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Original article

Reduction in ground reaction force variables with instructed barefoot running

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Abstract

Backgound: Barefoot (BF) running has recently increased in popularity with claims that it is more natural and may result in fewer injuries due to a reduction in impact loading. However, novice BF runners do not necessarily immediately switch to a forefoot strike pattern. This may increase mechanical parameters such as loading rate, which has been associated with certain running-related injuries, specifically, tibial stress fractures, patellofemoral pain, and plantar fasciitis. The purpose of this study was to examine changes in loading parameters between typical shod running and instructed BF running with real-time force feedback.

Methods: Forty-nine patients seeking treatment for a lower extremity injury ran on a force-sensing treadmill in their typical shod condition and then BF at the same speed. While BF they received verbal instruction and real-time feedback of vertical ground reaction forces.

Results: While 92% of subjects ($n = 45$) demonstrated a rearfoot strike pattern when shod, only 2% ($n = 1$) did during the instructed BF run. Additionally, while BF 47% ($n = 23$) eliminated the vertical impact transient in all eight steps analyzed. All loading variables of interest were significantly reduced from the shod to instructed BF condition. These included maximum instantaneous and average vertical loading rates of the ground reaction force ($p < 0.0001$), stiffness during initial loading ($p < 0.0001$), and peak medial ($p = 0.001$) and lateral ($p < 0.0001$) ground reaction forces and impulses in the vertical ($p < 0.0001$), medial ($p = 0.047$), and lateral ($p < 0.0001$) directions.

Conclusion: As impact loading has been associated with certain running-related injuries, instruction and feedback on the proper forefoot strike pattern may help reduce the injury risk associated with transitioning to BF running.

Keywords: Barefoot running; Ground reaction force; Loading rates; Mediolateral forces; Vertical stiffness

1. Introduction

Barefoot (BF) running has recently increased in popularity among runners with a perception that it is more natural and may result in fewer injuries. In fact, the top reason runners report for choosing to transition to BF or minimal running is the notion of injury prevention.[1] The potential for a lower risk of injury is postulated based on strengthening of the foot,[2] and changes in loading parameters due to alterations in running pattern associated with BF running.[3]

It has been documented that up to 89% of traditionally shod runners land on their heels or with a rearfoot strike (RFS).[4,5] This strike pattern is associated with an impact transient in the vertical ground reaction force (VGRF), followed by a propulsive peak. The impact transient appears as a distinct change in the positive slope of the VGRF trace, sometimes characterized by a local maximum or impact peak (VIP). The rate of development of the VGRF is referred to as the loading rate (Fig. 1A). High loading rates and impact transients have

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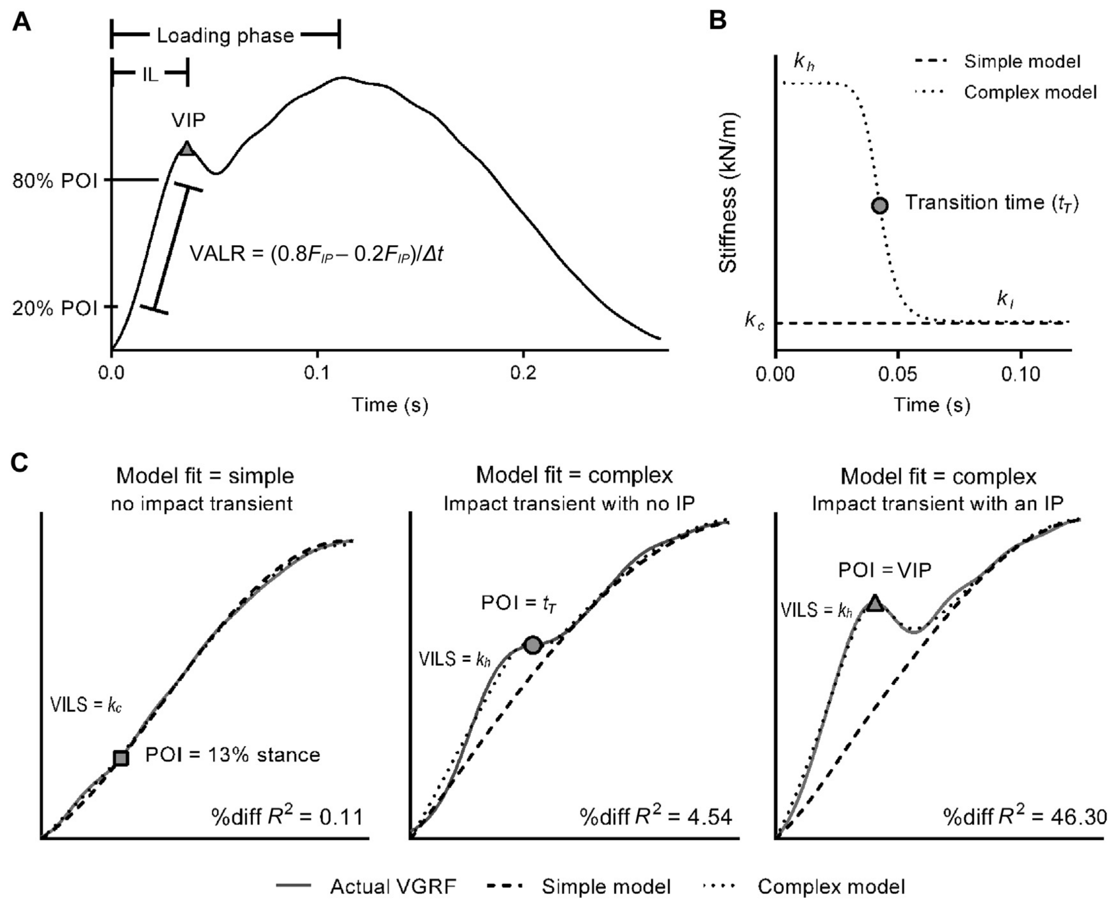

Fig. 1. Complex and simple model fits of the vertical ground reaction force (VGRF). (A) VGRF indicating how the average loading rate (VALR) is computed from 20% to 80% of the force at the point of interest (POI); in this case the vertical impact peak (VIP). (B) Stiffness curve for the simple and complex models. The simple model uses a constant stiffness, k_c, while the complex model has a time varying stiffness that starts with a high stiffness, k_h, during the initial loading (IL) phase and transitions to a low stiffness k_l. (C) Plots of the loading phase of the VGRF curve with both the simple and complex model overlaid for three cases. Case 1: the percent difference for the R^2 values between the two fits (%diff R^2) is <3.0; the simple model is chosen; stiffness during IL is set to k_c; loading rate is computed from 13% of stance. Cases 2 and 3: %diff $R^2 > 3.0$; the complex model is chosen; stiffness during IL is set to k_h. Case 2: There is no impact peak (IP); POI for computing loading rates is the transition time. Case 3: IP present; POI for computing loading rates is the VIP.

been associated with a number of common running-related injuries such as tibial stress fractures,[6] patellofemoral pain,[7] and plantar fasciitis.[8] Most habitual BF runners land on the ball of their foot, referred to as a forefoot strike (FFS) with their foot in a relatively flat orientation.[9] This pattern typically has a single propulsive peak in the VGRF, lacking a distinct vertical impact transient.[10] Elimination of this impact transient is accomplished by reducing vertical stiffness of the body. Vertical stiffness can be assessed using a simple mass spring model[11,12] which works well for an FFS pattern. However, when impact transients are present, a dual stiffness model, such as described by Hunter,[13] should be used.

The influence of strike pattern on the medial and lateral components of the ground reaction force (GRF) is not well established. The lateral GRF may contribute to pronation of the foot, which when excessive has been linked to lower leg and knee pain.[14,15] Changes in the lateral force may influence a runner's tendency to overpronate. Therefore evaluating the change in this parameter between shod and BF runners may lend future insight into the link between these running conditions and certain injuries.

BF running is also associated with a shorter stride and a higher cadence.[16,17] Higher cadence running has been reported to reduce loading at the hip and knee,[18] which may influence injury risk. A higher cadence also results in a shorter stance time with each footstrike. Therefore, it is expected that vertical and mediolateral impulses will also be reduced.

While habitual BF runners usually land with a midfoot strike (MFS) or FFS pattern,[3] novice BF runners may persist with an RFS pattern and experience higher loading rates than when shod.[3,16] It has been theorized that an RFS is uncomfortable or painful when running BF, thus encouraging runners to naturally transition to an FFS. However, the length of time it takes novice BF runners for this transition is unknown. In a recent study, 20 of 30 novice BF runners immediately transitioned to an FFS without instruction.[19] Despite this transition, two of 20 runners maintained high loading rates. Therefore, providing feedback and instruction early in the process may assist in reducing impacts and loading rates when first learning to transition to BF running.

The purpose of this study was to determine changes in loading parameters when habitually shod runners who exhibit an impact transient run BF while being given verbal instruction and real-time visual feedback of their VGRFs. We hypothesized that outcome variables derived from the GRF (vertical stiffness, vertical loading rates, mediolateral forces and impulses) will decrease when runners transition from typical shod running to BF running during a single session of training with feedback.

2. Methods

2.1. Participants

A total of 100 patients seeking treatment for a chronic lower extremity injury between 24 July, 2012 and 6 August, 2013 were considered for inclusion in this study. As this was research involving data collected solely for clinical purposes, the institutional IRB granted authorization and a waiver of informed consent. Patients experiencing acute pain were asked to reschedule their appointments until they would be able to run comfortably on a treadmill for up to 10 min. Patients were excluded if they were unable to run in the BF condition due to pain from aggravating their existing injury ($n = 2$). In order for speeds to be consistent between conditions, those who were uncomfortable running at their self-selected shod pace when running BF (and hence ran slower) were also excluded ($n = 33$). Throughout the analysis, a step was defined as having an impact transient if it demonstrated a change in vertical stiffness during the loading phase of stance (described in Section 2.3.2). We focused the study on runners who typically had an impact transient and would require a change in running pattern to eliminate it. Therefore, patients who lacked an impact transient in at least five of eight steps during the shod condition were excluded ($n = 16$). This yielded 26 females and 23 males with a mean age of 33.7 years (range: 15.0−69.9 years), an average height of 1.72 m (range: 1.50−1.93 m), and weight of 66.2 kg (range: 46.0−95.2 kg). All patients received an evaluation that included an initial force assessment on an instrumented treadmill where data were recorded, followed by a video analysis on a motorized treadmill in the clinic. In the clinic, patients were asked to rate any pain they experienced while running on a scale of 0−10. The average pain rating was 1.45 ± 1.7 on the right and 1.08 ± 1.98 on the left (mean ± SD).

2.2. Protocol

Patients first ran in their typical shod condition on a tandem force-sensing treadmill (AMTI, Watertown, MA, USA) at a self-selected comfortable pace, (range: 4.8−8.0 mph). This was immediately followed by a BF run at the same speed. During the BF run patients were made aware of the VGRF trace being displayed on a monitor and were given a few simple instructions. They were asked to make the vertical force trace “as smooth as possible”. If they did not automatically adopt an FFS pattern they were instructed to land as “softly as possible”, and to “land on the ball of the foot”. Using a Vicon motion capture system (Vicon Motion Systems Ltd., Oxford, UK), analog data were collected at 1200 Hz for approximately 15 s during both the shod and BF (after verbal instruction) running conditions.

2.3. Analysis

The GRF from both shod and BF running was used to determine vertical stiffness during initial loading (VILS), average and maximum instantaneous loading rate (VALR and VILR), peak medial and lateral forces (PMF, PLF), impulses in the vertical, medial and lateral directions (V-Imp, M-Imp, and L-Imp), step length (SL), and step rate (SR). All outcome variables were computed for eight steps on the right side and then averaged for each patient. The exception was SR, which is typically computed from both feet and was therefore determined for 16 steps: the same eight right steps and the corresponding left steps. Strike pattern was visually classified for each condition as either an FFS, MFS, or RFS by a single rater using video recorded at 100 Hz.

2.3.1. Impact transient definition

The vertical stiffness responsible for producing a given VGRF was used to distinguish between steps that had an impact transient and those that did not. Throughout this paper a step was defined as having a “transient” if there was a distinct, short-duration change in vertical stiffness during the loading phase of stance. This change in vertical stiffness corresponds to a short-duration change in the development of the VGRF, referred to as an “impact transient”. This is an important distinction from the typical definition of an impact transient that is often synonymous with an impact “peak”.

2.3.2. Vertical stiffness

Vertical stiffness was determined by fitting the VGRF with the spring-mass model of running,

$$\mathrm{VGRF}(t) = -k(t) \times x(t),$$

where t is time, $x(t)$ is the displacement of the center-of-mass,[20] and $k(t)$ is the vertical stiffness. Therefore, we fit VGRF from footstrike to peak VGRF with two models. The first was a simple model with constant stiffness k_c and the second a more complex model with non-constant stiffness that varied as a function of time.[13] Constant stiffness, k_c, was defined as peak VGRF divided by center-of-mass excursion.[12] The complex model fit VGRF in the least squares sense by estimating $k(t)$ with a 4-parameter logistic ogive function,[21]

$$k(t) = \frac{k_l - k_h}{1 + t/t_{T^m}} + k_l,$$

where k_h was a high stiffness during initial loading (IL) that transitioned at time, t_T, to a low stiffness value, k_l. The fourth parameter, m, analogous to slope, controlled the smoothness of the transition between k_h and k_l (Fig. 1B). Computations for the model fits were performed using custom code written in MATLAB (The MathWorks, Inc., Natick, MA, USA).

To determine which model, and hence which vertical stiffness described a given step, a comparison of the R^2 values was used. If the percent difference between R^2 values for the simple and complex models was less than 3.0%, the simple model was considered. This indicated the absence of an impact transient. Otherwise the more complex, dual stiffness model was deemed necessary (Fig. 1C). In this case, the step was classified as having an impact transient.

We focused our analysis on the IL and defined the vertical stiffness during IL (VILS). IL is defined as the time from contact to the impact transient, when it exists (Fig. 1A). This phase is of interest since it is used for computing loading rates, which are linked to a higher risk of certain running-related injuries. For the complex model, the stiffness during IL is equivalent to k_h. For the simple model, stiffness is constant throughout stance, therefore VILS is equivalent to k_c.

2.3.3. Loading rates

The loading rates were computed differently depending on the model used. When the complex, dual stiffness model was used, the point of interest (POI) from which to compute the loading rates was chosen as the VIP, when one existed. If there was no VIP, but the complex model was used, the transition time t_T was used as the POI. For the simple model, the POI was taken as 13% of stance since this has been reported to be the average location of the VIP when one is present[22] (Fig. 1C). The VALR was computed as the average slope from 20% to 80% of the VGRF at the POI[23] (Fig. 1A). The instantaneous loading rate (VILR) was maximum slope computed between each frame of the VGRF from contact until the POI.

2.3.4. Peak forces and impulses

Peak GRF in the medial and lateral directions were determined from the entire stance phase and reported in newtons (N). Impulses were computed as the area enclosed by the zero line and the ground reaction curve for each direction of interest in Ns. Lateral was defined as positive, with medial being negative.

2.4. Statistical analysis

Each step was classified using the model fit technique and outcome measures were computed on a step-by-step basis. Outcomes were averaged across eight steps within each subject and condition. However, we also tracked which model was fit to individual steps and whether a VIP was present. A paired t test was performed between the shod and the BF conditions after instruction for each outcome variable. A p value of less than 0.05 was used to indicate a statistical difference. Vertical stiffness was compared during IL only, and not to the peak VGRF. Pearson correlation coefficients were computed to establish the relationship between change in VILS and changes in the VALR and VILR. All computations were performed in MATLAB.

3. Results

Forty-five of 49 (92%) patients employed an RFS during their typical shod run, while two (4%) ran with an MFS and two (4%) ran with an FFS pattern. During the BF run, after instruction, 47 of 49 (96%) patients employed an FFS pattern, while only one (2%) used an MFS and one (2%) an RFS pattern.

3.1. Impact transients

An impact transient was identified in 384 steps (98% of steps) during the shod condition and only 99 (25% of steps) during the instructed BF condition. Of these, 34 shod and 58 BF steps did not have a VIP (Table 1). Eighty-six percent ($n = 42$) had an impact transient in all eight steps, while only 12% ($n = 6$) had an impact in seven of eight steps and 2% ($n = 1$) had an impact in six of eight steps. When running BF, with instruction, 47% of patients had no impact transient in any of the eight steps ($n = 23$), and only 14% ($n = 7$) had an impact transient in five or more steps.

3.2. Loading rates

Both instantaneous and average loading rates were significantly reduced ($p < 0.0001$) during the instructed BF run compared to the shod condition (Table 2). VALR during shod running ranged between 34.9 and 138.3 BW/s. All participants with the exception of one (>4 SD from the mean), reduced their VALR to between 15.4 and 36.8 BW/s during the instructed BF run (Fig. 2). On average, VILR and VALR were reduced by 51% and 57% respectively, from shod to instructed BF runs.

3.3. Vertical stiffness

Vertical stiffness during initial loading (VILS) was 33% lower for the instructed BF compared to the shod run (Table 2). Stiffness during initial loading was correlated to both VALR ($r = 0.726$, $p < 0.0001$) and VILR ($r = 0.658$, $p < 0.0001$). The proportion of patients with at least one step without an impact transient increased from 14% during the shod run to 92% during the BF run following instruction. When patients did have an impact transient in at least one step during BF running (53%), the average VILS was still reduced compared to the shod condition (Table 3).

The R^2 value describing the fit for the simple, constant stiffness model was higher on average in the BF condition (0.974 ± 0.024) compared to the shod (0.820 ± 0.121). The

Table 1
Impact transient prescience, time of occurrence, and frequency.

Condition	Impact transient?	Impact peak?	Point of interest	% Stance (mean ± SD)	Number of steps
Shod	Y	Y	Impact peak	12.5 ± 2.4	350
	Y	N	Transition time	16.9 ± 4.3	34
	N	N	13% of stance	N/A	8
Instructed barefoot	Y	Y	Impact peak	20.3 ± 6.8	41
	Y	N	Transition time	23.9 ± 5.7	58
	N	N	13% of stance	N/A	293

Table 2
Paired t test for comparison between shod and instructed BF conditions.

Variable	Mean ± SD		Paired t test		
	Shod	BF	Mean (|shod| − |BF|)[a]	p	95%CI
VALR (BW/s)	71.6 ± 26.6	26.7 ± 7.8	44.9	0.0000	36.9, 52.9
VILR (BW/s)	83.9 ± 29.0	36.5 ± 9.2	47.4	0.0000	38.4, 56.4
VILS (kN/m)	42.3 ± 11.2	27.6 ± 7.3	14.7	0.0000	11.9, 17.5
PLF (N)	69.5 ± 40.2	25.6 ± 19.2	44.0	0.0000	33.9, 54.1
PMF (N)	−77.4 ± 34.7	−64.1 ± 23.0	13.3	0.0014	−21.2, 5.4
V-Imp (Ns)	247.9 ± 45.8	231.3 ± 41.8	16.6	0.0000	12.9, 20.3
L-Imp (Ns)	2.10 ± 1.66	1.15 ± 1.42	0.95	0.0000	0.69, 1.21
M-Imp (Ns)	−6.02 ± 2.92	−5.61 ± 2.73	0.41	0.0474	−0.82, 0.00
SL (cm)	97.5 ± 11.7	92.1 ± 11.3	5.4	0.0000	4.1, 6.7
SR (steps/min)	163.2 ± 8.5	172.9 ± 10.8	−9.8	0.0000	−12.1, 7.4

Abbreviations: BF = barefoot; VALR = average loading rate of the vertical ground reaction force; VILR = maximum instantaneous loading rate of the vertical ground reaction force; VILS = stiffness during the initial loading phase; PLF = peak lateral ground reaction force; PMF = peak medial ground reaction force; V-Imp = impulse of the vertical ground reaction force during stance; L-Imp = impulse of the lateral ground reaction force during stance; M-Imp = impulse of the medial ground reaction force during stance; SL = step length; SR = step rate (cadence); CI = confidence interval.
[a] Note: For positive values shod was larger in magnitude than BF; for negative values the BF was larger in magnitude than Shod, on average.

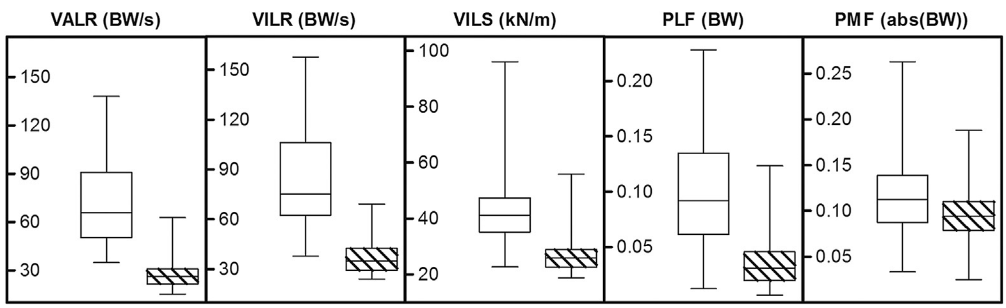

Fig. 2. Boxplots indicating differences between the shod condition and the instructed barefoot condition (hashed) for the average vertical loading rate (VALR), the maximum instantaneous loading rate (VILR), the stiffness during initial loading (VILS), the peak lateral ground reaction force (PLF), and the peak medial ground reaction force (PMF).

variable stiffness model provided a better fit than the simple model for the shod condition with an R^2 of 0.987 ± 0.008.

3.4. Peak GRF and impulses

Instructed BF running resulted in a significant decrease of 53% ($p < 0.0001$) in peak lateral GRF and 9.7% ($p = 0.0014$) in peak medial GRF (Table 2). Similar trends were seen with a large, statistically significant decrease in lateral impulse (44%, $p < 0.0001$) and a less drastic decrease in medial impulse (2.4%, $p = 0.0474$). Vertical impulse was also significantly decreased between shod and instructed BF running (6.6%, $p < 0.0001$).

3.5. Temporal spatial parameters

SR increased by an average of 10 steps/min (6.0%) during BF running compared to shod. This was complemented by a 5.5% decrease in step length for the BF condition. However, as step rate increased there was a decrease in vertical impulse ($r = -0.881$, $p < 0.0001$) and low stiffness, k_l ($r = 0.797$, $p < 0.0001$).

4. Discussion

We hypothesized that habitually shod runners who demonstrated an impact transient would reduce loading and vertical stiffness when running BF following verbal and visual feedback to encourage an FFS pattern. Previous studies demonstrated an increase in loading rates when BF runners persisted with an RFS pattern.[3,19,24,25] The important distinction in this study was that we did not allow runners to adapt to the BF condition independently. All eight variables of interest (VILS, VALR, VILR, PMF, PLF, V-Imp, M-Imp, L-Imp) were significantly reduced in this cohort of injured runners during the instructed BF run compared to the shod.

Table 3
Proportion of steps described with each model and the average of the corresponding vertical stiffness (mean ± SD).

Condition	Simple model			Complex model			
	k_c (kN/m)	n	% of steps	k_h	k_l	n	% of steps
Shod	25.40 ± 6.40	7	14.3 ± 4.7	42.49 ± 11.10	23.24 ± 3.90	49	98.0 ± 5.3
Barefoot	24.69 ± 3.90	45	81.4 ± 23.8	33.86 ± 8.20	25.09 ± 4.70	26	47.6 ± 29.4

Notes: n, number of patients who had at least one step described with the given stiffness; % of steps, average, across all patients who had at least one step with this model, of the percentage of steps described with the given stiffness.

4.1. Impact transients

The majority of patients who had impact transients during the shod run (384 of 392 steps) were able to eliminate or reduce the number of these during the instructed BF run (99 of 392 steps; Table 1). This is likely related to the fact that 96% of patients were able to convert to an FFS pattern while running BF given feedback and verbal instruction. These results are significant since previous studies suggest that novice BF runners may fail to adopt an FFS pattern independently.[3,16]

We introduced a robust method to classify the presence or absence of an impact transient. This differs from many studies which use the presence of an impact "peak" to signify that an impact transient exists.[16] The VIP is a local maximum that occurs prior to the overall peak, defined by de Wit as "the first vertical impact force peak". The presence or absence of this VIP influences the manner in which other variables, such as the loading rate and stiffness are computed. In steps where no impact "peak" was detected, our model determined that a high and low stiffness was required to adequately approximate VGRF (Table 1), and thus an impact "transient" existed. Despite lacking an impact "peak", these curves often exhibited higher stiffness during initial loading and higher loading rates of VGRF than steps fit with the simple model. Had we only searched for local maxima, or impact peaks, we would have underestimated the number of steps with an impact transient by 34 and 58 steps for the shod and instructed BF conditions, respectively. This would have resulted in an underestimate of VILS for these steps by 26% in the shod condition (25.0 *vs.* 38.2 kN/m) and 35% in the BF condition (25.7 *vs.* 34.7 kN/m). We recommend that the broader definition of "impact transient" defined in this study be adopted in future studies. While the term "impact peak" can still refer to a transient that exhibits a local maximum, a local maximum is not a necessary condition for existence of an impact transient.

4.2. Loading rates

The current study demonstrates that novice BF runners are successful at immediately reducing loading parameters when they are provided with instruction and real-time feedback on transitioning to an FFS pattern. We reported a large decrease in VALR and VILR during the BF run compared to the shod. This is contrary to other studies that reported increases in loading rates in novice BF[16] and minimalist[26] runners without instruction; suggesting that instruction may be playing a role. Kinematics indicate that these participants often continued to RFS while running BF or in minimal shoes. These studies, combined with additional literature,[3,27] demonstrate that not all runners convert to an FFS pattern when transitioning to BF running. Studies report that BF runners who RFS, have significantly higher loading rates than forefoot strikers and shod runners who RFS.[3,24,25] Increased loading rates of the VGRF have been linked to a number of common running-related injuries.[6–8] Additionally, it has been shown that RFS runners are 2.5 times as likely to had a retrospective, repetitive stress injury than FFS runners.[28]

The use of feedback and instruction likely encouraged an FFS pattern as runners transitioned to BF running. Evaluation of high-speed video indicated that 96% of novice BF runners were able to adopt an FFS pattern. This altered strike pattern likely contributed to decreased loading rates during the BF condition. Literature reporting loading rates between BF runners who employ an FFS pattern and shod runners is extremely limited.[25] Lieberman et al.[3] reported no overall difference in loading rates between BF FFS and shod runners. This is in contrast to the current study and a recent study by Shih et al.[24] where habitually shod RFS runners were asked to run BF and shod with both an FFS and RFS pattern. Confidence intervals on each condition indicated that loading rates were significantly reduced for an FFS pattern while BF and shod compared to an RFS pattern while shod. Importantly, participants in this study were told to use a specific strike pattern and given a brief time to practice an FFS before data were collected.

Considered collectively, the results of these recent studies imply that not all FFS patterns are equivalent. Not all BF runners who make initial contact with the ground on the front of the foot may necessarily be successful at significantly reducing loading rates. The marked decrease in loading rates observed in this study may be partly attributed to the use of real-time feedback. This allowed patients to focus not only on contacting the ground with the forefoot, but also to strive for a "soft" landing eliminating the impact transient in the VGRF. The fact that a runner can make an immediate alteration in their footstrike pattern does not necessarily mean that this change is permanent. Adopting an FFS pattern places greater demands on the calf musculature.[29] Thus, while patients were able to convert to an FFS pattern and significantly reduce their loading rates, additional training and conditioning of the lower leg and foot would be needed to maintain this pattern for longer duration runs. In addition, the increased load on the calf musculature, especially if it occurs suddenly, could put the runner at risk for muscle soreness or foot and ankle injuries resulting from overuse. This further supports the need for proper training, strengthening, and conditioning for a proper transition to an FFS pattern.

4.3. Vertical stiffness

Results of this study suggest that a runner who contacts the ground with less compliance, or a higher vertical stiffness during IL, will exhibit a faster rise in VGRF, increasing the loading rate. We found strong correlations between the change in VILS and the changes in VALR and VILR. Both VALR and VILR, as well as VILS significantly decreased from shod running to instructed BF running (Table 2). This suggests that runners decreased their vertical stiffness in order to eliminate impact transients and reduce loading rates.

The relationship between stiffness and loading rate is not always apparent in the literature due to the technique used to compute vertical stiffness. Most studies use a constant stiffness, which does not model the transient of the VGRF, thus ignoring the high stiffness during IL. For example, Shih et al.[24] reported a significant increase in loading rates between an FFS and an RFS pattern that had an impact transient early in stance. However stiffness was similar between groups. Similarly, Divert et al.[30] found no difference in vertical stiffness between shod and BF runners despite reporting that only three of 12 BF runners demonstrated an impact peak. This is because, in both studies, stiffness was assessed as an average value across the entire loading phase. The constant stiffness model misrepresents the actual vertical stiffness in cases where an impact transient exists. However, the method introduced by Hunter[13] and employed in this study is an important tool that provides a more accurate computation of stiffness, particularly during initial loading.

In the current study, we reported a significant reduction in VILS between the shod and BF conditions (mean $(|\text{shod}| - |\text{BF}|) = 14.7 \pm 9.8$ kN/m, $p < 0.00001$). However, had the simple constant stiffness model been applied to all steps, ignoring the impact transient, the results would have actually indicated the opposite. We would have seen an *increase* in VILS from the shod to instructed BF condition (shod: 23.2 ± 3.9 kN/m, BF: 24.9 ± 4.1 kN/m; mean$(|\text{shod}| - |\text{BF}|) = -1.7 \pm 2.2$ kN/m; $p < 0.00001$). This increase in VILS during the BF condition would greatly misrepresent what is actually occurring. In fact, the results showed that stiffness was greatly reduced during the BF condition (Table 3).

4.4. Peak GRF and impulses

The limited research investigating medial and lateral GRFs during BF running suggests that there is no change in peak medial and lateral GRF[31,32] or impulses[32,33] between shod and BF runners. These results were not specific to BF runners who employ an FFS pattern. The current study revealed a significant decrease in medial and lateral impulses and peak GRF in the instructed BF condition compared to shod. Although significant changes occurred in both directions, the largest of these were apparent in the lateral peak and impulse ($p < 0.0001$). Significant decreases were also seen in the V-Imp ($p < 0.0001$).

The exact mechanism for the large decrease in lateral loading is unclear, but is likely due to many interacting factors. Having patients land softly to reduce impact in the VGRF may have translated to similar reductions in the lateral direction. In early stance, there is typically a lateral GRF transient (Fig. 3), which tended to coincide roughly with the VIP. It occurred within ±5% of stance of the VIP in 73% of runners during the shod condition. Both the lateral and vertical GRF contribute to the external pronation moment that the foot must control. Therefore, reducing both the initial vertical and lateral GRF may reduce the pronatory moment. This may explain, in part, the reduction in pronation during BF running noted by Bonacci et al.[17] More research is required exploring foot kinematics in conjunction with GRF to gain a better understanding of the mechanisms behind this reduction in mediolateral loading.

We reported an increase in step rate between shod and instructed BF running for the same speed, therefore it is likely that there was a decrease in stance time during the BF condition. Since impulse is a measure of the cumulative force over the stance phase, it is not surprising that there would be a resulting decrease in vertical, medial and lateral impulses for the instructed BF run. This would result in a decrease in cumulative load for each step and an increase the number of cycles over a given distance. Reduction of step length, despite the increase in loading cycles, has been shown to decrease the risk of stress fractures.[34] Additionally, reducing step length results in a significant reduction in hip and knee loads as well as significant reductions in hip adduction.[18] Hip adduction has been related to a number of running-related injuries including stress fractures, iliotibial band syndrome,[35,36] and patellofemoral pain syndrome.[37] Together, these studies suggest that despite the increase in loading cycles, running with a shorter stride length likely reduces injury risk.

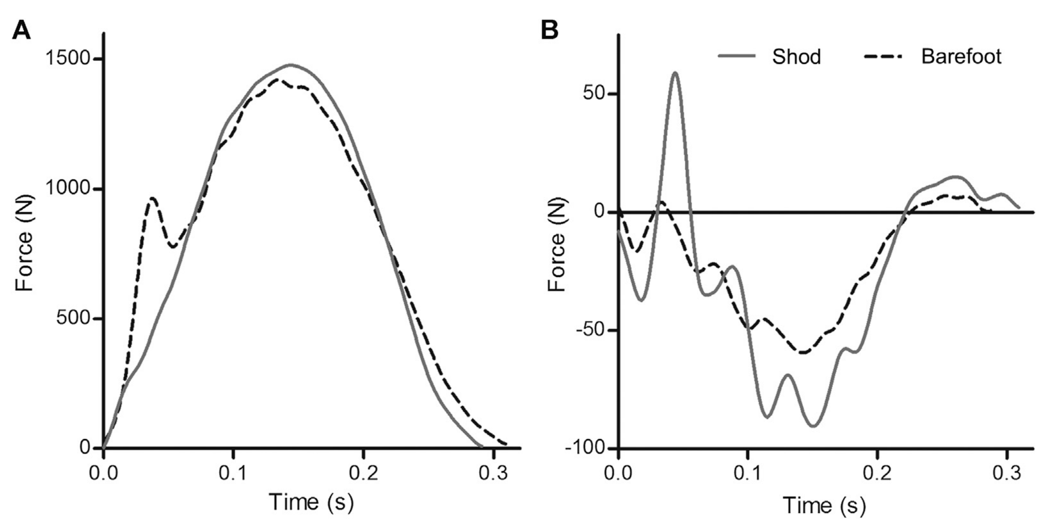

Fig. 3. Changes in the vertical (A) and mediolateral (B) ground reaction forces between shod rarefoot strike and instructed barefoot forefoot strike running. Note the elimination of the vertical impact transient and the significant reduction in the early and rapid lateral peak in the barefoot compared to the shod condition.

4.5. Temporal spatial parameters

Consistent with other studies,[16,17] we reported an increase in SR and decrease in SL from typical shod to instructed BF running. Increased SR and reduced step length has also been shown to reduce kinetic outcomes such as hip moments and energy absorption at the hip and knee,[18] as well as VILR and VALR.[38]

4.6. Limitations

This study was limited in a number of ways. Participants were all attending a clinic for treatment of a lower body injury. Therefore, some may not have exhibited a typical running pattern. Only participants who were comfortable running the same pace BF as that selected while shod were included in the dataset. While this was a necessary criterion due to the relationship between speed and loading parameters, it may have inflated the improvements measured between conditions. Additionally, multiple factors were altered from the initial condition to the BF run. Participants removed their shoes, converted to an FFS pattern, and were provided with feedback and instruction on how to achieve this. Without a control group it impossible to distinguish the influence of these individual factors on the outcomes of the study.

5. Conclusion

Based upon the results of this study, patients with lower extremity running-related injuries were able to significantly reduce their impact loading, as well as peak forces and impulses during a brief bout of instructed BF FFS running. As impact loading has been associated with some of the most common running-related injuries, this instruction may help to reduce the risk of these injuries in individuals transitioning to BF running. Subsequent research is required to directly evaluate the relationship between strike pattern and injury as well as further explore the impact of instruction.

References

1. Rothschild CE. Primitive running: a survey analysis of runners' interest, participation, and implementation. *J Strength Cond Res Natl Strength Cond Assoc* 2012;**26**:2021–6.
2. Brüggemann G-P, Potthast W, Niehoff A, Braunstein B, Assheuer J. Adaptation of morphology and function of the intrinsic foot and shank muscles to mechanical loading induced through footwear. In: Subic A, Ujihashi S, editors. *The impact of technology on sport*. Melbourne: Australasian Sports Technology Alliance Pty Ltd.; 2005.
3. Lieberman DE, Venkadesan M, Werbel WA, Daoud AI, D'Andrea S, Davis IS, et al. Foot strike patterns and collision forces in habitually barefoot versus shod runners. *Nature* 2010;**463**:531–5.
4. Hasegawa H, Yamauchi T, Kraemer WJ. Foot strike patterns of runners at the 15-km point during an elite-level half marathon. *J Strength Cond Res Natl Strength Cond Assoc* 2007;**21**:888–93.
5. Larson P, Higgins E, Kaminski J, Decker T, Preble J, Lyons D, et al. Foot strike patterns of recreational and sub-elite runners in a long-distance road race. *J Sports Sci* 2011;**29**:1665–73.
6. 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**:323–8.
7. Davis IS, Bowser BJ, Hamill J. Vertical impact loading in runners with a history of patellofemoral pain syndrome. *Med Sci Sports Exerc* 2010;**42**(Suppl. 15):682.
8. Pohl MB, Hamill J, Davis IS. Biomechanical and anatomic factors associated with a history of plantar fasciitis in female runners. *Clin J Sport Med Off J Can Acad Sport Med* 2009;**19**:372–6.
9. Lohman 3rd EB, Balan Sackiriyas KS, Swen RW. A comparison of the spatiotemporal parameters, kinematics, and biomechanics between shod, unshod, and minimally supported running as compared to walking. *Phys Ther Sport Off J Assoc Chart Physiother Sports Med* 2011;**12**:151–63.
10. Nilsson J, Thorstensson A. Ground reaction forces at different speeds of human walking and running. *Acta Physiol Scand* 1989;**136**:217–27.
11. Blickhan R. The spring-mass model for running and hopping. *J Biomech* 1989;**22**:1217–27.
12. McMahon TA, Cheng GC. The mechanics of running: how does stiffness couple with speed? *J Biomech* 1990;**23**(Suppl. 1):65–78.
13. Hunter I. A new approach to modeling vertical stiffness in heel-toe distance runners. *J Sports Sci Med* 2003;**2**:139–43.
14. Rodrigues P, TenBroek T, Hamill J. Runners with anterior knee pain use a greater percentage of their available pronation range of motion. *J Appl Biomech* 2013;**29**:141–6.
15. Willems TM, De Clercq D, Delbaere K, Vanderstraeten G, De Cock A, Witvrouw E. A prospective study of gait related risk factors for exercise-related lower leg pain. *Gait Posture* 2006;**23**:91–8.
16. De Wit B, De Clercq D, Aerts P. Biomechanical analysis of the stance phase during barefoot and shod running. *J Biomech* 2000;**33**:269–78.
17. Bonacci J, Saunders PU, Hicks A, Rantalainen T, Vicenzino BT, Spratford W. Running in a minimalist and lightweight shoe is not the same as running barefoot: a biomechanical study. *Br J Sports Med* 2013;**47**:387–92.
18. Heiderscheit BC, Chumanov ES, Michalski MP, Wille CM, Ryan MB. Effects of step rate manipulation on joint mechanics during running. *Med Sci Sports Exerc* 2011;**43**:296–302.
19. Cheung RTH, Rainbow MJ. Landing pattern and vertical loading rates during first attempt of barefoot running in habitual shod runners. *Hum Mov Sci* 2014;**34**:120–7.
20. Cavagna GA. Force platforms as ergometers. *J Appl Physiol* 1975;**39**:174–9.
21. Ratkowsky DA, Reedy TJ. Choosing near-linear parameters in the four-parameter logistic model for radioligand and related assays. *Biometrics* 1986;**42**:575–82.
22. Willy RW, Pohl MB, Davis IS. Calculation of vertical load rates in the absence of vertical impact peaks. In: *Proceedings of the 32nd American Society of Biomechanics*; 2008. Ann Arbour, MI, USA.
23. Lafortune MA, Hennig EM. Cushioning properties of footwear during walking: accelerometer and force platform measurements. *Clin Biomech* 1992;**7**:181–4.
24. Shih Y, Lin KL, Shiang TY. Is the foot striking pattern more important than barefoot or shod conditions in running? *Gait Posture* 2013;**38**:490–4.
25. Hall JPL, Barton C, Jones PR, Morrissey D. The biomechanical differences between barefoot and shod distance running: a systematic review and preliminary meta-analysis. *Sports Med Auckl NZ* 2013;**43**:1335–53.
26. Willy RW, Davis IS. Kinematic and kinetic comparison of running in standard and minimalist shoes. *Med Sci Sports Exerc* 2014;**46**:318–23.
27. Gruber AH, Silvernail JF, Brueggemann P, Rohr E, Hamill J. Footfall patterns during barefoot running on harder and softer surfaces. *Footwear Sci* 2013;**5**:39–44.
28. Daoud AI, Geissler GJ, Wang F, Saretsky J, Daoud YA, Lieberman DE. Foot strike and injury rates in endurance runners: a retrospective study. *Med Sci Sports Exerc* 2012;**44**:1325–34.
29. Williams 3rd DSB, Green DH, Wurzinger B. Changes in lower extremity movement and power absorption during forefoot striking and barefoot running. *Int J Sports Phys Ther* 2012;**7**:525–32.

30. Divert C, Mornieux G, Freychat P, Baly L, Mayer F, Belli A. Barefoot-shod running differences: shoe or mass effect? *Int J Sports Med* 2008;**29**:512−8.
31. Braunstein B, Arampatzis A, Eysel P, Brüggemann G-P. Footwear affects the gearing at the ankle and knee joints during running. *J Biomech* 2010;**43**:2120−5.
32. Morley JB, Decker LM, Dierks T, Blanke D, French JA, Stergiou N. Effects of varying amounts of pronation on the mediolateral ground reaction forces during barefoot versus shod running. *J Appl Biomech* 2010;**26**:205−14.
33. Divert C, Mornieux G, Baur H, Mayer F, Belli A. Mechanical comparison of barefoot and shod running. *Int J Sports Med* 2005;**26**:593−8.
34. 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**:2177−84.
35. Noehren B, Davis I, Hamill J. ASB clinical biomechanics award winner 2006 prospective study of the biomechanical factors associated with iliotibial band syndrome. *Clin Biomech Bristol Avon* 2007;**22**:951−6.
36. Ferber R, Noehren B, Hamill J, Davis IS. Competitive female runners with a history of iliotibial band syndrome demonstrate atypical hip and knee kinematics. *J Orthop Sports Phys Ther* 2010;**40**:52−8.
37. Noehren B, Hamill J, Davis I. Prospective evidence for a hip etiology in patellofemoral pain. *Med Sci Sports Exerc* 2013;**45**:1120−4.
38. Hobara H, Sato T, Sakaguchi M, Sato T, Nakazawa K. Step frequency and lower extremity loading during running. *Int J Sports Med* 2012;**33**:310−3.