Medial longitudinal arch biomechanics evaluation during gait in subjects with flexible flatfoot

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Purpose: Medial longitudinal arch (MLA) strengthening has been considered an important part of successful flatfoot treatment. But, to date, the biomechanical loading behavior of the medial arch in flatfoot has not been evaluated. This study aimed to evaluate the MLA moment, MLA deformation angle, foot kinematics and ground reaction forces (GRF) in both normal foot and flatfoot groups. Methods: Each participant's foot was classified according to arch type using foot prints and radiographs. Twenty-eight non-obese adults (13 flatfeet and 15 normal feet) were involved. The biomechanics data were collected in a 3D motion analysis laboratory. The MLA biomechanics were calculated. Hindfoot and forefoot kinematics were also analyzed. Results: The flatfoot group had a significantly greater peak eversion MLA moment ($p = 0.005$) and a smaller peak MLA deformation angle ($p < 0.05$) during specific subphases. The peak of hindfoot plantarflexion ($p < 0.05$) and internal rotation ($p < 0.05$) and the peak of forefoot abduction ($p < 0.05$) in the specific subphases were greater in the flatfoot group. The flatfoot group also had significantly smaller peak vertical GRF ($p < 0.05$) during late stance and larger peak medial GRF ($p < 0.05$) during mid stance. Conclusions: This study found a significantly greater eversion deforming force acting at the MLA structure, greater hindfoot and forefoot motion, less MLA flexibility and abnormal GRF in a flatfoot group during walking, which reflected the deficit of foot function in a flatfoot group.

Key words: biomechanics, gait, medial longitudinal arch, flatfoot

1. Introduction

Flexible flatfoot is a reversible complex foot deformity, which is comprised of the heel valgus, lower medial longitudinal arch (MLA) and abductory foot twist. Subtalar joint (STJ) is the main joint to evaluate flatfoot pathomechanics. The medial deviation of subtalar joint axis location is used to determine the pronation pathological force in flatfoot [11]. But the precision in the determination of the STJ axis needs clinical practice and experience. In addition to the STJ axis, the stability of the MLA is also an important foot structure in flatfoot motion. The talonavicular joint

(TNJ) is the keystone joint of the MLA structure located at the apex of the MLA and is a dual member of the subtalar complex joint and transverse tarsal joint [21]. Arangio and colleagues [1] found that 5 degrees of STJ pronation increased TNJ moment by 47%. The complexity of this articulation is called the acetabulum pedis [19], [21]. Ligamentous laxity is one of the related conditions with the flatfoot [19]. The hyper-flexibility of the acetabulum pedis allows an excessive motion of talar head, calcaneus and navicular bones [8]. This can cause peritalar subluxation, an unlocked midtarsal joint, and unstable foot in pronation posture. The lever arm of the MLA apex is increased and a larger moment is applied on the

joint. This leads to ongoing injury of the weak spring ligament complex (SLC) and then posterior tibialis tendon insufficiency develops with further foot destabilization [10]. As part of shock absorption and rigid lever assistance of the foot during locomotion the function of TNJ and MLA should be evaluated in flatfoot gait [16], [18].

MLA structure evaluation has been used in clinical practice to evaluate foot function, such as navicular height, navicular drop test, and navicular drift test during static and dynamic conditions [12]. Additionally, motion analysis of the MLA, MLA movement and TNJ moment, was evaluated during walking to determine the relationship to the tarsal joints in normal foot [1], [15], [24].

In flatfoot, Tome and colleagues [25] suggested that in addition to the hindfoot angle, the MLA angle and forefoot kinematics were also important parameters to develop a treatment plan for posterior tibial tendon dysfunction (PTTD) in patients. They found a significantly greater hindfoot eversion, MLA and forefoot abduction angles across the specific phase of stance in PTTD patients in comparison with the normal group. Besides that, the load at the MLA of flatfoot is essential, too. Arangio and colleagues [2] evaluated the TNJ moment with a mathematical model in a cadaveric flatfoot model. They found a larger significant TNJ moment in weakened MLA stabilizers during talar dome loading.

At present, a study of the dynamic MLA biomechanics in flatfoot has not been reported. This study experimentally tested the hypothesis that there would be differences of the MLA biomechanics (MLA moment and MLA deformation [MLAD] angle), hindfoot and forefoot kinematics, and ground reaction forces between normal foot and flatfoot groups in each gait subphase including loading response, mid stance, terminal stance, and pre-swing.

2. Methods

2.1. Participants

Twenty-eight participants who were 18 to 50 years old with a Body Mass Index less than 25 were recruited from the local population in southern Thailand. Potential participants who had any neuromuscular-skeletal diseases were excluded. The foot arch types of normal foot and flatfoot groups were classified by footprint and foot radiographs [14]. The foot-

print was determined by the arch index which is the area of the midfoot divided by the whole foot area excluding the digits. Two views of foot radiographs were taken: anterior-posterior and lateral views. The calcaneal pitch (CIA) and calcaneal-first metatarsal (C1MA) angles were measured in the lateral view. The talonavicular coverage (TNC) and talus to the second metatarsal (T2MA) angles were measured in the anterior-posterior view. In this study, the normal foot criteria reference range was defined following the Leslie et al. study [13] and within one standard deviation. Only one foot of each participant was evaluated. Informed consent was obtained from each participant before the study. Ethical approval for this study was obtained from the Institutional Review Board (Ethics ID: 55-299-25-6-3).

2.2. Instrumentation

This study used a three-dimensional motion analysis system (Vicon Mx, Vicon Motion System Ltd., Oxford, England) with 10 cameras (MX T20) with a sampling frequency of 100 Hz to capture the foot kinematics. Three force platforms (AMTI, OR6, USA) with sampling at 1,000 Hz were used to measure the ground reaction forces (GRF).

2.3. Procedures

Reflective markers with a diameter of 9 mm were mounted onto the participants' feet with double-sided adhesive tape based on the positions defined by the Oxford Foot Model [22], [25]. Additional markers to evaluate the MLA were also mounted (Fig. 1a). All participants attended a single session test by walking barefoot at a self-selected comfortable speed along a 10 meter walkway. During static capture, the participants were asked to stand in the relaxed position. Each participant had to become familiar with the walkway before data capture began. Three successful trials of only one foot on one force platform were collected for data analysis.

2.4. Mechanical evaluation

The MLA biomechanics in this study were based on the motion of two virtual vector segments. The proximal vector segment was $\overrightarrow{NC}$ and the distal vector segment was $\overrightarrow{NM}$ as shown in Fig. 1a. The three-

dimensional MLAD angle was the difference of the MLA angle ($\theta_{MLA,i}$) between each gait subphase from the standing position [25], which was calculated using equations (1) and (2). The 3D MLAD was used to determine the flexibility of the arch and windlass mechanism [15], [23].

$$\theta_{MLA,i} = \cos^{-1}\left(\frac{\overrightarrow{NC} \cdot \overrightarrow{NM}}{\left|\overrightarrow{NC}\right|\left|\overrightarrow{NM}\right|}\right) \qquad (1)$$

where *i* is the specific time in the gait cycle, $\overrightarrow{NC}$ is a vector of the virtual segment from navicular tuberosity to the medial aspect of the calcaneus, and $\overrightarrow{NM}$ is a vector of the virtual segment from navicular tuberosity to the first metatarsal head.

$$\Delta\theta_{MLA,i} = \theta_{MLA,i} - \theta_{MLA,0} \qquad (2)$$

where $\Delta\theta_{MLA,i}$ is the MLA deformation angle, *i* is the specific time in gait cycle, and 0 is the standing position.
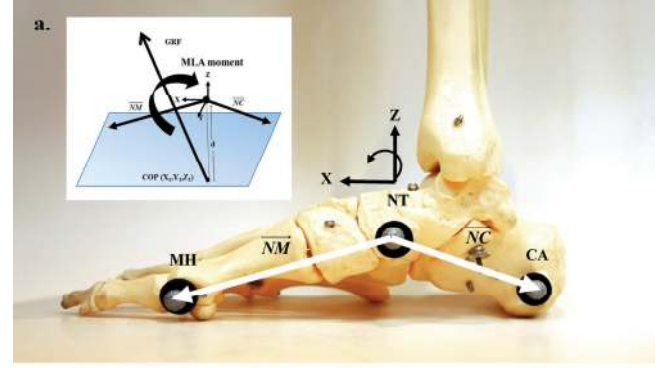
The biomechanical loading at the MLA apex, the MLA moment, was represented by $\overrightarrow{M}_{MLA}$. The $\overrightarrow{M}_{MLA}$ calculation was developed with the instantaneous analysis assumption in three planes of motion following equation (3). Instantaneous analysis assumption in our study considered a snapshot motion in each specific gait subphase cycle. The external GRF acted around the navicular tuberosity (NT) point which is the apex of the MLA consisting of the MLA stabilizers.

$$\overrightarrow{M}_{MLA} = \vec{d} \times \overrightarrow{GRF} \qquad (3)$$

where $\overrightarrow{M}_{MLA}$ is the net moment acting at the apex of MLA which is the result of the cross product between $\vec{d}$ and $\overrightarrow{GRF}$, $\vec{d}$ is the displacement from the NT point to the location of the center of pressure (COP), and $\overrightarrow{GRF}$ is the ground reaction force (as shown in the inset of Fig. 1a). The $\overrightarrow{M}_{MLA}$ was considered in three planes. The $\overrightarrow{M}_{MLA}$ acting about the *y*, *x*, and *z* axes were the moments in the sagittal, frontal, and transverse planes, respectively.

In this study, the foot kinematics were collected by the four-segment rigid body model of the foot (Oxford Foot Model), including the hallux, forefoot, hindfoot, and tibia [22]. The Oxford Foot Model defines the forefoot segment by the markers on the base and head of the first and fifth metatarsus. For the hindfoot segment, it is defined by the markers on the posterior heel, lateral calcaneus, and sustentaculum tali. For the tibia, it is defined by the markers on the tibial, fibular, and medial malleoli. The positive directions of rotation in this study were defined as downward, supination, internal rotation, dorsiflexion, inversion, and adduction (Fig. 1b).
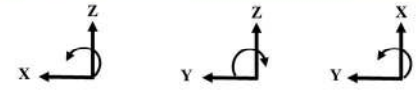
Joints/Planes	Sagittal	Frontal	Transverse
MLA moment	Downward	Inversion	Internal rotation
Hindfoot angle relative to tibia	Dorsiflexion	Inversion	Internal rotation
Forefoot angle relative to hindfoot	Dorsiflexion	Supination	Adduction

Fig. 1. Marker locations and rotational direction of the right foot
(a) The medial longitudinal arch (MLA) marker locations
are the medial aspect of the calcaneus (CA),
navicular tuberosity (NT), and first metatarsal head (MH).
The white arrows represent the $\overrightarrow{NC}$ and $\overrightarrow{NM}$ vector segments.
(Inset) The graphic demonstrates vectors
for MLA moment calculation;
(b) The positive direction of rotation of the kinematics
and kinetics parameters

2.5. Data analysis

All kinematics data were filtered using the Woltering filter with the predicted mean square error (MSE) of 10. This study evaluated the peak kinematics that included the hindfoot motion relative to the tibia angle, forefoot motion relative to the hindfoot angle and the MLA deformation angle, and the peak kinetics included the MLA moment and ground reaction forces. Each parameter between the groups was analyzed during 100% of the stance and also in each subphase: loading response (0–16% of stance), mid

stance (17–48% of stance), terminal stance (49–81% of stance), and pre-swing (82–100% of stance) [4], [7]. The data were then statistically compared between groups with unpaired *t*-test via Prism 5 (GraphPad, San Diego, CA) and the significance level was set at *p* value < 0.05.

3. Results

There were significant differences between flatfoot and normal foot groups in both kinematics and kinetics during specific subphases of the gait cycle. The data were presented in the mean difference between the groups and standard deviation among the data. The demographic data and the spatio-temporal parameters of all participants have been concluded in Table 1.

Table. Participants' characteristics data (mean ± SD)

Characteristic	Normal foot group (*N* = 15)	Flatfoot group (*N* = 13)
Age (years)	32.7 ± 8.9	24.9 ± 3.3
Gender (Female/Male)	14/1	10/3
Foot side (Right/Left)	11/4	10/3
Height (m)	1.6 ± 0.1	1.6 ± 0.1
Body mass (kg)	52.2 ± 4.8	59.1 ± 7.6
Body Mass Index (kg/m^2)	21.1 ± 1.6	22.7 ± 1.9
Footprint arch index	0.25 ± 0.02	0.35 ± 0.03
Radiography (degrees)		
Calcaneal inclination angle	21.2 ± 1.7	13.9 ± 2.7
Calcaneal-first metatarsal angle	133.4 ± 2.9	147.0 ± 4.9
Talonavicular coverage angle	18.2 ± 3.6	23.7 ± 7.7
Talus to second metatarsal angle	18.6 ± 4.5	27.4 ± 11.8
Spatio-temporal parameters		
Gait velocity (m/s)	1.12 ± 0.1	1.09 ± 0.1
Stride length (m)	1.19 ± 0.1	1.20 ± 0.1
Cadence (step/min)	113.51 ± 7.8	109.19 ± 7.6

3.1. Medial longitudinal arch (MLA) biomechanics

During static capture, the flatfoot had a significantly greater MLA angle compared with the normal foot (144.3 ± 2.1 vs. 131.1 ± 3.3 degrees, *p* = 0.002). The MLAD angle changed along the stance and peaked at terminal stance. Only the peak MLAD angle during terminal stance in the flatfoot group had a smaller deformation (–0.46 ± 4.4 vs. 2.8 ± 3.2 degrees, *p* = 0.038) compared with that in the normal foot group (Fig. 2).
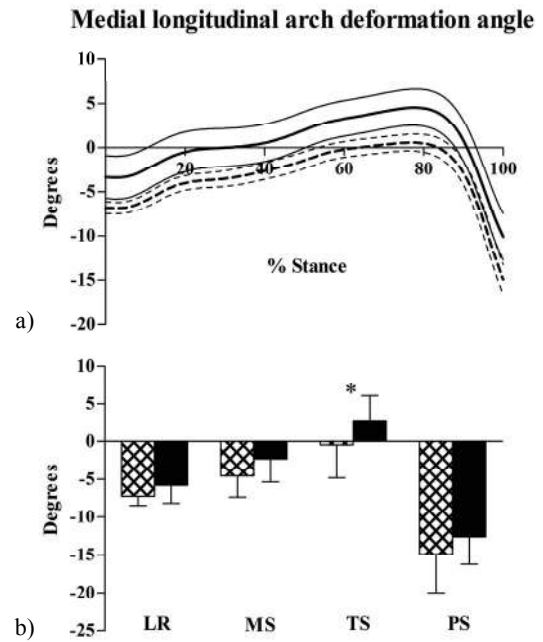
Medial longitudinal arch deformation angle

Fig. 2. Medial longitudinal arch deformation (MLAD) angle. (a) The average MLAD angle with 95% confidence interval during stance between normal foot (solid line) and flatfoot (dash line) groups; (b) Comparison of peak MLAD angle in each phase of gait between 2 groups (normal group in black bar and flat group in cross-hatching bar): LR = loading response (0–16% stance), MS = mid stance (17–48% stance), TS = terminal stance (49–81% stance), and PS = pre-swing (82–100% stance). The positive values represent the greater MLAD angle in comparison with static condition, while the negative values represent the smaller MLAD angle in comparison with static condition

The MLA moment in the sagittal plane demonstrated that the two simple foot segments of forefoot and hindfoot were elevated after loading response and peaked at terminal stance. The MLA moment in the frontal plane started with an increasing eversion moment from heel strike through loading response. The arch apex showed a decreased eversion moment during mid stance, a peaked eversion moment at terminal stance, and finally ended with an increased inversion moment. For the MLA moment in the transverse plane, the external moment was applied at the MLA during early stance and then the external moment converted to an internal moment during terminal stance. The difference in the peak MLA moment between the flatfoot and normal foot groups was observed in the frontal plane during mid stance. There was a significant increase of the eversion moment in the flatfoot group (–0.54 ± 0.06 vs. –0.47 ± 0.07 N·m/kg, *p* = 0.005) during mid stance which is consistent with the average MLA moment (Fig. 3, middle). There were no significant differences of the MLA moment in the sagittal and transverse planes during stance between the two groups.

Sagittal plane

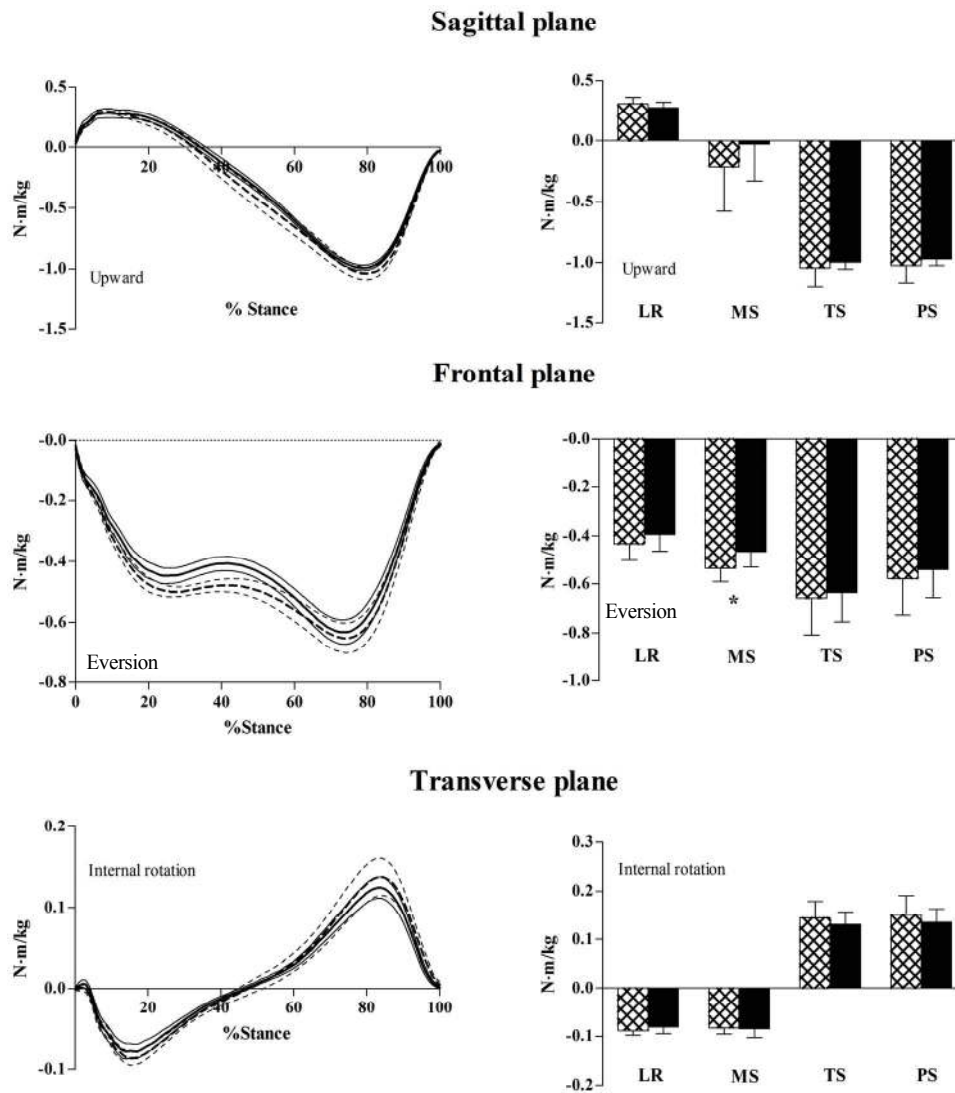## Frontal plane

Transverse plane

Fig. 3. Medial longitudinal arch (MLA) moment in sagittal, frontal and transverse planes.
(Left) The average MLA moment with 95% confidence interval during stance
between normal foot (solid line) and flatfoot (dash line) groups. (Right) Comparison of peak MLA moment
in each phase of gait between 2 groups (normal group in black bar and flat group in cross-hatching bar):
LR = loading response (0–16% stance), MS = mid stance (17–48% stance),
TS = terminal stance (49–81% stance), and PS = pre-swing (82–100% stance)

3.2. Hindfoot motion

Significant differences in hindfoot motion between the two arch types were observed mainly in the sagittal and transverse planes. In the sagittal plane, the flatfoot group had a significant increase in the peak plantarflexion angle during loading response (-7.0 ± 0.9 vs. -3.9 ± 2.6 degrees, $p < 0.001$) and a significant decrease of the peak dorsiflexion angle during mid stance (-2.5 ± 6.5 vs. 3.6 ± 5.9 degrees, $p = 0.017$) which is consistent with the average hindfoot motion (Fig. 4, top). In the transverse plane, there was a significant increase in the peak internal rotation angle in the flatfoot group during stance (loading response: 23.0 ± 7.8 vs 3.2 ± 20.0 degrees, $p = 0.003$; mid stance: 14.0 ± 12.0 vs. -1.0 ± 19.0 degrees, $p = 0.018$; terminal stance: 21.0 ± 9.3 vs. 5.8 ± 18.0 degrees, $p = 0.013$; pre-swing: 22.0 ± 7.5 vs. 1.4 ± 22.0 degrees, $p = 0.003$) which is consistent with the average hindfoot motion (Fig. 4, bottom). The flatfoot group also had prolonged hindfoot eversion. The hindfoot of the flatfoot group converted to inversion during the late stance at 58.31% of stance while the normal group converted at 56.64% of stance.

Sagittal plane

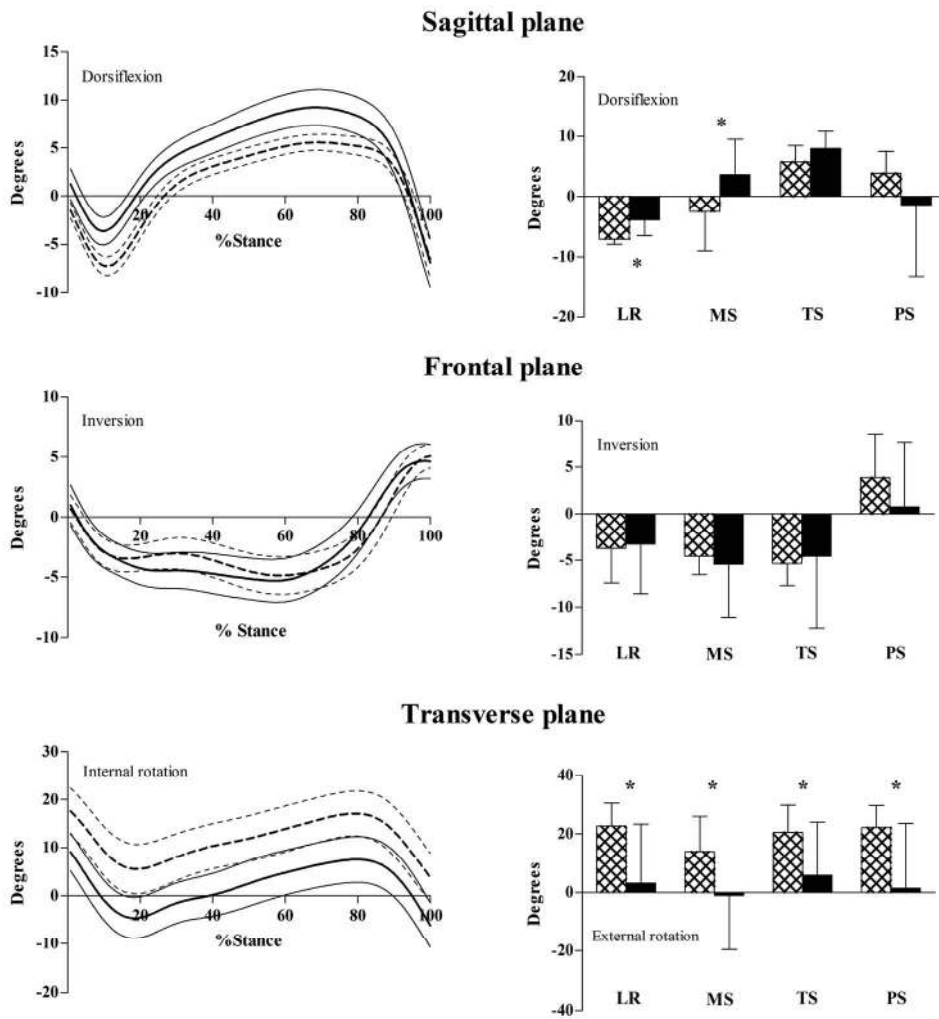## Frontal plane

Transverse plane

Fig. 4. Hindfoot motion in sagittal, frontal and transverse planes. (Left) The average angle
with 95% confidence interval during stance between normal foot (solid line) and flatfoot (dash line) groups.
(Right) Comparison of peak angle in each phase of gait between 2 groups
(normal group in black bar and flat group in cross-hatching bar):
LR = loading response (0–16% stance), MS = mid stance (17–48% stance),
TS = terminal stance (49–81% stance), and PS = pre-swing (82–100% stance)

3.3. Forefoot motion

A significant difference in forefoot motion between the two arch types was observed only in the transverse plane. The flatfoot group had a significant increase in the peak abduction angle during pre-swing (-1.3 ± 9.6 vs. 7.3 ± 7.5 degrees, $p = 0.016$), which is consistent with the average forefoot motion (Fig. 5, bottom).

3.4. Ground reaction forces

The first peak of vertical GRF and anterior-posterior GRF were not significantly different between the two groups. As shown in Fig. 6 (top), the flatfoot group had a significantly shallower trough (0.83 ± 0.05 vs. 0.78 ± 0.06 of normalized body weight, $p = 0.012$) and significantly smaller second peak (1.08 ± 0.03 vs. 1.12 ± 0.05 of normalized body weight, $p = 0.004$). The maximum medial-lateral GRF increased significantly in the flatfoot group (0.07 ± 0.01 vs. 0.05 ± 0.02 of normalized body weight, $p = 0.047$). When we considered the relationship with the subphases of gait, the peaks of vertical GRF in the flatfoot group were significantly less than in the normal foot group (terminal stance: 1.08 ± 0.03 vs. 1.12 ± 0.05, $p = 0.004$; pre-swing: 1.02 ± 0.04 vs. 1.06 ± 0.03 of normalized body weight, $p = 0.008$). However, the peak of medial-lateral GRF was significantly greater during mid stance (0.07 ± 0.01 vs. 0.05 ± 0.02 of normalized body weight, $p = 0.047$) when compared with the normal group (Fig. 6).

Sagittal plane

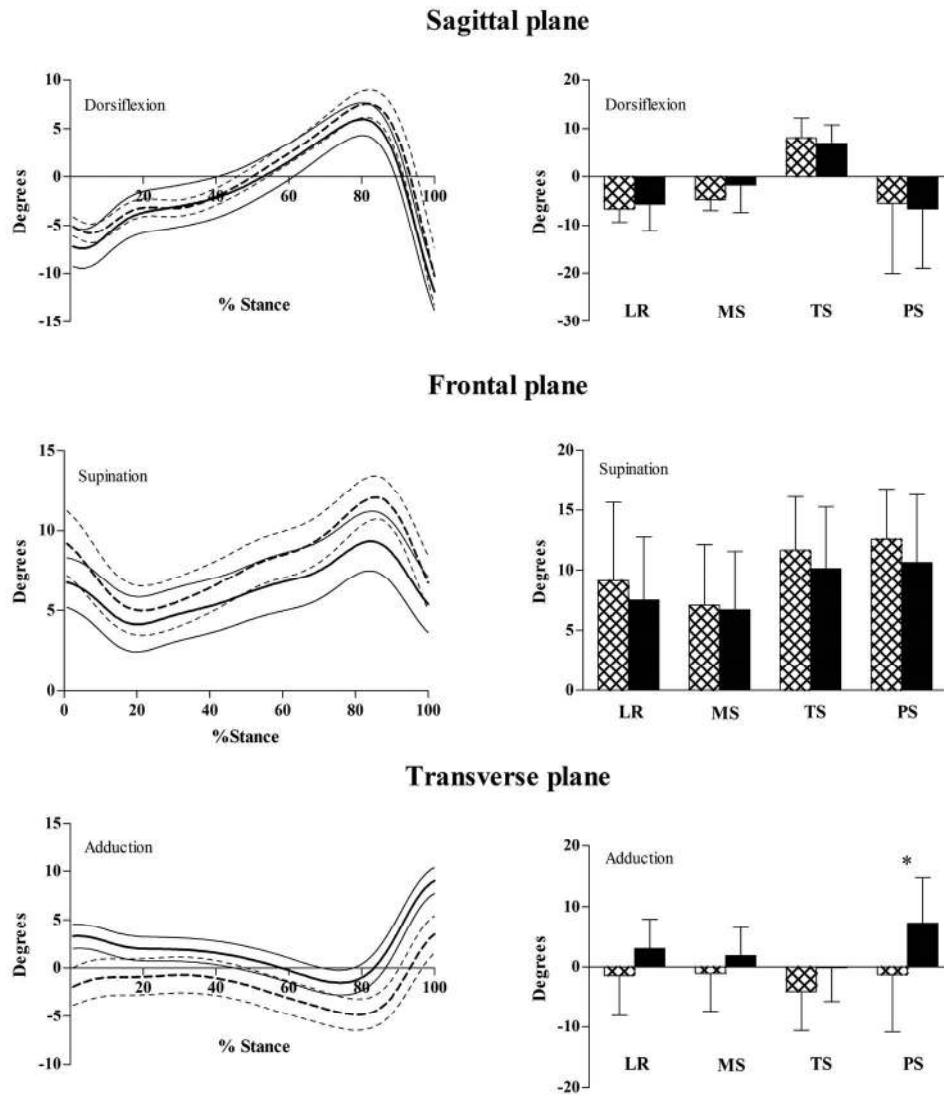## Frontal plane

Transverse plane

Fig. 5. Forefoot motion in sagittal, frontal and transverse planes. (Left) The average angle with 95% confidence interval during stance between normal foot (solid line) and flatfoot (dash line) groups. (Right) Comparison of peak angle in each phase of gait between 2 groups (normal group in black bar and flat group in cross-hatching bar): LR = loading response (0–16% stance), MS = mid stance (17–48% stance), TS = terminal stance (49–81% stance), and PS = pre-swing (82–100% stance)

4. Discussion

The results demonstrated significant differences of gait biomechanics between two arch foot types which reflected the deficit of foot functions while walking: the shock absorption ability, foot stability, and propulsion function. The proposed MLA biomechanics evaluation was also able to determine these significant differences between the groups in specific subphases.

From the literature, the previous studies assumed the MLA structure as part of midfoot joint segment in the various complex foot models [4], [20], [24]. This study proposed alternative parameters to evaluate MLA structure and the MLA biomechanics including the MLAD angle and MLA moment. The MLAD angle determined the sagittal plane movement of the MLA angle, which is one of the indicators for windlass mechanism function evaluation [5]. The magnitude of deformation indicated the flexibility of the MLA [23]. The MLA moment represented the net moment at the apex of the MLA, the talonavicular joint, which was the result of a combination movement of the subtalar and midfoot joints. The proposed MLA moment graphs showed a similar pattern with the previous studies [20], [24]. From the calculation, the MLA moment in the sagittal plane revealed a "flattening out" of both forefoot and hindfoot. This was affected by

Vertical direction

Medial-lateral direction
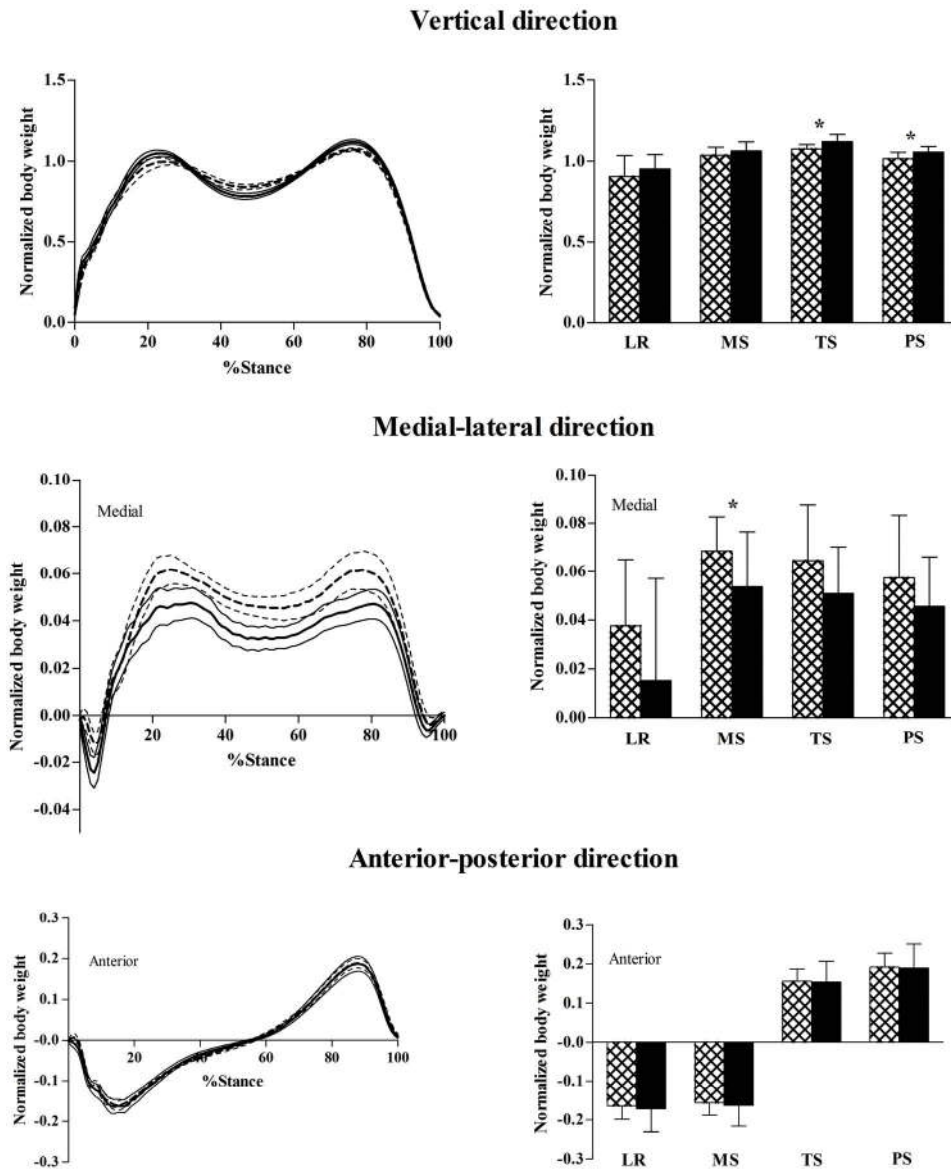

Anterior-posterior direction

Fig. 6. Ground reaction force (GRF) in vertical, medial-lateral and anterior-posterior directions.
(Left) The average forces with 95% confidence interval during stance between normal foot (solid line)
and flatfoot (dash line) groups. (Right) Comparison of peak forces in each phase of gait between 2 groups
(normal group in black bar and flat group in cross-hatching bar): LR = loading response (0–16% stance),
MS = mid stance (17–48% stance), TS = terminal stance (49–81% stance), and PS = pre-swing (82–100% stance)

the external GRF during early stance. This was in contrast with Takashima's study [24] and Saraswat's study [20], which considered only the forefoot motion in their calculation and not the hindfoot. Thus the change in their MLA moment did not appear during the heel strike until it reached the loading response subphase. However, a change in the MLA moment in our study occurred during the late stance and peaked at terminal stance. The hindfoot and forefoot segments of our study were pushed upward and apart resulting in the upward MLA moment due to the forefoot action which was consistent with earlier studies [20], [24]. For a more comprehensive study, the foot function

should be evaluated in six degrees of freedom. Thus, the frontal and transverse planes were included in this study. We found some relationships between the MLA moment in sagittal and transverse planes. During early stance, there was a vertical compression from the body weight which caused the MLA structure to flatten and this corresponded to external rotation. During late stance, the forefoot was pushed against the floor and the external GRF stretched the MLA structure causing an upward and internal rotation moment. This also corresponded with the pattern of foot motion during gait cycle. Moreover, the MLA moment in the frontal plane from our calculation showed only a nega-

tive value for the eversion moment because the direction of foot progression was displaced through a negative direction of global coordinate axis.

During the loading response, the normal foot has normal hindfoot eversion-inversion adjustment to unlock the midtarsal joint doing shock absorption. This function is aided by the flexibility of the MLA from the reverse windlass mechanism [3]. Our results showed that flatfeet had significantly larger hindfoot eversion and internal rotation relative to the tibia which was consistent with the results of previous studies [4]. These reflected the defect of shock absorption ability in the flatfeet gait. The larger magnitude of the MLAD angle in the flatfoot group also represented an increase of arch flexibility during plantigrade walking on the floor [25].

Mid stance is the subphase of gait when the body force vector moves across the stationary foot as single limb support [18]. The foot must be stable. A shallow trough of the vertical GRF larger than 0.7 of normalized body weight in the flatfoot group showed the difficulty of the body to move across the foot which was similar to the finding reported in a study by Pauk [17]. Under the tension from vertical compression during body progression, the deficient SLC is unable to hold the navicular bone against the talus bone in a close-packed position [19]. Furthermore, the tibialis posterior muscle, the dynamic MLA stabilizer, and arch forming muscle are not active during mid stance [18]. An increase of the eversion moment at the MLA and a larger maximum medial GRF caused the foot further instability [10].

The most important part of the late stance phase is the propulsion function. The highest leg muscle activity occurs during this period of both arch forming and arch deforming muscles [18]. These cause the talus bone to move externally and upward thereby locking the midtarsal joint that aids the windlass mechanism to develop a rigid lever for propulsion. Our study found greater hindfoot internal rotation accompanied by greater hindfoot eversion and forefoot abduction in the flatfoot group. These cause difficulty of the foot to resupinate and invert the foot for propulsion as described in earlier researches [4]. The prolonged foot pronation also promoted insufficient propulsion [3]. The magnitude of the MLAD angle was smaller when compared with the normal foot group which represented the powerlessness of the MLA stabilizers to raise the arch in preparation for propulsion. During late stance, the GRF moves across the metatarsal joints transmitting energy to promote body progression. The smaller second peak of vertical GRF in the flatfoot group

that was found in this study also diminished the efficiency of the progression [3], [17].

There were several limitations and concerns in the results of this study based on foot classification criteria that we modified for the purpose of our study. We focused on the foot alignment in the sagittal and transverse planes. The normal feet were defined as one standard deviation of the sample of population. The true normal foot was possibly not recruited from this non-restrict criteria. Although this protocol was perhaps not a good representation to classify foot type, this protocol quantitatively provided the foot classification in this study. The heterogeneity of genders between groups might influence the results. The higher incidence of ligamentous laxity joints in females [6], [17], compared with males, would obscure the true biomechanical change between arches of foot types. Some results also differed from previous studies because our reference position was the relaxed standing position [9] and our participants were not diagnosed as posterior tibialis tendon dysfunction. The MLAD angle in the present study was measured only in a sagittal plane; therefore, the deformation in a frontal plane of MLA angle needs to be studied further. Also, this study did not include the electromyography to complete the gait mechanics evaluation. Therefore, an explanation related to muscle activities was not confirmed in our study.

5. Conclusion

There were significant abnormal gait mechanics during specific subphases that reflected the dysfunction of flatfoot. The proposed MLA moment calculation in this study showed a consistency in the results with previous literature and the calculation can be used in the biomechanical MLA apex loading evaluation in three-dimensional motion of flatfoot while walking. Applying this proposed MLA moment calculation can be used as an alternative parameter to evaluate the arch of foot function in addition to the general flatfoot biomechanics evaluation.

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