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The effect of changing mediolateral center of pressure on rearfoot eversion during treadmill running

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ABSTRACT

Introduction: Atypical rearfoot eversion is an important kinematic risk factor in running-related injuries. Prominent interventions for atypical rearfoot eversion include foot orthoses, footwear, and taping, yet a running gait retraining is lacking. Therefore, the aim was to investigate the effects of changing mediolateral center of pressure (COP) on rearfoot eversion, subtalar pronation, medial longitudinal arch angle (MLAA), hip kinematics and vertical ground reaction force (vGRF).

Methods: Fifteen healthy female runners underwent gait retraining under three conditions. Participants were instructed to run normally, on the lateral (COP lateral) and medial (COP medial) side of the foot. Foot progression angle (FPA) was controlled using real-time visual feedback. 3D measurements of rearfoot eversion, subtalar pronation, MLAA, FPA, hip kinematics, vGRF and COP were analyzed. A repeated-measures ANOVA followed by pairwise comparisons was used to analyze changes in outcome between three conditions. Data were also analyzed using statistic parameter mapping.

Results: Running on the lateral side of the foot compared to normal running and running on the medial side of the foot reduced peak rearfoot eversion (mean difference (MD) with normal 3.3°, $p < 0.001$, MD with COP medial 6°, $p < 0.001$), peak pronation (MD with normal 5°, $p < 0.001$, MD with COP medial 9.6°, p=<0.001), peak MLAA (MD with normal 2.3°, $p < 0.001$, MD with COP medial 4.1°, $p < 0.001$), peak hip internal rotation (MD with normal 1.8°, $p < 0.001$), and peak hip adduction (MD with normal running 1°, $p = 0.011$). Running on the medial side of the foot significantly increased peak rearfoot eversion, pronation and MLAA compared to normal running.

Significance: This study demonstrated that COP translation along the mediolateral foot axis significantly influences rearfoot eversion, MLAA, and subtalar pronation during running. Running with either more lateral or medial COP reduced or increased peak rearfoot eversion, peak subtalar pronation, and peak MLAA, respectively, compared to normal running. These results might use as a basis to help clinicians and researchers prescribe running gait retraining by changing mediolateral COP for runners with atypical rearfoot eversion or MLAA.

1. Introduction

Running-related injuries (RRIs) are a problem both for athletes and individuals who want to improve their physical fitness and health by running [1]. The etiology of RRIs is known to be multifactorial in nature; running biomechanics certainly play a role [1,2]. Abnormal foot kinematics such as increased peak rearfoot eversion, decreased peak ankle dorsiflexion and increased ankle plantar-flexion range of motion are considered to play an important role in the high incidence of RRIs [2–4]. Of these, rearfoot eversion has received increased attention in biomechanical studies due to its potential effect on the incidence of RRIs. Recent studies report atypical rearfoot eversion as a contributing factor predisposing runners to RRIs, specifically increased peak rearfoot eversion and duration of rearfoot eversion are a risk factor for patellar tendinopathy and medial tibial stress syndrome, decreased rearfoot eversion a risk factor for iliotibial band syndrome, and increased

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rearfoot eversion at touchdown for Achilles tendinopathy [2,5]. Female runners with atypical rearfoot eversion can be more prone to RRIs, specifically increased peak rearfoot eversion is related to tibial stress fracture [6] and decreased peak rearfoot eversion is associated with iliotibial band syndrome in recreational female runners [2].

Atypical rearfoot eversion can change lower limb closed kinematic chain movements during gait [7]. These lower limb kinematic changes may lead to the development of lower limb injuries not only in the foot and ankle but also in the other proximal parts, including the knee, hip, and pelvis [8]. Rearfoot eversion is generally calculated to predict subtalar pronation as the subtalar coordination axis is not aligned with the foot coordination axes, and besides, no anatomical landmark exists on the talus. However, there are some biomechanical models such as Human Body Model (HBM) that calculate subtalar pronation as a standard outcome. Rearfoot eversion is highly associated with the medial longitudinal arch angle (MLAA) [9]. Previous studies report larger MLAA (lower plantar arch height) as a risk factor for medial tibial stress syndrome in runners [5,10]. Normal rearfoot eversion and MLAA enhance shock absorption and the ability of proper force transition to the more proximal segments from touchdown to toe-off [11]. By contrast, an abnormal rearfoot eversion may disturb plantar pressure distribution, which results in improper force transition and consequently potential overuse injuries [12]. Hence many biomechanical studies on sports-related injuries have explored the mechanics of rearfoot eversion and/or MLAA during gait and modifying atypical rearfoot eversion.

Center of pressure (COP) is the application point of the ground reaction force (GRF) under the foot [13]. The COP moves from heel to toe on the lateral aspect of the foot and just before push-off it quickly moves medially. The COP is associated with biomechanical alterations in the lower limb joints; previous studies underlined the high potential of COP manipulation in altering gait characteristics, joint biomechanics, and modifying foot malalignment [14,15]. A more lateral position of the COP results in a smaller frontal plane moment arm of the GRF around the ankle, thereby decreasing the rearfoot tendency toward eversion [16]. Some studies investigating mediolateral COP manipulation using lateral wedge insoles in order to reduce knee adduction moment reported changes in rearfoot eversion too [17,18]. This suggests that changes in mediolateral COP during running may have the potential to change rearfoot eversion.

Gait retraining is a novel and increasingly common way of inducing the body or a segment to change a movement pattern or a segment's motion direction [19]. Previous studies reported promising results when gait retraining was used to modify biomechanical risk factors. Step rate, step width, step length, vertical loading rate, and foot strike pattern are some of the most common parameters retrained to modify biomechanical risk factors associated with RRIs [20–22]. These studies demonstrated an effective modification of several biomechanical risk factors such as hip rotation and adduction, knee abduction/adduction, ankle dorsiflexion/plantar flexion, tibial acceleration, and ground reaction force characteristics. Running retraining is also reported as the most successful strategy to modify abnormal kinematics [21]. To our knowledge, no study has specifically addressed rearfoot eversion using running gait retraining.

Our first aim is therefore to investigate the effect of intentional changes of mediolateral COP on rearfoot eversion. The secondary aim is to evaluate the effect of intentional changes of mediolateral COP on subtalar pronation, MLAA, hip kinematics and vGRF. We hypothesized that running with more lateral COP reduces peak rearfoot eversion, subtalar pronation, and MLAA, and running with more medial COP increases these factors. As foot acts as a shock absorber, manipulation of mediolateral COP could negatively influence shock absorption, potentially leading to an increase in loading rate and peak vGRF. We, therefore, hypothesized that manipulation of mediolateral COP increases vertical average loading rate (VALR) and peak vGRF. From a clinical perspective, the results of this study may have potential implications for modifying abnormal rearfoot eversion using mediolateral COP modification during management of RRIs.

2. Methods

2.1. Study design

This is a cross-sectional pilot study conducted to investigate the effects of changing mediolateral COP on rearfoot eversion, subtalar pronation, and MLAA. Since our previous study showed that foot progression angle (FPA) affects rearfoot eversion [23], we controlled FPA using real-time visual feedback when performing COP tasks.

2.2. Setting

Data were collected at the Motion Lab of the Center for Rehabilitation, University Medical Center Groningen, The Netherlands between January and April 2019.

2.3. Participants

Seventeen female runners recruited by advertisements from local running clubs and the University of Groningen participated in this study voluntarily. Inclusion criteria were: female, age 18–40, running experience of minimum 1 year, running distance >10 km/week, ability to run with rearfoot strike, free of self-reported lower limb injuries or pain over the previous six months, no musculoskeletal disorders, no abnormal foot arch specified using the navicular drop test (5 mm < normal<10 mm), and no abnormal static rearfoot eversion ($0° < \text{normal} < 4°$) [24] prior to data collection. Two participants were excluded because of flat foot and/or excessive static rearfoot eversion. Fifteen volunteers who met the inclusion criteria comprised the participants of this study. Ethical approval was obtained from the local ethics committees (METc 2018/086) of University Medical Center Groningen, and all participants gave written informed consent and completed a questionnaire on demographic information before motion analysis testing.

2.4. Instrumentation

An instrumented split-belt treadmill with two integrated 3D force plates of the Gait Real-time Analysis Interactive Lab (GRAIL) system (Motekforce Link, The Netherlands) was used for running assessments. Ground reaction force (GRF) data were recorded at 1000 Hz, synchronized with a 10-camera integrated motion capture system (Vicon Bonita 10; Vicon Motion Systems, Oxford, UK). Selected kinematics and kinetics were further processed in D-Flow (v. 3.28; Motekforce Link, The Netherlands) at a sampling frequency of 100 Hz. Real-time filtering of the marker data was processed using a low-pass 2-order zero-phase Butterworth filter with a cut-off frequency of 6 Hz.

2.5. Marker placement and baseline measurement

Markers were placed on the participant's body by the same investigator (SHM). Twenty-six markers were placed according to the human body model 2 (HBM2) (Fig. 1) [19]. Additionally, 8 markers were placed on both feet to compute rearfoot eversion and MLAA. These markers were placed at the posterior part of calcaneus, medial side of calcaneus, navicular bone tuberosity, and first metatarsal head. Four holes were cut in the shoes to uncover these parts of the foot in order to attach markers directly to the skin. All participants wore the same brand of neutral shoes (Dr Comfort, refresh, USA) with the same neutral insole. Before running, participants familiarized themselves with the environment and treadmill. Next, after a 5-minute warm-up period a 20-second baseline dataset was collected. Because our previous study indicated that changes in FPA affect rearfoot eversion [23], we aimed to control FPA when

Fig. 1. Marker placement. Twenty-six markers were attached to the body according to the HBM model; 8 markers were attached to the feet, to be used for calculating rearfoot eversion and MLAA.

running. To determine normal FPA, the midstance FPA (the value at 50 % of the stance phase) was averaged for the first 20 strides. FPA was the angle between the line attaching the marker on the superior calcaneus to the second metatarsal head marker and the longitudinal axis of the treadmill.

2.6. Feedback on FPA

A custom-made application was developed on the D-flow software to produce real-time feedback for FPA. A clock with a red pointer, which was the FPA indicator (degrees), was designed and projected on the screen to reflect FPA during midstance in real time (Fig. 2). A 5° target range whose middle point was set to normal FPA was shown on the clock (Fig. 2, green part). Participants were asked to place the red pointer within the target area when running, thereby turning the target area green (positive feedback). If the red pointer was placed outside the target range, the area became red (negative feedback). The red pointer was fixed on FPA in midstance and updated at each step. Participants

Fig. 2. Picture representing real-time visual feedback for changing FPA. The training process: real-time visual feedback is provided to the subject via the big screen. The red pointer represents the FPA of the right foot that is fixed in midstance (50 % stance phase) and updated at each step. The target area is a wedge with a 5° range, with its middle point specifying the subject's normal FPA. The aim is to turn the target area green (positive feedback) by keeping the red pointer (FPA) inside the target area. If the red pointer leaves the target area, the target area turns red (negative feedback). (For interpretation of the references to colour in the Figure, the reader is referred to the web version of this article).

were instructed to run for 2 min with FPA feedback and asked to maintain their FPA within the targeted area. All participants were right-leg dominant, therefore, tests were performed on right foot.

2.7. Instruction for changing mediolateral COP

To change the COP toward lateral and medial, participants were instructed to apply their plantar pressure on the lateral and medial edges of the shoes, respectively, in such a way that they still felt comfortable while running. Likewise, participants were asked to simultaneously follow their normal FPA as instructed above while running. Participants were then given a 2-minute practice to change their plantar pressure laterally and medially while following their real-time FPA. Participants were asked to run for each task and after a 1-minute run, 20-second data were collected. The order of the experimental tasks was randomized. Participants did not receive any feedback on their mediolateral COP at midstance, therefore, they did not know whether they successfully changed their mediolateral COP. Running speed was set at 8 km/h for all conditions and all participants. The speed was selected based on pilot testing. At this speed, participants could easily concentrate on changing COP and FPA during running. This became more difficult to perform at a higher speed.

2.8. COP calculation

To calculate the mediolateral COP a vector was created between the superior heel and second toe markers. The angle of this vector relative to the longitudinal lab coordination system was used to create a rotation matrix. The translated COP was corrected with this rotation matrix to get the COP relative to the foot. For each step the mediolateral COP in midstance (the single frame at 50 % of stance phase) was used for further analysis. The mediolateral COP was averaged for the first 5 steps of baseline measurement. To select steps for lateral and medial COP conditions, the first 5 steps whose COP value were 5 mm (1SD of COP according to a previous study [25], so a correct response to the instruction) smaller (for medial) or larger (for lateral) than baseline-averaged COP were selected.

2.9. Outcomes

The rearfoot segment coordinate system was established according to International Society of Biomechanics (ISB) recommendations and calculated as rotation of the local calcaneus coordination system relative to the fixed laboratory coordinate system using the rotation sequence defined by ISB [26]. Subtalar pronation/supination is a standard measure of HBM2 calculated using the Isman and Inman method [27] and further descriptions by van den Bogert et al. [19]. According to van den Bogert et al. (1994) [28] "the subtalar joint (STJ) center is defined 12 mm below the ankle joint center, implemented as a displacement along the vertical axis of the foot and scaled by tibia length as follows: Tibia length (distance between lateral knee and ankle markers) /0.375*12. The Z-axis is parallel to the vector from second metatarsal to superior heel markers and points posteriorly. The X-axis is based on the Z-axis and a temporal Y-axis defined between the lateral and medial malleolus pointing to the left. The X-axis is the cross product of the temporal Y-axis and the Z-axis, and will point dorsally. The Y-axis is the cross product of the Z- and X-axis. According to Isman and Inman, the STJ is inclined 42 degrees from the horizontal plane, and deviates 23 degrees medially from the sagittal plane". MLAA was measured based on the angle formed between three markers: medial aspect of calcaneus marker, navicular bone tuberosity marker, and first metatarsal head marker. Hip kinematics are standard measures of HBM2 computed as explained by van den Bogert et al. [19]. A custom MATLAB script (Version R2018a, Natick, MA, USA) was used to analyze data. Touchdown and toe-off were determined using GRF data with a threshold of 10 N vGRF. Kinematic and GRF data were filtered using low-pass zero phase 2-order Butterworth filters with a cutoff frequency of 6 Hz. Outcomes of five steps were calculated and averaged. Kinematic data were time-normalized to 100 % of stance phase. Peak angles were expressed as the maximum angle during the stance phase. Timing of peak angles was expressed as percentage of the stance phase. Angular excursions were expressed as range of motion from touchdown to peak angle. The VALR was calculated as the average slope of the vGRF during 20 %–80 % of the non-normalized stance time from foot strike to vertical impact peak. In cases of a missing vertical impact peak, the force value at 13 % of stance phase was used [29].

2.10. Statistical analysis

Kinematic data were compared between conditions at touchdown and at their peak. Also, time to peak and excursion were compared between conditions as well as peak vGRF and VALR. Using IBM SPSS version 23 (IBM Corp., Armonk, NY, USA), a one-way repeated-measures ANOVA followed by a Bonferroni adjustment was performed to statistically explore differences between conditions: baseline, lateral COP, and medial COP trials. Shapiro-Wilk tests and QQ were used to assess the normal distribution hypothesis. The significance level was set at 0.05.

For each outcome, SPM analyses with a repeated measures ANOVA were used to examine any statistical differences between the three conditions for the entire stance phase. If applicable, post-hoc paired t-tests were performed to compare condition pairs. A Bonferroni correction was applied to adjust α for multiple post-hoc comparisons. All SPM analyses were conducted in MATLAB (Version R2018a, Natick, MA, USA) using the open-source software package spm1D 0.4 (www.spm1d.org).

3. Results

Table 1 shows the characteristics of 15 participants of this study. All assumptions for repeated-measures ANOVA were met (no significant outliers, normal distribution, and sphericity). Participants were instructed to follow their normal FPA in midstance when running on the lateral and medial side of the foot, resulting in a 0.5° ($p = 0.051$) and 0.4° ($p = 0.106$) difference relative to normal running, respectively (Fig. 3). Lateral and medial COP in midstance resulted in 6 ($p < 0.001$) and −5.9 ($p < 0.001$) mm differences relative to the COP in normal running, respectively (Fig. 3).

Table 2 shows the results of one-way repeated-measures ANOVA for all measured outcome variables. Running on lateral side of the foot significantly reduced peak rearfoot eversion (mean difference (MD) with normal running = 3.3°, $p < 0.001$), rearfoot eversion at touchdown (MD = 2.1°, $p = 0.002$), eversion excursion (MD=-1.2°, $p = 0.020$), peak pronation (MD=-5.0°, $p < 0.001$), pronation at touchdown (MD=-4.1°, $p = 0.004$), peak MLAA (MD=-2.3°, $p < 0.001$), MLAA excursion (MD=1.7°, $p < 0.001$), peak hip internal rotation (MD = 1.8°, $p = 0.001$), hip internal rotation at touchdown (MD = 2.7°, $p < 0.001$), and hip adduction at touchdown (MD = 1.2°, $p = 0.011$) relative to normal running. By contrast, running on the medial side of the foot significantly increased peak rearfoot eversion (MD=-2.7°, $p < 0.001$), rearfoot eversion at touchdown (MD=-2.2°, $p = 0.002$), peak pronation

Table 1
Participant characteristics.

Variable	Mean (SD)	Range
Age, y	27.5 (6.3)	21–40
Height, cm	170 (5)	164–182
Weight, kg	61.4 (6.1)	50–72
Running experience, y	6.3 (4.4)	2–17
Weekly distance, km	32.7 (17.4)	10–65
Navicular drop (mm)	7.1 (1.1)	6–9
Static rearfoot eversion (°)	2.0 (0.8)	1–3

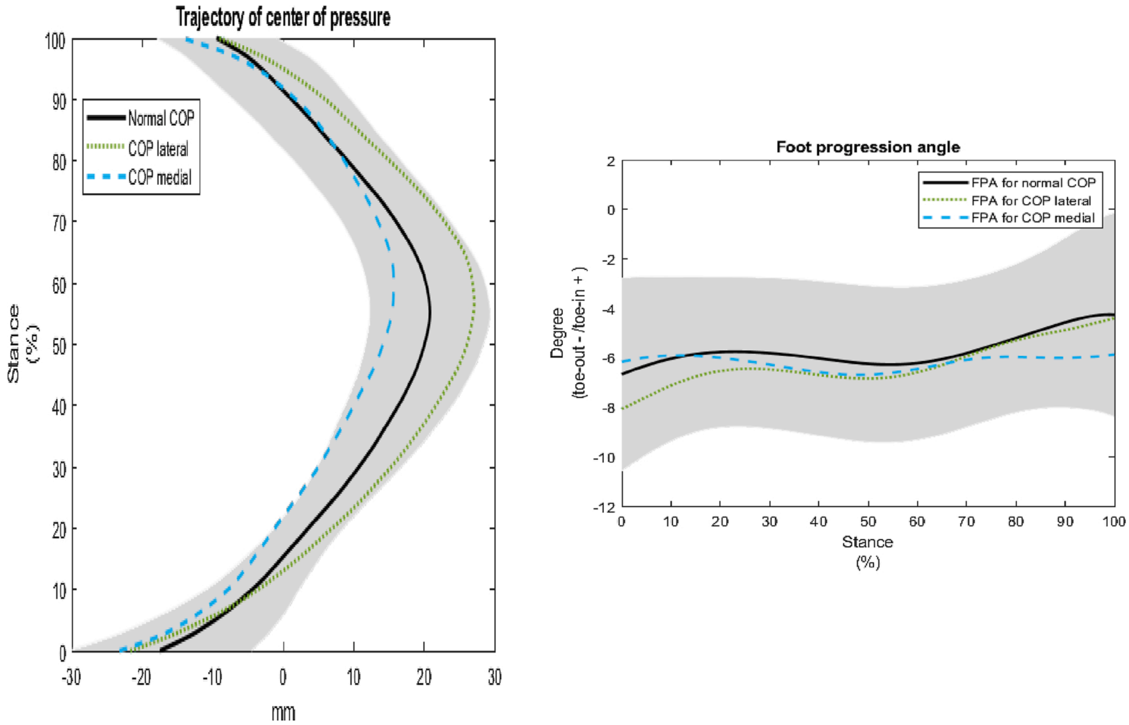

Fig. 3. Ensemble average curves of mediolateral COP and foot progression angle for three conditions during 0% to 100 % stance phase. Solid lines are normal COP, dotted lines lateral COP, dashed lines medial COP. Shaded area represents ±1 SD of the normal COP condition.

(MD = 4.6°, $p < 0.001$), time to peak pronation (MD = 5.4, $p = 0.010$), pronation at touchdown (MD = 3.6°, $p = 0.001$), peak MLAA (MD = 1.8°, $p < 0.001$), MLAA excursion (MD = 1.3°, $p = 0.005$), and hip adduction at touchdown (MD = 1°, $p = 0.045$) relative to normal running. Peak vGRF (lateral vs. normal COP MD = 0.1, $p = 0.088$; medial vs. normal COP MD = 0.0, $p = 0.337$) and VALR (lateral vs. normal COP MD = 0.2, $p = 0.999$; medial vs. normal COP MD = 0.6, $p = 0.640$) were not significantly different between conditions.

Fig. 4 shows the results of the SPM analyses for the kinematic outcomes between three conditions. The SPM analyses show that running while lateralizing COP decreases rearfoot eversion between 0–80 % of the stance phase compared with normal running. In contrast, running while medializing COP increases rearfoot eversion between the entire stance phase compared to normal running. Running while lateralizing or medializing COP decreases or increases subtalar pronation during the entire stance phases compared to normal running, respectively. Running while lateralizing COP decreases MLAA between 10–77 % of the stance phase compared with normal running. Running while medializing COP increases MLAA between 32–82 % of the stance phase compared to normal running. Running while lateralizing COP decreases hip internal rotation between 0–97 % of the stance phase compared with normal running while running with medializing COP does not significantly change hip internal rotation during the stance phase compared to normal running. Hip adduction only changed significantly between lateralizing COP and normal running between 30–31 % of the stance phase.

4. Discussion

This study aimed to investigate the effects of intentionally changing mediolateral COP on rearfoot eversion, subtalar pronation/supination, MLAA, hip kinematics and vGRF during running. The results confirm our hypotheses, showing that manipulation of mediolateral COP while keeping the FPA constant significantly affects rearfoot eversion, MLAA, subtalar pronation, and hip internal rotation and adduction. Specifically, running with more lateral COP reduced peak rearfoot eversion, peak subtalar pronation, peak MLAA, and peak hip internal rotation and adduction compared to normal COP. Running with more medial COP increased peak rearfoot eversion, peak subtalar pronation, and peak MLAA compared to normal COP. Peak vGRf and VALR were not significantly different between conditions. This study established an applicable running gait retraining basis for modifying rearfoot eversion, subtalar pronation and MLAA. The SPM analyses showed that lateralizing and medializing COP (as much as 6 mm at midstance) decreases and increases rearfoot eversion and subtalar pronation during at least 80 % of the stance phase, respectively. Our findings are promising as the findings of previous studies show that a 2−3° deviation in peak rearfoot eversion is sufficient to predispose runners to RRIs such as Achilles tendinopathy, iliotibial band syndrome, patellar tendinopathy, posterior tibial tendon dysfunction and medial tibial stress syndrome [2,5]. Although whether the opposite is also true remains unknown, it is speculated that lateralizing COP during running even slightly reduces excessive rearfoot eversion, subtalar pronation and MLAA and with that may prevent RRIs.

Lateral and medial shifting of the COP reduced and increased rearfoot eversion and MLAA excursion, respectively. Greater rearfoot eversion excursion may delay foot re-supination, an important mechanism for locking the tarsal joint in late stance phase, which helps the foot turn to a rigid lever. This is important for the forward propulsion during pre-swing. It is stated that the longitudinal foot arch plays a leading role in the transition of the weight from the rearfoot to the lateral side of the foot [30]. We found an increased time to peak for subtalar pronation in the COP medial condition relative to normal running. It seems that medializing COP needs more control and is more difficult to perform, possibly due to tighter tissues in the medial side of the foot.Our results also showed that lateralizing COP when running has the potential to reduce rearfoot eversion and subtalar pronation at touchdown, which is promising as moderate evidence suggests that larger rearfoot eversion at touchdown is a risk factor for runners with Achilles tendinopathy [2].

Our results showed that lateralizing COP, besides reducing rearfoot eversion, significantly reduces peak hip adduction, peak hip internal rotation and hip internal rotation at touchdown compared to normal

Table 2
Results of one-way repeated-measures ANOVA analysis of measured variables[a] for each running condition.

Variable	Running condition			One-way repeated measures results			P-value between groups, 95 %(CI)		
	Normal COP	Lateral COP	Medial COP	F-value	P-value	Eta squared	Normal & lateral	Medial & normal	Medial & lateral
FPA in midstance (°)[b]	*−6.1 (2.9)*	*−6.6 (2.8)*	*−6.5 (2.7)*	*2.54*	*0.080*	*0.18*	*p = 0.051 (-0.01, 1.12)*	*p = 0.106 (-0.90, 0.07)*	*p = 0.670 (-0.16, 0.45)*
ML COP in midstance (mm)	*19.8 (8.3)*	*25.8 (8.1)*	*13.9 (8.1)*	*1709.09*	*< 0.001*	*0.99*	*p < 0.001 (-6.49, -5.49)*	*p < 0.001 (-6.42, -5.43)*	*p < 0.001 (-12.5, -11.26)*
Peak rearfoot eversion (°)	−8.5 (2.2)	−5.2 (2.5) *†	−11.2 (2.4) ‡	65.63	< 0.001	0.82	p < 0.001 (-4.26, -2.24)	p < 0.001 (-3.81, -1.61)	p < 0.001 (-7.90, -4.01)
Time to peak rearfoot eversion (% stance)	46.3 (2.2)	47.7 (2.9)	48.1 (3.2)	2.46	0.104	0.15	p = 0.365 (-3.71, 0.91)	p = 0.260 (-0.86, 4.46)	p = 0.999 (-1.53, 2.33)
Rearfoot eversion at TD (°)	3.2 (2.1)	5.3 (2.0) *†	1.0 (2.8) ‡	24.13	< 0.001	0.63	p = 0.002 (0.80, 3.31)	p = 0.002 (0.84, 3.64)	p < 0.001 (2.07, 6.51)
Rearfoot eversion excursion (°)	11.7 (3.5)	10.5 (3.3) *†	12.2 (3.2)	10.03	0.001	0.42	P = 0.020 (0.17, 2.22)	P = 0.443 (-0.37, 1.31)	P = 0.007 (0.44, 2.89)
Peak subtalar pronation (°)	4.4 (4.5)	−0.6 (5.0) *†	9.0 (5.9) ‡	71.85	< 0.001	0.84	p < 0.001 (3.47, 6.50)	p < 0.001 (2.67, 6.43)	p < 0.001 (6.67, 12.40)
Time to peak pronation (% stance)	70.1 (15.5)	73.1 (19.5)	75.5 (18.9) ‡	6.96	0.004	0.33	p = 0.138 (-6.72, 0.72)	p = 0.010 (1.23, 9.57)	p = 0.356 (-1.52, 6.32)
Subtalar pronation at TD (°)	−2.3 (4.9)	−6.4 (5.9) *†	1.3 (6.9) ‡	28.24	< 0.001	0.67	p = 0.004 (1.34, 6.8 5)	p = 0.001 (1.54, 5.80)	p < 0.001 (4.37, 11.17)
Subtalar pronation excursion (°)	6.7 (4.2)	5.8 (3.2)	7.6 (4.0)	3.26	0.053	0.19	p = 0.827 (-1.24, 3.02)	p = 0.584 (-0.88, 2.63)	p = 0.051 (0.45, 3.50)
Peak MLAA (°)	6.2 (2.2)	3.9 (2.3) *†	8.0 (2.2) ‡	101.50	< 0.001	0.88	p < 0.001 (1.75, 2.82)	p < 0.001 (1.12, 2.65)	p < 0.001 (3.16, 5.19)
Time to peak MLAA (% stance)	54 (8.1)	54.3 (7.9)	54.2 (5.8)	0.02	0.980	0.00	p = 0.999, (-6.10, 5.43)	p = 0.999, (-4.07, 4.47)	p = 0.999, (-3.30, 3.04)
MLAA at TD (°)	−1.0 (2.3)	−1.6 (2.4) †	−0.4 (2.1)	5.72	0.008	0.29	p = 0.103 (-0.10, 1.32)	p = 0.450 (-0.45, 1.58)	p = 0.030 (0.10, 2.25)
MLAA excursion (°)	7.2 (1.7)	5.5 (1.7) *†	8.5 (1.8) ‡	44.76	< 0.001	0.76	p < 0.001 (0.91, 2.44)	p = 0.005 (0.40, 2.23)	p < 0.001 (2.10, 3.90)
Peak hip internal rotation (°)	7.5 (5.3)	5.7 (5.2) *†	8.1 (5.4)	24.82	< 0.001	0.64	p = 0.001 (0.80, 2.87)	p = 0.167 (-0.18, 1.34)	p < 0.001 (1.32, 3.51)
Time to peak hip internal rotation (% stance)	52.9 (39.0)	64.5 (37.8)	52.9 (34.9)	1.74	0.194	0.11	p = 0.492, (-32.87, 9.81)	p = 0.999, (-23.0, 23.0)	p = 0.061, (-23.50, 0.44)
Hip internal rotation at TD (°)	4.8 (5.6)	2.1 (5.7) *†	5.7 (5.6)	47.30	< 0.001	0.77	p < 0.001 (1.67, 3.60)	p = 0.162 (-0.27, 2.11)	p < 0.001 (2.64, 4.47)
Hip internal rotation excursion (°)	2.8 (2.2)	3.6 (3.1) †	2.4 (2.5)	4.65	0.018	0.25	p = 0.311, (-2.05, 0.45)	p = 0.999, (-1.36, 0.67)	p = 0.007 (-1.97, -0.31)
Peak hip adduction (°)	14.5 (3.2)	13.3 (3.7) *	13.8 (3.9)	7.25	0.003	0.34	p = 0.011 (0.26, 2.14)	p = 0.081 (-1.42, 0.07)	p = 0.387, (-0.36, 1.39)
Time to peak hip adduction (% stance)	33.6 (4.1)	34.9 (5.6)	34.7 (5.8)	1.92	0.165	0.12	p = 0.233, (-3.08, 0.54)	p = 0.695, (-1.25, 3.39)	p = 0.999, (-1.63, 1.23)
Hip adduction at TD (°)	7.5 (2.5)	6.8 (2.1)	6.5 (2.6) ‡	4.42	0.021	0.24	p = 0.142, (-0.18, 1.60)	p = 0.045 (-1.84, -0.02)	p = 0.999, (-1.08, 0.65)
Hip adduction excursion (°)	7.0 (2.5)	6.5 (3.1)	7.3 (2.4)	2.67	0.087	0.16	p = 0.237, (-0.21, 1.18)	p = 0.999, (-0.61, 1.11)	p = 0.233, (-0.30, 1.79)
Peak vertical GRF (N/BW)	2.1 (0.16)	2.0 (0.18)	2.0 (0.18)	4.24	0.040	0.23	p = 0.088, (-0.01, 0.13)	p = 0.337, (-0.11, 0.03)	p = 0.383, (-0.02, 0.06)
Vertical average loading rate (BW/s)	30.4 (5.3)	30.6 (5.2)	29.8 (5.9)	0.76	0.477	0.05	p = 0.999, (-1.41, 1.20)	p = 0.999, (-2.74, 1.57)	p = 0.640, (-2.28, 0.80)

[a] Values expressed as mean (SD), [b] toe-out is negative, FPA foot progression angle, ML mediolateral, COP center of pressure, TD touchdown, MLAA medial longitudinal arch angle, GRF ground reaction force; * significant difference between normal and lateral COP $p < 0.05$, † significant difference between lateral and medial COP $p < 0.05$, ‡ significant difference between normal and medial COP $p < 0.05$.

COP. Rearfoot eversion has been associated with hip internal rotation in the closed lower limb kinematic chain [31]. Therefore, it seems plausible that change in rearfoot eversion is accompanied by change in hip internal rotation. In line with that, the results of the SPM analyses showed decreased rearfoot eversion and hip internal rotation during 0–80 % and 0–97% of the stance phase, respectively, when lateralizing COP compared to normal COP. However, our results did not show any significant differences in hip internal rotation and adduction when medializing COP compared to normal COP. While our results showed that the averaged mediolateral COP change at midstance in the analyzed data is identical in both medial and lateral direction (6 mm), medializing COP did not significantly affect hip internal rotation and adduction. It seems that change in medial COP needs to be larger to affect hip kinematics. Both excessive hip internal rotation and hip adduction are contributing factors in the development of iliotibial band syndrome and patellofemoral pain syndrome [2,32]. Hence it seems that lateralizing COP has also the potential to modify these two abnormal kinematics. A gait retraining intervention might be more effective when individuals show both excessive rearfoot eversion and hip internal rotation.

COP manipulation has been indicated as a conservative treatment for patients with knee osteoarthritis [33]. It has been demonstrated that lateralizing COP using moveable convex elements attached to the shoe is accompanied with decreased knee adduction moment in patients with knee osteoarthritis [33]. These passive half dome shoes most likely increase rearfoot eversion as was shown in a study reporting that wearing lateral wedged foot orthotic devices increases rearfoot eversion [17]. Also, studies have shown that medializing COP actively reduces knee adduction moment [15,34]. However, they did not report whether medializing COP actively changes rearfoot eversion.

Manipulating foot pressure during locomotion may impact muscle

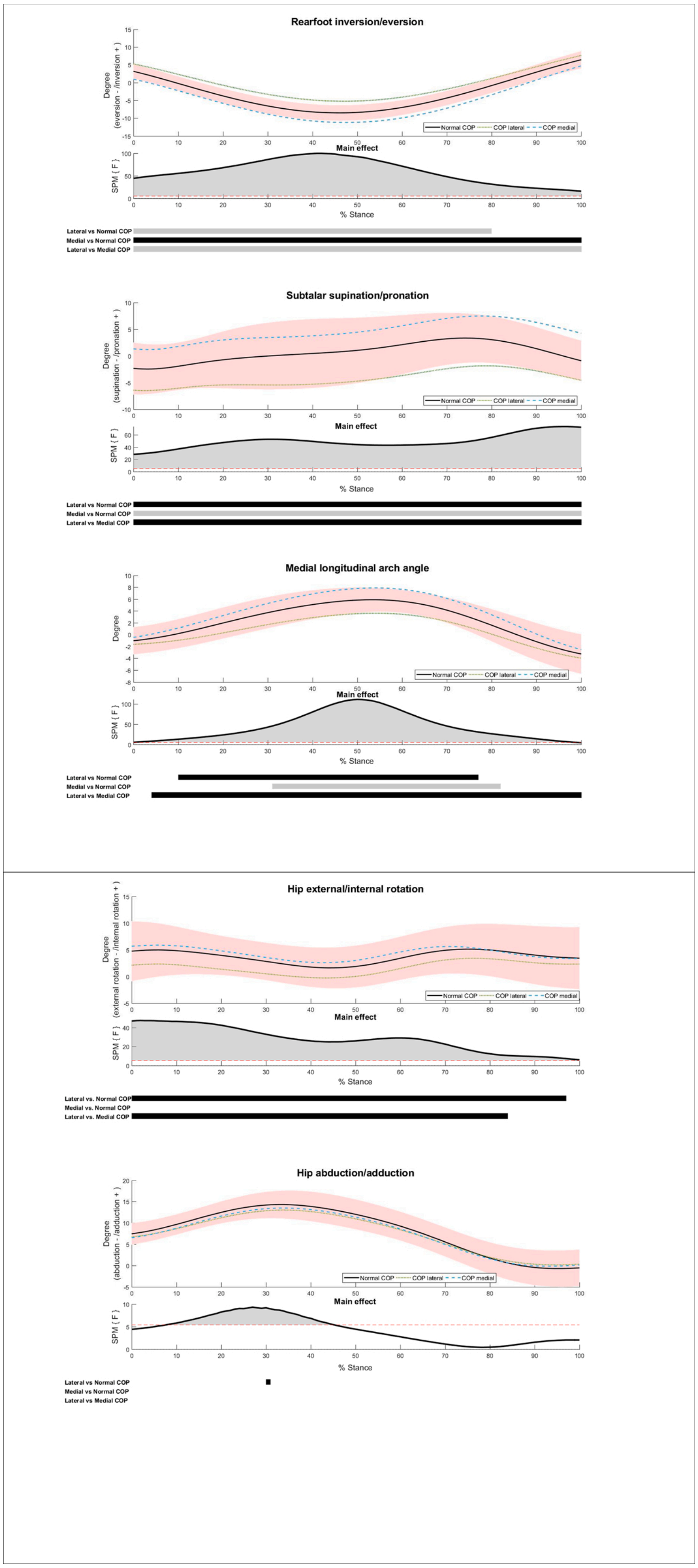

Fig. 4. Graphical representation of between-condition comparison for the kinematic outcomes. Red shaded area represents ±1 SD of the normal COP condition. SPM (F) denotes the F value; the horizontal dashed red line denotes the critical thresholds for statistical significance; the grey area indicates significant main effects in the corresponding portion of the stance phase ($p < 0.05$). The horizontal bars indicate the corresponding portion of the stance phase with statistically significant differences between condition pairs ($p < 0.05/3 = 0.017$): lateral vs. normal COP, medial vs. normal COP, and lateral vs. medial COP. (For interpretation of the references to colour in the Figure, the reader is referred to the web version of this article).

activation. For example, lateralizing foot pressure may impact the posterior tibialis muscle which is responsible for controlling subtalar pronation, hence may be leading to balancing muscle activation as well as strengthening these muscles. A prospective study reported that runners who have a more lateral COP develop less Achilles tendinitis, plantar fasciopathy and medial tibial stress syndrome [35]. Prospective studies showed that increased pressure underneath the medial side of the foot and larger rearfoot eversion during running are associated with exercise-related lower leg pain and medial tibial stress syndrome [5,36]. Their data along with our findings highlight the potential for lateral COP modification as an alternative to using foot orthoses, motion control shoes, and taping in individuals with excessive rearfoot eversion for normalization of rearfoot kinematics. However, it should be taken into account that lateralizing COP may lead to increased pressure underneath the lateral border of the foot, resulting in increased risk of injuries such as stress fracture in the foot [37]. It has been demonstrated that a more laterally directed force displacement is associated with lower leg overuse injuries [38]. Therefore, clinicians and researchers are advised to consider the pros and cons of mediolateral COP for modifying rearfoot eversion when planning any prevention or treatment program.

One major concern regarding rearfoot eversion is that there is no standardized range for rearfoot eversion during running to determine the extent to which rearfoot eversion falls into typical or atypical movements. Studies investigating rearfoot eversion for lower limb injuries mainly compared rearfoot eversion between non-injured and injured (history of injury) individuals. Therefore, this makes rearfoot eversion classification difficult. One possible solution could be to link the static measurement of rearfoot eversion or medial longitudinal arch angle (using standardized methods) to rearfoot eversion during running, However, these static measurements cannot always be reliable alternatives to rearfoot eversion during running [39].

We assessed FPA during running using real-time visual feedback as one of our previous studies showed that FPA impacts rearfoot eversion [23]. Changing FPA internally is accompanied with reduced rearfoot eversion; in contrast, changing FPA externally increases rearfoot eversion. FPA modification is widely used as a strategy to reduce knee joint load in patients with knee osteoarthritis [40,41]. FPA is associated with changes in lower limb biomechanics during gait. For example, a study showed that greater hip external rotation is associated with greater FPA (toe-out) [42]. FPA has the ability to easily adapt itself to the biomechanical changes imposed to lower extremity and also to specify the lever arm of the GRF during gait [43]. This information highlights the key role of FPA in the kinetic chain and also its potential to change during gait modification suggesting that FPA should be controlled during gait retraining. Hence in our study, participants were verbally instructed to run on the lateral and medial side of the foot while following their normal FPA (in midstance) projected to the screen in real time. Our observations indicated that lateralizing or medializing foot pressure during running changes the normal FPA if no feedback is given. Participants were nonetheless able to adopt their normal FPA using the real-time visual feedback. In fact, the FPA display in the real-time feedback during familiarization helped subjects adapt to the experiment, because the pointer was aligned with the subject's FPA so it could be easily perceived.

Embedded sensor insole systems show promising results for giving real-time feedback on COP [44–46]. These approaches may be used to manipulate mediolateral COP to a given quantity (both in-lab and out-of-lab running) to modify atypical rearfoot eversion or MLAA during running. According to our results, it seems that lateralizing COP at midstance as much as 6 mm can reduce peak rearfoot eversion, subtalar pronation and MLAA as much as 3°, 5°, and 2°, respectively; medializing COP acts in the opposite direction with similar values. This information can be used by clinicians and researchers when planning to modify rearfoot eversion using feedback on mediolateral COP. To see the effectiveness of gait retraining on rearfoot eversion over time or to be sure that change in rearfoot eversion is at the desired level, tracking rearfoot eversion might be useful. As application of 3D motion capture systems is not easy in clinical practice. 2D measurement of rearfoot eversion using smartphone application can be used as a surrogate to 3D measurement [47].

4.1. Limitations and recommendations for future studies

Results of this study should be interpreted with some caution since there were some limitations. Our study included a relatively small sample size of only healthy female runners. This choice was made based on the differences in biomechanical characteristics between genders [48]. Therefore, our results should be interpreted carefully and cannot be generalized to male runners and/or injured runners. Further research is needed to investigate whether our results have the same effect on male and injured runners with atypical rearfoot eversion, subtalar pronation, and/or MLAA. Since all runners ran with rearfoot strike, our results may not be reproducible while running with either mid foot or fore foot strike. Running speed was set at 8 km/h, so it is not clear whether the same results would be found at higher or slower speeds. Participants ran on a dual-belt treadmill which may affect step width [49]. As step width can affect kinematics, our results might not be reproducible during single-belt treadmill or overground running. Kinematic data were collected at 100 Hz while higher sampling rate is more common in studies conducted on running kinematics (e.g. 200 Hz). As increase in sampling rate will increase the number of data points within each step for the kinematic signals, this may affect the outcome accuracy. Future research should investigate the potential positive and negative consequences of mediolateral COP modification on other lower limb biomechanical parameters during running. This will help clinicians better understand for whom changing mediolateral COP to modify atypical rearfoot eversion is applicable or more effective and for whom it is not. Moreover, further studies are warranted to investigate the viability of mediolateral translation of COP during running in the long term and whether modifying COP in runners with atypical rearfoot eversion helps to prevent or manage RRIs or indeed leads to less injuries.

5. Conclusion

The present study demonstrated that translation of COP along the mediolateral foot axis at midstance has a significant effect on rearfoot eversion, MLAA, subtalar pronation and peak hip adduction during running. Running with more lateral COP reduced peak rearfoot eversion, peak MLAA, and peak subtalar pronation compared to normal running, while running with more medial COP increased these variables. Mediolateral COP manipulation at midstance did not significantly change vGRF and VALR compared to normal running. These results might serve as a basis to help clinicians and researchers prescribe gait modifications for runners with atypical rearfoot eversion or MLAA.

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Declaration of Competing Interest

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