

Medial longitudinal arch collapse of flexible flatfoot during running

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ABSTRACT

Background: Uncertainty regarding an actual amount of the medial longitudinal arch (MLA) collapse during running among the flexible flatfoot and normal foot individuals has been existed. Additionally, it was unclear whether the alterations of muscle activation and lower limb motion enhanced the MLA collapse in the flexible flatfoot individuals.

Objective: This cross-sectional study aimed to compare the dynamic MLA collapse during running between the flexible flatfoot and the normal foot individuals. It also sought to investigate whether the greater MLA collapse in the flexible flatfoot individuals was related to the alteration of kinematics, kinetics, and muscle activation.

Materials and methods: Sixteen-volunteered males participated in this study. Based on their static navicular drop test, they were assigned into either the flexible flatfoot or the control groups totally 8 participants in each group. They performed barefoot running at a speed of 3.5 ± 0.25 m/s while the three-dimensional motion of the lower body, electromyography and ground reaction force were recorded synchronously. A subject-specific foot model was constructed from the additional reflective markers attached on their feet to determine the maximal dynamic navicular drop (ND_{Dmax}) during the stance phase.

Results: The flexible flatfoot group had greater MLA collapse during midstance and propulsion which coincided with greater hip and knee flexions, and ankle dorsiflexion. The 95% CI of their ND_{Dmax} was in the ranges of 27.18 to 37.68 mm while that of the control group ranged between 20.71 mm and 27.82 mm. The flexible flatfoot group had lesser activations of the hip, knee, and ankle extensors to counteract the flexion moments. They also had larger ankle inversion ($p=0.04$) without greater activation of peroneus longus.

Conclusion: Greater MLA collapse in the flexible flatfoot individuals at midstance and propulsion was associated with greater hip and knee flexions, and ankle dorsiflexion. It might be the consequence of the lower activation of their leg extensors.

Introduction

Flexible flatfoot is a less rigid medial longitudinal arch (MLA) which collapses from weight bearing and recoils back after unloading.¹ The MLA is formed by the calcaneus, talus, navicular, three cuneiform bones, and the first, second, and third metatarsals. The role of MLA is shock absorption which attenuates ground reaction forces transferred to the lower limb.^{2,3} Malfunction of plantar intrinsic and extrinsic foot muscles leads to unstable and malalignment of MLA.⁴ Reduction of the abductor hallucis muscle activation from nerve blockage⁵ and fatigue exercise also resulted in MLA collapse.⁶ Greater MLA collapse may

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lead to the exercise-induced leg pain⁷ and plantar fasciitis in runners.⁸

Nevertheless, the evidence regarding the contribution of muscle function to MLA collapse has been unclarified. There was reported that runners with greater MLA collapse and foot pronation have larger plantar intrinsic foot muscle, the flexor digitorum brevis and flexor digitorum longus.⁹ However, the size of their peroneus longus (PL)^{9,10} did not differ from that of normal foot runners. Hence, it was likely that other factors besides foot muscle imbalance or overactivation of PL relate to the MLA collapse during running.

Abnormal motion of the proximal joint may enhance the MLA collapse. Nevertheless, there were controversial data¹¹ from the difference in the study interventions such as footwear condition,¹¹ locomotion speed,¹²⁻¹⁶ foot classification,¹⁷ and foot modelling.^{9,12-14} During walking, the FF adolescents¹³ and adults¹⁴ had greater plantar flexion, smaller dorsiflexion,¹² and eversion¹² of the tibiotalar joint. In contrast, the low, compared to the high, MLA runners exhibited greater ankle dorsiflexion¹⁵ and knee flexion¹⁶ without the difference in rearfoot eversion during barefoot running. Moreover, the similar static pronated feet presented different dynamic arch heights at the stance phase.^{17,18}

Therefore, this study aimed to determine the amount of MLA collapse that occurred during running in the flexible flatfoot and normal foot individuals. Moreover, it also investigated whether the difference in this dynamic MLA collapse between groups was related to the alteration of kinematics, kinetics, and muscle activation.

In this study, the MLA collapse was determined from the navicular height relative to the plantar surface of foot during the stance phase of running using the subject-specific foot model. The difference between groups was analyzed across the entire stance phase of running using Statistical Parametric Mapping. We hypothesized that the flexible flatfoot individuals had greater dynamics MLA collapse compared to normal foot individuals. Furthermore, the alteration of the lower limb motion and muscle activation related to the difference in dynamics MLA collapse between groups.

Materials and methods

Participants

Based on the peak hindfoot inversion from walking trials of flat and normal foot individuals,¹² the effect size of 1.94 and 95% power to detect the difference between group at 0.05 alpha level, the sample size of each group should be at least 7 (G^*power 3.1.9.7).¹⁹ Totally 16 male volunteers, aged 18-22 years to account for the foot growth, participated in the study. Inclusion criteria were (1) the right foot dominant and (2) engaging in 10 km running or less, 2-3 times per week for at least 1 year. Exclusion criteria were having (1) any lower extremity injuries or neurological conditions within 6 months prior to the study and (2) leg length discrepancy greater than 10 mm. They were allocated to either the flexible flatfoot

(FF) or the control (CON) groups based on their static navicular drop (ND_s). ND_s was the difference in navicular heights measured in sitting and standing positions on the first visit for group allocation. The most prominent part of navicular tuberosity was marked by a trained researcher and used for the navicular height and ND_s evaluations. The test-retest intraclass correlation, ICC [3, 1], from the repeated ND_s equaled 0.895 and its 95% CI was in the range of 0.780-0.955. The standard error of the measurements was 1.15 mm.

ND_s of young adults (24.13 ± 3.41 years of age) with low MLA ($ND_s > 10$) had an excellent correlation with the foot posture index-6, the clinical assessment for foot posture, at 0.818 (Pearson's product moment correlation coefficient, r).²⁰ The test-retest intraclass correlation (ICC) of ND_s was reported in the range of 0.950-0.955.²⁰ ND_s of the middle-aged adults (42.7 ± 7.3 yrs) also had moderate to strong correlation with the radiography ($r = 0.61-0.89$).²¹ The average ND_s of the healthy male (age 19.6 ± 2.8 yrs; body height of 172.9 ± 5.3 cm) detected from radiography equaled 0.7 ± 0.2 cm.²² ND_s of the 5-8-year-old children, who were identified as having flexible flatfoot from abnormal radiological findings, equaled 15.56 ± 1.93 mm.²³ However, MLA of children mature with age. ND_s of adults with low MLA (mean 11.83 ± 1.68 mm) strongly correlated with the foot posture index-6 score ≥ 6 which indicated foot pronation.²⁰ Moreover, the 2-D dynamic navicular drop, occurred during walking, was in the range of 1.7 to 8.7 mm (95% CI).²⁴ After the initial ND_s screening of 31 volunteers, therefore, participants who had ND_s greater than 10 mm were assigned to the FF group and those who had ND_s less than 9 mm were assigned to the CON group, total 8 participants in each group. The Research Ethics Committee approved (COA64/044) the study, and the informed consents were obtained from all participants.

Measuring muscle activation

A wireless electromyogram (EMG) system (Noraxon Ultium®, Noraxon USA Inc., Scottsdale, AZ) was synchronized with infrared cameras and force plates to record muscle activation during running at a sampling rate of 2000 Hz. Based on SENIAM (www.seniam.org), adhesive Ag/AgCl surface electrodes (BlueSensor P, Ambu Inc., Malaysia) were attached to skin, with the interelectrode distance of 3.4 cm, above peroneus longus (PL), gastrocnemius medialis (GM), vastus lateralis (VL), vastus medialis oblique (VM), semitendinosus (ST), tensor fasciae latae (TFL), and gluteus medius (GLM) muscles. For tibialis anterior (TA), electrodes were placed at 47.5% between the tip of fibular head and the medial malleolus.²⁵ Before running, EMGs of each muscle were recorded during the maximal voluntary contraction (MVC) following protocol in table 1. After setting up position, participants were encouraged to perform movement against the resistance from the researcher with their maximal effort for 3-5 seconds, total 3 trials for each muscle testing. A 2-minute rest was allowed between the tests.

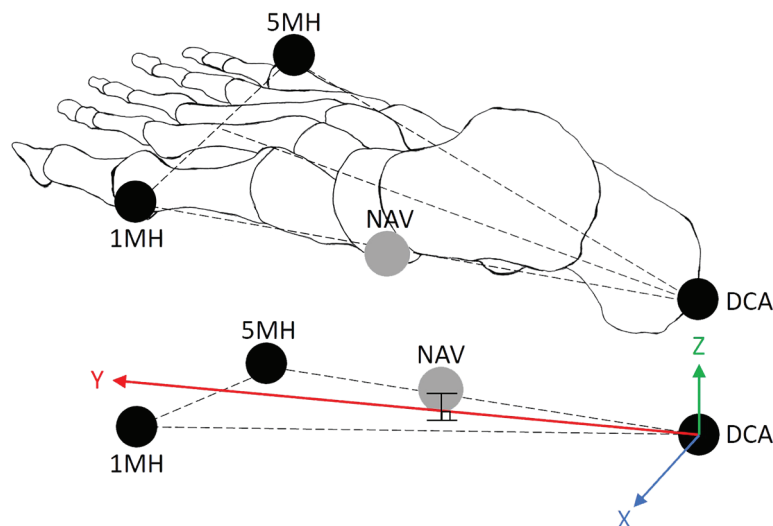
Table 1. MVC protocol and testing position for each muscle.

Muscle(s)	Movement	Position
TA	Ankle dorsiflexion with inversion	Seated, ankle plantarflexed to 10 degrees. ²⁵
PL	Ankle eversion with plantarflexion	Supine, hip, and knee flexed to 0 degrees.
GM	Plantarflexion	Prone, hip and knee flexed to 0 degrees, ankle dorsiflexed to 90 degrees.
VL&VM	Knee extension	Seated, hip flexed to 90 degrees and knee flexed to 60 degrees. ²⁶
ST	Knee flexion	Prone, knee flexed to 60 degrees. ²⁷
TFL	Hip abduction with flexion and internal rotation	Semi-prone, hip flexed to 45 degrees, abducted to 30 degrees and knee flexed to 90 degrees.
GLM	Hip abduction with extension and external rotation	Side-lying, hip flexed to 0 degree, abducted to 25 degrees and knee flexed to 90 degrees.

Three-dimensional motion capture and foot modelling

Seven near infrared cameras (Vicon Motion System Ltd., Oxford, UK) captured running motion of participants at 200 Hz. They were synchronized with three force plates (Kistler Corporation, Winterthur, Switzerland) which collected ground reaction forces (GRFs) at 1000 Hz. Retro-reflective markers were attached to the skin of lower extremities following the lower body Plug-in-Gait marker set. An additional foot markers of the foot model were attached at the medial aspect of the 1st metatarsal head (1MH), the lateral aspect of the 5th metatarsal head (5MH), the navicular tuberosity (NAV) and the distal of the calcaneus (DCA) to compute navicular height during running.²⁸ The root mean square errors of this model for detecting the navicular height of healthy adults over the stance phase of barefoot walking was 0.25 mm compared to wearing a ski boot with rigid marker set.²⁸ After static calibration, participants performed a 20-meter barefoot running on synthetic track at the speed of 3.5 ± 0.25 m/s. This running speed range was chosen because it was in the average running speed range of amateur runners²⁹ and the preferred running speed range of recreational active

runners.³⁰ Running speed was monitored by the infrared timers (Fitness Technology, KMS, Australia). After practice, three running trials which the foot landed in the middle of the force plate were captured. Next, kinematic, and kinetic variables were computed using inverse dynamics (Vicon Nexus®, version 2.8.2). The foot model was constructed to determine the dynamic navicular height (NH_d), which was defined as the perpendicular distance from NAV to the plane formed between 1MH, 5MH, and DCA (Figure 1), using customized software (Matlab R2022a, The Mathworks Inc., Natick, USA). The origin of the foot model was located at the calcaneus and the directions of the local X, Y, Z axes were shown (Figure 1). The local coordinate of NAV was determined from its global coordinate and the rotational matrix between the global and the local axes. The maximum dynamic navicular drops (ND_{Dmax}), representing maximal MLA collapse, was defined as the difference between NH_d at heel strike and the minimal NH_d at stance phase of running. The percentage change of ND_{Dmax} was the difference between the ND_{Dmax} and ND_s as the percent change from ND_s .

**Figure 1.** The dynamic foot model and location of model's markers.

Measuring strengths of dorsiflexor and plantarflexor

The peak torques of dorsiflexor and plantarflexor were measured using an isokinetic dynamometer (Biodex System 4 Pro, Biodex medical systems, USA) to detect the difference in muscle strength among the groups. The participants sat on the dynamometer chair with backward tilt around 70 degrees. Their right hip and knee were flexed at 110 degrees and 70-80 degrees, respectively, until the tibia was parallel to the ground. Then, the right foot was fixed with the tested footplate, and the lateral malleolus was aligned with the dynamometer's axis of rotation. Next, participants performed 5 repetitions of plantarflexion and dorsiflexion with their maximal effort using the concentric-concentric mode at the speed of 30°/sec. The maximal plantarflexion and dorsiflexion torque were normalized by body weight (%BW) and reported as the plantarflexion and dorsiflexion peak torque. The test-retest intraclass correlation coefficients (ICC) of ankle dorsiflexion and plantarflexion were 0.93 and 0.91, respectively.³¹

Data processing

The EMG signals collected during running were rectified and normalized to their maximal EMG amplitude obtained from MVC. Next, the smoothing technique, a moving root mean square with an exponential weight method was applied to the normalized EMG. Thereafter, it was filtered by a low-pass infinite impulse response

filter with the passband frequency below the signal's mean frequency (Matlab R2022a, The Mathworks Inc., Natick, USA). The normalized EMG, NH_D , kinematics and kinetics at each percentage of the stance phase were determined from cubic spline interpolation. In addition, the normalized EMGs at each percent stance were summed up throughout the stance phase to represent total muscle activation, iEMG.

Statistical analysis

Normal distributions of variables were observed using the Shapiro–Wilk test. For the normal distributed variables, the between-group difference and the association between variables were evaluated using the unpaired T-test and Pearson's product moment correlation, respectively. The Mann-Whitney U test was applied to the non-normal distributed variables, the ST iEMG (SPSS for Windows, version 25, IBM Inc., Armonk, NY, USA). Classical statistical parametric mapping (SPM), the unpaired T-test from the open-source package spm1d version M.0.4.11(spm1d.org, © T. Pataky), was deployed on all trials of the time domain data to assess the between group difference at each percent stance using MATLAB (Matlab R2022a, The Mathworks Inc., Natick, USA).³² The statistical significance was set at $p < 0.05$.

Results

There were no between-group differences in running speed, dorsiflexion and plantarflexion peak torques (Table 2).

Table 2. Characteristics of the participants, static navicular drop, running speed and peak torques.

	CON (N=8)		FF (N=8)		p value
	Mean±SD	95% CI	Mean±SD	95% CI	
Age (year)	19.75±0.70	19.15-20.34	19.12±1.12	18.18-20.06	0.205
Weight (kg)	61.62±5.68	56.87-66.37	62.16±4.13	58.70-65.61	0.832
Height (cm)	1.70±0.04	1.67-1.74	1.72±0.07	1.66-1.79	0.570
ND _s (mm)	7.13±1.31	6.04-8.23	13.54±1.43	12.34-14.74	0.000**
Running speed (m/s)	3.53±0.11	3.44-3.63	3.47±0.10	3.39-3.56	0.267
T _{DF} (%BW)	48.51±9.74	38.36-54.66	47.68±7.46	41.44-53.92	0.791
T _{PF} (%BW)	103.50±24.03	83.40-123.59	124.81±32.79	97.39-152.22	0.160

Note: CON: control group, FF: flexible flatfoot, ND_s: static navicular drop, kg: kilogram, cm: centimeter, mm: millimeter, m/s: meter/second, T_{DF}: dorsiflexion peak torque, T_{PF}: plantarflexion peak torque, BW: body weight,

** statistically significant difference at $p < 0.01$ using unpaired T-test.

There was a negative correlation between the percentage change of ND_{Dmax} and plantarflexion peak torque ($r=-0.438$, $p=0.045$, $N=16$) from the pooled data of both groups (Figure 2A). Comparing between groups, the FF group had lower dynamic navicular height

(Figure 2B) which resulted in greater ND_{Dmax} (Figure 2C). The 95% confidence interval (CI) of ND_{Dmax} of the FF group was in the range of 27.18-37.68 mm while that of the CON group was in the range of 20.71-27.82 mm.

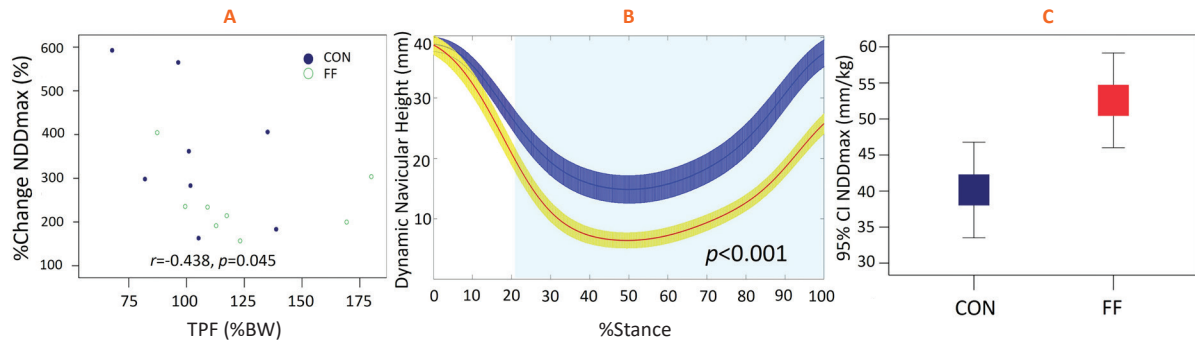


Figure 2. The maximum dynamic navicular drops (ND_{Dmax}) and the dynamic navicular height (NH_d) during the stance phase. A: correlation between the percentage change of ND_{Dmax} and plantar flexion peak torque (TPF) from the pooled data, B: Ensemble graph of NH_d at each percent of the stance phase, C: The 95% CI of ND_{Dmax} solid line: mean, error bar: SEM, blue line: control group, red line: FF group, CON: control, FF: flexible flatfoot. The shaded area shows the significant difference between groups using the SPM analysis with unpaired T-test.

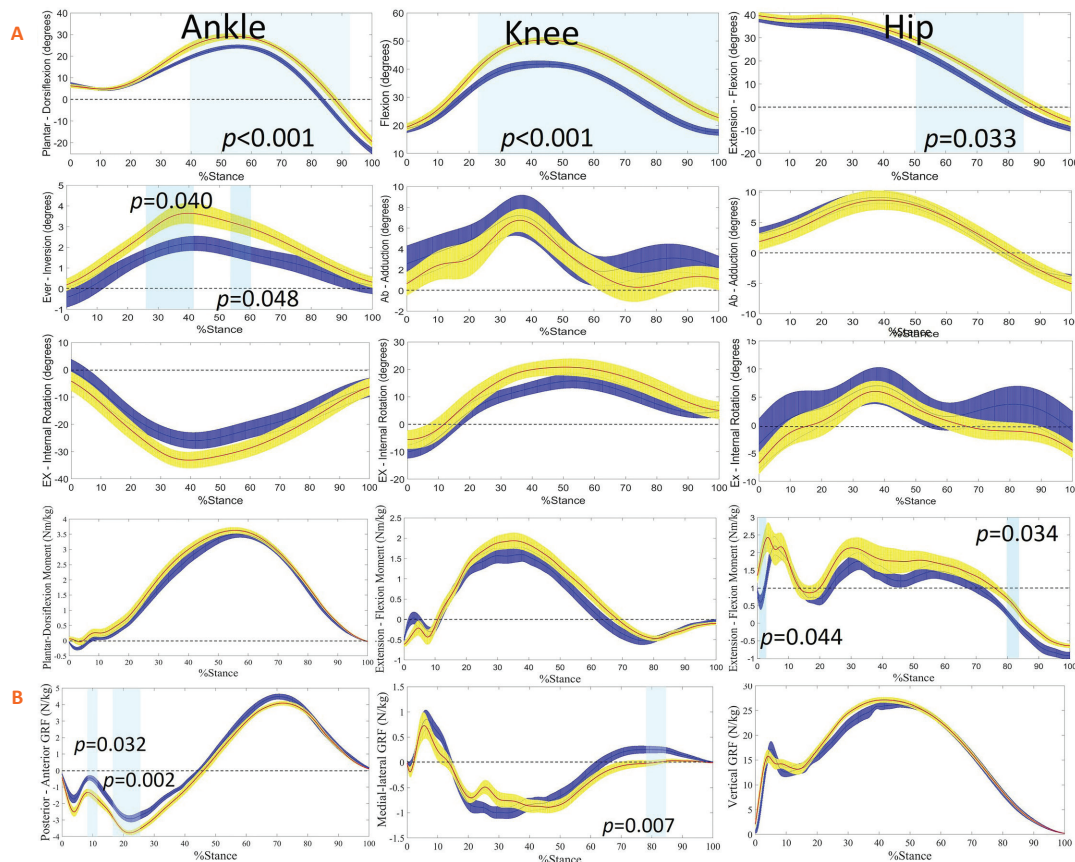


Figure 3. Ensemble graphs of joint angles, moments, and GRF during the stance phase. A: joint angles and moments, B: GRF were plotted against the stance phase of running. Positive values of joint angle and moment are dorsiflexion/flexion, inversion/adduction, and internal rotation. Positive values of GRF are anterior, lateral and superior, solid lines: mean, error bars: SEM, blue lines: control group, red lines: FF group. Shaded areas show significant difference between groups using the SPM analysis with unpaired T-test.

The dynamic navicular drop of the FF group was greater than that of the CON group (Figure 2B) during midstance and propulsion (20-100% stance). This incidence coincided with their greater knee flexion angle (Figure 3A). The greater ankle dorsiflexion and hip flexion angles also existed in the FF group during propulsion. Moreover, the FF participants plantarflexed their ankle with a smaller angle during push-off. The FF participants also had a greater hip flexion moment at initial impact and a smaller hip extension moment during push-off. A larger ankle inversion existed in the FF group during the initial and mid propulsions. Nevertheless, there were no

between group differences in the adduction-abduction and the internal-external rotation angles of the knee and the hip. Compared to the CON participants, the larger GRF applied to the FF participants posteriorly during the midstance while the smaller GRF applied laterally during push-off (Figure 3B).

The lesser VL, VM, GLM activations (Figure 4) occurred in the FF group at initial propulsion (45-50% stance) and their lesser GM activation also presented at push-off (75% stance). Moreover, TA, GM, GLM and ST iEMGs of the FF group were lesser than those of the normal foot group (Figure 5).

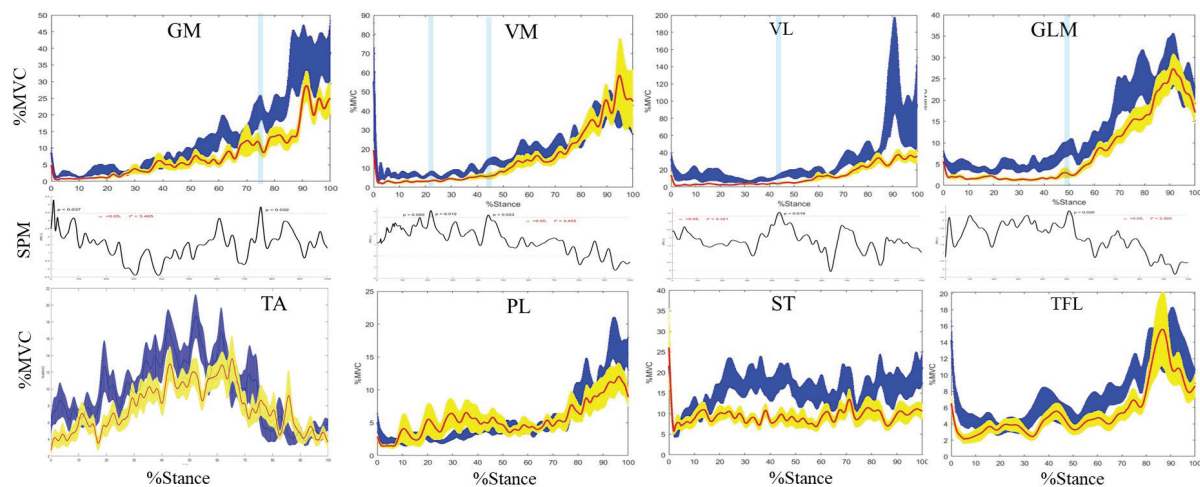


Figure 4. Ensemble graphs of muscle activation during the stance phase of running. Solid lines: mean, error bars: SEM, blue lines: control group, red lines: FF group. Shaded areas show the significant difference between groups using SPM analysis with unpaired T-test.

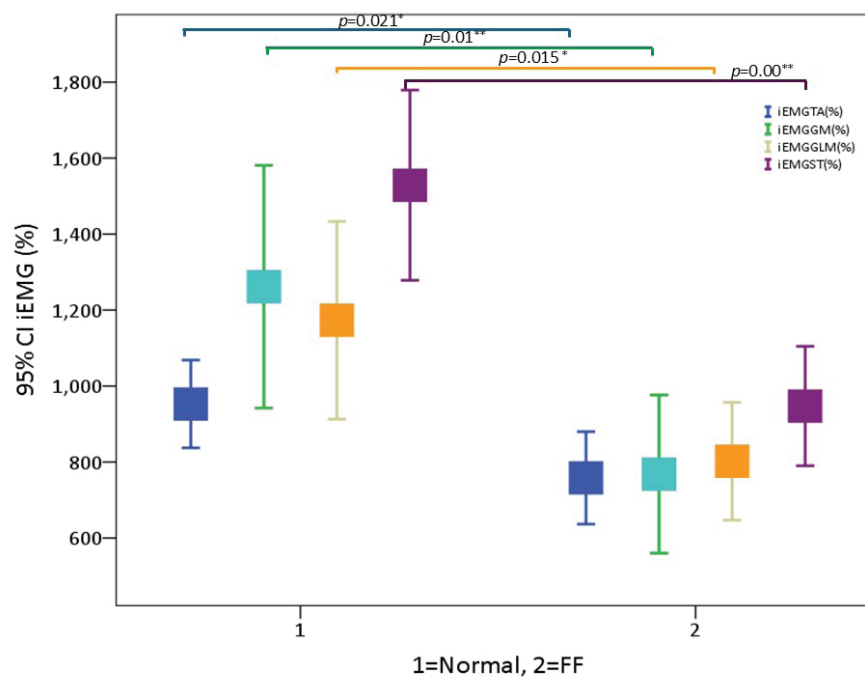


Figure 5. The 95% CI of iEMG. The differences between groups of TA, GM and GLM iEMGs were assessed by the unpaired T-test while that of ST was detected from the Mann-Whitney U test.

Discussion

This study investigated whether the alterations of muscle activation and limb motion in the FF individuals associated with the greater MLA collapse during running. Our 3-D foot model supported the hypothesis that the FF participants had greater dynamic MLA collapse during running compared to the normal foot participants (27.18-37.68 mm VS 20.71-27.82 mm, the 95% CI of ND_{Dmax}). Moreover, this ND_{Dmax} from the normal foot's running was greater than the ND_{Dmax} range of 1.7-8.7 mm from 2-D walking analysis.²⁴ However, running at the same speed as walking also increased NH_D by 3.5 mm.³³ The alteration of lower limb kinematics,³³ an elevation of the foot arch from greater hallux dorsiflexion,³⁴ the stiffer arch and an increase in MLA angle during propulsion,³⁵ and a larger fall of MLA with running may relate to the greater NH_D during running.³⁶ Moreover, our study showed that the greater percentage change from static to dynamic navicular drops during running was correlated with lower plantarflexion peak torque. Therefore, the weakness of plantar flexor could increase MLA collapse during running.

The FF group had greater MLA collapse during midstance and propulsion which coincided with greater ankle dorsiflexion, knee, and hip flexions. It supported the previous finding¹⁵ that FF individuals exhibited greater ankle dorsiflexion during propulsion. However, neither excessive ankle eversion nor substantial PL activation existed in the FF group. In contrast, they had larger ankle inversion during midstance and propulsion. Greater activation of tibialis posterior and lesser activation of PL were also found in the flat-arched, compared to the normal-arched, groups during the propulsion of walking.³⁷ Therefore, PL may not be a major contributor to the dynamic MLA collapse in FF individuals.

Greater dynamic MLA collapse in the FF group could be the consequence of greater lower extremities collapse as well as an insufficient activation of leg extensors. Their larger hip flexion during propulsion and smaller hip extension moment during push-off resulted from the greater hip flexion moment at initial impact and lesser activation of hip extensors, GLM and ST, respectively. Similarly, the larger posterior GRF at midstance accompanying with the lesser activation of knee extensors, VL and VM also led to the greater knee flexion. The former study also reported the increasing of peak knee flexion from the reduction of knee stiffness in the low arch runners, whereas the activation of knee extensor, VL, raised knee stiffness.³⁸ In addition, the knee and hip stiffnesses had positive relationship with ankle stiffness.³⁹ Moreover, the greater dorsiflexion angle occurred in the FF group during propulsion should be the consequence of their lesser plantar flexor activation, GM, because the anterior and vertical GRFs were not significantly different between groups.

The limitation of this study was that skin artifacts could induce marker vibration during running. The skin marker set of the similar MLA model underestimated the MLA deformation about 2 degrees compared to the radiography-based MLA deformation.⁴⁰ Hence, the exact

amount of navicular drop occurs during running could be a bit greater than the computed navicular drop of our study.

Conclusion

The FF participants had greater MLA collapse during running which associated with the lower activation of leg extensors to counteract the hip, knee, and ankle flexions at midstance and propulsion. Greater hip flexion moment at the initial impact and the lesser activations of GLM, and ST enhanced larger hip flexion in the FF group. The larger posterior GRF and lesser activations of VL and VM resulted in their greater knee flexion. Lesser GM activation in the FF participants related to greater dorsiflexion even though both FF and normal foot groups had comparable plantar flexor strength. Nevertheless, greater percentage change of dynamic from static MLA collapse was correlated with lower plantarflexion peak torque.

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Conflict of Interest

The authors declare no conflict of interest.

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