FLEX CEUs

Barefoot Running Considerations

Variation in Foot Strike Patterns among Habitually Barefoot and Shod Runners in Kenya

Abstract

Runners are often categorized as forefoot, midfoot or rearfoot strikers, but how much and why do individuals vary in foot strike patterns when running on level terrain? This study used general linear mixed-effects models to explore both intra- and inter-individual varia-tions in foot strike pattern among 48 Kalenjin-speaking participants from Kenya who varied in age, sex, body mass, height, running history, and habitual use of footwear. High speed video was used to measure lower extremity kinematics at ground contact in the sagittal plane while participants ran down 13 meter-long tracks with three variables independently controlled: speed, track stiffness, and step frequency. 72% of the habitually barefoot and 32% of the habitually shod participants used multiple strike types, with significantly higher levels of foot strike variation among individuals who ran less frequently and who used lower step frequencies. There was no effect of sex, age, height or weight on foot strike angle, but individuals were more likely to midfoot or forefoot strike when they ran on a stiff surface, had a high preferred stride frequency, were habitually barefoot, and had more experience run-ning. It is hypothesized that strike type variation during running, including a more frequent use of forefoot and midfoot strikes, used to be greater before the introduction of cushioned shoes and paved surfaces.

Introduction

Runners are commonly categorized according to strike type (also known as footfall pattern), and it is widely observed that more than 85% of habitually shod runners typically rearfoot strike (RFS), in which the heel is the first part of the foot to contact the ground [1,2]. In con-trast, some runners (many of them elite athletes) have been observed to forefoot strike (FFS), in which the ball of the foot lands before the heel, or to midfoot strike (MFS), in which the heel and ball of the foot land almost simultaneously [3]. In addition, numerous studies have found that barefoot and minimally shod runners are more likely than habitually shod runners to MFS or FFS [4–13]. However, some habitually barefoot individuals have been observed to primarily RFS when they run [14], and people in minimal shoes are more likely to run with a RFS than those who are barefoot [15].

Differences in strike patterns have led to numerous hypotheses about their relative costs and benefits. Although FFS and RFS landings do not differ in terms of economy [16–19], FFS and some MFS landings differ from RFS landings in generating no discernible impact peak in the vertical ground reaction force just after contact. Whether the rate of loading and magnitude of impact peaks contribute to repetitive stress injuries is debated [20–23], but impact peaks can be uncomfortable, often causing barefoot runners to avoid RFS landings on hard surfaces without a cushioned shoe [7–10, 20–23].

Regardless of the advantages and disadvantages of FFS, MFS and RFS landings, one issue that has been insufficiently considered is variation, both within and between individuals. How much do runners vary their strike patterns, and what causes this variation? Although runners tend to be characterized as either rearfoot, midfoot or forefoot strikers, it is likely that most use all three kinds of strikes but in different proportions and contexts. All people FFS when running up a steep incline, and the tendency to RFS is often greater when descending [24,25]. In addition, runners are more likely to MFS or FFS as they increase speed [26]. Additional factors that may affect strike type include training and skill, fatigue, the presence of shoes, shoe design, and substrate characteristics such as stiffness, slipperiness, unevenness and roughness. For example, habitually shod people who normally RFS typically switch to a FFS when asked to run barefoot on hard surfaces such as asphalt, but often continue to RFS when running barefoot on less stiff surfaces such as grass or cushioned mats [7–9]. Evidence that minimally shod individuals are more than twice as likely to RFS as barefoot individuals [7–9,15] suggests that sensory feedback from the foot strongly influences strike type.

The goal of this study therefore was to explore how much runners vary strike type on level surfaces, and to test some of the factors that may contribute to this variation. Conceptually, the factors that influence strike type variation can be classified into three non-mutually exclusive categories: *intrinsic*, *extrinsic* and *acquired*. Intrinsic factors relevant to strike type are characteristics of the runner that are not under control such as height, sex, age, and body mass. The dominant extrinsic factors relevant to strike type are characteristics of a runner's environment that potentially affect kinematics such as the nature of the substrate (e.g., surface stiffness, slope, unevenness, slipperiness) as well as footwear characteristics such as heel cushioning that may affect how the body interacts with the ground. Speed can also be an extrinsic factor depending on circumstance (e.g., when a runner is required to run faster or slower, as in this experiment). Finally, acquired factors are characteristics that a runner develops or learns. Some acquired characteristics, such as running history, footwear history, physical fitness, strength, and existing injuries, are often a product of an individual's background. Others such as step frequency may be modifiable characteristics—skills—that are acquired through cultural processes such as coaching, imitation, practice, and experimentation [27].

Using this conceptual framework, we tested two general hypotheses about intra- and inter-individual strike type variation among a diverse sample of individuals who varied in several intrinsic, extrinsic and acquired characteristics, and for whom we experimentally modified several extrinsic and acquired variables including surface stiffness, speed, and step frequency. The first general hypothesis (H1) is that extrinsic, intrinsic, and acquired factors influence the degree of intra-individual variation in strike type. Specifically, because shoes slow the rate of impact loading, limit exteroreception, and mitigate the effects of substrate variations on the foot and the rest of the body, H1a predicts that individuals who are barefoot will use more varied foot strike patterns than individuals who are shod. In addition, because speed and surface stiffness may affect aspects of kinematics and kinetics relevant to strike [7–9, 26], H1b predicts that runners are likely to use more varied strike patterns on soft surfaces and at slower speeds.

The second general hypothesis (H2) is that a combination of extrinsic, intrinsic, and acquired factors are predictive of foot strike angle as well as strike type both within and between individuals. Specifically, we predict that runners are more likely to FFS as they increase speed, increase step frequency, and run on harder surfaces. In addition, because impact peaks can cause discomfort and might contribute to overuse injuries, especially in unshod individuals, we predict that runners who are habitually barefoot and run more regularly over longer distances are more likely to FFS independent of other intrinsic factors such as sex, age, and body shape and size variation.

Materials and Methods

Study Design

Although kinematic variables such as foot strike are often compared between groups that differ in terms of footwear use (e.g., [4–8,11,12,15]), the hypotheses this study tests require a combined within- and between- subjects experimental design. In particular, we asked subjects who varied in terms of the intrinsic and acquired factors described above to perform a set of trials that independently varied three external factors: speed, surface stiffness and step frequency. Although this study design requires repeated measurements, which can be accounted for statistically using General Linear Mixed Models [36], it avoids potential sampling problems, such as heterogeneity within and between groups as well as assignment bias.

Participants

Because this study explores both intra- and inter-individual variation, it is necessary to test the above hypotheses with an appropriate population that varies considerably in a range of extrinsic, intrinsic, and acquired factors including footwear use. Almost all people in developed nations are habitually shod, and although barefoot running has recently gained popularity in countries such as the US, few if any of these barefoot enthusiasts grew up unshod, and some may have consciously adopted a running form advocated by books or websites. At the other end of the continuum, most habitually barefoot populations do not practice much long distance running. For this reason, we chose to focus on Kalenjin-speaking communities from Western part of Kenya, an area of special relevance for the questions posed by this study because of the considerable variation in footwear usage and running habits within this population, which includes many of the world's best distance runners, most of whom grew up barefoot [27,28].

48 Kalenjin individuals (Table 1) were recruited from the region around Eldoret in the Uasin Gishu and Nandi Counties of Kenya. 38 participants (19 M, 19 F) were adolescents

Table 1. Levene's Test of unequal variance for nominal comparisons of foot strike angle (FSA).

Comparison	F-ratio	p-value
Sex (male vs female)	0.1349	0.7151
Footwear (bare vs shod)	2.5124	0.1197
Surface (hard vs soft)	6.1117	**0.0152**
Habitually barefoot	0.1062	0.7458
Habitually shod	0.081	0.7775

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between the ages of 13 and 17 from three schools. 19 students (10 male, 9 female) aged 13–17 attended a school in a rural part of the Nandi South District where almost all the students are primarily barefoot and very physically active [28]. The school is not directly accessible by road, and these students walk or run barefoot an average of 7.5 ± 3.0 km/day to travel to and from school [29]. A few of these students wear shoes a few hours a week when they attend church and other special occasions, but they are otherwise almost always barefoot (see below). We also recruited 9 female students aged 13–17 from a girl's secondary school in Kobujoi, Kenya, and 10 male students aged 14–16 from a boy's secondary school in Eldoret, Kenya. These students board at school and wear thick-soled leather shoes for most of the day, and either rubber sports shoes (plimsoles) or cushioned athletic shoes (trainers) during athletic activities. Finally, we recruited 10 habitually barefoot, male adults aged 23–60 from the Nandi South District, Kenya. These men walk long distances regularly, some still run several kilometers per week, and most of them ran long distances when they were younger.

Individuals who had current lower extremity injuries or evident illness were excluded. In order to avoid biased samples in terms of fitness, we asked the teachers at the three schools to select only students who were "average" in terms of sports ability, thus excluding participants who either exceptional or poor in athletics.

Ethics Statement

Approval for the human experimental study described in this paper was granted by the Harvard University Committee on the Use of Human Subjects (protocol F23121), and by the Moi University Medical Institutional Research And Ethics Committee (protocol 00695). As approved by the aforementioned committees, written informed consent for minors was provided by their teachers; informed consent was provided orally by adults who were unable to read and documented with their signature.

Anthropometrics and Background Information

Basic anthropometrics were collected from all participants including height, body mass, and leg length (from the greater trochanter to the base of the heel). An orthopedic doctor (POM) examined all participants for lower extremity injuries. All participants (some of whom were not literate) were asked how far they walk and run on average each day, their regular physical activities, and what kinds of footwear they use. All questions were asked on two different occasions, either in Kalenjin or Kiswahili; one of the questioners (MS) speaks Kalenjin, knows the region intimately, and was able to evaluate how far each participant had to walk or run every day. Answers were then averaged. Since footwear usage and running history could not be quantified precisely as continuous variables, answers to these questions were binned into four rank order categories. Footwear score categories were: 1, almost always shod (less than 10% outdoor activity spent barefoot or in minimal shoes); 2, usually shod (mostly wear shoes, but do sports either barefoot or in minimal shoes); 3, mixed (sometimes walk, run or do physical activity in normal shoes and sometimes barefoot or in minimal shoes); 4, mostly barefoot (more than 80% of walking, running and physical activity done either barefoot or in minimal shoes). Running history categories were: 1, little (run less than 5 km/week); 2, occasional (run 5–10 km a week on an occasional but non-regular basis; 3, moderate (run 5–10 km a week on a regular basis); high (run $>$10 km a week on a regular basis).

Experimental Trials

Participants were asked to wear whatever footwear they normally use (if applicable), and to wear shorts or skirts that could be rolled above the knee. In order to record 2-dimensional kinematics in lateral view, reflective tape markers were placed on the following locations on one side of the body: the greater trochanter, the center of the knee (in between the lateral femoral epicondyle and the lateral tibial plateau), the lateral malleolus, the lateral surface of the 5th metatarsal head, and the lateral tuber calcaneus. Participants were then photographed with a visual scale in lateral and frontal position with a numeric identification. All participants were then instructed to run around an open field at a "pace they would choose if running a long distance" for approximately 5 minutes, at which point step frequency was then measured using an adjustable metronome (Matrix, New Market, VA, USA) fitted with an earpiece. Preferred step frequency (PSF) was recorded only for step frequencies that did not deviate by more than 4 steps/minute over a minimum of 30 seconds.

After warm-up, each participant's kinematics was immediately recorded in lateral view on two adjacent tracks approximately 13–15 m in length. The "hard" track was the unaltered, grass-free, compact surface of a field, similar to the stiffness of a dirt road's surface, and typical of the surfaces on which the participants normally run when traveling or doing athletics. A "soft" track was excavated parallel to the hard track by digging down 10 cm with a pickaxe, tamping down the earth, and then raking the dirt to create a smooth, soft surface. Penetrometer measurements repeated on each track (AMS Corp, American Falls, ID) indicate that the average compression strength of the hard track (3.85 kg/cm^2±0.29 S.D.) was 5.5 times greater than the soft track (0.70 kg/cm^2±0.27 S.D.). The soft track was raked between each set of trials, and re-excavated regularly to maintain a similar compliant surface for all participants. Small flags were used to mark the borders of the two tracks. A high-speed video camera (Casio EX-ZR100) was positioned at 0.7 m height approximately 4 m lateral to the 10 m point on the track, providing an additional 3–5 m of track beyond the field of the camera. All sequences were recorded at 240 frames per second.

For each trial, participants were asked to run down the track while looking forward and without decelerating until they had passed a marker positioned approximately 3 meters beyond the camera's field of view. Participants were asked to run down both the hard and soft tracks at approximately 3.0 m/sec ("slow") and 4.0 m/sec ("fast") at several step frequencies: the previously determined preferred step frequency (PSF), and at 150, 170 and 190 steps/min. As a result, each participant ran a minimum of 16 conditions. Step frequency was controlled using a small, lightweight digital metronome either handheld or clipped onto clothing (Seiko DM50, SeikoUSA, Mahwah, NJ). For each trial, the participant was familiarized with the frequency and then asked to try to maintain that frequency for the entire length of the track. There was no landing target on the track in order to avoid having participants alter their gaits by either shortening or lengthening their steps as they passed the camera's field of view. If the marked foot did not land in front of the camera, the trial was repeated without explaining the reason for repetition until a minimum of two trials were recorded for each speed, step frequency, and track. Following these trials, we administered the mile run test to the adolescent participants from each school according to methods outlined by the FITNESSGRAM test [30]. To avoid influencing how participants ran, we asked no questions about running form before or after the trials, and neither the participants nor their teachers were informed of the experiment's objectives.

Kinematic Analysis

All video sequences were converted to stacks of TIFF files and analyzed using ImageJ, version 1.46r (http://imagej.nih.gov/ij). A visual scale was determined for each participant using the measured distance between the lateral malleolus and knee markers. Since running speed was not controlled precisely during the experiment, running speed for each trial was quantified by measuring the horizontal translation of the marker on the greater trochanter between two homologous points during a stride cycle (e.g., toe-off to toe-off, or foot strike to foot strike for the same foot) relative to time (calculated from the number of frames divided by frame rate).

Foot strike was measured using only high-speed sequences in which the marked foot landed in front of the camera permitting a clear view of the foot's lateral margin, which has been shown to yield high accuracy and reliability [31]. Foot strike angle (FSA) was quantified as a continuous variable by measuring the orientation of the calcaneus and 5th metatarsal head markers relative to horizontal at the first frame of contact minus the same angle measured at foot flat [7]. Since FSA is a continuous variable but foot strike itself is a nominal variable, strike types were also classified using the following criteria: FFS, angles above 0.3°; MFS, angles between 0.3° and -5.6°; RFS, angles below -5.6°. In order to avoid classifying RFS and FFS landings as MFS landings, these cutoff values are more conservative than those used by Altman and Davis [31]. The correlation between strike type and FSA was 0.95 ($p<0.0001$).

Step frequency was quantified as the number of frames between the foot strike used to measure strike type and the previous strike multiplied by the number of seconds sampled per frame times 60. In order to quantify variations in the position of the foot at landing caused by variations in stride length, two measures of foot position relative to the rest of the lower extremity (overstride): overstride relative to the knee was measured as the projected anteroposterior distance of the lateral malleolus relative to the center of the knee at foot strike; overstride relative to the hip was measured as the projected anteroposterior distance of the lateral malleolus relative to the greater trochanter at foot strike. Several sagittal plane angles were measured at the moment of foot strike and at midstance (determined as the temporal midpoint between foot strike and toe-off). Knee angle was measured as the angle between the lines from the knee to the greater trochanter and the knee to the lateral malleolus; ankle angle was measured as the angle between the lines from the knee and to the lateral malleolus and from the lateral malleolus to the lateral MTP joint. Since this angle is affected by heel height, ankle angle was corrected by the angle measured during standing. Although hip angle is often measured as the orientation of the line from the knee to the greater trochanter relative to trunk angle, hip angle was measured as the orientation of the line from the knee to the greater trochanter relative to earth horizontal thus avoiding the effects of variations in trunk angle; similarly, trunk angle was measured as the angle between the greater trochanter and the center of the neck relative to earth horizontal.

Because each participant ran different numbers of trials, all kinematic measurements were averaged for each individual and condition. Measurement reliability was quantified by taking the same set of angular measurements from one individual on five separate occasions [11]. The average standard deviation was 0.32° with a range of 0.18°-0.49°. In addition, a test-retest sensitivity analysis conducted by taking all measurements twice from the same trial, yielding a correlation coefficient of 0.927.

Statistical analyses

As described above, this study tests hypotheses about levels of variation in foot strike (H1) as well as the factors that influence this variation (H2). In order to test the effects of nominal and continuous variables on overall levels of variation, as predicted by H1, we used two different methods. First, we used Levene's Test to compare measured variance of the two foot strike variables, FSA and strike type, in relation to the three nominal variables studied: sex, footwear condition (barefoot versus shod), and surface stiffness (hard versus soft). To test if footwear condition affects foot strike variability on the two surfaces, Levene's Test was also used to compare foot strike variance on the hard versus soft trackways within the barefoot and shod participants. To account for repeated measures, these tests used the mean variance of each individual. A Chi-squared analysis was also used to test if the proportion of individuals who varied their strike type differed between the barefoot and shod participants. Second, to test if there is a relationship between levels of variation in foot strike and intrinsic, extrinsic and acquired factors that are continuously distributed, we used a bivariate General Linear Mixed Model (GLMM) to calculate the residuals of the regression between FSA and each predictor variable using a subject identifier as the random effect to account for the non-independent error generated by repeated measures on the same individuals [32]. We then used a second GLMM to regress the absolute value of these residuals against the relevant predictor variable. A slope (the coefficient of the GLMM) significantly different from zero indicates a significant increase or decrease in variation with respect to the predictor variable. Since GLMMs assume that variables are normally distributed and in comparable units, non-normally distributed variables (assessed using a Shapiro-Wilk test) were log-transformed, and then all variables were converted to Z-scores.

In order to test Hypothesis 2, we used multivariate GLMMs to model the effects of the intrinsic, extrinsic, and acquired variables on foot strike across treatments. In the first GLMM, the dependent variable was FSA was regressed against the fixed effects included several variables classified as extrinsic (substrate stiffness), intrinsic (age, sex, height, body mass), and acquired factors (footwear history and running history; preferred step frequency; and the speed at which the participants could run a mile, a proxy for overall physical fitness). The first GLMM took the following form:

$$\begin{aligned}\mathrm{FSA} = {} & \beta_1\mathrm{Surface} + \beta_2\mathrm{Age} + \beta_3\mathrm{Sex} + \beta_4\mathrm{Height} + \beta_5\mathrm{Body\ Mass} + \beta_6\mathrm{Footwear\ History} + \\ & \beta_7\mathrm{Running\ History} + \beta_8\mathrm{Preferred\ Step\ Frequency} + \beta_9\mathrm{Mile\ Time} + ZU + \epsilon\end{aligned}$$

A second GLMM (S1 Table) was also calculated to test the effects of kinematics on foot strike. In this GLMM, the dependent variable was FSA and the fixed effects were aspects of kinematics (speed, step frequency, trunk angle, hip angle, knee angle, and overstride relative to the knee). This took the following form:

$$\begin{aligned}\mathrm{FSA} = {} & \beta_1\mathrm{Speed} + \beta_2\mathrm{Step\ Frequency} + \beta_3\mathrm{Trunk\ Angle} + \beta_4\mathrm{Hip\ Angle} + \beta_5\mathrm{Knee\ Angle} + \\ & \beta_6\mathrm{Ankle\ Angle} + \beta_7\mathrm{Overstride} + ZU + \epsilon\end{aligned}$$

In both models, β_i is the fixed-effect coefficient for the *i*th predictor, Z is the design matrix for the random grouping variable, U is a vector of random effects, and ϵ is residual model error. A subject identifier was used as the random grouping effect to account for repeated measures on the same individuals. In addition, all variables were transformed to Z-scores, and non-normally distributed variables (assessed using a Shapiro-Wilk test) were log-transformed.

A few of the variables (including the primary outcome measurement, FSA) were not normally distributed after transformation to Z-scores, and some variables are highly collinear (e.g., height and body mass, speed and step frequency). Therefore, to account for non-normality and isolate the potential effects of multicollinearity on significance testing, we also used a non-parametric residual randomization method to calculate p-values in the GLMMs. Residual randomization is a type of permutation test in which statistical significance is tested by permuting the residuals of a model rather than the observations [33]. In this study, residual randomization is used to evaluate the significance of each variable's effect independently, removing partial effects of other collinear variables (see S1 File).

Results

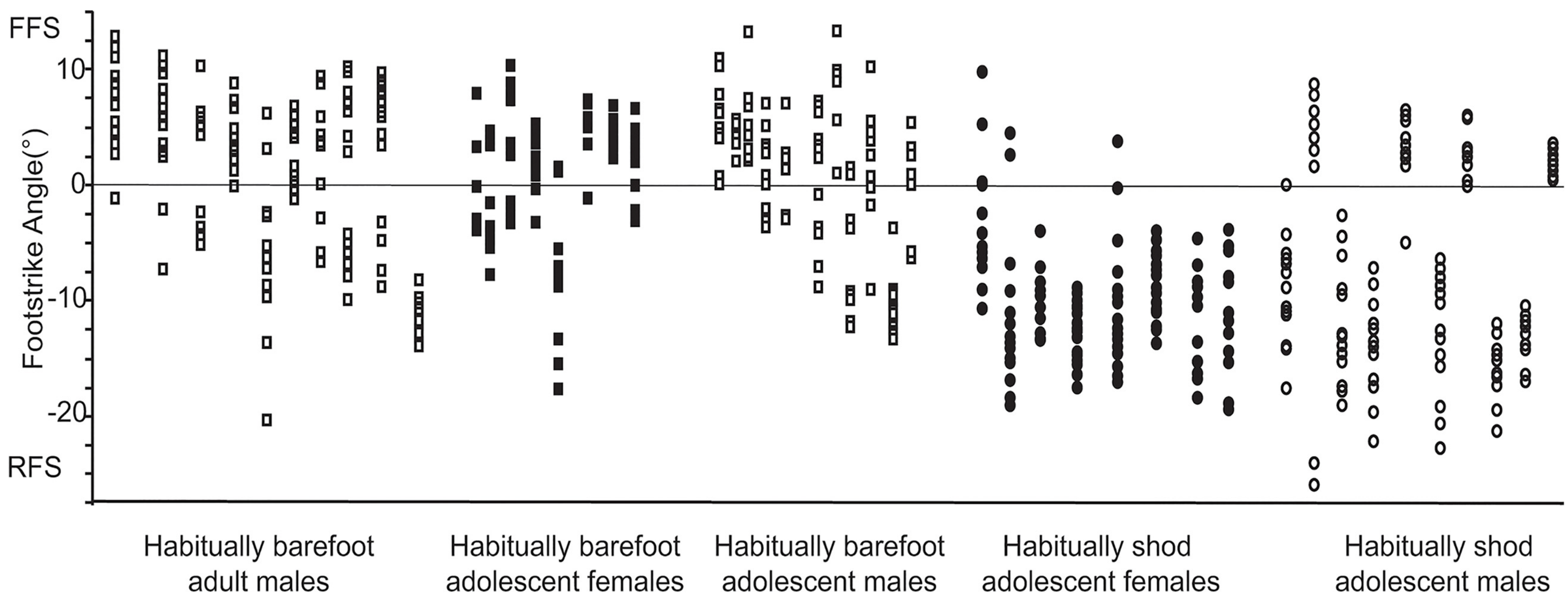
Because this study aimed to sample a wide range of variation both within and between different groups, we begin with a summary of the variation sampled. In terms of external factors, average height was 160.9 ±9.5 cm (range 142–177) in the barefoot population and 162.3±8.1 cm (range 147–174) in the shod population ($p = 0.517$); average body mass was 47.0±8.4 kg (range 32–62) in the barefoot population and 54.0±8.1 kg (range 37–65) in the shod population ($p = 0.038$); average age was 24.8±15.1 years (range 13–37) in the barefoot population, and 15.4 ±1.07 years (range 13–18) in the habitually shod population. In terms of acquired variables measured, the average footwear score in the barefoot population was 3.34±1.2 (range 2–4), higher ($p<0.001$) than the average shod population score of 1.20±0.4 (range 1–2); the average running history score in the barefoot population was 2.86±1.3 (range 1–4), also higher ($p<0.001$) than the average shod population score of 1.37±0.6 (range 1–3); however, in both populations there were individuals who ran more than 10 km per week and those who ran infrequently (<5 km/week). The average mile time of the barefoot individuals was 1:38 faster than the shod individuals ($p = 0.003$), with ranges of 5:19–7:42 and 5:29–12:20, respectively. Preferred step frequency averaged 172.6±7.7 steps/minute (range 152–185) in the barefoot population, higher ($p<0.001$) than the average in the shod population of 159.2±8.0 steps/minute (range 150–172).

Fig 1, which graphs the FSA and strike type of every trial of every subject, highlights the considerable variation in foot strike observed within subjects as well as between groups. Although average FSA was 1.1° ±5.3 among the habitually barefoot individuals and -8.3° ±6.1 among the habitually shod individuals, indicating that average strike type for each group was a FFS and RFS, respectively, a slight majority of individuals (56%) used more than one strike type. The average intra-individual variance for FSA was 20.65° ±3.12 s.e.

The first hypothesis, H1, tested predictions of higher levels of variation in foot strike with respect to several nominal variables including sex, footwear use (barefoot versus shod) and surface stiffness (hard versus soft trackways), as well as in continuous variables such as speed, running history, and footwear history. Levene's Tests of nominal comparisons for FSA, summarized in Table 1, indicate that variation in FSA was not greater in men than women, or among individuals who were barefoot than shod. However, a Chi-square test revealed that the percentage of barefoot individuals who used more than one strike type (72%) was significantly greater than the percentage of shod individuals who used varied strike types (32%) (Pearson $\chi^2 = 7.78$ (1); $p = 0.005$). Although the entire study sample had significantly more foot strike variation when running on soft than on hard surfaces ($p<0.05$), this difference was not significant within the barefoot population or within the shod population.

Since most of the variables analyzed in this study are continuous, GLMMs were used to calculate the residuals of mixed-effect regressions between FSA and each predictor variable (Table 2). Of the continuously distributed extrinsic variables studied, speed had no significant effect on variation in FSA, but individuals used more variable foot strikes ($p = 0.05$) when they ran with lower step frequencies. Finally, of the acquired factors measured, variation in FSA was homogenous for preferred step frequency, footwear history or mile time on the degree of foot strike variation, but individuals who ran more frequently had considerably less variation in foot strike variation ($p<0.0001$) than individuals who ran less.

The second hypothesis, H2, focused on what extrinsic, intrinsic and acquired factors influence FSA in the population studied. We predict that FSA would be significantly affected by speed, step frequency, and surface stiffness, as well as footwear and running history. The hypothesis was tested using a GLMM, summarized in Table 3, with FSA as the response variable, and in which the fixed effects included age, sex, height, body mass, substrate stiffness,



FFS
10
0
-10
-20
RFS
Footstrike Angle(°)
Habitually barefoot adult males
Habitually barefoot adolescent females
Habitually barefoot adolescent males
Habitually shod adolescent females
Habitually shod adolescent males

Fig 1. Variation in foot strike angle (FSA). Every FSA measured for every participant, noting which are forefoot (FFS), midfoot (MFS) and rearfoot (RFS) strikes. Note the greater degree of variability in the habitually barefoot individuals.

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footwear history, running history, preferred step frequency, and the speed at which the participants could run a mile. As the coefficients (which represent the slope of the relationship between FSA and each predictor variable) in Table 3 indicate, none of the intrinsic variables (sex, age, body mass) have an effect on FSA at conventional levels of significance ($p<0.05$), but there were marked, significant effects on FSA (in order of t-value) from preferred step frequency ($p = 0.001$), footwear history ($p = 0.011$) and track surface ($p = 0.02$); running history ($p = 0.086$) and mile time ($p = 0.083$) trended toward conventional levels of significance. In other words, individuals were more likely to have a higher FSA and thus FFS or MFS independent of speed and anthropometric characteristics if they used a higher step frequency, rarely used shoes, and ran on a soft surface; and they had a greater tendency to FFS or MFS if they had more experience running and had faster mile times.

Figs 2 and 3 further explore the conventional bivariate associations between averaged FSA and selected intrinsic, extrinsic and acquired variables. In terms of speed, the habitually barefoot participants ran about 18% faster than the habitually shod participants, leading to a significant correlation ($r = 0.49$; $p = 0.004$) between speed and FSA within the population as a whole, but not within the habitually barefoot ($r = 0.04$; $p = 0.83$) and habitually shod ($r = 0.16$; $p = 0.51$) groups (Fig 2A). Measured step frequency was uncorrelated with FSA either within or between groups (Fig 2B), but preferred step frequency correlated strongly with FSA in the population as a whole ($r = 0.692$; $p<0.001$) and within the habitually shod runners ($r = 0.652$; $p = 0.002$), and approached conventional levels of significance within the habitually barefoot groups ($r = 0.333$; $p = 0.07$) (Fig 2C). To assess the effects of surface stiffness on foot strike, Fig 2D graphs the difference in average FSA on the hard versus soft tracks, with a value of zero indicating no difference, and positive or negative values indicating a greater tendency to RFS or FFS on soft surfaces, respectively. As this analysis shows, habitually barefoot individuals were more likely to RFS on the soft track with average FSAs that were 1.88°± 0.85 (s.e.) more dorsiflexed (t-test = 1.71, $p = 0.04$); in contrast, habitually shod individuals were more likely to FFS with average FSAs that were 2.16°± 0.95 (s.e.) more plantar flexed (t-test = -5.83, $p<0.001$).

As noted above, it was not possible to quantify running history and footwear history as continuous variables, but Fig 3 summarizes the relationship between binned categories of these

Table 2. GLMM analysis of variation in foot strike angle (FSA) relative to continuously distributed predictor variables.

Variable	Coefficient	S.E.	t-value	p-value
Intrinsic factors				
Age	-0.025	0.419	-0.83	0.41
Weight	0.000	0.119	0.00	1.00
Height	-0.019	0.413	0.47	0.65
Extrinsic factors				
Speed	0.015	0.021	0.72	0.47
Step freq	0.052	0.026	1.97	**0.05**
Acquired factors				
PSF	-0.007	0.026	-0.28	0.78
Footwear history	-0.013	0.026	-0.50	0.62
Running history	-0.160	0.039	-0.42	**<0.001**
Mile time	0.045	0.059	0.76	0.45

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Table 3. GLMM analysis of effects of intrinsic, extrinsic and acquired variables on foot strike angle (FSA)*.

Variable	Coefficient Estimate	Std. Error	t-value	Standard parametric p-value	Residual Randomization p-values
Surface	0.1224	0.0474	2.585	**0.0101**	**0.015**
Age	0.5341	0.6462	0.8265	0.4167	0.303
Sex	0.401	0.2931	1.3681	0.184	**0.089**
Height	0.0211	0.1568	0.1348	0.8939	0.791
Body mass	-0.1226	0.1857	-0.6602	0.5154	0.304
Footwear History					**0.011**
Footwear 2	-1.0245	0.4338	-2.3615	**0.0267**	
Footwear 3	-1.5676	0.6289	-2.4928	**0.02**	
Footwear 4	-0.9965	0.6289	-1.5847	0.1261	
Running History					**0.086**
Running 2	0.1368	0.3383	0.4043	0.6895	
Running 3	1.2861	0.57	2.2562	**0.0334**	
Running 4	1.412	0.5843	2.4167	**0.0236**	
Preferred Step Frequency	0.6667	0.2063	3.2326	**0.0035**	**0.001**
Mile Time	0.2808	0.2296	1.2226	0.2333	**0.083**

*Fixed effects multiple R-squared: 0.76, Fixed effects adjusted R-squared: 0.75.

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acquired variables and FSA. As Fig 3A shows, individuals who spend more time barefoot show considerable variation in FSA but tend to have more positive FSAs, whereas individuals who are more habitually shod have more negative FSAs (ANOVA, $p<0.001$). Similarly, individuals who run more have significantly higher FSAs, reflecting a higher percentage of FFS (Fig 3B). Because running and footwear history are not independent in this population, we tested the effects of multicollinearity using partial correlation analysis. The partial correlation of running history with FSA holding constant the effects of footwear history is 0.31 ($p = 0.04$), and the partial correlation of footwear history with FSA holding constant running history is 0.45 ($p = 0.001$), indicating that both of these acquired factors contribute independently to foot strike variation.

Finally, since the focus of this study was on variation in FSA, a second GLMM was computed to explore the effects of running kinematics on FSA. The results of this analysis (S1 Table), indicate that FSA was most influenced by ankle angle, overstride relative to the knee, and trunk angle ($p = 0.001$), and was not strongly associated with speed and hip angle. An ANOVA, however, revealed some significant differences in kinematics between the habitually barefoot and shod individuals. In particular, the habitually barefoot individuals had 8% higher average preferred step frequencies (172 vs 159 steps/min, $p<0.0001$); tended to land with 5–6° more flexed knees and hips ($p<0.001$); had approximately 50% less overstride relative to the knee ($p = 0.0007$); and had 4° more vertical trunks ($p = 0.002$). Some of these differences may be attributable to speed, which was 0.6 m/s higher in the habitually barefoot individuals, largely because the habitually unshod adolescent males ran approximately 0.5 $m{\bullet}s^{-1}$ faster (4.04 $m{\bullet}s^{-1} \pm 0.31$) than the population mean of 3.69 $m{\bullet}s^{-1} \pm 0.44$, and the habitually shod adolescent females ran significantly slower (3.04 $m{\bullet}s^{-1} \pm 0.32$) (ANOVA, $p<0.0001$). Group means for these kinematic variables are summarized in S2 Table.

Discussion

The most basic result of this study is that under varied running conditions foot strike angle and type can be variable both within and between individuals, especially among habitually barefoot

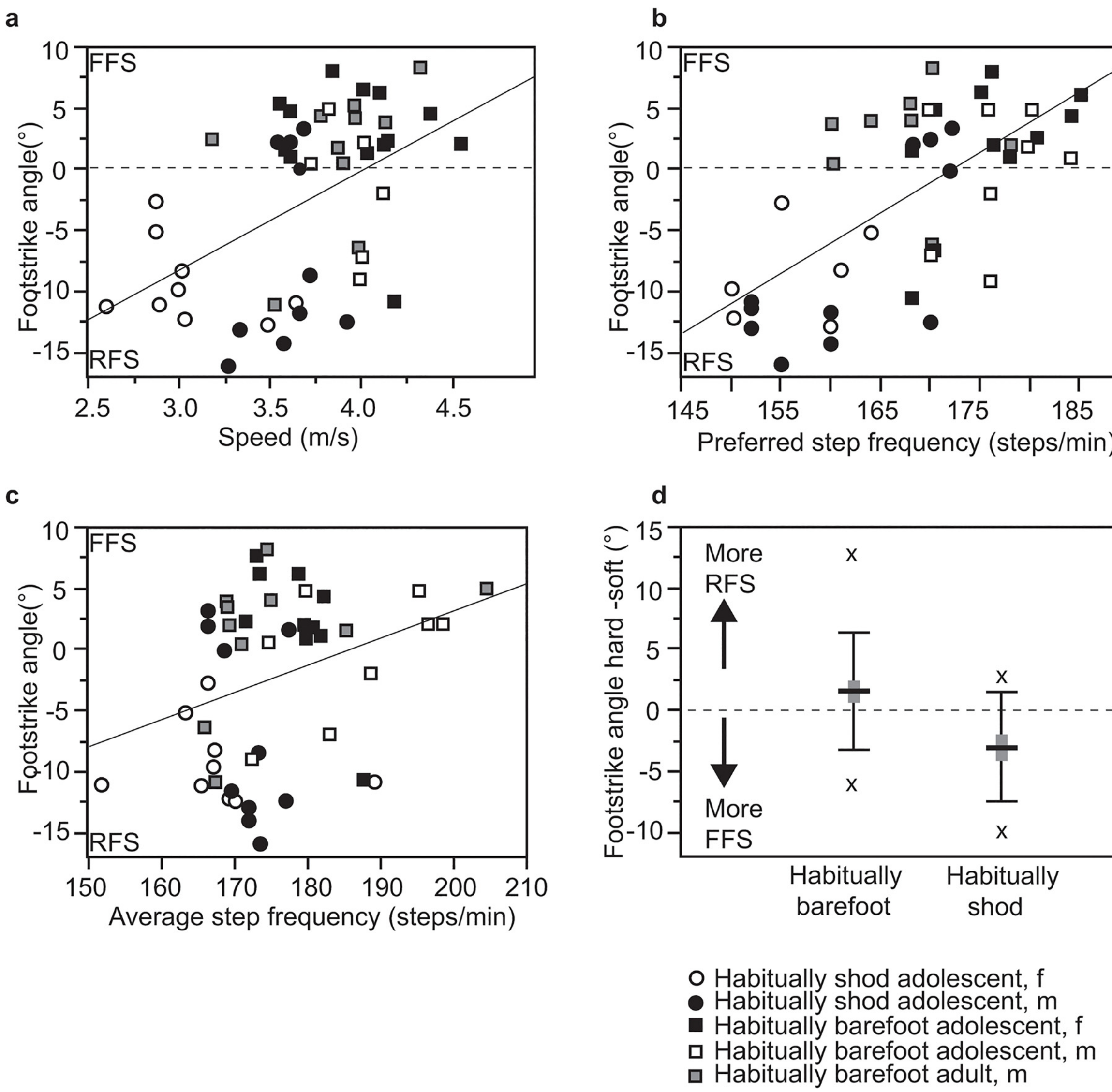

Fig 2. Sources of variation in foot strike angle (FSA). a) Regression of speed versus FSA; b) regression of measured step frequency versus FSA; c) regression of preferred stride frequency versus FSA; d) Box (standard error) and whisker (standard deviation) plot of difference in FSA on hard versus soft tracks for habitually barefoot and shod individuals (more positive values indicate more dorsiflexed FSA on soft surface; more negative values indicate more plantar flexed FSA on soft surface); x marks indicate maximum and minimum values.

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individuals. This variation is highlighted by the plot of every FSA recorded in the study (Fig 1), which shows that the average intra-individual standard deviation of FSA was 4.12°, and that while a majority of participants (56%) used a combination of FFS, MFS and RFS landings, 72% of the barefoot runners and 32% of the shod runners used multiple strike types.

This study tested two general hypotheses regarding the effects of intrinsic, extrinsic, and acquired factors on variations in observed in foot strike. The first general hypothesis—that certain factors influence the degree of variation in strike type—was supported (see Table 1 and Table 2). Although none of the intrinsic factors measured (height, sex, age, and body mass) affected the degree of variation in FSA, several extrinsic and acquired factors did influence FSA variation. In particular, there was a significantly greater degree of FSA variability within individuals who used lower step frequencies and who typically ran less. In addition, although FSA

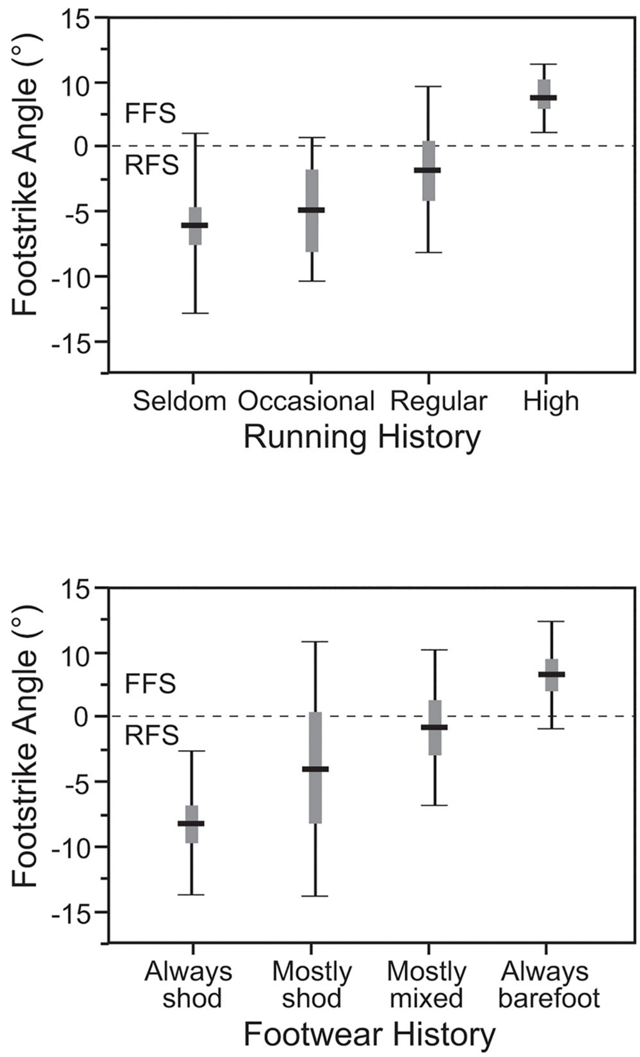

Fig 3. Foot strike angle (FSA) and running history and footwear history. Box (standard error) and whisker (standard deviation) plots of average FSA (°) for individuals categorized by running history (a) and by footwear history (b). See text for explanation of how participants were binned into categories. In both analyses, $p<0.001$ (oneway ANOVA).

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variation was not affected by footwear history, individuals who were barefoot had significantly more variable foot strike types than those who were wearing shoes. The explanation for this seemingly contradictory result is that average FSA among barefoot individuals was 1.11° ± 5.3, whereas the average FSA among those who were habitually shod was -8.3° ± 6.1 (t-test, $p<0.001$). Consequently, individuals who were barefoot were more likely to not only FFS and MFS, but also to sometimes land with negative values (a RFS), while habitually shod individuals were less likely to land with flat or plantar flexed feet. Note also that there was no effect of preferred step frequency, mile time, surface stiffness, or speed on the degree of FSA variation.

The second general hypothesis tested was that a combination of intrinsic, extrinsic, and acquired factors would influence FSA, hence strike type. In particular, it was predicted that participants would be more likely to shift to more positive FSA values, hence a higher frequency of MFS or FFS landings, when they ran at faster speeds, higher step frequencies and on harder surfaces, and that participants who were more experienced runners or were habitually barefoot would also be more likely to MFS or FSS. These hypotheses were all supported. In particular, the GLMM (Table 3) revealed significant effects of track stiffness, preferred step frequency, footwear history, running history and mile time speeds. Put simply, individuals were less likely to RFS when they ran on a harder track (Fig 2D), preferred higher step frequencies (Fig 2C), were able to run faster, were experienced runners (Fig 3A), and were habitually barefoot (Fig 3B). In contrast, there was no effect on FSA from age, sex, body mass, height, or speed (Fig 2A).

These results are consistent with a previous, smaller comparison of barefoot and shod Kalenjin individuals that sampled a more limited range of faster speeds [7], as well as studies that compare running form among populations that vary in footwear use [11,15] or in which habitually shod individuals have been studied both barefoot and shod [8,9,34]. Although barefoot individuals sometimes RFS, they are more likely to FFS and MFS depending on conditions and experience; in contrast, habitually shod individuals are more likely to RFS under a range of conditions.

Although not a focus of this study, the results presented here confirm those of previous studies that compared kinematics and kinetics between barefoot and shod runners [4–12,15]. In general, the habitually barefoot participants landed with more flexed knees and hips, they had slightly more vertical trunks, they preferred higher step frequencies, and they were less likely to overstride (S2 Table). When a GLMM was used to tease apart which of these variables were associated with variations in FSA, the strongest predictor was ankle angle, with significant associations also evident for overstride and trunk angle (S1 Table).

Before considering the implications of these results, it is worth summarizing the study's limitations. One problem is the limited range of subjects, conditions, and factors sampled. We were unable to include adult women, and the sample sizes for each group were necessarily limited by time and opportunity. Broadening the sample in terms of age, sex, running experience, and footwear history would likely reveal additional variability. In addition, the experimental design did not look at fatigue, which can increase the likelihood of using a RFS [35], and only a few external factors hypothesized to influence kinematics (notably speed, step frequency and surface stiffness) were manipulated. Future studies would benefit from examining substrate factors such as slipperiness, smoothness, inclines, and changes in direction of the sort that runners encounter when they run on trails and other variable environments that, until relatively recently, were the primary contexts in which people ran. Another necessary limitation of the study was to measure only sagittal plane kinematics using video without collecting information on ground reaction forces and muscle function. A final concern was the participants' ability to run normally. Although the experiment was not conducted in the laboratory on a treadmill, running at different speeds on a track with markers taped to one's joints while trying to adapt one's step frequency to a metronome is an unusual experience that can interfere with normal running form. This concern, however, applies to all studies of running kinematics and kinetics, and it is arguable that the conditions tested here are a step in the direction of understanding variability in running form beyond the laboratory and among individuals who are not just habitually shod from developed countries. Although such people are the focus of most research, they are unusual from an evolutionary perspective [36].

These limitations aside, the study's results have some relevance for current discussions about running form. Most importantly, very few studies on running biomechanics have sampled runners who are not habitually shod and in the natural settings in which people used to locomote rather than in controlled laboratory conditions, primarily on treadmills or over force plates [7,11,13,14]. It is reasonable to hypothesize that these modern contexts limit variation in foot strike as well as other aspects of kinematics. The results presented here raise the possibility that for much of human evolution foot strike patterns were more variable. The two most obvious factors that have potentially contributed to less variation in how people run is increased use of flat, paved surfaces and treadmills for running, and the prevalence of running shoes with elevated, cushioned heels that have been available only since the 1970s [7]. Just as cushioned heels facilitate RFS landings on hard surfaces, it is reasonable to hypothesize that the barefoot individuals measured in this study were more likely to run with a RFS on the soft trackway because softer substrates, like cushioned heels, make RFS landings more comfortable by lowering the rate of loading of the impact peak [7]. Although soft and smooth surfaces no doubt existed in the past such as along lakeshores and in sandy environments, most people typically walked and ran on compacted soil with rocks, vegetation, and features that increase substrate complexity and stiffness. Walking and running without shoes on these surfaces unquestionably elicits much more varied and extreme stimuli from sensory nerves on the glabrous surface of the sole. It is therefore reasonable to hypothesize that people ran with more varied kinematics prior to the invention of shoes, which probably occurred in the last 40,000 years [37].

Another factor that may have affected variation in running form is skill. Since the running boom that began in the 1970s, there has been an increase in running among amateurs, who usually get less coaching and train less intensively than athletes who are professional or on teams [38]. One hypothesis that merits further testing is that untrained, amateur runners in developed nations are more likely to run like the habitually shod Kalenjin studied here, with a relatively slower step frequency and a greater proclivity to land with a dorsiflexed foot, hence a RFS. This observation leads to the hypothesis that a contributing factor to Kalenjin excellence in distance running might be that most elite Kalenjin runners grew up running long distances without shoes on a regular basis in the same conditions as the habitual barefoot participants analyzed in this study [27]. Although habitually barefoot people from the Daasanach tribe in northern Kenya were observed to mostly run with a RFS at slow speeds (2.1–3.0 m/s) a possible explanation for this different result, apart from speed, is that these individuals live in a hot, sandy desert and do not run often much [14]. Other studies of adults from habitually barefoot and minimally shod populations found that individuals (especially men) were more likely to FFS or MFS [7,11,13].

Finally, what do these results mean for habitually shod individuals who run mostly on pavement and treadmills, and wonder how to make sense of diverse arguments about minimal shoes, cushioning, and strike type? First, the restricted variation in strike type among habitually shod runners today may be a recent phenomenon, and it would be useful to test if runners adopt more variation when running on trails rather than on pavement or treadmills. In addition, although rearfoot and forefoot striking are both normal, everything involves trade-offs. RFS landings have the potential advantages of being comfortable in shoes or on soft surfaces, they require less calf and foot muscle strength, and they lessen external moments acting around the ankle [39]. Their potential disadvantages are that they cause impact peaks whose rate and magnitude are hypothesized by some researchers to be related to some repetitive stress injuries, they increase external moments around the knee, and certain kinematic patterns associated with (but not exclusive to) RFS gaits, such as overstriding and extended knees at landing, are implicated in some repetitive stress injuries [40]. More research is needed to evaluate the costs and benefits of different strike types, but one hypothesis that also needs to be explored is that running with more kinematic variation, as perhaps occurs during trail running, is more natural and may also be beneficial.

Supporting Information

S1 File. Residual Randomization Methods.
(DOCX)

S1 Table. GLMM analysis of effects of kinematic variables on Foot strike Angle (FSA)*.
(DOCX)

S2 Table. Group means and standard deviations for major variables studied.
(DOCX)

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16 Weeks of Progressive Barefoot Running Training Changes Impact Force and Muscle Activation in Habitual Shod Runners

Abstract

Short-term effects of barefoot and simulated barefoot running have been widely discussed in recent years. Consequences of adopting barefoot running for a long period, including as a training approach, still remain unknown. The present study evaluated the influence of 16 weeks of progressive barefoot running training on impact force and muscle activation in habitual shod runners. Six habitual shod runners (3 men and 3 women, 29.5 ± 7.3 years) were tested barefoot (BF) and shod (SH), before and after 16 weeks of progressive barefoot running training. Tests consisted of running on instrumented treadmill at 9 km/h, for 10 minutes in each experimental condition. Nine data acquisitions (10 s) of vertical ground reaction force (VGRF) and electromyographic (EMG) signal were conducted in each experimental condition for each test. BF training was effective to alter VGRF and EMG parameters of running in habitual shod runners, regardless of footwear condition (SH or BF). The magnitude of first peak of VGRF (Fy1) and the impulse of the first 50 ms decreased after training for BF and SH ($p<0.01$). The activation reduced from PRE to POST training for four muscles in BF running ($p<0.001$), whereas only muscle gastrocnemius lateralis decreased significantly its activation ($p<0.01$) in SH running. A 16-week progressive barefoot running training seems to be an effective training strategy to reduce impact force, improve shock attenuation and to decrease muscle activation intensity, not only in BF running, but also in SH running, although BF condition seems to be more influenced by BF training.

Introduction

Research interest and participation in barefoot (BF) running has increased remarkably in recent years [1–4]. Many reasons seem to drive people to BF running, however, this popularity is mainly based on the belief that BF alters biomechanical parameters of running, improving impact forces attenuation, increasing performance and reducing injury risk [2,3,5–8].

In short-term, the effects of BF running on biomechanical parameters have been previously described. Changes in spatiotemporal variables [7,9–11], foot strike pattern [6,12–14] and joint kinematics [13,15–18] have been reported for BF running in both habitual shod and barefoot runners. In runners without experience in BF, impact forces seem to be increased during barefoot locomotion, suggesting increased risk of injuries compared to shod (SH) condition [5,9,19–22]. Probably as consequence of this increased external load, literature also reports greater amplitudes of muscle pre-activation [17], increased muscle activation [17,18,23] and altered muscle coordination in BF running [17,18,23]. Such data could mean that the absence of footwear could also represent a risk for runners and, additionally, a less efficient running economy [17,23], although the real influence of BF condition on running economy remains unclear.

Despite the potential benefits of BF running, literature lacks studies investigating the long-term effects of barefoot running. The few data available on literature about this issue are related to studies that investigated habitual SH runners under short periods of familiarization or running training programs based on simulated barefoot (through minimalist shoes) [24–28]. Evidence shows that 4–12 weeks of simulated barefoot running induced to reduced plantar pressure [26], changes in muscle activation [27], a mid/forefoot strike pattern [24,26] and improvements in running economy [29]. Although there have been studies on how people switch running form and shoes, as far as we know, no study investigated the long-term effects of BF running or the use of this strategy as training approach for habitual SH runners. Additionally, none of these studies investigated the chronic influence of BF running on impact forces and shock attenuation. Thus, the investigation of impact forces and lower limb muscles involved in running becomes crucial for the understanding of barefoot adaptation's process in long-term, as well as of the use of this mechanical condition as training approach.

Shot-term studies in habitual BF runners suggest the human body could adapt to BF situation and get benefits from the chronic use of this way of locomotion [6,10,12,13,16,30]. Improved mechanical load control [6,10,12], reduced muscle activation and improved running economy [16,30] have been observed in habitual BF runners. Experienced BF runners presented improvements in shock attenuation, as smaller incidence of first peak of VGRF or reduced magnitude of impact peak of VGRF during unshod [6,10,12,13]. Additionally, experienced BF runners present alterations in muscles activation pattern and decreased activation intensity of some muscles, what could mean less energy cost [10,16,30]. Thereby, BF running arises as a possible training strategy to improve mechanical load control and muscle activation [6,31–33].

Therefore, the purpose of this study was to analyze the influence of 16 weeks of progressive BF running training on kinetics and activation of selected muscles of lower limbs in habitual SH runners. For this, parameters of vertical ground reaction force (VGRF) and electromyographic (EMG) signal obtained during BF and SH running will be compared before and after 16 weeks of progressive BF training. Improvements in shock attenuation and decreased muscle activation intensity are expected after training.

Material and Methods

Participants

This research was approved by the Ethics Committee of the School of Physical Education and Sport of the University of São Paulo (Protocol N˚ 17816613.9.0000.5391, approved on January 3rd 2013) prior to recruitment of participants. Investigation was conducted according to the principles expressed in the *Declaration of Helsinki*. All participants read and signed an informed consent term. Experimental design was approved by the local ethics committee. The authors confirm that all ongoing and related trials for this intervention are registered (ClinicalTrials.gov

Identifier: NCT02815826). The S1 and S2 Files present the Trend Checklist and the study protocol approved for this research.

This prospective study was performed from September 2012 to June 2013. After advertising the research runners' communities, thirty three participants enrolled to the study but 13 were excluded (Fig 1). Twenty eligible participants (13 men and 7 women; 33.2 ± 6.4 years; 72.6 ± 14.2 kg; 1.72 ± 0.11 m) were recruited from a community of runners at the University of São Paulo, in Brazil. A questionnaire was used to collect information about running experience, average weekly running distance and previous lower limb injuries. Participants should be 18–40 years old, be experienced in running, but without experience in minimalist/barefoot running, had a minimum of 6 months of regular running training and a minimum of 6 months of experience in running on treadmills. Participants were not included if they had suffered any orthopedic injury in the last 12 months. Additionally, participants who presented habitual forefoot strike pattern, completed less than 80% of training and/or suffered any injury during training were excluded. Participants reported 5.6 years of experience in regular running training (0.5–22 years), weekly volume of 44.2 km (25–100 kilometers per week) and 4 training sessions per week (3–5 sessions per week).

Intervention

According to literature [34–37], the transition from SH to BF running must be done through gradual changes of volume and intensity of stimulus. Thus, barefoot training was based on the weekly training volume (WTV) of each participant. The BF training volume and surfaces of training were controlled.

During the 16 weeks of training, participants kept their normal running training routine (wearing shoes), while they were introduced progressively to BF condition. Three training sessions were performed per week. Barefoot training started with 5% and ended with 20% of their WTV being performed without shoes (Table 1). Soft surfaces (i.e. sand and grass) were adopted in the beginning of training (week 1 to 8). From week 9 to 16, participants mixed soft with harder surfaces, including treadmill and asphalt, to accomplish the training. Training sessions were planned and prescribed by professionals, researches and participants together. All training sessions were supervised by the researchers.

Experimental protocol

Participants ran, before and after intervention, on a treadmill under two experimental conditions: barefoot and shod. Experimental condition order was randomized to avoid learning effects.

Each session test started with participants performing a maximum voluntary isometric contraction (MVIC) test for each muscle of interest [38,39]. The MVIC protocol consisted of 4 movement trials for each muscle: 2 submaximal trials of 10 seconds; 1 maximal trial of 5 seconds; and 1 maximal trial of 10 seconds. Then, a 5-minute period of warm-up at self-selected speed was performed on a treadmill. After that, participants ran (at 9km/h) during 10 minutes on an instrumented treadmill in both barefoot and shod conditions. Participants had a 2-minute interval between each trial while experimental condition was changed. The VGRF of both legs and EMG signal of tibialis anterior (TA), gastrocnemius lateralis (GL), long head of biceps femoris (BCF), rectus femoris (RF) and vastus lateralis (VL) of the right leg of each participant were obtained. These muscles were chosen due to their importance and contribution to running [40–42]. For shod trial, runners wore their own habitual running shoes. All shoes were in good conditions of use and had similar characteristics of construction.

Fig 1 presents the CONSORT flowchart of the study's enrollment and follow-up.

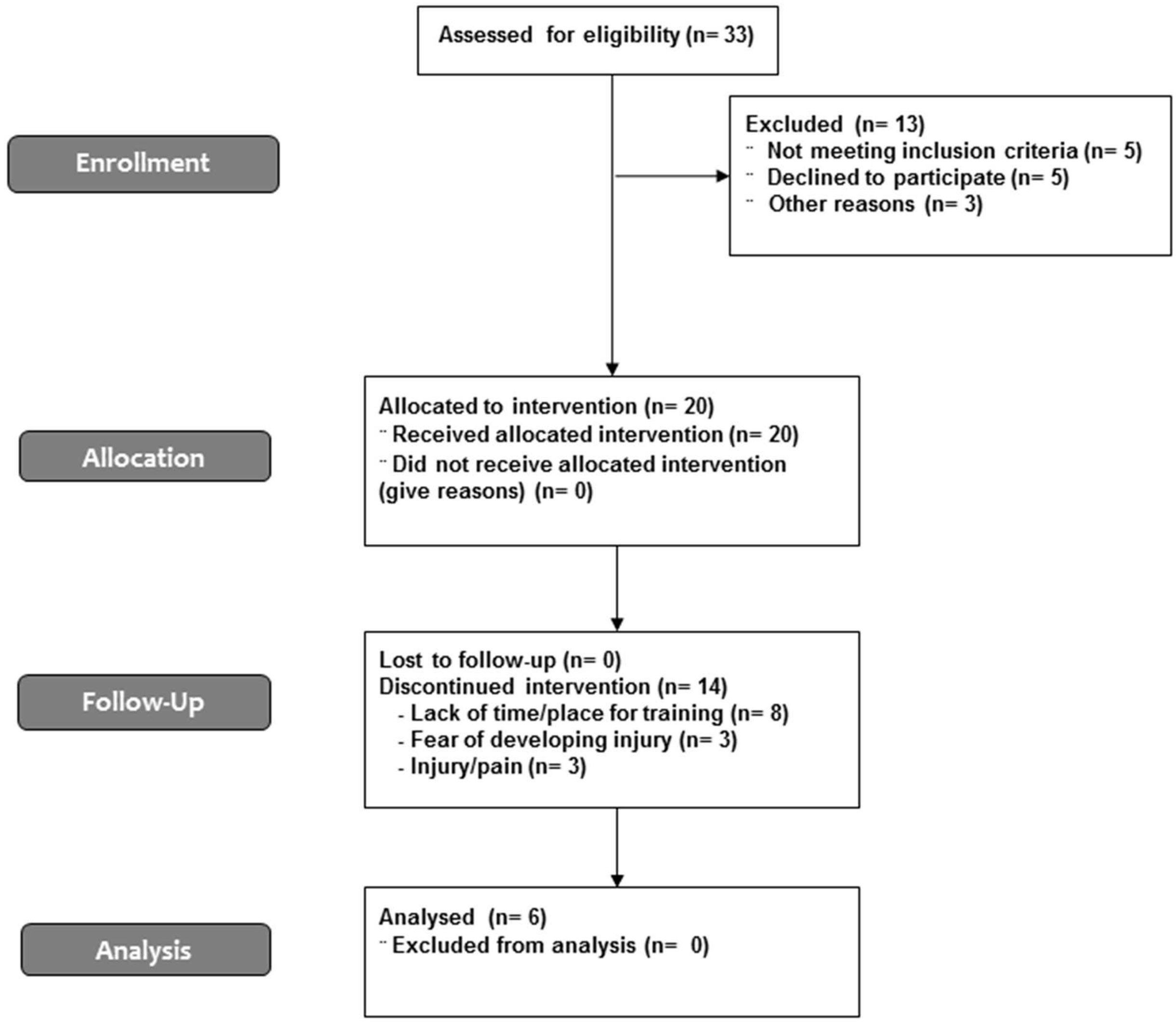

Fig 1. CONSORT flowchart of enrollment and follow-up.

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Equipment and data acquisition

The VGRF data was obtained by the Gaitway Instrumented Treadmill System (9810S1), composed by an instrumented treadmill with two piezoeletric platforms assembled on its surface (Trotter Treadmill Model 685, 01–06560201), an Analog/Digital (A/D) conversor (Keithley MetraByte DAS–1402) and the Gaitway Software (Versão 1.0x). The EMG signal was measured by the Lynx-EMG System 1000 (Lynx Electronic Technology LTDA.), composed by data

Table 1. Barefoot running training progression (in % of weekly training volume—WTV).

Period (weeks)	Barefoot Training
1^{st} to 4^{th}	Until 5%–walking in soft surfaces
5^{th} to 8^{th}	5% to 10%–walking and light running (6–8 km/h) in soft surfaces
9^{th} to 12^{th}	10% to 15%–light running (7–8 km/h) in mixed surfaces
13^{th} to 16^{th}	15% to 20%–moderate running (8–10 km/h) in mixed surfaces

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acquisition EMG1000-VxRy module, an Analog/Digital (A/D) converter and the Lynx-AqDados program. Bipolar surface electrodes "Double" (Hal Industry and Trade LTDA), AgCl, were placed on muscle bellies and connected to active preamplifiers AX1010 (Lynx Electronic Technology LTDA.). Electrodes placement in each muscle occurred according to the criteria established by SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles). Nine acquisitions (10 seconds each) of VGRF and EMG signal were recorded over the 10 minutes of test in each experimental condition, with sampling rate of 2600 Hz. An average of 20 steps (10 right and 10 left) were obtained in each trial acquisition.

Signal processing and statistical analysis

The VGRF data was low pass filtered by a Butterworth filter (4th order, 90 Hz cutoff frequency). The start and end of each left and right step was determined using 30N threshold. VGRF was normalized by individual body weight, and time was normalized by total support time (0 to 100% of the support, 0.1% lag). The EMG signal was filtered by a digital Butterworth band pass filter of 4th order (cutoff frequency from 20 to 450Hz) and notch filters of 60Hz, 120Hz and 180Hz. After these procedures, RMS was calculated and data was normalized by the maximum voluntary isometric contraction (MVIC), obtained at the beginning of the test session, prior to the running test. The signal obtained between the 4^{th} and 8^{th} second of the last maximal trial of each muscle was used for normalization of EMG signal obtained during running. Examples of raw GRF and EMG data are available as Supporting Information (S3 File).

For VGRF analysis, the following variables were selected: magnitude of first peak of VGRF (Fy1); time to achieve first peak of VGRF (tFy1); loading rate (LR1), calculated by the ratio Fy1/tFy1; and impulse during the first 50 ms of stance (Imp50), calculated from the area under the curve GRF x Time, until 50 ms. Muscle activation intensity was assessed through calculation of the RMS (Root Mean Square) of EMG signal. This procedure was done for each muscle analyzed, during stance phase, for shod and barefoot running.

Data normal distribution was checked with the Kolmogorov-Smirnov test, while homoscedasticity was tested by Levene test. A two-way ANOVA for repeated measures was performed to compare shod and unshod running (condition), as well as pre and post intervention (moment). The Student-Newman-Keuls test was performed as post hoc test. The level of significance was set at 5%. The statistical analysis was performed with SigmaStat 3.5 (Systat Software Inc., USA). Statistical data reports are available as Supporting Information (S4 File).

Results

Participants

Of the 20 participants recruited for the study, only 6 runners (3 men and 3 women, 29.5 ± 7.3 years, 64.1 ± 11.0 kg, 1.68 ± 0.14 m) completed the study protocol and were included in the final analysis. Despite dropouts, the sample baseline characteristics remained similar. Dropouts from the study occurred due to: lack of time/place for training sessions (n = 8), fear of developing injury (n = 3) and injury/pain (n = 3), although one of these injuries was not related to BF training. Large samples are not common in researches involving a new, tough and long training program, as the present study. Although the reduced sample size, a sensitivity power analysis test was performed (alpha = 0.05; power = 0.8; number of measurements considered for each participant = 9; by G*Power v.3.1.9.2 free software, Dusseldorf, Germany) and a medium effect size (0.41) was observed [43–45].

Table 2. Summary statistics (mean, standard deviation, F-value and p-value of interactions) of GRF data for shod (SH) and barefoot (BF) running, before (PRE) and after (POST) training.

VARIABLES	SH		BF		*F-value (interaction)*	*p-value (interaction)*
	PRE	POST	PRE	POST		
Fy1 (BW)	1.44 ± 0.06[a]	1.15 ± 0,06[a d]	1.63 ± 0.06[b]	0.89 ± 0.06[b d]	***12.616***	***0.016****
tFy1 (ms)	34.10 ± 2.23	33.60 ± 2.23	20.30 ± 2.23	18.80 ± 2.23	*0.0579*	*0.819[+]*
LR1 (BW/s)	33.41 ± 2.36[c]	28.96 ± 2,36	62.65 ± 2.36[b c]	29.14 ± 2.36[b]	***37.816***	***0.002****
Imp50 (BW.ms)	38.70 ± 0.98[a]	32.10 ± 0.98[a]	45.50 ± 0.98[b]	32.70 ± 0.98[b]	***9.801***	***0.026****

*: significant interaction between shoe condition and moment.

[+]: significant main effect of moment.

[a]: difference between PRE and POST in SH running (post hoc).

[b]: difference between PRE and POST in BF running (post hoc).

[c]: difference between SH and BF at PRE moment (post hoc).

[d]: difference between SH and BF at POST moment (post hoc).

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Ground reaction force

Significant interactions between condition (SH / BF) and moment (PRE / POST) were observed for all VGRF parameters, except for tFy1 (Table 2). Time to reach first peak (tFy1) presented only main effect of condition (p = 0.008), being 42.18% smaller for BF condition.

Post hoc test revealed differences between conditions and moments for variables analyzed. For Fy1, differences occurred between PRE and POST for both conditions of running (SH and BF) (p = 0.007 and p<0.001, respectively) and between SH POST and BF POST (p = 0.025) (Fig 2). In SH running, Fy1 decreased 20.1% from PRE to POST, while Fy1 presented a reduction of 45.4% from PRE in BF running. The Fy1 in BF POST was 22.6% smaller than SH POST.

About LR (Fig 3), differences occurred between conditions before training (SH PRE and BF PRE) (p = 0.001) and between PRE and POST for BF running (p<0.001). Before training, LR in SH running was 46.7% smaller than in BF running. Additionally, LR reduced about 53.5% its value from PRE to POST in BF running.

Differences between PRE and POST in both conditions (p = 0.027 for SH and p<0.001 for BF) (Fig 4) were observed for Imp50. The Imp50 was 17% smaller after training in SH running. Similarly, Imp50 was 28.1% smaller after intervention in BF running. Additionally, a statistical trend of difference (p = 0.085) between SH and BF was observed before training. BF PRE was 17.57% higher than SH PRE. Probably, the post hoc test adopted in this study was not powerful enough to reveal statistical difference as sample size was reduced. This result is a reasonable explanation for the interaction observed for Imp50.

Muscle activation

Significant interactions were observed for muscle activation. The muscle activation intensity decreased for most muscle as response to training, regardless footwear condition (p<0.01). Differences in RMS between the moments analyzed are presented on Figs 5 and 6. Only GL altered significantly its activation intensity during stance phase for SH running, decreasing 63% the RMS value from PRE to POST (Fig 5). All muscles, except BCF, reduced their activation intensity during stance phase of BF running after intervention (Fig 6). The TA decreased 69% the RMS from PRE to POST. Similarly, GL and VL showed a decrease of 66% and 65%, respectively, in their RMS when after BF training. A decrease of 45% in the RMS was also observed for RF in POST. The TA (111%), VL (131%) and BCF (115%) had greater values of

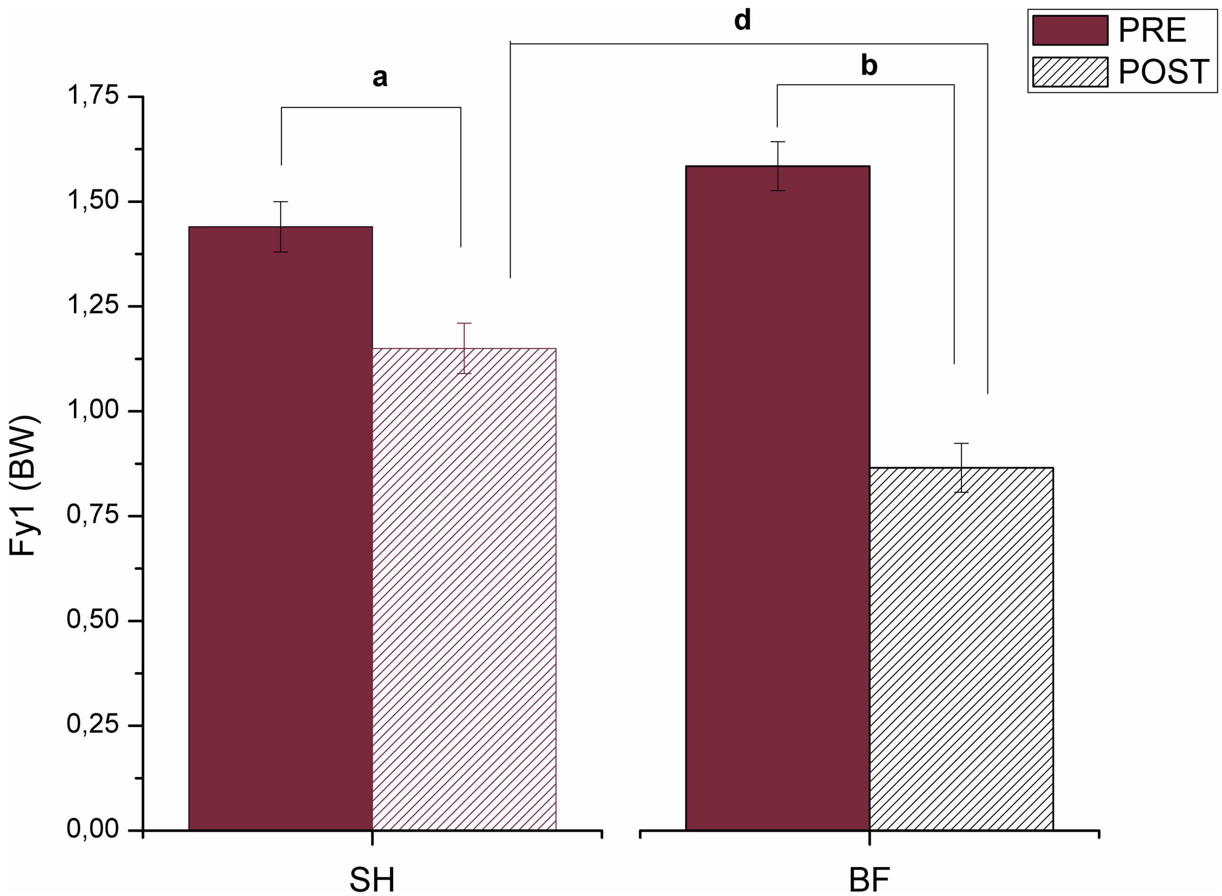

Fig 2. Mean and standard deviation values for the magnitude of first peak (Fy1) during running shod (SH) and barefoot (BF), in both PRE and POST training, where ([a]) means difference between PRE and POST in SH running; ([b]) means difference between PRE and POST in BF running; and ([d]) means difference between SH and BF at POST moment.

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RMS during stance phase in BF compared to SH running before training. After intervention, all muscles had similar activation intensity for both SH and BF, except BCF (157% greater RMS for BF then to SH). Fig 7 presents average VGRF curves and raw EMG signal obtained from GL during stance phase of one participant during BF running, before and after intervention, together with an illustrative sequence of running cycle during stance phase. S1 Table presents the summary statistics (mean, standard deviation and p-value of interactions) of RMS data.

Discussion

This study aimed to investigate the effects of a 16-week progressive barefoot running training program on kinetics and EMG signal of lower limbs muscles in SH and BF running. Improved mechanical load control and decreased muscle activation intensity were expected after intervention, in both footwear conditions (SH and BF). The current investigation is, as far as we know, the first research to access the long-term progressive use of unshod running training.

The main finding of this study is that 16 weeks of progressive BF training induced changes to kinetic and EMG parameters of running regardless footwear condition, although the more substantial influence in muscle activation has occurred in BF condition. Another key finding of this research is that the human body was capable to adapt to unshod intervention. After

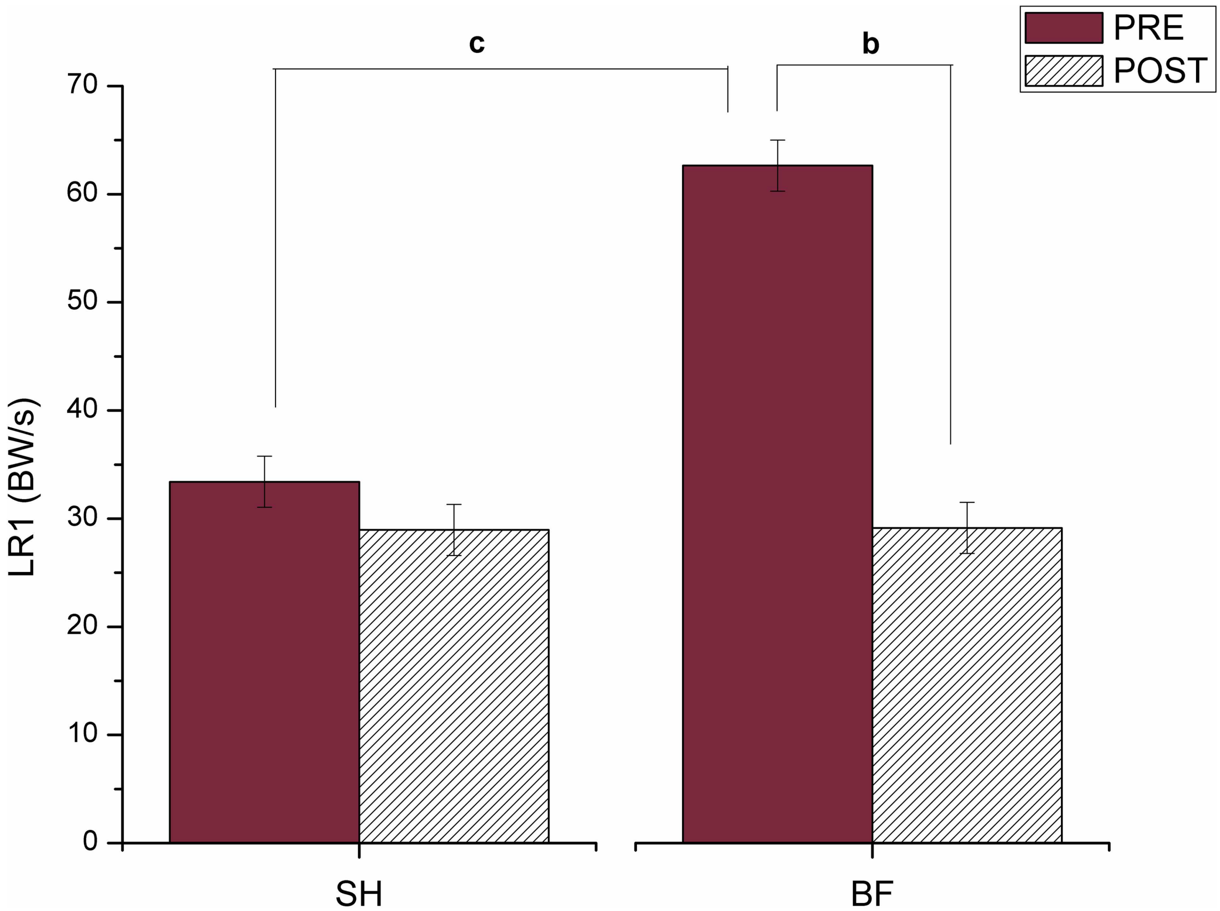

Fig 3. Mean and standard deviation values for the loading rate of first peak (LR) during running shod (SH) and barefoot (BF), in both PRE and POST training, where ([b]) means difference between PRE and POST in BF running; and ([d]) means difference between SH and BF at POST moment.

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training, similar or, even, lower impact force and muscle activation were observed for BF running compared to SH.

Results show that BF running is characterized by less efficient shock attenuation than SH condition in habitual shod runners, as described by previous research [9–11,13,22]. The higher value of LR for BF running before training suggests increased impact forces for this mechanical condition. Recent studies has shown that injured runners present higher values of LR [46,47]. Thus, since this VGRF variable is highly associated with some running injuries [46,48–51], BF running could characterize a initially harmful situation to habitual SH runners [1,9,20,35,36,49]. As expected, EMG signal followed the same behavior observed for external load. Corroborating to previous studies [10,17,23,52] our results showed that habitual SH runners presents higher muscle activation intensity under BF condition. Differences between BF and SH for RMS before training were marked in muscles associated with shock absorption, such as TA, VL and BCF [40,41,50]. This finding indicates the muscle behavior observed in this study was possibly a response to the greater impact forces presented by habitual SH runners during BF condition. In the presence of higher mechanical load, muscles may increase their activation to help in shock absorption. According to literature, greater muscle activation is related to injuries [51,53–56], high cost energy and less efficient

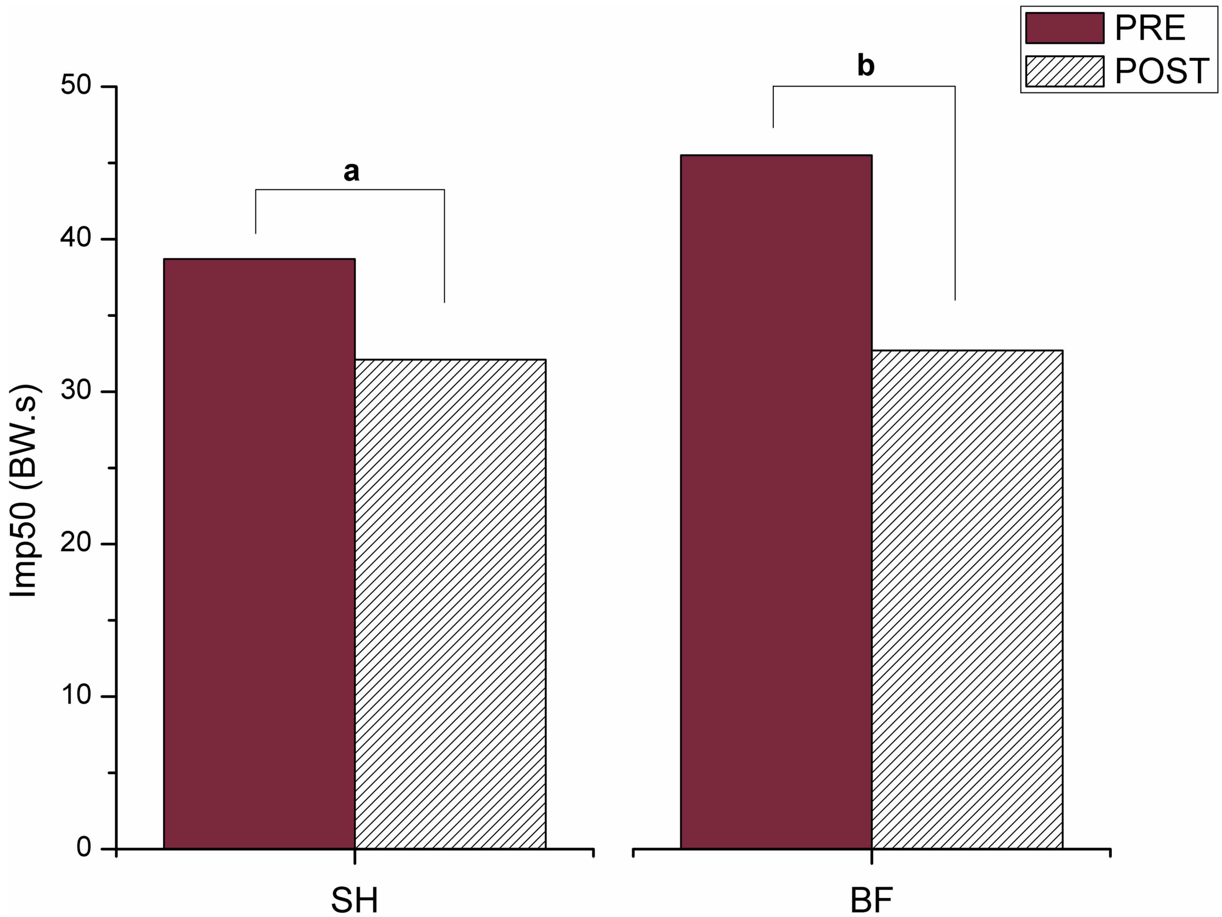

Fig 4. Mean and standard deviation values for the Impulse during the first 50 ms of stance (Imp50) during running shod (SH) and barefoot (BF), in both PRE and POST training, where ([a]) means difference between PRE and POST in SH running; and ([b]) means difference between PRE and POST in BF running.

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running economy [10,17,23,25,30,57]. Therefore, habitual SH runners in their first attempt in this condition could have their protection and performance impaired.

The BF training induced to similar values of LR and Imp50 for BF and SH running, whereas BF condition presented smaller magnitude of impact peak of VGRF (Fy1) after training. Results show that runners adapted to BF condition potentially experiences diminished impact forces during BF running compared to SH running, corroborating to the findings reported by Divert et. al. [10], Lieberman et. al. [6] and Squadrone et. al. [12]. Accordingly, few differences were observed in muscle activation intensity between BF and SH running after intervention. Both ways of running presented similar RMS values after training for all muscles, except BCF. As such, runners adapted to the absence of footwear may be as efficient in BF as in SH running [57]. Additionally, BF running could be seen as a favorable training context to habitual BF runners, where they experiences the similar muscle activation of SH condition, but with decreased impact forces.

Both SH and BF running showed reduced values for variables related to shock attenuation, as Fy1 and Imp50, after training. These results suggest the unshod training improves mechanical load control and shock attenuation in BF and, also, in SH running. Considering these VGRF variables represent the impact forces and energy absorbed by human body structures,

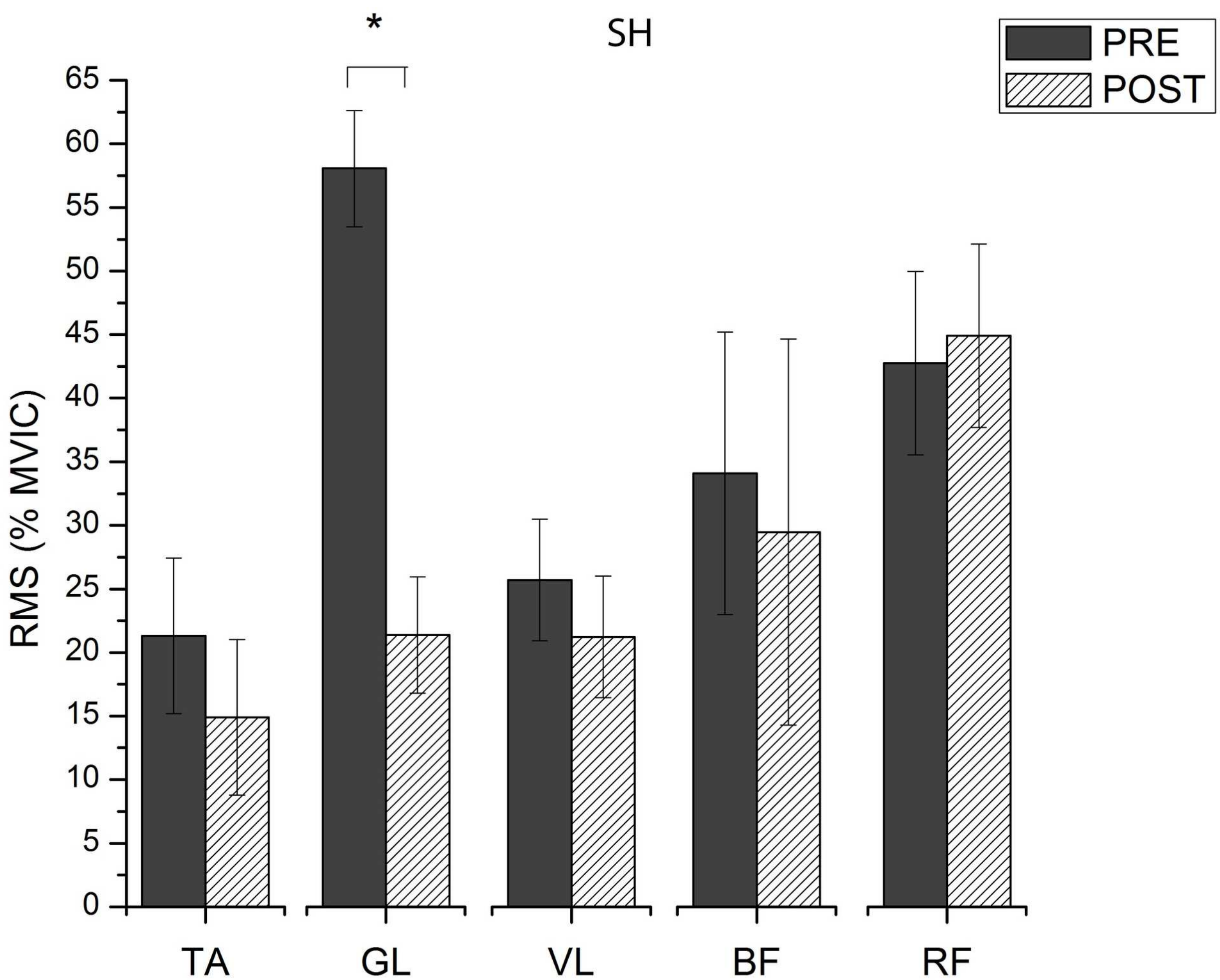

Fig 5. RMS values (% of MVIC) during stance phase of SH running, before (PRE) and after (POST) training.

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some studies associate them to running injuries [48–51]. Supported by these studies, results suggest an improved protection and reduced injury risk after BF training for both ways of running (BF and SH). Nevertheless, it is important to highlight the association between GRF variables and running injuries is still controversial and our assertion is a mere analysis of the potential of risk.

According to the literature, reduced impact forces are related to switching from rearfoot to a mid/forefoot strike pattern and to alterations in spatiotemporal parameters (as stride length and frequency) induced by BF condition [6,10,12,13,24,58]. As SH running also reduced impact forces after BF training, our results suggest these alterations induced by BF condition might be incorporated by runners during SH running. As expected, changes were observed for muscle activation in both SH and BF running as response to the 16-week progressive BF training. Almost all muscles reduced their activation intensity from PRE to POST training for both conditions. However, this reduction was statistically significant for 4 muscles in BF running, whereas only GL was significantly influenced by training in SH running. Although kinematic data was not measured, reduction in activation intensity of muscle may reflect the absence of impact shock to absorb induced by possible switching from rearfoot to a mid/forefoot strike pattern. Such results suggest neuromuscular adaptations in response to training, that could be

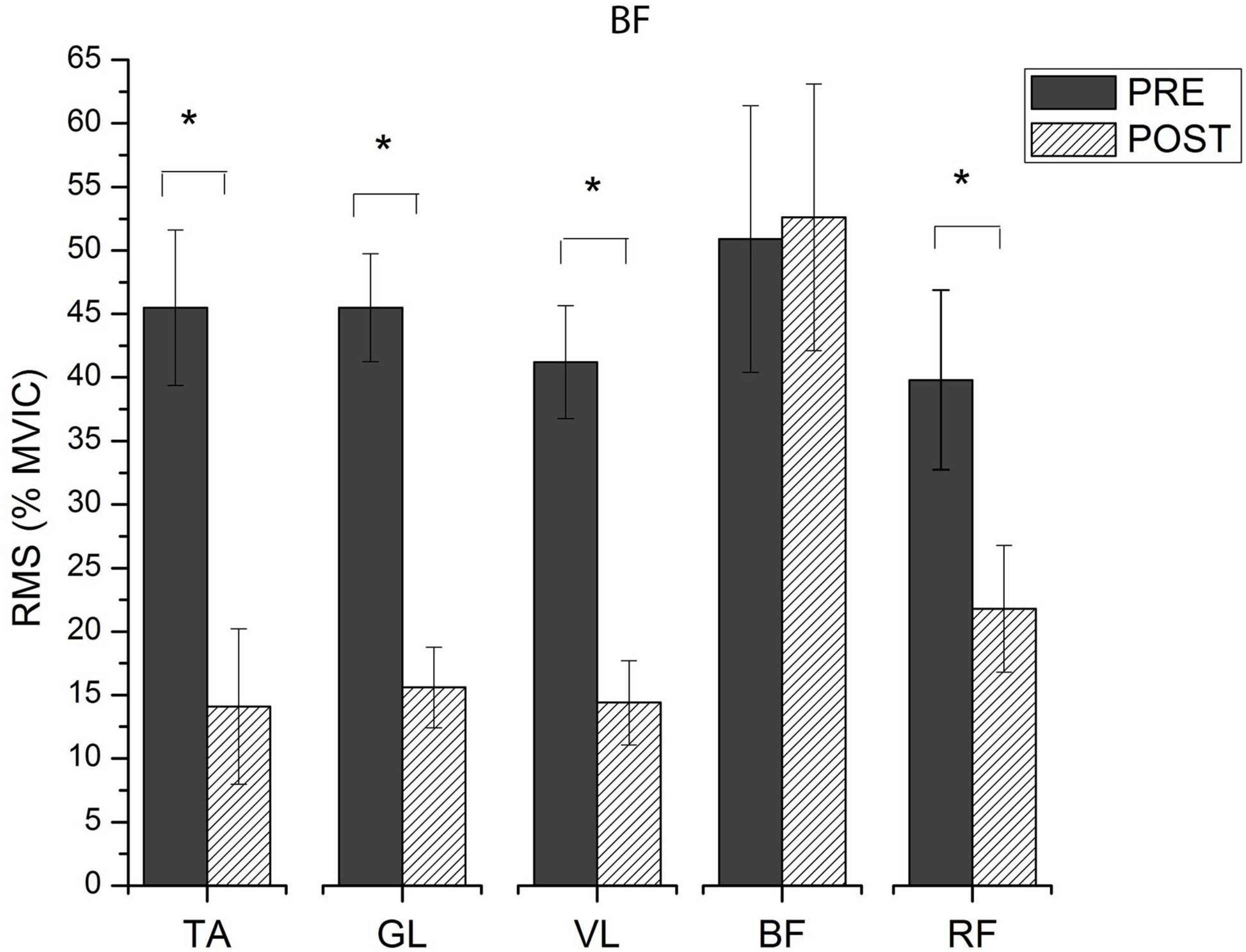

Fig 6. RMS values (% of MVIC) during stance phase of BF running, before (PRE) and after (POST) training.

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related to more efficient muscle recruitment pattern and improved modulation of muscle activation in both passive and active phase of running [10,17,23,41,59,60]. As muscles play an important role as shock absorbers during running [40,41,50], the reduced RMS may reflect the neuromuscular response of Central Nervous System (CNS) to the diminished impact forces observed after training, mainly in BF running. Another possible reason for the reduced intensity of muscle activation after training may be an improvement in the stretch-shortening cycle. Researchers report the kinematics and foot strike pattern induced by unshod running possibly improves the use of storage elastic energy [6,16,57]. Results also suggest the chronic adaptations in muscle activation intensity to the 16 weeks of progressive unshod training seem to be more expressive in BF running. Due to movement specificity and learning effects (since all runners were habitual SH), BF running may have been more sensitive to our intervention than SH running. It is important to notice that many of these changes on GRF and EMG variables in habitual SH runners may be also observed for barefoot simulated running, achieved by minimalist shoes, but in different magnitudes [12,14,17,58,61,62]. The strike pattern is another relevant aspect that must be considered on the study of BF running training. Although strike pattern may limit BF effects [8,63], it seems to be also determined by footwear condition [64,65]. Due to its complexity, strike pattern appears as an issue that should be investigated more deeply.

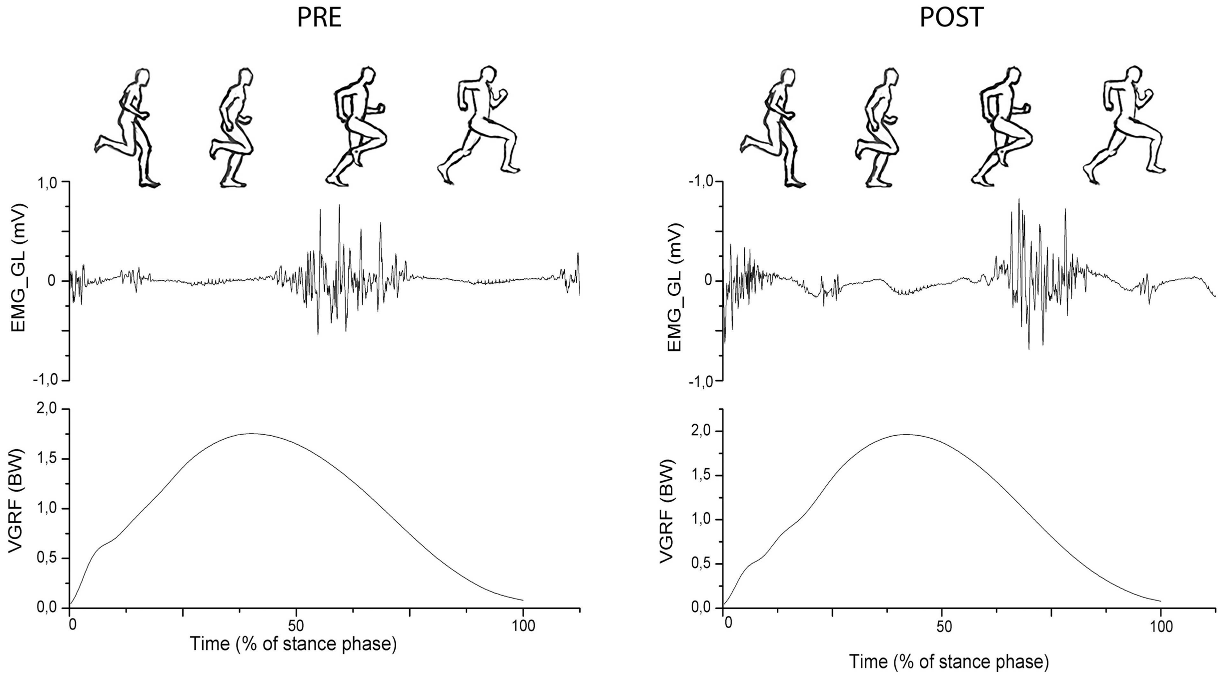

Fig 7. Illustrative average VGRF curves, raw EMG signal of m. gastrocnemius lateralis (GL) and running cycle of stance phase, for one participant, during BF running in before (PRE) and after (POST) training.

doi:10.1371/journal.pone.0167234.g007

Some limitations should be considered in interpretations of findings. First, the sample size that completed the study protocol may restrict interpretation of results. Indeed, the final sample size is small, but different scenery would not be possible with the experimental design adopted. The difficulty of maintaining participants involved in an experimental protocol based on arduous and long training, as barefoot running for 16 weeks, must be highlighted and considered. Notwithstanding, effect size was calculated to express the reliability and sensitivity of our results. The effect size is a complementary statistical tool usually adopted in order to reveal the size of effects [66], being particularly meaningful for our study to assure more reliability and certainty to our results, even with a small sample. Another limitation is that running tests were performed on treadmill, what could change running style and mechanical responses [41,51]. To minimize this limitation, runners experienced in treadmills were recruited and familiarization period was provided. The analysis of EMG signal, considering the entire stance phase, also appears as a limitation of the study. Nevertheless, our intention was to obtain information about the global effort exerted by muscles during running phase. The absence of a control group also appears as a limitation of the study. To reduce this limitation, all participants kept their normal running training routine, without significant change in training volume and intensity, in order to guarantee that the inclusion of BF intervention would be the only modification in their training periodization. Additionally, the mechanisms behind the changes observed in this study were not accessed. Further investigations about the influence of BF training on different biomechanical parameters, including kinematics and foot strike pattern, are encouraged. Finally, our results are protocol dependent and should be extrapolated to other situations carefully. Other populations and different BF interventions may induce distinct mechanical responses for the conditions tested.

Thus, this research provides information to better understand the adaptation's process to BF condition and about the consequences of adopting BF running as training approach. A progressive BF training was effective to alter VGRF and EMG parameters of running in habitual shod runners, regardless footwear condition (SH or BF). Results suggest the use of BF condition could be an efficient training strategy to reduce impact forces and to decrease muscle activation intensity, not only in BF running, but also in SH running, although changes in muscle activation has been more expressive for BF condition. The BF condition arises as an option and practicable training approach to improve mechanical load control and to enhance muscle recruitment for both SH and BF running.

Conclusion

A 16-week progressive barefoot (BF) training altered running kinetics and changed variables of ground reaction force (GRF) related to external forces in habitual shod runners. Additionally, muscle activation intensity of habitual shod runners was influenced by BF training. Alterations occurred in both shod (SH) and barefoot (BF) running. Hence, a progressive BF running training could be used as strategy to improve mechanical load control and shock attenuation in running, regardless footwear condition. Moreover, an intervention based on BF condition reduces muscle activity intensity in long-term, mainly in BF running, what could be a possible and useful approach to improve running economy.

Supporting Information

S1 File. Trend Statement Checklist of this paper.
(PDF)

S2 File. Study protocol submitted and approved by local ethic committee.
(PDF)

S3 File. Data used in the statistical analysis.
(XLSX)

S4 File. Statistical data reports from SigmaStat 3.5.
(RTF)

S1 Table. Summary statistics (mean and standard deviation) of RMS data (% MVIC) during stance phase for shod (SH) and barefoot (BF) running, before (PRE) and after (POST) training ($p<0.05$).
(DOCX)

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Comparison of Minimalist Footwear Strategies for Simulating Barefoot Running: A Randomized Crossover Study

Abstract

Possible benefits of barefoot running have been widely discussed in recent years. Uncertainty exists about which footwear strategy adequately simulates barefoot running kinematics. The objective of this study was to investigate the effects of athletic footwear with different minimalist strategies on running kinematics. Thirty-five distance runners (22 males, 13 females, 27.9 ± 6.2 years, 179.2 ± 8.4 cm, 73.4 ± 12.1 kg, 24.9 ± 10.9 km.week^{-1}) performed a treadmill protocol at three running velocities (2.22, 2.78 and 3.33 m.s^{-1}) using four footwear conditions: barefoot, uncushioned minimalist shoes, cushioned minimalist shoes, and standard running shoes. 3D kinematic analysis was performed to determine ankle and knee angles at initial foot-ground contact, rate of rear-foot strikes, stride frequency and step length. Ankle angle at foot strike, step length and stride frequency were significantly influenced by footwear conditions ($p<0.001$) at all running velocities. Posthoc pairwise comparisons showed significant differences ($p<0.001$) between running barefoot and all shod situations as well as between the uncushioned minimalistic shoe and both cushioned shoe conditions. The rate of rear-foot strikes was lowest during barefoot running (58.6% at 3.33 m.s^{-1}), followed by running with uncushioned minimalist shoes (62.9%), cushioned minimalist (88.6%) and standard shoes (94.3%). Aside from showing the influence of shod conditions on running kinematics, this study helps to elucidate differences between footwear marked as minimalist shoes and their ability to mimic barefoot running adequately. These findings have implications on the use of footwear applied in future research debating the topic of barefoot or minimalist shoe running.

Introduction

The last few years, barefoot and barefoot-like running has been widely discussed as a natural alternative to traditional shoe running in recreational sports [1,2]. The long-believed benefits of stable and cushioned running shoes are questioned by findings that show lower prevalence of foot disorders [3,4], improved running economy [5,6] and lower impact forces in barefoot runners [5,7,8]. These effects are probably due to alterations in lower extremity running biomechanics. Numerous studies [5–8] have shown a higher rate of rear-foot strikes (RFS) during running with shoes whereas barefoot running produces more forefoot strikes during initial ground contact. According to Lieberman et al. [7], this is mainly caused by cushioning of the heel, which allows "a runner to rear-foot strike comfortably" by reducing peak ground reaction forces. However, forefoot running patterns are not only a result of missing shoe cushioning. They also occur more frequently at increased running speeds, are influenced by the running surface, and are dependent on individual habituation [7,9–11]. Hence, the reported kinematic and kinetic characteristics of barefoot running [1,7,8] are more likely due to a more plantarflexed footstrike than to the footwear condition [12]. Although the forefoot ground contact and lower impact forces are also often believed to be associated with a reduced injury risk, no conclusive evidence exists on the influence of regular barefoot running on lower extremity injury rates [9,13–16].

Running with bare feet is sometimes restricted by hard and unsafe ground conditions or low temperatures. In recent years, the development of barefoot-like footwear with reduced cushioning and/or high flexibility has gained increasing attention among numerous manufactures. In the literature, shoes with minimal cushioning and weight, and/or increased sole flexibility are typically referred to as "minimalist shoes", "lightweight shoes" or "barefoot shoes". The effectiveness of minimalist footwear for simulating barefoot running is mostly unclear due to inconsistent findings in the literature. Squadrone and Gallozzi [17] found similar ankle angles at initial foot ground contact during barefoot running and running with uncushioned minimalist shoes. Both conditions were significantly different from standard shoe running. Bonacci et al. [18] reported significant differences between cushioned minimalist shoes with ultraflexible soles and barefoot condition in knee flexion and ankle dorsiflexion during initial ground contact. Taking the discrepant findings into account, it seems reasonable that shoe cushioning plays an important role in the simulation of barefoot running. First data on the influence of different midsole thicknesses compared to no cushioning (barefoot) were previously shown regarding joint stiffness, vertical ground reaction force and strike index [19]. However to our knowledge, no study has yet compared the effects of minimalist shoes with different characteristics regarding cushioning and weight on running kinematics in one study protocol. The differentiation between these effects may help to understand the relevant factors of barefoot running simulation.

The objective of this study was to determine the influence of shoe cushioning and flexibility on treadmill running ankle and knee kinematics in habitual shod runners. Two varying minimalist shoe models of different cushioning were compared with barefoot and standard footwear conditions at three running speeds. Considering previous findings, we hypothesize that kinematics during running with uncushioned minimalist shoes are closer to barefoot conditions than cushioned minimalist shoes.

Methods

This study had a randomized crossover design and took place in a University Biomechanics Laboratory. Ethical approval for the study was obtained from the ethics committee of the medical association Hamburg (protocol no. PV4271). Prior to the study all participants provided their written informed consent to participate in this study. The study followed the principles of the Helsinki Declaration.

For inclusion, participants had to be recreational runners, running at least 12 km per week, between 18 and 45 years of age and free of orthopedic, neurological or musculoskeletal disorders for the past six months. Participants were not allowed to have any experience with minimalist running shoes. Both, habitually forefoot and rear-foot strikers were considered for participation.

In this study, four different conditions were applied in random order: barefoot running, standard running shoe running, cushioned minimalist shoe running and uncushioned minimalist running shoe running (Fig 1). The order was counterbalanced between the first thirty-two participants and partly balanced between the last three participants. All shoes were commercially available. An Asics GT-2160 (ASICS, Kobe, Japan) was used as standard running footwear. It has an ethylene-vinyl acetate midsole, an arch support, 12 mm heel-forefoot offset and a weight of 314 g (woman's shoe, US size 6.5). As a representative of cushioned minimalist footwear, a Nike Free 3.0 (NIKE, Beaverton, OR, USA) with a 4 mm heel-forefoot offset, no arch support and a weight of 189 g was used. A Leguano (LEGUANO, St. Augustin, Germany) was used for uncushioned minimalist footwear. It has a polyvinyl chloride midsole, 0 mm heel-forefoot offset, no arch support and a weight of 137 g. Cushioning properties of shoes were measured using a drop tester, designed according to the American Society for Testing and Material (ASTM's) "Standard Test Method for Shock Attenuating Properties of Materials Systems for Athletic Footwear" (F1976, ASTM International, West Conshohocken, PA, USA). An indenter of 35 mm diameter with a load cell completed 10 impacts on the heel of one shoe of each footwear condition (US size 6.5). For the standard running shoe the peak impact force was 750 N with a maximum impact depth of 7.70 mm. The cushioned minimalist shoe produced a peak impact force of 845 N and a maximum impact depth of 7.49 mm. Peak impact force of the uncushioned minimalist shoe was 2200 N and the maximum impact depth 1.85 mm.

The primary outcome was ankle angle at footstrike. Secondary outcomes were knee angle at footstrike, rate of rear-foot strike (RFS), step length and stride frequency. Kinematic analysis was performed using a three-dimensional 8-camera infrared motion analysis system operating at 200 Hz (VICON, Oxford, UK). The cameras were placed around a treadmill (Ergo-Fit TRAC 4000, ERGO-FIT GmbH & Co. KG, Pirmasens, GERMANY) for data collection with minimized marker occlusions. According to the Plug-in-Gait model (VICON, Oxford, UK), sixteen retro-reflective markers (14 mm diameter) were located bilaterally at anatomical bony landmarks of the pelvis, thigh, knee, shank, ankle, and foot as used in a prior study [20]. To enable calculation of knee and ankle joint angles, the following anthropometric measures were obtained: bilateral leg length, knee width, ankle width, height, and body mass.

After randomization of footwear conditions, participants ran each condition at three different velocities (v_1 = 2.22 $m \cdot s^{-1}$, v_2 = 2.78 $m \cdot s^{-1}$, v_3 = 3.33 $m \cdot s^{-1}$). All markers remained on the identical position, except for the foot markers. They were adjusted for each condition on the surface of the shoe in reference to the foot. Calcaneal and second metatarsal marker were kept at the same height and level of the shoe. The same distance to the ground was determined by the use of a caliper. Additionally, standing calibrations were taken separately for each footwear condition. This individual capture of calibration trials were used to create a biomechanical model of the lower body (Plug-In-Gait).

An accommodation to the treadmill and a warm-up period by walking in a self-selected velocity was conducted. Participants were asked to indicate readiness and the treadmill was accelerated to 2.22 $m \cdot s^{-1}$ with a rate of 0.2 $m \cdot s^{-2}$. Thirty seconds afterwards, data recording started for fifteen seconds over two consecutive sessions during each trial. The second recording was taken as a backup. After data collection for the first velocity, participants were given a one-minute rest. The same procedure was applied for 2.78 $m \cdot s^{-1}$ and 3.33 $m \cdot s^{-1}$. Then subsequently, the

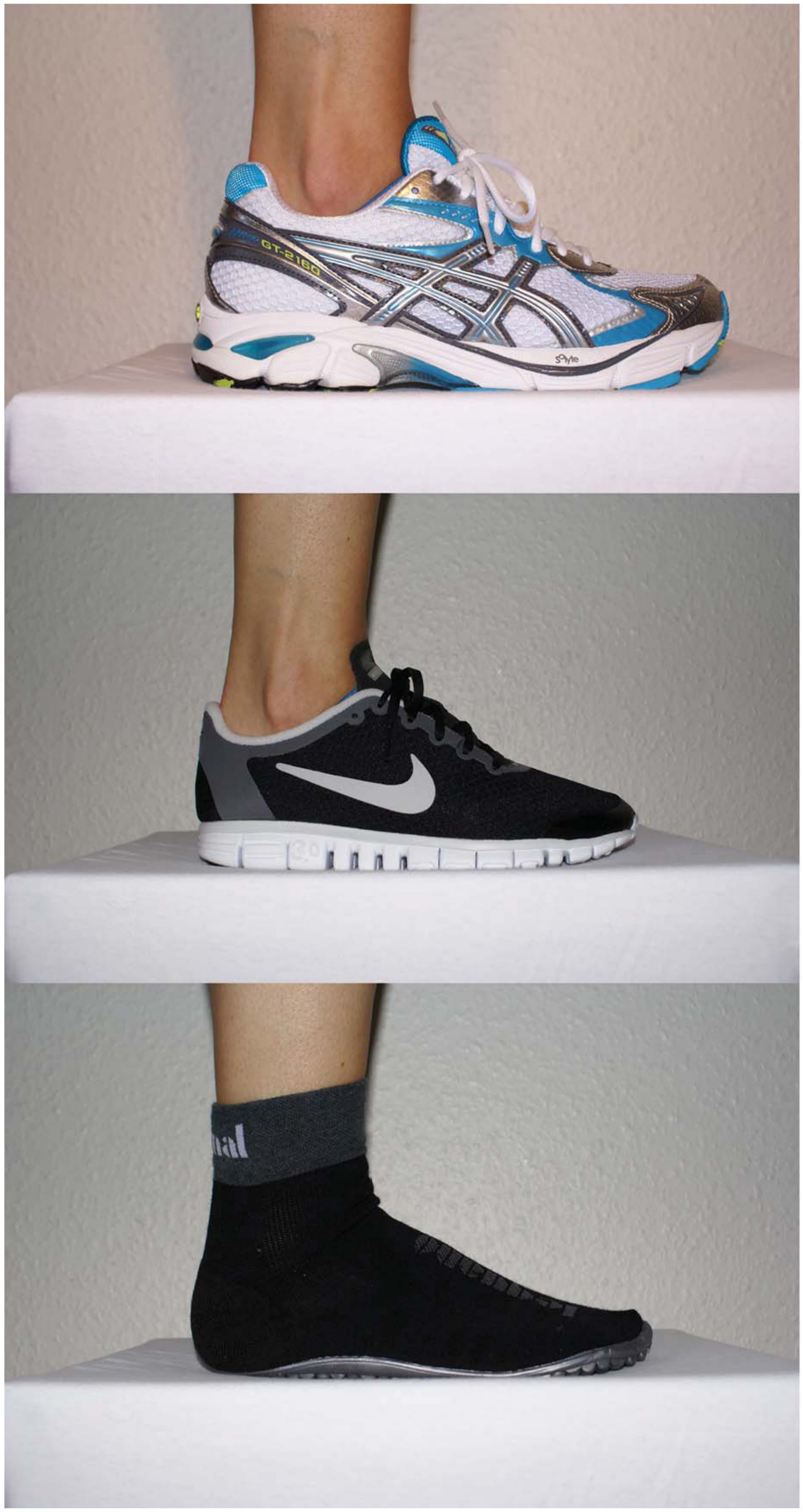

Fig 1. Shoe conditions. (top image = Asics GT-2160, center image = Nike free 3.0, lower image = Leguano).

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footwear condition changed according to the randomization protocol and the test procedure was repeated equally for each condition.

Kinematic data was filtered using a Woltring filtering routine (mean square error = 15). All data processing was done using Vicon Nexus 1.7.1 and Polygon 3.5.1 (VICON, Oxford, UK). Footstrike was defined when the vertical velocity of the distal heel marker changed from negative to positive. This method was recently described as the most valid and reliable method for the kinematic identification of foot strike [21].

Outcomes of interests were ankle and knee angles in the sagittal plane during the phase of initial ground contact. Ankle dorsiflexion and knee flexion angles were recorded during the whole gait cycle and analyzed only during the last 10% of gait cycle (before the identified ground contact). The time interval was utilized in order to address the sources of error that occur when the velocity of the distal heel marker is used to identify the initial ground during different footwear conditions with and without cushioning. Gait cycle data were compared to neutral standing position. A virtual biomechanical model was developed for each subject and condition. Rate of RFS was determined visually by examination of a lateral high-speed video independently by two investigators.

Ankle and knee kinematics, step length and stride frequency were analyzed for ten consecutive gait cycles. Individual knee and ankle angle kinematic data for each leg were processed using Matlab software (Mathworks, Natick, MA, USA). Trials were normalized to 100% of gait cycle. The Vicon motion capture system is a reliable tool for the analysis of gait kinematics [22,23].

To determine differences between shoe conditions, we calculated mixed models [24] for interesting metric dependent variables ankle and knee angle at footstrike, step length, and stride frequency. To adjust for the cluster structure, participants were included as a random factor. The interesting main effect of shoe condition was included as a fixed effect as well as the factors running velocity and leg side. Tentatively, two-way interactions between Shoe×Sex and Shoe×Side (left/right) were added and kept in all models if significant. Furthermore, a Bonferroni post hoc test was conducted between shoe conditions. Cohen's d was calculated by using the difference between the means divided by the pooled standard deviation. A generalized estimating equation model for a repeated measures logistic regression was calculated for the dichotomous variable "rate of RFS". For shoe comparisons odds ratios are presented. The SPSS statistical package Version 21 (SPSS, Armonk, NY, USA) was used for all statistical procedures.

Results

Thirty-five recreational distance runners took part in the study (22 males, 13 females, age = 27.9 ± 6.2 years, height = 179,2 ± 8,4 cm, mass 73.4 ± 12.1 kg, mileage = 24.9 ± 10.9 $km \cdot week^{-1}$). All participants were habitual shod runners who were used to treadmill running. Two participants were habitual forefoot runners.

Kinematic parameters for all running conditions are shown in Table 1. Footwear conditions and running velocity significantly ($p<0.001$) influenced ankle angles, stride frequency and step length (Table 2). Ankle angles differed with statistical significance ($p<0.001$) between all shoe conditions for each velocity except for comparison of cushioned minimalist and standard shoe condition ($p = 0.674$) (Table 3). Running barefoot reduced the dorsiflexion by 1.73° (95% CI 0.99°;2.48°) compared to uncushioned minimalist shoes, 5.52° (95% CI 4.77°;6.27°) compared to cushioned minimalist shoes and 5.68° (95% CI 4.96°,6.47°) compared to standard shoes. The uncushioned minimalist running condition produced a 3.78° (95% CI 3.04°;4.53°) lower dorsiflexion during foot landing than the cushioned minimalist running. Additionally, running velocity ($p<0.001$), body weight ($p<0.05$) and weekly mileage ($p<0.05$) significantly influenced

Table 1. Group mean (SD) temporal-spatial and kinematic parameters for 2.22, 2.78 and 3.33 $m^{\cdot}s^{-1}$.

	Barefoot	Uncushioned minimalist shoe	Cushioned minimalist shoe	Standard running shoe
		2.22 $m^{\cdot}s^{-1}$		
Ankle angle at footstrike (°)	6.90 (5.95)	8.69 (6.12)	11.66 (4.88)	11.14 (4.16)
Knee angle at footstrike (°)	10.77 (5.26)	10.53 (4.71)	10.07 (4.24)	10.02 (4.51)
Stride frequency (steps$^{\cdot}$minute^{-1})	160.87 (5.46)	158.14 (6.06)	155.70 (7.78)	154.47 (5.14)
Step length (cm)	82.98 (2.82)	84.44 (3.25)	85.80 (3.83)	86.41 (2.92)
Rate of rear-foot strikes (%)	62.9	74.3	90.0	94.3
		2.78 $m^{\cdot}s^{-1}$		
Ankle angle at footstrike (°)	5.70 (6.46)	7.39 (6.19)	11.57 (4.74)	11.33 (4.24)
Knee angle at footstrike (°)	9.77 (6.99)	10.83 (4.48)	10.27 (5.26)	10.65 (5.24)
Stride frequency (steps$^{\cdot}$minute^{-1})	167.09 (8.18)	164.36 (7.44)	161.68 (7.52)	158.68 (5.98)
Step length (cm)	99.98 (4.91)	101.61 (4.60)	103.30 (4.85)	105.18 (3.96)
Rate of rear-foot strikes (%)	55.7	68.6	92.9	94.3
		3.33 $m^{\cdot}s^{-1}$		
Ankle angle at footstrike (°)	4.68 (7.23)	6.40 (6.80)	10.56 (5.23)	11.85 (4.12)
Knee angle at footstrike (°)	12.56 (5.73)	12.52 (5.27)	12.03 (5.16)	11.40 (4.89)
Stride frequency (steps$^{\cdot}$minute^{-1})	174.85 (9.90)	170.80 (8.52)	168.60 (8.43)	164.84 (7.44)
Step length (cm)	114.74 (6.37)	117.38 (5.83)	118.92 (5.93)	118.15 (6.37)
Rate of rear-foot strikes (%)	58.6	62.9	88.6	94.3

SD standard deviation

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ankle angle at footstrike (Table 2). There was no statistically significant effect of shoe conditions on the knee angle at footstrike (p = 0.239). Effects on the knee angle at footstrike were found for velocity ($p<0.001$) and sex ($p<0.05$). Females produced higher knee angels at footstrike compared to males.

The repeated measures logistic regression analysis showed that rate of rear-foot strikes was not significantly influenced by velocity (p = .294), sex (p = .415) or leg side (p = .234). Significantly different RFS were shown for the different footwear conditions ($p<.001$). During all velocities, the RFS was highest for standard shoe running, followed by cushioned minimalist shoe, uncushioned minimalist shoe and barefoot conditions (Table 1). Statistically significantly different odds ratios were found between barefoot and both cushioned shoe conditions (2.22 $m^{\cdot}s^{-1}$ OR = .188 (95% CI: .075, .471) and OR = .103 (95% CI: .033, .314)) as well as between uncushioned minimalist and both cushioned shoe conditions (2.22 $m^{\cdot}s^{-1}$ OR = .321 (95% CI: .284, 1.207) and OR = .175 (95% CI: .056, .549)). Running barefoot and with uncushioned minimalist shoes did not differ for the rate of RFS (2.22 $m^{\cdot}s^{-1}$ OR = .586 (95% CI: .075, .471)).

Regarding temporal-spatial outcomes, running barefoot, subjects took the smallest steps with the highest stride frequency compared to uncushioned minimalist ($p<.001$), cushioned minimalist ($p<.001$) and standard shoes ($p<.001$). Stride frequency was higher and step length

Table 2. Mixed model effects (p-values) for included factors.

	Footwear	Running Velocity	Leg side	Footwear* Velocity
Ankle angle at footstrike (°)	<.001	.001	.699	.026
Knee angle at footstrike (°)	.239	<.001	.157	.285
Stride frequency (steps.minute-1)	<.001	<.001	.611	<.001
Step length (cm)	<.001	<.001	.622	<.001

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Table 3. Differences (95% CI) and p-values of pairwise comparisons between footwear conditions.

	Ankle angle at footstrike (°)			Knee angle at footstrike (°)			Stride frequency (steps.minute-1)			Step length (cm)		
	Diff. (CI)	P-value	Cohen's d	Diff. (CI)	P-value	Cohen's d	Diff. (CI)	P-value	Cohen's d	Diff. (CI)	P-value	Cohen's d
Barefoot vs Standard running shoe	-5.68 (-6.426; -4.935)	<.001	-1.032	.346 (-.319; 1.010)	.308	.062	8.275 (7.624; 8.926)	<.001	.945	-5,156 (-5,586;-4,726)	<.001	-.357
Barefoot vs Uncushioned minimalist shoe	-1.73 (-2.479;-.988)	<.001	-.946	-.258 (-.923; .407)	.446	.045	3.171 (2.520; 3.823)	<.001	.589	-1,910 (-2,340; -1,480)	<.001	-.244
Barefoot vs cushioned minimalist shoe	-5.52 (-6.267;-4.772)	<.001	-.461	.310 (-.357; .976)	.362	-.047	5.613 (4.962; 6.264)	<.001	.337	-3,442 (-3,872; -3,013)	<.001	-.136
Uncushioned minimalist shoe vs cushioned minimalist shoe	-3.79 (-4.534;-3.039)	<.001	-.464	.568 (-.098; 1.235)	.095	.103	2.441 (1.790; 3.093)	<.001	.268	-1,532 (-1,962; -1,103)	<.001	-.107
Uncushioned minimalist shoe vs standard running shoe	-3.95 (-4.692;-3.201)	<.001	-.527	.604 (-.061; 1.268)	.075	.123	5.104 (4.452; 5.755)	<.001	.615	-3,246 (-3,676;-2,816)	<.001	-.222
Cushioned minimalist shoe vs standard running shoe	-.16 (-.908;.587)	.674	-.037	.036 (-.631; .702)	.917	.019	2.662 (2.011; 3.313)	<.001	.317	-1,714 (-2,143; -1,284)	<.001	-.117

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shorter during running with uncushioned minimalist shoes compared to cushioned minimalist shoes (<.001). The standard running shoe condition led to the highest step length and smallest stride frequency. Running velocity also influenced stride frequency and step length significantly ($p<0.001$) (Table 2).

Discussion

The objective of this study was to identify minimalist footwear characteristics responsible for the simulation of barefoot running kinematics. In a random and counterbalanced order cushioned and uncushioned minimalist shoes were compared to standard cushioned shoe and barefoot conditions. The study's hypothesis was that kinematics during running with uncushioned minimalist shoes are closer to barefoot conditions than cushioned minimalist shoes.

In agreement with other studies [20,25,26], we found significant differences in ankle kinematics and step length as well as stride frequency between barefoot running and all shod running conditions. The most remarkable differences were observed between barefoot and cushioned shoe conditions. The main finding of this study was that minimalist shoes differ in their ability to simulate barefoot running. All outcome measures except for the knee angle were significantly different between cushioned and uncushioned minimalist shoes.

Minimalist footwear has been designed in order to replicate barefoot running and is increasingly used by recreational runners [27]. While the impact of barefoot running on biomechanics is widely discussed, it has not yet been defined which shoe characteristics adequately meet the criteria to mimic barefoot running biomechanics. Hence, current minimalist shoe models differ in their cushioning and flexibility characteristics and produce uncertainty regarding the comparability of running in barefoot-simulating footwear and real barefoot running. Our results show that the effectiveness of minimalist footwear for simulating barefoot running kinematics seems to be influenced by the cushioning properties. The findings are in accordance with the findings of Squadrone & Gallozzi [17], who used a minimalist shoe (Vibram five-fingers) similar to the one used in our study (no cushioning, 0 mm heel-forefoot offset). Contrary to our finding, the authors observed no differences in the ankle dorsiflexion angle at foot strike between barefoot and minimalist shoe running. The findings reported by Bonacci et al [18], who used the same cushioned minimalist shoe (Nike Free 3.0), are comparable to our results concerning ankle kinematics. They reported significant differences in knee and ankle kinematics between minimalist shoe and barefoot running conditions. However, the comparability between both studies is further limited due to different populations used. While Squadrone & Gallozzi [17] investigated habitually barefoot runners, Bonacci et al [18] analyzed subjects that were highly trained but habitually shod. Taking these considerations and our results into account, one can say that footwear with less heel-forefoot offset and less cushioning seem to be more capable of replicating barefoot running than shoe models without these characteristics.

In this study, running shod led to increased ankle angles at footstrike compared to barefoot running. These findings are in agreement with several other studies [7,20,26,28]. The lack of differences in knee angles, however, are inconsistent compared to other research [18,25]. This might be explained with the effect of gender on the knee angle shown in this study or the different populations investigated. Our participants were recreational and habitually shod runners. Other studies compared habitually shod and habitually barefoot runners [7], highly trained runners [18], exclusively male runners [25] or runners that were just included when being habitual shod heelstriker [29]. The lower ankle dorsiflexion angles in our study indicate a flatter foot at landing for barefoot and uncushioned shoe running. Hence, it is not surprising that both running conditions significantly decreased the rate of rear-foot strikes among participants. Nevertheless, it should also be noted that during barefoot and minimalist running, the

RFS was still present in more than 50% of the participants. The flatter foot placement at initial contact is a typical characteristic of barefoot running [7,28,30]. It is generally believed that this is a common strategy in order to generate lower impact forces during initial ground contact [28]. Our data indicate that the lack of cushioning might be predominantly responsible for this effect. However, it should also be considered that this landing pattern seems to depend on the running surface and speed as well as on the subject [30].

Furthermore, our research showed an increase of stride frequency and a decrease of step length when running barefoot. These findings have been reported in many other studies for healthy adult [9], adolescent [5] and infantile [20,31] populations. They are probably a consequence of a smaller impact force during landing [17] but might also be explained by a more cautious gait due to higher proprioception [32]. It has been previously shown that taking smaller steps reduces the impact force peak and loading rates [33] and may prevent impact-related injuries [34].

Some limitations should be considered in interpretation of findings. First of all, neither participants, nor researchers were masked to the running condition, which may have induced bias towards the benefits of a particular running condition. Nevertheless, no information was given to participants on the study hypothesis. Furthermore, the marker placement on the shoe surface causes the second metatarsal head marker to be slightly more superior compared to the attachment directly on the skin. Other studies [29,35] addressed this problem by cutting windows into the shoe's upper material or using sandals [36]. We adjusted the superior-inferior position of the heel marker and used separate calibrations for each condition. The most important limitation in this study is the lack of ground reaction force data allowing direct conclusions on running kinetics. Therefore, the discussion of impact forces during landings in this study remains mainly speculative. Our study also lacks the ability to make conclusions about the footwear's influence on injury risk or prevention. In contrast to the widely discussed beneficial effects of minimalist footwear, two recent studies show first evidence about an increased injury risks due to minimalist footwear training [15,16]. In accordance with other studies [37,38], we conclude that well-powered prospective studies are needed to elucidate relationship between the influence of shoes and running injuries.

Conclusion

In this study, running kinematics of healthy long distance runners were influenced by footwear and running velocity. Ankle dorsiflexion angles and rate of rear-foot strikes were lowest during the barefoot running condition and increased with augmented cushioning properties of footwear. Running kinematics for uncushioned minimalist shoes were closer to barefoot running kinematics than those of cushioned minimalist shoes. The results indicate that cushioning plays an important role for simulating barefoot running kinematics. These findings have implications on the use of footwear used in future research debating the topic of barefoot or minimalist shoe running.

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Footwear Decreases Gait Asymmetry during Running

Abstract

Previous research on elderly people has suggested that footwear may improve neuromuscular control of motion. If footwear does in fact improve neuromuscular control, then such an influence might already be present in young, healthy adults. A feature that is often used to assess neuromuscular control of motion is the level of gait asymmetry. The objectives of the study were (a) to develop a comprehensive asymmetry index (CAI) that is capable of detecting gait asymmetry changes caused by external boundary conditions such as footwear, and (b) to use the CAI to investigate whether footwear influences gait asymmetry during running in a healthy, young cohort. Kinematic and kinetic data were collected for both legs of 15 subjects performing five barefoot and five shod over-ground running trials. Thirty continuous gait variables including ground reaction forces and variables of the hip, knee, and ankle joints were computed for each leg. For each individual, the differences between the variables for the right and left leg were calculated. Using this data, a principal component analysis was conducted to obtain the CAI. This study had two main outcomes. First, a sensitivity analysis suggested that the CAI had an improved sensitivity for detecting changes in gait asymmetry caused by external boundary conditions. The CAI may, therefore, have important clinical applications such as monitoring the progress of neuromuscular diseases (e.g. stroke or cerebral palsy). Second, the mean CAI for shod running (131.2 ± 48.5; mean ± standard deviation) was significantly lower ($p = 0.041$) than the CAI for barefoot running (155.7 ± 39.5). This finding suggests that in healthy, young adults gait asymmetry is reduced when running in shoes compared to running barefoot, which may be a result of improved neuromuscular control caused by changes in the afferent sensory feedback.

Introduction

Falls are one of the main causes for fatal injury and hospitalization in older adults [1–3]. Identifying factors that contribute to falls has become an important objective in clinical geriatric research. The absence of footwear was identified as an important risk factor for the occurrence of falls in elderly adults [4]. The reduced risk of falls reported in the mentioned study concurs

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Abbreviations: CAI, Comprehensive Asymmetry Index; EV, Eigenvalue; PCA, Principal Component Analysis; PC-vector, Principal Component Vector.

with other studies that assessed the effect of footwear on the likelihood of falls or balance [5–7]. In addition to mechanical factors potentially causing a reduced risk of falls when wearing footwear [8], it is also possible that footwear may alter the type or amount of afferent sensory feedback causing improved neuromuscular control. If footwear does in fact improve neuromuscular control, then such an influence might already be present in young, healthy adults, long before it may become clinically relevant in the prevention of falls. A feature that is often used to assess neuromuscular control of motion is the level of asymmetry between the contra-lateral limbs during gait. In fact, in many neurophysiological disorders such as stroke [9, 10], Parkinson's disease [11], or cerebral palsy [12], gait asymmetry can be seen as one of the indicators of the severity of the condition.

One challenge when assessing gait asymmetry in healthy, young adults is that the kinematic and kinetic differences between the left and right lower limbs are rather small compared to the inherent movement variability. In addition, one could argue that gait asymmetry is a characteristic that applies to several body segments simultaneously [13–15], especially when investigating changes caused by external boundary conditions such as footwear. Therefore, a new asymmetry index, a *comprehensive asymmetry index* (CAI), is required that is especially sensitive to changes in gait asymmetry caused by external boundary conditions. Three actions can be taken in order to increase the sensitivity of the CAI: First, all available kinematic and kinetic data should be incorporated to provide an all-encompassing assessment of an individual's lower limb gait asymmetry. This allows considering the moving human body as a whole system rather than analysing individual variables [16, 17]. Second, the waveforms of all gait variables should be normalized to their standard deviation waveform to account for asymmetry caused by the natural variability of the movement. This should be done since previous studies indicated that gait asymmetry may only be relevant when it exceeds the inherent variability of a gait variable [13, 18]. Third, a principal component analysis (PCA) can be used to filter out the covariate structure of gait asymmetry [16, 19]. This is based on the assumption that gait asymmetry observed in one variable can only occur if it is accompanied by asymmetries in other variables [19]. To give a simplified example: contra-lateral asymmetries in the knee joint angle can only occur within a given motion task, if ankle and/or hip angles change accordingly.

In summary, a CAI with enhanced sensitivity to detect gait asymmetry changes is required in order to investigate whether footwear influences the level of asymmetry between the contralateral limbs during gait. A reduction in gait asymmetry may support previous research indicating that footwear improves neuromuscular control. The new CAI should be tested on a highly automated movement, i.e. running, rather than more complex movements in which higher cognitive functions are more likely to interfere with the movement pattern and may potentially affect gait asymmetry.

Therefore, the objectives of the study were (a) to develop a *comprehensive asymmetry index* (CAI) that can be used to study changes in gait asymmetry caused by external boundary conditions such as footwear, and (b) to use the CAI to investigate whether footwear influences gait asymmetry during running in a healthy, young cohort. Based on the aforementioned studies, it was hypothesized that footwear decreases gait asymmetry as compared to barefoot running.

Methods

Study participants

Fifteen subjects were recruited for this study, seven females and eight males: age: 25.4 (SD 4.4) years; height: 1.74 (SD 0.07) m; mass: 71.2 (SD 8.4) kg. The subjects were healthy, with no neuromuscular or neurological disorders, and had no lower-extremity pain at the time of testing. All study participants provided written informed consent in accordance with the University of

Calgary's policy on research using human subjects. The study protocol was approved by the Conjoint Health Research Ethics Board of the University of Calgary.

Data collection

Kinematic and kinetic data were collected while the subjects performed for each leg five barefoot and five shod heel-toe over-ground running trials (running speed: 4.00 ± 0.6 ms^{-1}). A standard, neutral running shoe, without unique design features that potentially could have influenced gait asymmetry, was provided for each subject (New Balance 506; New Balance Athletic Shoe Inc., USA). A running trial was considered successful when the subject's foot that was being tested landed within the edges of a force platform (Kistler Instrumente AG, Switzerland). The force platform was used to record ground reaction forces (GRFs) at a sampling rate of 2,400 Hz. At the same time, kinematic data were collected by means of a marker-based motion capture system having eight synchronized, digital, high-speed, infrared cameras (Motion Analysis Corporation, USA). Twenty-two retro-reflective markers were mounted on each study participant. Marker locations included the right and left anterior superior iliac spine, the right and left posterior superior iliac spine, and proximal, lateral, and distal aspects of the thigh and shank. To describe the foot motion, markers were placed at proximal and distal, and lateral locations of the test shoe and on corresponding locations on the bare foot. For the purpose of a neutral standing trial, additional markers were also placed on (and after the neutral trial removed from) the right and left greater trochanters, the medial and lateral knee joint, and the medial and lateral malleoli to define joint centres. A sampling rate of 240 Hz was used to record the trajectories of the markers.

Data pre-processing

Cortex motion analysis software (Motion Analysis Corporation, USA) was used to reconstruct the trajectories of the 22 markers for each running trial. A fourth-order, low-pass, Butterworth filter was applied to the kinematic and kinetic data to filter out movement artefacts and measurement noise with cut-off frequencies of 6 Hz for kinematic data and 50 Hz for kinetic data [20]. Standard motion analysis software *(*KinTrak 7.0; Human Performance Laboratory, Calgary, Canada) was used to compute 30 time-continuous gait variables. The 30 variables included joint angles, joint moments, and joint angular velocities of the ankle, knee, and hip, as well as ground reaction forces in all three planes of motion: frontal, sagittal, and transverse (Table 1). Joint moments and GRFs were normalized to body weight. All variables were resampled to 101 time points representing 0 to 100% of the stance phase.

Comprehensive asymmetry index

The following data-processing steps were conducted for each subject and shoe condition (i.e. barefoot and shod). First, the mean waveform for each of the 30 variables was calculated based

Table 1. Gait variables.

Segment	Variables (frontal, sagittal, and transverse planes)		
Hip joint	Angles [°]	Moments [BWm]	Angular velocities [$°s^{-1}$]
Knee joint	Angles [°]	Moments [BWm]	Angular velocities [$°s^{-1}$]
Ankle joint	Angles [°]	Moments [BWm]	Angular velocities [$°s^{-1}$]
Centre of pressure		Ground reaction forces [BW]	

Time-continuous gait variables that were computed over the stance phase for each subject, leg, and shoe condition. These variable types were used for the comprehensive asymmetry index.

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on the five collected trials. Second, the mean waveform for each variable was divided by the average of the corresponding standard deviation waveforms. This was done to normalize the variables to account for asymmetry caused by the natural variability of the movement [13, 18]. Third, all normalized waveforms were vectorized into a 3,030-dimensional (30 variables x 101 time points) row vector, $\boldsymbol{q}$, by horizontally appending the waveforms. Hence, $\boldsymbol{q}_{left\_leg}$ and $\boldsymbol{q}_{right\_leg}$ incorporated all available information about an individual's movement during the stance phase. Finally, a difference vector, $\Delta\boldsymbol{q} = \boldsymbol{q}_{right\_leg}$—$\boldsymbol{q}_{left\_leg}$, between the multi-dimensional row vectors of the right and left legs was calculated for each participant and shoe condition. The difference vector $\Delta\boldsymbol{q}$ quantified all measured aspects of asymmetry of the participants' gait. Therefore, the vector norm of $\Delta\boldsymbol{q}$ (i.e. the Euclidean distance from the origin to $\Delta\boldsymbol{q}$) may serve as a single CAI of the study participants' overall gait asymmetry.

However, $\Delta\boldsymbol{q}$ is a complex high-dimensional (3,030 dimensions) construct. It is possible that some components of $\Delta\boldsymbol{q}$ contain artefacts that appear to indicate asymmetry. These artefacts are actually the result of random fluctuations of the data due to the natural variability of the movement. The expected gait asymmetry changes within an individual were rather small and the signal-to-noise ratio is unfavourable. Relevant changes in the gait pattern and, therefore, in gait asymmetry between shoe conditions in one variable have to be interrelated with changes in the asymmetry of other variables [19]. It was speculated that the use of a PCA would allow increasing the sensitivity of the CAI to detect small changes in gait asymmetry. For the PCA, an input matrix $\boldsymbol{M}$ was created containing the difference vector for each individual with each shoe condition:

$$\boldsymbol{M} = \begin{bmatrix} \Delta\boldsymbol{q}_1 \\ \vdots \\ \Delta\boldsymbol{q}_{30} \end{bmatrix} \tag{1}$$

The input matrix contained 3,030 columns (30 variables x 101 time points) and 30 rows (15 subjects x 2 shoe conditions). The PCA comprised the following steps: (1) calculation of the covariance matrix of $\boldsymbol{M}$; and (2) calculation of the eigenvectors and eigenvalues of the covariance matrix [21]. The eigenvectors represent the orthogonal principal component vectors (PC-vectors), $\boldsymbol{p}$. The PC-vectors are defined by the direction of the highest correlated variance in the data. Since in the current study the input matrix for the PCA contained the difference vectors (right-left) for each of the individuals, the variance in the matrix and the definition of the PC-vectors were due to the asymmetry of the individuals' gait.

The eigenvalue (EV) spectrum was assessed to determine a suitable number k of PC-vectors for the definition of the CAI. Within the first 15 EVs a drop is visible between EV8 and EV9 (Fig 1). Therefore, the first eight PC-vectors (k = 8) were expected to provide the best compromise between retaining as much correlated asymmetry as possible and filtering out uncorrelated noise [16].

The difference vectors $\Delta\boldsymbol{q}$ were then represented in a subspace spanned by the eight selected PC-vectors by projecting each difference vector $\Delta\boldsymbol{q}$ onto the PC-vectors:

$$\mathrm{P}_{\mathrm{si}} = \Delta\boldsymbol{q}_s \cdot \boldsymbol{p}_i \tag{2}$$

where $\boldsymbol{s}$ indicates the study participants and $\boldsymbol{i}$ represents the number of the PC-vector. A subject- and condition-specific CAI was then calculated as the Euclidean distance from the origin using the projections (P_{si}):

$$CAI_s = \sqrt{\sum_{i=1}^{k} (P_{si})^2} \quad (3)$$

Sensitivity analysis and statistics

To assess the sensitivity of the CAI, it was determined whether the difference vectors by themselves would be able to confirm the hypothesized difference in gait asymmetry between shod and barefoot running and how the CAI depended on the number k of PC-vectors used. Therefore, different variations of the CAI for each individual and shoe condition were calculated: (1) CAIs without PCA, using the vector norm (i.e. Euclidean distance) of the raw $\Delta\boldsymbol{q}$ only; (2) CAIs with PCA, based on all possible numbers of PC-vectors (k = 1. . .30). A paired samples t-test ($p \leq 0.05$; IBM SPSS Statistics 20, IBM Corporation, USA) was then used to assess the significance of the difference between the different mean CAIs for barefoot and shod running.

Relevant asymmetry variables

The relevant asymmetry variables and their correlations were identified by analysing the loadings of the eight PC-vectors. The loading magnitude indicates the amount of variance in a variable that is captured by the corresponding PC-vector [22]. Since this variance was caused by gait asymmetry, variables with higher loadings contributed more to an asymmetrical gait. The loadings were multiplied with their corresponding EVs to weight the loadings according to the amount of variance/asymmetry covered by each PC-vector.

Results

The eight PC-vectors that were used for the calculation of the CAI contained 76.4% of the overall asymmetry in all gait variables (Fig 1). The subject-specific CAIs for barefoot running ranged from 103.9 to 210.9, whereas the range for shod running was from 48.4 to 212.1 (Fig 2).

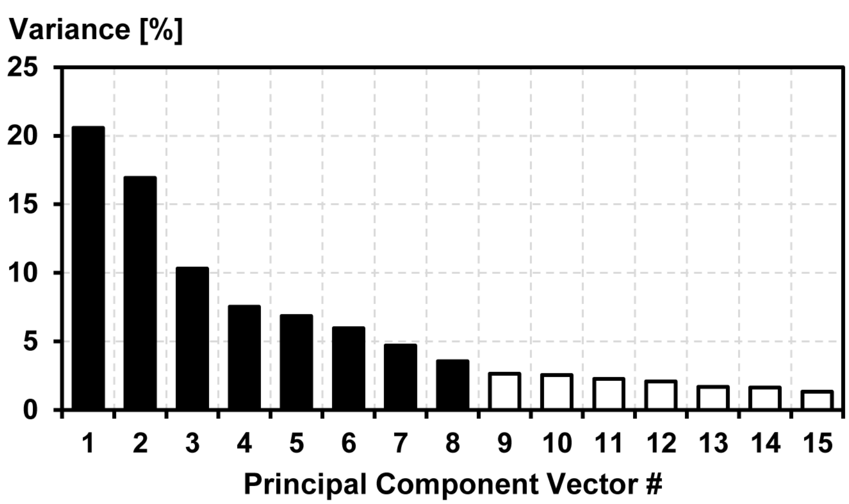

Fig 1. Eigenvalue spectrum. Eigenvalue spectrum of the first 15 principal component vectors that was used to determine the number of principal component vectors for the definition of the *comprehensive asymmetry index* (CAI). After the first eight eigenvalues (black bars) a drop can be seen. Hence, the first eight principal component vectors (k = 8) were used for the definition of the CAI.

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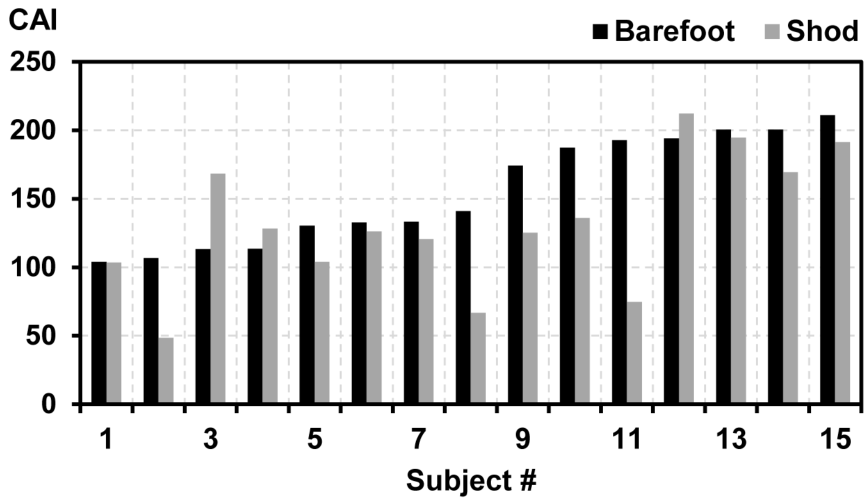

Fig 2. Subject-specific *comprehensive asymmetry index* (CAI) for barefoot and shod running. Study participants are arranged by increasing CAI for barefoot running. All CAIs calculated using eight principal component vectors (k = 8).

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Averaged over all participants the CAI (k = 8) for running barefoot was 155.7 ± 39.5 (mean ± standard deviation) and for running in the shoe condition was 131.2 ± 48.5 (Table 2). The difference between the two conditions was significant (p = 0.041). Comparing barefoot and shod running using the CAI calculated as the direct Euclidean distance of the raw $\Delta\boldsymbol{q}$ to the origin (i.e. without filtering out uncorrelated asymmetries by the PCA) revealed no significant difference (p = 0.067; Table 2). The evaluation of how the CAI depended on the number k of PC-vectors used for the definition of the CAI showed that $k \leq 3$ was not sufficient to detect significant asymmetry differences between barefoot and shod running (Table 2). For $4 \leq k \leq 8$ and $12 \leq k \leq 13$ the differences between the mean CAIs for barefoot and shod running were significant.

The relevant asymmetry variables (i.e. variables with the highest PC-vector loadings) were mainly located in the ankle and knee joint (Fig 3). The frontal knee angle had the highest PC-vector loading (1.73) followed by the frontal ankle moment (1.50) and the frontal ankle angle (1.39). The PC-vector loadings showed correlations particularly between the frontal ankle angle/moment and the frontal knee angle/moment (PC-vector 1, PC-vector 2).

Discussion

The current study had two main outcomes. First, a novel approach to quantify gait asymmetry was proposed that combined correlated asymmetries in multiple gait variables into one *comprehensive asymmetry index*, the CAI. The sensitivity analysis suggested that considering correlated asymmetries improves the sensitivity for detecting changes in gait asymmetry caused by external boundary conditions. This would be particularly useful when assessing the progression of clinical conditions such as cerebral palsy or the progress of rehabilitation treatments. The proposed method allowed to examining the structure of gait asymmetry by assessing the individual loadings of principal component vectors. Again, this has potential for clinical gait analysis and may contribute to a better understanding of the specific manifestations of a patient's underlying condition, for example, in stroke and cerebral palsy patients. Second, the result of the CAI supported the hypothesis that even in healthy, young adults, gait asymmetry is reduced when running in shoes compared to running barefoot. This suggests that footwear seems to

Table 2. Mean *comprehensive asymmetry indexes* (CAI) for barefoot and shod running.

k	Mean CAI Barefoot	Mean CAI Shod	p-Value
Δq	177.7 (SD 33.7)	157.9 (SD 39.1)	**0.067**
1	61.5 (SD 40.1)	68.6 (SD 48.2)	**0.302**
2	106.3 (SD 46.4)	84.0 (SD 43.5)	**0.084**
3	123.2 (SD 43.3)	98.6 (SD 41.2)	**0.060**
4	136.0 (SD 39.3)	104.0 (SD 45.4)	**0.020**
5	141.4 (SD 40.8)	113.9 (SD 48.6)	**0.045**
6	147.0 (SD 42.9)	121.4 (SD 48.4)	**0.042**
7	152.9 (SD 40.7)	126.3 (SD 47.6)	**0.031**
8	155.7 (SD 39.5)	131.2 (SD 48.5)	**0.041**
9	157.8 (SD 39.8)	135.1 (SD 46.8)	**0.061**
10	161.0 (SD 39.2)	136.9 (SD 47.0)	**0.052**
11	163.3 (SD 38.1)	139.6 (SD 46.2)	**0.059**
12	165.9 (SD 37.3)	142.0 (SD 43.5)	**0.042**
13	167.2 (SD 37.2)	144.2 (SD 42.9)	**0.050**
14	168.4 (SD 37.2)	146.2 (SD 42.8)	**0.064**
15	169.8 (SD 36.4)	147.4 (SD 42.8)	**0.060**
16	171.2 (SD 35.2)	148.7 (SD 42.4)	**0.058**
17	171.8 (SD 35.3)	150.1 (SD 42.6)	**0.069**
18	172.6 (SD 35.6)	151.0 (SD 42.8)	**0.072**
19	173.4 (SD 35.9)	151.8 (SD 42.3)	**0.073**
20	174.2 (SD 35.1)	152.8 (SD 42.1)	**0.072**
21	174.5 (SD 35.3)	153.7 (SD 42.0)	**0.078**
22	175.2 (SD 35.4)	154.3 (SD 41.6)	**0.075**
23	175.4 (SD 35.4)	155.3 (SD 41.1)	**0.083**
24	176.1 (SD 34.6)	155.7 (SD 40.9)	**0.071**
25	176.5 (SD 34.5)	156.2 (SD 40.8)	**0.071**
26	176.8 (SD 34.5)	156.7 (SD 40.4)	**0.072**
27	177.1 (SD 34.4)	157.1 (SD 40.0)	**0.072**
28	177.4 (SD 34.1)	157.4 (SD 39.7)	**0.069**
29	177.6 (SD 33.8)	157.6 (SD 39.5)	**0.067**
30	177.7 (SD 33.7)	157.9 (SD 39.1)	**0.067**

Mean *comprehensive asymmetry indexes* (CAI) and p-values (paired samples t-test) for comparisons between barefoot and shod running based on different CAIs calculated with the raw difference vector (Δq) and different numbers of principal component vectors (k = 1. . .30).

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affect certain aspects of the neuromuscular control system that are involved in the coordination of the movements of left and right lower limbs.

Comprehensive asymmetry index

The development of the CAI was motivated by the goal to provide a comprehensive asymmetry index with enhanced sensitivity for changes in gait asymmetry. Considering this main goal and the way it was implemented led to advantageous and disadvantageous characteristics of the proposed method, which will be discussed in the following paragraphs.

Since the CAI is a single value representing the totality of gait asymmetry of an individual (based on the measured variables), it facilitates direct comparisons between individuals with

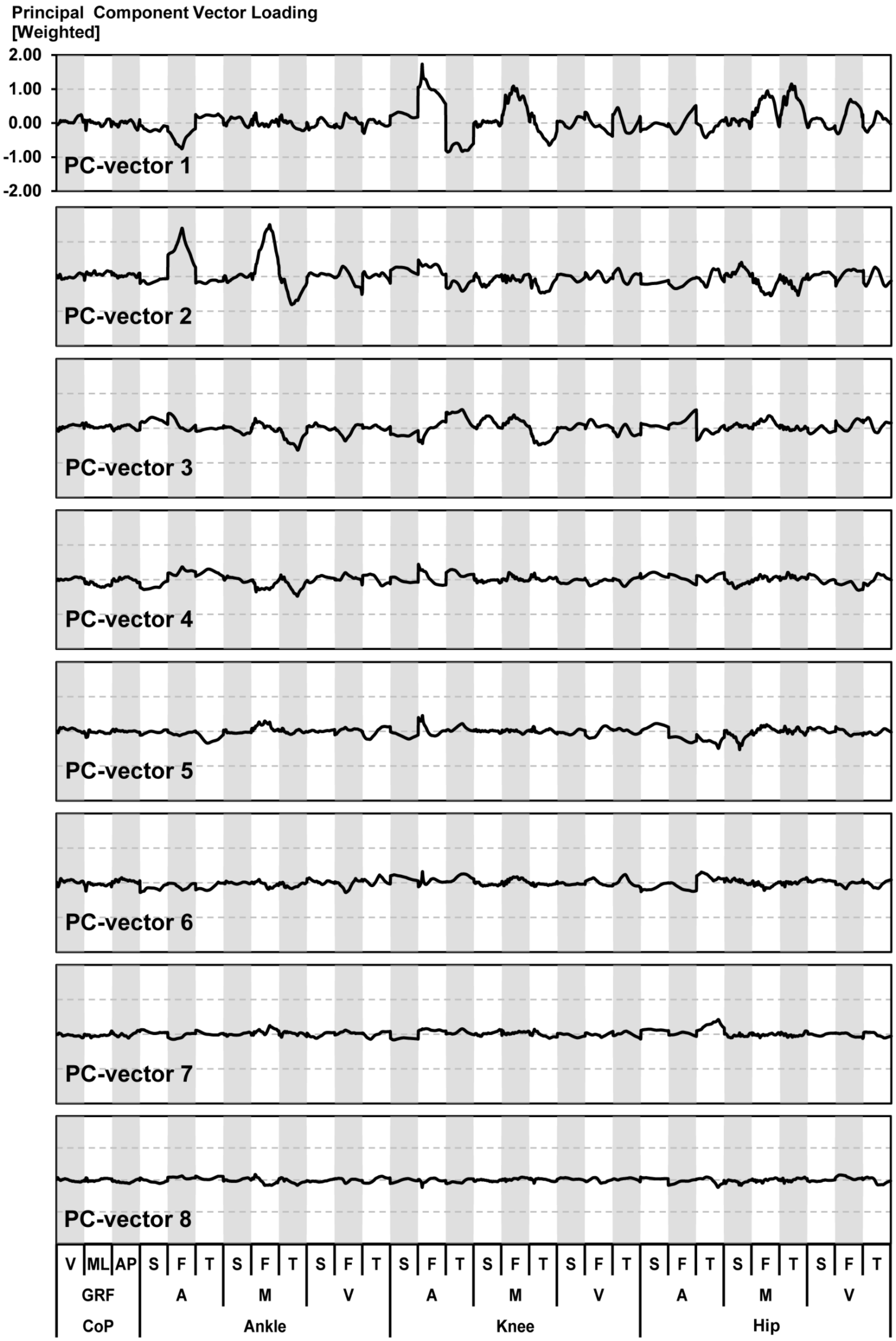

Fig 3. Weighted loadings of the first eight principal component vectors. These eight principal component vectors (PC-vectors) were used to calculate the *comprehensive asymmetry index* (CAI). Y-axes indicate the magnitude of the loading. X-axes represent the analysed biomechanical variables: V-Vertical; ML-Medial lateral; AP-Anterior posterior; GRF-Ground reaction force; CoP-Centre of pressure; S-Sagittal plane; F-Frontal plane; T-Transverse plane; A-Angle; M-Moment; V-Velocity.

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respect to overall gait asymmetry. The CAI offers no advantage, however, when it is necessary to quantify gait asymmetries of isolated variables (e.g. sagittal knee joint angle) at a specific time-point (e.g. at mid stance). In this case, other methods may provide a faster and more precise assessment of gait asymmetry [15, 23–26]. It is important to realize that CAIs can only be compared among individuals when they have been calculated using the same variables. Another limitation of the current method is that it is possible that unique gait asymmetries present in only one individual may not contribute sufficiently to be represented in the lower order PC-vectors. Therefore, if this method is applied as a diagnostic tool to quantify asymmetry in an individual patient, then both the PCA-filtered and direct Euclidean distance-based CAI should be assessed to ensure that the patient does not exhibit an unusual asymmetry pattern.

The results of the sensitivity analysis (Table 2) suggested that the PCA acted as a filter separating correlated from uncorrelated gait asymmetry variables [16]. Correlated asymmetries are more likely to contain actual differences in the movement pattern while uncorrelated asymmetries are more likely to contain a high proportion of noise [19]. Another advantage of determining the correlation structure of gait asymmetry using a PCA is that the resultant PC-vector loadings show the relevant asymmetry variables and their correlations. In fact, investigating the relevant asymmetry variables and their correlations suggested that the ankle and knee joint seemed to have the highest importance for the generation and compensation of gait asymmetry (Fig 3). Gait variables of the hip seemed to be less involved. Determining the relevant asymmetry variables and their correlation has potential for clinical gait analysis and may contribute to a better understanding of the specific manifestations of a patient's underlying condition.

PCA has been used before when investigating gait asymmetry [14, 15, 24]. However, to the best knowledge of the authors, it has not yet been applied in the all-encompassing form that was set up in this study.

The CAI was based on data measured with a 3D motion capture system and a force platform during over-ground running. This experimental setup limits the amount of strides that can be measured and may also reduce the applicability of the CAI to monitor gait asymmetry in specific cases (i.e. a laboratory setting is required). Therefore, future studies should investigate the sensitivity of the CAI to detect gait asymmetry changes using data acquired with wearable sensors (e.g. accelerometers) to increase the amount of data that can be collected and the applicability of the CAI.

Because of the small sample size (15 study participants) and the recruitment of healthy individuals only, a systematic discussion of CAI values is not possible, and an actual non-pathological asymmetry range was not identified. Further studies should determine specific pathological and non-pathological ranges, as well as investigate how limb dominance, gender, or other external boundary conditions affect the CAI.

Effect of footwear on gait asymmetry

Gait asymmetry in a healthy population has been documented in several studies [14, 15, 27]. Previous research has also reported an impact of footwear on the running kinematics and kinetics of healthy adults [28–30]. From a purely mechanical perspective, one would expect that wearing footwear, which may not be manufactured perfectly symmetrical, would either not affect or increase gait asymmetry. However, as pointed out in the introduction, previous studies indicated that footwear may improve neuromuscular control of motion. This might lead to a decrease in gait asymmetry as suggested by Vagenas and Hoshizaki [31] based on a limited set of isolated kinematic variables of the foot. The findings of the comprehensive analysis of this study support this hypothesis (Table 2).

Improved motor control mechanisms associated with wearing footwear might be a result of altered cutaneous sensory information of the plantar or dorsal surface of the feet [32–34]. Two recent review studies attested to the significance of plantar sensory feedback for the control of movement and supported the utilization of textured materials for improving perceptual-motor performance [35, 36].

The magnitude of the effect of footwear on gait asymmetry was subject-dependent (Fig 2). In fact, a few study participants (3 out of 15) even demonstrated an increase in gait asymmetry when running in shoes. De Wit et al. [28] reported a subject-depended impact of footwear on the kinematics and kinetics during running. However, it remains unknown which mechanisms cause these subject-dependent responses to footwear. One mechanism might be related to subject-specific sensitivity thresholds of the plantar or dorsal surface of the feet that may influence the afferent feedback to the neuromuscular control system [33].

Conclusion

Footwear seems to reduce gait asymmetry during running in healthy, young individuals. Changes in the afferent sensory feedback to the neuromuscular control system may be a possible explanation for this observation.

Supporting Information

S1 File. Supplementary Data. Subject demographics, eigenvalue spectrum, subject-specific comprehensive asymmetry index (CAI) for barefoot and shod running calculated using the raw $\Delta \boldsymbol{q}$ and different numbers of principal component vectors (k = 1. . .30), and weighted loadings of the first eight principal component vectors.
(XLSX)

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