Biomechanics of Running Shoes. Champaign: Human Kinetics; 1986: 63–116
9 *Derrick TR*, *Dereu D*, *McLean SP*. Impacts and kinematic adjustments during an exhaustive run. Med Sci Sports Exerc 2002; 34: 998–1002
10 *Derrick TR*, *Hamill J*, *Caldwell GE*. Energy absorption of impacts during running at various stride lengths. Med Sci Sports Exerc 1998; 30: 128–135
11 *Derrick TR*. The effects of knee contact angle on impact forces and accelerations. Med Sci Sports Exerc 2004; 36: 832–837
12 *Hamill J*, *Derrick T*, *Holt K*. Shock attenuation and stride frequency during running. Hum Movement Sci 1995; 14: 45–60
13 *Harriss DJ*, *Atkinson G*. Ethical standards in sport and exercise science research: 2016 update. Int J Sports Med. 2015; 36: 1121–1124
14 *Johnston C*, *Taunton J*, *Lloyd-Smith D*, *McKenzie D*. Preventing running injuries. Practical approach for family doctors. Can Fam Physician 2003; 49: 1101–1109
15 *Kadaba MP*, *Ramakrishnan HK*, *Wootten ME*. Measurement of lower extremity kinematics during level walking. J Orthopaed Res 1990; 8: 383–392
16 *Lafortune MA*. Three-dimensional acceleration of the tibia during walking and running. J Biomech 1991; 24: 877–886
17 *Lieberman DE*, *Venkadesan M*, *Werbel WA*, *Daoud AI*, *D'Andrea S*, *Davis IS*, *Mang'Eni RO*, *Pitsiladis Y*. Foot strike patterns and collision forces in habitually barefoot versus shod runners. Nature 2010; 463: 531–535
18 *Lieberman DE*. What we can learn about running from barefoot running: An evolutionary medical perspective. Exerc Sport Sci Rev 2012; 40: 63–72
19 *Lysholm J*, *Wiklander J*. Injuries in runners. Am J Sports Med 1987; 15: 168–171
20 *McNair PJ*, *Marshall RN*. Kinematic and kinetic parameters associated with running in different shoes. Br J Sports Med 1994; 28: 256–260
21 *Milner CE*, *Ferber R*, *Pollard CD*, *Hamill J*, *Davis IS*. Biomechanical factors associated with tibial stress fracture in female runners. Med Sci Sports Exerc 2006; 38: 323–328
22 *Mizrahi J*, *Verbitsky O*, *Isakov E*, *Daily D*. Effect of fatigue on leg kinematics and impact acceleration in long distance running. Hum Movement Sci 2000; 19: 139–151
23 *Nigg BM*, *Cole GK*, *Bruggemann GP*. Impact forces during heel-toe running. J Appl Biomech 1995; 11: 407–432
24 *Nigg BM*, *Liu W*. The effect of muscle stiffness and damping on simulated impact force peaks during running. J Biomech 1999; 32: 849–856
25 *Pain MTG*, *Challis JH*. The role of the heel pad and shank soft tissue during impacts: a further resolution of a paradox. J Biomech 2001; 34: 327–333
26 *Pohl M*, *Hamill J*, *Davis I*. Biomechanical and anatomic factors associated with a history of plantar fasciitis in female runners. Clin J Sports Med 2009; 19: 372–376
27 *Richards CE*, *Magin PJ*, *Callister R*. Is your prescription of distance running shoes evidence based? Br J Sports Med 2009; 43: 159–162
28 *Robbins SE*, *Hanna AM*. Running-related injury prevention through barefoot adaptations. Med Sci Sports Exerc 1987; 19: 148–156
29 *Verbitsky O*, *Mizrahi J*, *Voloshin A*, *Treiger J*, *Isakov E*. Shock transmission and fatigue in human running. J Appl Biomech 1998; 14: 301–311
30 *Voloshin A*, *Mizrahi J*, *Verbitsky O*, *Isakov E*. Dynamic loading on the human musculoskeletal system – effect of fatigue. Clin Biomech 1998; 13: 515–520
31 *Voloshin A*, *Wosk J*, *Brull M*. Force wave transmission through the human locomotor system. J Biomech Eng 1981; 103: 48–50
32 *Wright IC*, *Neptune RR*, *van den Bogert AJ*, *Nigg BM*. Passive regulation of impact forces in heel-toe running. Clin Biomech 1998; 13: 521–531
33 *Yan AF*, *Sinclair PJ*, *Hiller C*, *Wegener C*, *Smith RM*. Impact attenuation during weight bearing activities in barefoot vs. shod conditions: A systematic review. Gait Posture 2012; 38: 175–186
34 *Yeung SS*, *Yeung EW*, *Gillespie LD*. Interventions for preventing lower limb soft-tissue running injuries. Cochrane Database of Systematic Reviews 2011 Art. No.: CD001256