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ORIGINAL ARTICLE

Joint Contact Forces with Changes in Running Stride Length and Midsole Stiffness

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Abstract
The purpose of this study was to determine if lower extremity joint loading was influenced by stride length or shoe midsole cushioning. Ten subjects completed 10 trials of overground running at an average speed of 4.43 m/s in each of three conditions: normal running, running with a stride length (SL) reduced by 10% of normal, and running with a cushioned midsole stiffness (i.e., mechanical impact reduction of 13.7–10.9 g). Reaction forces calculated from inverse dynamics were summed with muscle forces estimated from a musculoskeletal model using static optimization to obtain joint contact forces at the hip, knee and ankle joints. Peak components of the contact forces [axial, anterior–posterior, and medial–lateral (ML)] were examined using parametric statistics ($\alpha = 0.05$). Reducing stride length resulted in significant decreases in absolute peak ankle contact forces in the axial direction (normal: -14.5 ± 1.5 BW; reduced SL: -14.0 ± 1.6 BW) and the ML direction (normal: 0.67 ± 0.23 BW; reduced SL: 0.61 ± 0.21 BW). Reducing stride length also reduced the peak absolute axial forces at the knee (normal: -10.6 ± 1.3 BW; reduced SL: -9.8 ± 1.2 BW) and the hip (normal: -7.26 ± 2.24 BW; reduced SL: -6.75 ± 2.10 BW). The cushioned shoe did not statistically reduce the peak absolute contact forces from the normal stride condition at any of the joints. Post hoc stress analysis suggested that the observed changes in anterior hip force would increase stress more than any of the other statistically significant results. Reductions in stride length appear to decrease some joint contact variables but cushioning in the heel region of the shoe does not.

Keywords Gait · Injury · Compressive force · Shear force · Cushioning · Strain

Introduction

Excessive loading of the lower extremity can result in damage to the skeletal system and methods to reduce these loads may result in decreased injury rates. The incidence of stress fractures has been reported to be 21% among track runners [5] and 8–13% for recreational runners [6]. Stress fractures account for up to 20% of injuries seen in sports medicine clinics [16]. An incidence of 4.0% was found in military recruits during 12 weeks of training, with the highest injury rates correlating with the weeks of the highest training volume [1]. A systematic review of military literature indicated a stress fracture incidence of 3% in males and 9% in females [27]. A 10-year study of 6000 athletes at the University of Minnesota found an overall incidence of stress fractures of 1% but the incidence was higher in female athletes (1.9%) and distance runners (3.2%) [2]. When viewed on a continuum rather than the ultimate fracture, bone stress injury can account for 20% of injuries seen in a sports medicine clinic [24].

Current knowledge of the etiology of stress injury is equivocal. Davis et al. [12] concluded that average vertical loading rate was greater in female runners classified with medically diagnosed injuries. However, Edwards [14] suggested that overuse injuries in bone are not the result of high rates of loading, but rather a mechanical fatigue phenomenon highly dependent on loading magnitude. Meardon et al. [18] found that runners with a history of stress fracture had smaller bone geometry and greater bending moments in the tibia during midstance—much later in the running cycle

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than impact. These results suggest that additional research is needed to clearly understand the details of this injury.

Two mechanisms that have the potential to reduce joint contact forces in runners are stride length and shoe cushioning manipulation. Derrick et al. [13] found that impacts were greatly influenced by stride length, whereby increasing stride length from −20% of preferred to +20% of preferred increased tibial accelerations from 5.7 to 11.3 g. Although impacts can be reduced with stride length, it is unclear if this technique would reduce lower extremity joint contact forces. Furthermore, shorter strides will require an increased number of strides for a given distance, which may also increase the potential for injury. Edwards et al. [15] used a probabilistic stress fracture model to show a 10% decrease in stride length would reduce the probability of tibial stress fracture for a given mileage, despite an increase in the number of loading cycles; therefore, reducing stride length has the potential to reduce injury.

The effect of increasing midsole cushioning on stress injury potential is less clear. This may be, in part, due to the difficulty in altering shoe midsole cushioning without changing other aspects of shoe performance. It is likely that a softer midsole can reduce the impact experienced by the runner even if the change is not reflected in the ground reaction force [20, 22, 25]. This reduced impact does not necessitate a reduced injury potential but there is correlational evidence suggesting a relationship between impacts and injury [12]. Meardon et al. [19] examined soft (9.3 g), medium (10.3 g) and hard (12.0 g) midsole shoes and found that increasing midsole cushioning reduced the ankle and, to a lesser extent, knee joint peak absolute axial contact forces, but contact forces at the hip remained similar across conditions.

The purpose of this study was to determine how altered stride length and heel region shoe midsole cushioning affected joint contact forces. This research combined with previous results is expected to help narrow the list of crucial variables that affect bone and joint loading in runners. We defined joint loading using the peak contact force at the hip, knee and ankle joints. We hypothesized that reducing stride length by 10% would decrease the magnitude of peak joint contact forces and decreased midsole cushioning in the heel region would have lesser or no measurable effects on the peak contact forces.

Methods

Participants

Upon entering the laboratory, all subjects read and signed an Informed Consent Document approved by the Human Subjects Research Office at Iowa State University. Ten males (69.2 ± 6.5 kg, 1.78 ± 0.05 m, 22.2 ± 3.2 years) participated in the study. All were experienced competitive runners, nine of which were current or former college cross-country runners, with an average weekly mileage of 63 ± 20 miles. Subjects were limited to those who could wear a size 9 or 11 shoe. None of the subjects had lower extremity injuries at the time that would affect their ability to run.

Protocol

All subjects wore the Adidas 1.1 running shoe with adjustable cushioning. Each shoe contained a small microprocessor, a sensor, and a motorized cable system in the heel of the shoe. Through the use of a magnetic "key" the cable length can be lengthened or shortened to create a softer or firmer cushion as desired. Prior to data collection, the Adidas 1.1 shoes were mechanically tested with an impact testing device (Exeter Research, Exeter, NH) according to a standardized protocol [4] to verify cushioning differences between midsole conditions. The Adidas 1.1 shoes showed a peak value of 13.7 g for the normal setting. The cushioned shoe setting tested at 10.9 g—a 20% reduction in the peak g value from the normal setting.

Height, total body mass, thigh length, mid-thigh circumference, calf length, calf circumference, foot length, malleolus height, malleolus width, and foot breadth were recorded for each subject to build a rigid body model to be used with inverse dynamics analysis [26]. Retroreflective markers were placed on the anterior dorsifoot, heel, medial and lateral malleoli, anterior calf, medial and lateral knee, anterior thigh, right and left greater trochanter, right and left ASIS, and lower back. Prior to data collection, subjects were allowed to self-select their preferred running speed. Subjects were asked to run down a 30 m runway at a comfortable training speed towards a force platform imbedded in the floor, and land on the force platform with the right foot without targeting the force platform. Trials were accepted if the speed was ± 5% of their preferred running speed. Average self-selected speed was 4.43 ± 0.48 m/s. Subjects completed ten trials for each of the three conditions. The three conditions were a normal stride running condition (NS), a condition with a stride length reduced by 10% (RSL), and a condition with a cushioned stride via a soft midsole (CS). The shorter stride lengths were achieved through the use of floor markings [15]. All 30 trials were completed in a single session.

Data Collection

Marker coordinate data were collected with a Peak Motus 3D optical capture system (Vicon Peak, Centennial, CO) with a sampling frequency of 120 Hz. During dynamic trials, force platform data were collected concurrently with a sampling frequency of 1200 Hz using a strain gage force platform (model OR6-7-2000, AMTI).

Data Reduction

Marker coordinates and force platform data were low-pass digitally filtered at 12 and 30 Hz, respectively, using a fourth-order Butterworth filter. Running velocity was calculated as the average horizontal velocity of the lower back marker during the stance phase of the running cycle. The synchronized raw motion capture and force platform data were then exported to Matlab (R2018b, Mathworks, Natick, MA) for data processing. Post hoc footstrike analysis [7] was performed to determine if the runners utilized a forefoot (initial contact in distal two-thirds of the foot) or rearfoot strike pattern initial contact in the proximal third of the foot.

Inverse Dynamics

Dynamic motion capture data were referenced to a static trial for the calculation of segment coordinate systems for the foot, calf, thigh, and pelvis [23]. Three-dimensional helical angles were calculated for each segment and projected onto the segment coordinate systems. Three-dimensional internal joint moments and reaction forces were calculated in the global coordinate system using inverse dynamics with rigid body assumptions and then rotated into their respective proximal segment coordinate systems.

Muscle Forces

The muscle, joint, and function definitions of the musculoskeletal model from Arnold et al. [3] were used and implemented in Matlab (R2018b, Mathworks, Natick, MA). This allowed for the estimation of 44 lower extremity muscle moment arms, muscle orientations, and maximum dynamic muscle forces. Muscle forces were adjusted by physiological cross-sectional area, muscle fiber length, and muscle fiber velocity. Model dimensions were scaled by subject segment lengths and muscle forces were scaled by body mass.

Muscle forces during running were estimated using static optimization techniques. The cost function (u) was the sum of squared muscle stresses [17]:

$$u = \sum_{i=1}^{44} \left(f_i/\mathrm{PCSA}_i\right)^2,$$

where f_i is the force generated by the ith muscle, and PCSA_i is the physiological cross-sectional area of the ith muscle. The optimization was constrained so that the resulting sagittal plane hip, knee and ankle moments, the frontal plane ankle moment, and transverse plane hip moment from the inverse dynamics were equal to the moments created by the muscles. This method previously produced muscle force patterns similar to typical electromyographic signals in forefoot and rearfoot strike runners [21].

Contact Forces

Three-dimensional joint contact forces were calculated as the sum of reaction forces from the inverse dynamics and the muscle forces from the musculoskeletal model optimization procedure. Axial, mediolateral, and anteroposterior joint contact forces applied to the proximal segment were transformed into the proximal segment coordinate system.

Statistics

Values were averaged across trials for each condition. A repeated-measures MANOVA was performed in SPSS Statistics 25 to compare peak three-dimensional joint contact forces at all 3 joints. Significance was set to $\alpha = 0.05$. If the MANOVA was significant, univariate ANOVAs were subsequently evaluated at a 0.05 level. A Sidak adjustment for pairwise comparisons was used. When sphericity was violated, the Greenhouse–Geisser correction was used. Cohen's effect sizes (d) that were corrected for dependent measures [9] were used to estimate the size of the effect (small: 0.2–0.5; moderate: 0.5–0.8; large: > 0.8). In addition, ground reaction forces were presented for descriptive purposes and paired t tests ($\alpha = 0.05$) were used to test differences between peak values.

Results

Ground Reaction Forces

Footstrike pattern was not controlled and it was determined that four of the ten runners were forefoot strikers with the initial contact made in the distal two-thirds of the foot [8]. There was a noted similarity between the general shapes of the vertical ground reaction force (VGRF) curves from each condition (Fig. 1). Peak VGRF was significantly different between conditions with the normal stride condition having the greatest value [2.98 ± 0.30 BW (body weight)] followed by the cushioned shoe condition (2.95 ± 0.31 BW) and the reduced stride condition (2.84 ± 0.30 BW).

Contact Forces at the Ankle

Approximately 19% of the resultant joint contact force at the ankle was the result of joint reaction forces with the remaining force the result of muscular activity. Joint contact forces at the ankle were directed downward, forward and,

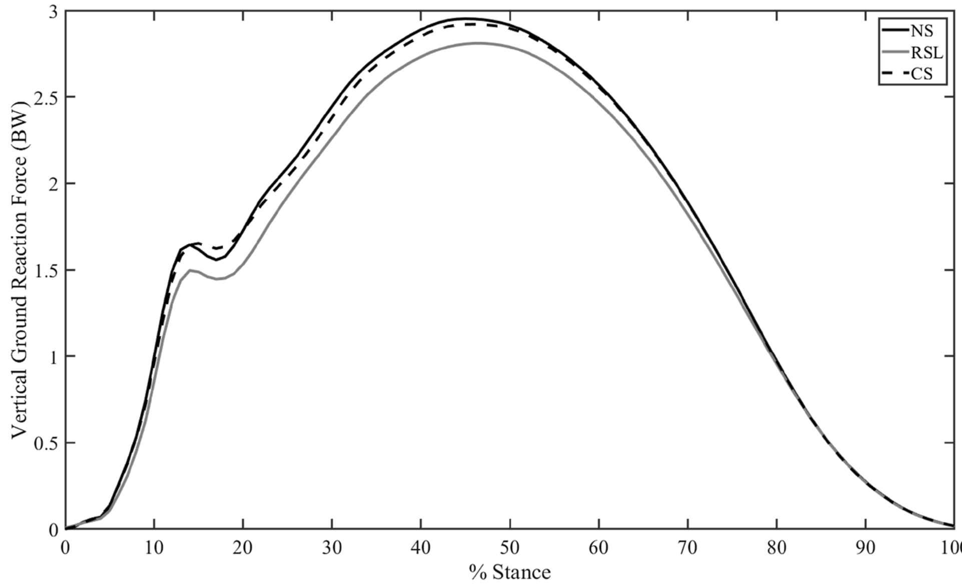

Fig. 1 Unsmoothed vertical ground reaction forces (VGRF) for the normal stride (NS), reduced stride length (RSL), and cushioned shoe (CS) conditions. Values are normalized to a percent of the stance phase of the running cycle and divided by body weight (BW)

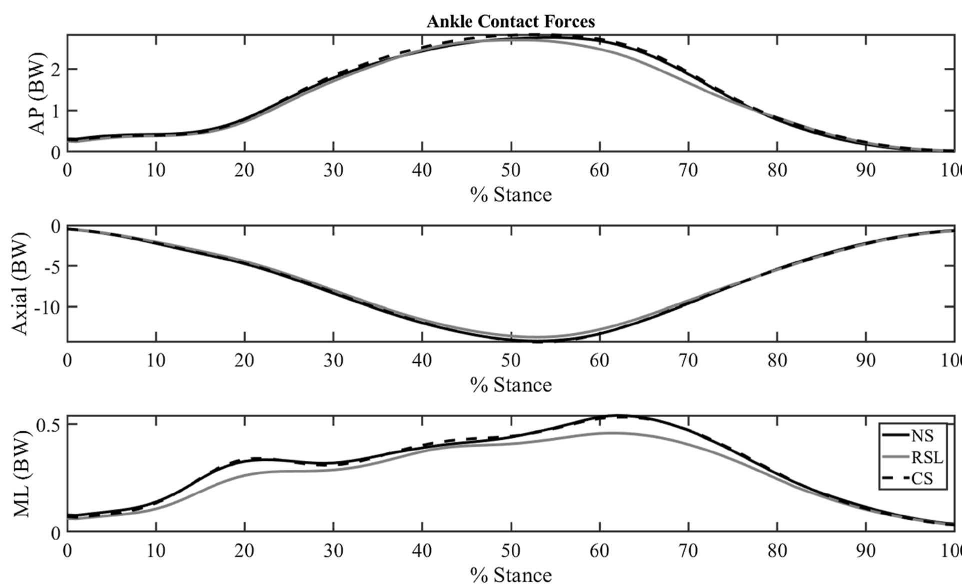

Fig. 2 Ensemble curves of ankle joint contact forces in the proximal segment coordinate system for the normal stride (NS), reduced stride length (RSL) and cushioned shoe (CS) conditions. Positive values indicate forces that are directed upward, forward and lateral. Curves are normalized to a percent of the stance phase of the running cycle and divided by body weight (BW)

slightly lateral in the leg coordinate system (Fig. 2). Peak values occurred at or slightly after midstance with magnitudes greatest in the axial direction (~ 14 BW). Univariate tests (Table 1) indicated statistically significant differences in the axial and medial–lateral (ML) directions. Post hoc tests indicated these differences were due to peak absolute axial forces that were less in the reduced stride length condition compared to the either the normal stride condition (RSL: -14.0 ± 1.6 BW vs. NS: -14.5 ± 1.5 BW; $P = 0.041$, $d = 0.75$) or the cushioned shoe condition (RSL: -14.0 ± 1.6 BW vs. CS: -14.5 ± 1.6 BW; $P = 0.002$, $d = 1.40$). In addition, the peak ML forces that were less in the reduced stride length condition compared to the normal condition (RSL: 0.61 ± 0.20 BW vs. NS: 0.67 ± 0.23 BW; $P = 0.021$, $d = 0.87$).

Table 1 Mean and standard deviations for the peak components of the ankle joint contact force during normal running (NS) reduced stride length running (RSL) and cushioned shoe running (CS)

ANKLE	NS	RSL	CS
Peak AP Force	2.98	2.85	3.05
(BW)	(0.45)	(0.46)	(0.48)
Peak Axial Force	– 14.5	– 14.0	– 14.5
(BW)[a,b]	(1.5)	(1.6)	(1.6)
Peak ML Force	0.67	0.61	0.66
(BW)[a]	(0.23)	(0.20)	(0.21)

[a]RSL different from NS

[b]RSL different from CS

Contact Forces at the Knee

Approximately 25% of the resultant joint contact force at the knee was the result of joint reaction forces with the remaining force the result of muscular activity. Joint contact forces at the knee were applied to the top of the tibia and directed downward, forward and slightly medial (Fig. 3) in the thigh coordinate system. Peak values occurred around midstance with magnitudes greatest in the axial direction (~ 10 BW). Univariate tests (Table 2) indicated statistically significant differences in the axial direction. Post hoc tests indicated these differences were due to peak absolute axial forces in the reduced stride length condition (RSL: -9.8 ± 1.2 BW) that were lesser compared to both the normal condition (NS: -10.6 ± 1.3 BW; $P=0.003$, $d=1.28$) and the cushioned shoe condition (CS: -10.6 ± 0.7 BW; $P=0.002$, $d=1.33$).

Contact Forces at the Hip

Approximately 28% of the resultant joint contact force at the hip was the result of joint reaction forces with the remaining force the result of muscular activity. Joint contact forces at the hip were measured such that they were applied to the top of the femur and directed downward and lateral in the pelvis coordinate system (Fig. 4). Contact forces in the anterior–posterior (AP) direction were directed anterior during the first half of stance and posterior during the second half. Univariate tests (Table 3) indicated statistically significant differences in the anteriorly directed peak (Peak AP Force 1). Post hoc tests indicated this difference was due to the peak anteriorly directed force in the reduced stride length condition (RSL: 1.17 ± 0.30 BW) that was less than the cushioned shoe condition (CS: 1.28 ± 0.37 BW; $P=0.014$, $d=0.97$). Post hoc test indicated all three comparisons were statistically significant in the axial direction (NS: -7.26 ± 2.24 BW vs. RSL: -6.75 ± 2.10 BW; $P=0.027$, $d=0.83$), (NS: -7.26 ± 2.24 BW vs. CS: -7.55 ± 2.17 BW; $P=0.024$, $d=0.86$) and (RSL: -6.75 ± 2.10 BW vs. CS: -7.55 ± 2.17 BW; $P=0.016$, $d=0.94$).

Table 2 Mean and standard deviations for the peak components of the knee joint contact force during normal running (NS) reduced stride length running (RSL) and cushioned shoe running (CS)

KNEE	NS	RSL	CS
Peak AP Force	−3.98	−3.87	−4.02
(BW)	(0.96)	(1.03)	(1.05)
Peak Axial Force	−10.6	−9.8	−10.6
(BW)[a,b]	(1.3)	(1.2)	(0.7)
Peak ML Force	−0.92	−0.87	−0.90
(BW)	(0.66)	(0.70)	(0.67)

[a]RSL different from NS

[b]RSL different from CS

Discussion

Reducing stride length and increasing shoe cushioning represent two potential mechanisms to decrease lower extremity loading in runners. In this study, we defined lower extremity loading by the peak joint contact forces occurring at the hip, knee and ankle during the stance phase of gait. Other measures of loading would be expected to elicit different conclusions.

Our hypothesis that shoe cushioning would not affect joint loading was, in part, due to the nature of the shoe and the method of assessing joint loading. A compilation of 142

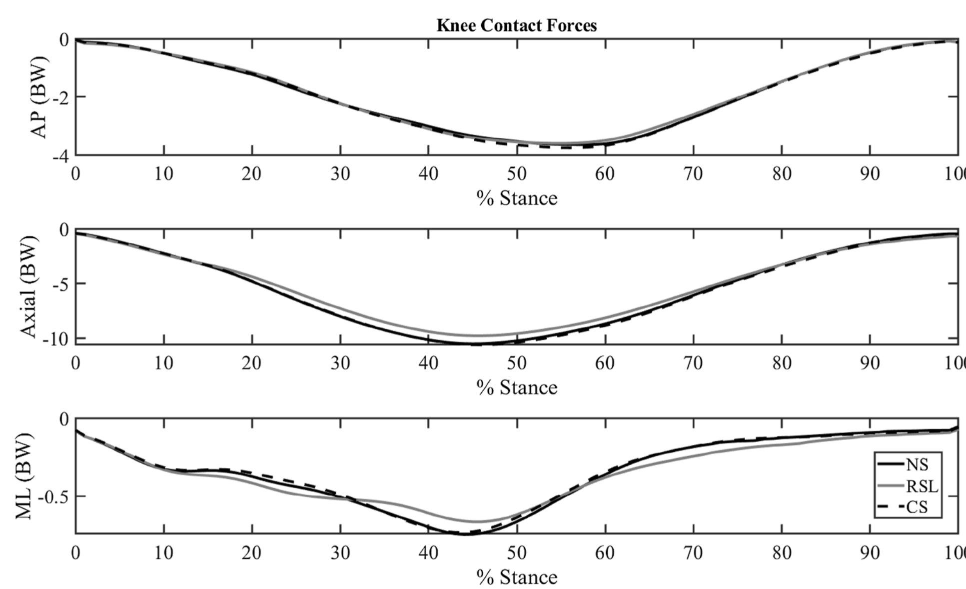

Fig. 3 Ensemble curves of knee joint contact forces in the proximal segment coordinate system for the normal stride (NS), reduced stride length (RSL) and cushioned shoe (CS) conditions. Positive values indicate forces that are directed upward, forward and lateral. Curves are normalized to a percent of the stance phase of the running cycle and divided by body weight (BW)

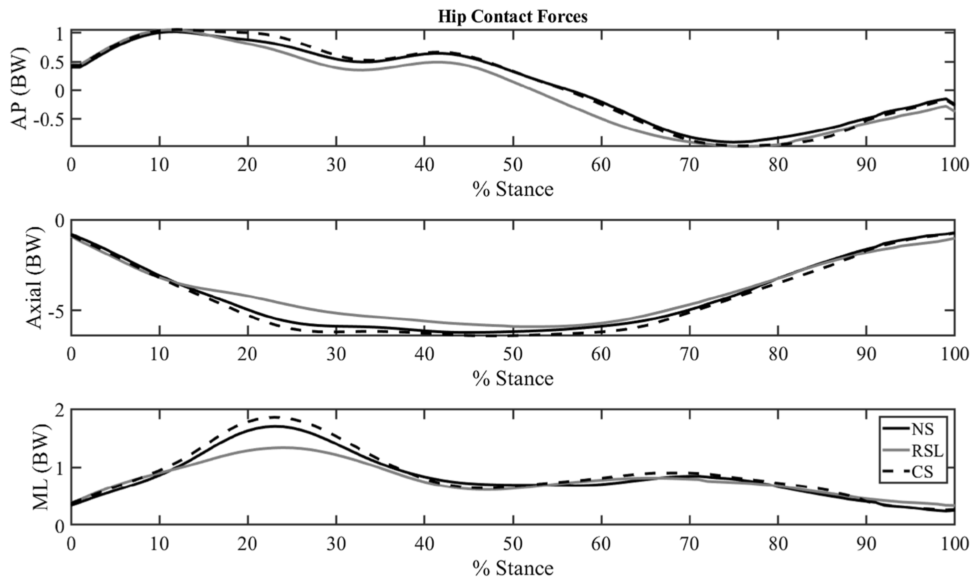

Fig. 4 Ensemble curves of hip joint contact forces in the proximal segment coordinate system for the normal stride (NS), reduced stride length (RSL) and cushioned shoe (CS) conditions. Positive values indicate forces that are directed upward, forward and lateral. Curves are normalized to a percent of the stance phase of the running cycle and divided by body weight (BW)

Table 3 Mean and standard deviations for the peak components of the hip joint contact force during normal running (NS) reduced stride length running (RSL) and cushioned shoe running (CS)

HIP	NS	RSL	CS
Peak AP Force 1	1.18	1.17	1.28
(BW)[b]	(0.26)	(0.30)	(0.37)
Peak AP Force 2	− 0.96	− 1.07	− 1.03
(BW)	(0.24)	(0.32)	(0.25)
Peak Axial Force	− 7.26	− 6.75	− 7.55
(BW)[a,b,c]	(2.24)	(2.10)	(2.17)
Peak ML Force	2.29	2.14	2.38
(BW)	(0.75)	(0.76)	(0.76)

[a]RSL different from NS

[b]RSL different from CS

[c]NS different from CS

commercially available running shoes tested in the same laboratory showed a range of 7.2–13.5 g using the same protocol. From these data it can be concluded that the normal setting for the shoe used in this study would be considered hard, while the cushioned setting would be considered moderate. The cable used to stiffen the midsole of the shoe only extends through the heel region, and therefore, loading of the body past the impact phase (first 50 ms) would not be expected to change unless general kinematics were affected by the heel stiffness. The exception to this was at the hip joint, where there was a peak AP force that occurred at 10–20% stance. Since peak contact force in this study generally occurred around midstance, the altered heel cushioning would be expected to have minimal effect. Other assessment methods, such as peak tibial acceleration from inertial measurement units, might be expected to produce lower impacts in the cushioned shoe. Unpublished results from our lab using these shoes indicated that the hard shoe setting would result in a 1 g increase in peak tibial accelerations over the normal setting. It is still a matter of debate whether impact loading rate or peak loading magnitude is the best measure to assess the probability of obtaining a stress injury [14].

Our hypotheses that stride length reduction would decrease lower extremity loading was supported. Reducing stride length significantly decreased the axial and AP directed knee joint contact forces relative to the normal condition. In addition, peak forces in the reduced stride length condition were significantly lower than the cushioned shoe condition for axial ankle and knee joint contact forces as well as for AP and hip joint contact forces. On the other hand, peak forces in the cushioned shoe condition were not statistically lower than the normal stride or reduced stride condition for any comparison.

Although statistical differences were associated with large effects (> 0.80) in all cases except the comparison between NS and RSL in the ankle compressive force ($d = 0.75$), it is not clear if these represent clinically relevant reductions in load. Indeed, the relationship between axial and shear contact forces (both AP and ML) and resulting bone strain is highly complex [13, 14]. Shear forces applied to the joint exert considerable bending moments, while axial forces tend to compress. Bending moments will produce compression on one side of the bone while producing tension on the other side, and most of the bone stress during locomotion is due to bending [13]. It has been estimated that 72% of the peak tibial stress during walking at the periphery of the tibial cross section located at 62% of tibial length (measured from the knee) is the result of bending moments [13]. Post-hoc analysis can give a rough estimate of the relative

Table 4 The influence of statistically significant changes in joint contact force and their effects on normal stress during normal running (NS) reduced stride length running (RSL) and cushioned shoe running (CS)

Joint	Direction	Significance	ΔForce (BW)	$\Delta\sigma$ (MPa)
Ankle	Compressive	NS > RSL	0.52	1.13
Ankle	Compressive	CS > RSL	0.54	1.17
Ankle	Lateral Shear	NS > RSL	0.03	3.74
Knee	Compressive	NS > RSL	0.79	1.72
Knee	Compressive	CS > RSL	0.84	1.82
Hip	Anterior shear	CS > RSL	0.11	9.12
Hip	Compressive	NS > RSL	0.51	0.85
Hip	Compressive	CS > RSL	0.80	1.33
Hip	Compressive	CS > NS	0.29	0.48

contributions of each significant result (Table 4). A simple model, in which the contact force is applied to the proximal or distal end of a hollow elliptical tube that represents the femur or the tibia, can be used to estimate stress on the mid-diaphysis periphery. For this analysis, mid-diaphyseal cross-sectional geometry and segment lengths were taken from the literature for the tibia [10] and femur [11]. The findings illustrated in Table 4 suggest that the change in hip joint anterior shear force would create an increase of 9.12 MPa of compressive stress on one side of the bone cross section and reduce the tensile stress by 9.12 MPa on the opposite side. Viewed in this manner this may be the most clinically relevant finding from this study, because it created the greatest change in bone stress.

Multi-scale modeling is necessarily compromised by the number of assumptions made in each model. Some of these assumptions are trivial, while others are not. In this study, muscle forces are not measured directly but estimated using optimization techniques. This is necessary because of the lack of a more direct measurement and the redundant nature of muscles. There are an infinite number of sets of muscle forces that will produce the joint torques estimated by the inverse dynamics. The cost function is used to select the most realistic set of muscle forces but there is no guarantee this is the actual muscle forces selected by the individual runner. The procedures in this research have several advantages that make this assumption more realistic. (1) Unlike raw electromyography, the maximum dynamic muscles forces are length, velocity and physiological cross-sectional area adjusted and virtually all muscles are accounted for rather than just surface muscles. (2) Muscle forces are not used in the end analysis. Contact forces are more robust than individual muscle forces, because they are the sum of all agonistic and antagonistic activity. (3) The repeated-measures study design is utilized so that any "miscalculation" of muscle forces would have to be inconsistent across conditions to affect the results.

Conclusion

Reducing stride length while running appears to be a productive strategy for decreasing some of the joint contact forces in the lower extremity. Conversely, increasing midsole cushioning in the heel region does not appear to reduce joint contact forces. Magnitude of the changes in the components of the joint contact forces needs to be evaluated with reference to the potential stresses that they may cause in the bone. Shear forces acting on long bones have the potential to increase bending moments at the mid-diaphysis and, therefore, increase peripheral stress beyond that of applied axial compressive forces.

References

1. Almeida SA, Williams KM, Shaffer RA, Brodine SK. Epidemiological patterns of musculoskeletal injuries and physical training. Med Sci Sports Exerc. 1999;31(8):1176–82.
2. Arendt E, Agel J, Heikes C, Griffiths H. Stress injuries to bone in college athletes: a retrospective review of experience at a single institution. Am J Sports Med. 2003;31(6):959–68.
3. Arnold EM, Ward SR, Lieber RL, Delp SL. A model of the lower limb for analysis of human movement. Ann Biomed Eng. 2010;38(2):269–79.
4. ASTM F. Standard test method for cushioning properties of athletic shoes using an impact test. American Society for Testing and Material. 1976.
5. Bennell KL, Malcolm SA, Thomas SA, Wark JD, Brukner PD. The incidence and distribution of stress fractures in competitive track and field athletes: a twelve-month prospective study. Am J Sports Med. 1996;24(2):211–7.
6. Brunet ME, Cook SD, Brinker MR, Dickinson JA. A survey of running injuries in 1505 competitive and recreational runners. J Sports Med Phys Fit. 1990;30(3):307–15.
7. Burdett RG. Forces predicted at the ankle during running. Med Sci Sports Exerc.1982;14(4):308–16.
8. Cavanagh PR, Lafortune MA. Ground reaction forces in distance running. J Biomech. 1980;13(5):397–406.
9. Cohen J. Statistical power analysis for the behavioral sciences. L. Lawrence Earlbaum Associates: Hillsdale; 1988. p. 2.
10. Croker SL, Clement JG, Donlon D. A comparison of cortical bone thickness in the femoral midshaft of humans and two non-human mammals. HOMO – J Comp Hum Biol. 2009;60(6):551–65.
11. Crowninshield RD, Brand RA. A physiologically based criterion of muscle force prediction in locomotion. J Biomech. 1981;14(11):793–801.
12. Davis IS, Bowser BJ, Mullineaux DR. Greater vertical impact loading in female runners with medically diagnosed injuries: a prospective investigation. Br J Sports Med. 2016;50(14):887–92.
13. Derrick TR, Hamill J, Caldwell GE. Energy absorption of impacts during running at various stride lengths. Med Sci Sports Exerc. 1998;30(1):128–35.
14. Edwards WB. Modeling overuse injuries in sport as a mechanical fatigue phenomenon. Exerc Sport Sci Rev. 2018;46(4):224–31.
15. Edwards WB, Gillette JC, Thomas JM, Derrick TR. Internal femoral forces and moments during running: implications for stress fracture development. Clin Biomechan. 2008;23(10):1269–78.

16. Edwards WB, Taylor D, Rudolphi TJ, Gillette JC, Derrick TR. Effects of stride length and running mileage on a probabilistic stress fracture model. Med Sci Sports Exerc. 2009;41(12):2177–84.
17. Glitsch U, Baumann W. The three-dimensional determination of internal loads in the lower extremity. J Biomech. 1997;30(11–12):1123–31.
18. Meardon SA, Willson JD, Gries SR, Kernozek TW, Derrick TR. Bone stress in runners with tibial stress fracture. Clin Biomech. 2015;30(9):895–902.
19. Meardon SA, Willson JD, Kernozek TW, Duerst AH, Derrick TR. Shoe cushioning affects lower extremity joint contact forces during running. Footwear Sci. 2018;10(2):109–17.
20. Miller RH, Hamill J. Computer simulation of the effects of shoe cushioning on internal and external loading during running impacts. Comput Methods Biomech Biomed Eng. 2009;12(4):481–90.
21. Nigg BM, Herzog W, Herzog W. Biomechanics of the musculo-skeletal system, vol 192. New York: Wiley; 1999.
22. Rooney BD, Derrick TR. Joint contact loading in forefoot and rearfoot strike patterns during running. J Biomech. 2013;46(13):2201–6.
23. Sasimontonkul S, Bay BK, Pavol MJ. Bone contact forces on the distal tibia during the stance phase of running. J Biomech. 2007;40(15):3503–9.
24. Scott SH, Winter DA. Internal forces of chronic running injury sites. Med Sci Sports Exerc. 1990;22(3):357–69.
25. Shorten M, Mientjes MI. The ‘heel impact’ force peak during running is neither ‘heel’ nor ‘impact’ and does not quantify shoe cushioning effects. Footwear Sci. 2011;3(1):41–58.
26. Söderkvist I, Wedin PÅ. Determining the movements of the skeleton using well-configured markers. J Biomech. 1993;26(12):1473–7.
27. Vaughan CL, Davis BL, Jeremy CO. Dynamics of human gait. Champaign: Human Kinetics; 1992. p. 17–22.