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Effect of foot orthoses on the medial longitudinal arch in children with flexible flatfoot deformity: A three-dimensional moment analysis

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ABSTRACT

Foot orthoses are often used to correct altered gait patterns. The purpose of this study was to investigate how foot orthoses can modify the magnitude of three dimensional moments of ankle, knee, and hip joints during a stride of gait in children with flexible flat feet. Bilateral gait data were collected from fourteen male children (age 10.2 ± 1.4 years) suffering from flat feet syndrome. In order to obtain the kinematics data, a Vicon system with six cameras (100 Hz) was used and two Kistler force plates (1000 Hz) to record the kinetics data under each leg. Arc support foot orthoses were used as an intervention. Paired-sample *T*-test was used for within-group comparisons ($\alpha = 0.05$). The results of data analysis showed that foot orthoses can decrease the ankle evertor moment, knee and hip abductor moments and hip flexor moment in dominant lower limb. In non-dominant lower limb, using the orthoses can decrease evertor and internal rotator moments at the ankle, flexor and internal rotator moments at the knee and extensor moment at the hip, while it can increase dorsiflexor moment at the ankle. The findings imply that effects of orthoses on three dimensional moments differ in dominant and non-dominant lower limbs. Furthermore, results demonstrated that dominant and non-dominant lower limbs would also show different responses to the same intervention.

1. Introduction

A low medial longitudinal arch during weight bearing position is a morphological characteristic of flexible flat foot (FF) [1]. It is demonstrated that 4% of children at the age of 10 years suffered from FF and 10% of these individuals showed flexible FF that needs treatment to prevent secondary deformity in their adulthood [2–6]. For example, Kothari et al. reported that reduced medial arch height of the foot was associated with increased odds of knee symptoms and hip/back symptoms in children [6].

Previous literature has reported that individuals with a FF are at increased risk for sustaining ankle sprains, knee injuries, and other overuse injuries such as metatarsal stress fractures, patellofemoral pain syndrome [7,8], and decreased in static dynamic stability [9]. Abnormal biomechanical changes in FF patients create a greater peak plantarflexor ankle moment and invertor moment during stance phase of walking [10]. Razeghi et al. reported higher patellar mechanism angle and also mean and peak forces across the knee extensor mechanism in FF children compared to normal group [11]. These individuals also demonstrated increased ankle stiffness during dorsiflexion regardless of demographic factors [12], and reduced midfoot frontal plane range of motion during pre-swing [13]. Twomey and McIntosh stated that the increased external hip rotation and greater external foot progression angle in the low arched foot highlight the need to examine gait comprehensively to diagnose causes and effects of the observed differences and hence determine appropriate treatment [14]. Furthermore, Lin et al. reported that the temporal spatial variables and the kinematics of the coronal and horizontal planes of the ankle and knee were significantly different in the moderate and severe flexible FF groups when compared with the normal arched group [15]. Those with FF walked more slowly and recorded greater abduction and external rotation in both the ankle and knee joints, particularly during stance. Therefore, FF must be considered not only as a problem of static alignment of the ankle and foot complex but also as a dynamic functional abnormality of the lower limb [15]. However, a recent study showed no temporo-spatial differences between low arched and normal arched children during gait at a self-selected speed [16].

Meanwhile, the standard conservative treatment for symptomatic FF deformity is orthotic therapy with the aid of insoles or ankle foot orthoses or exercise therapy [17,18]. Foot orthoses (FO) are often used

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to correct altered gait pattern [19,20]. Some in vivo studies have failed to demonstrate any beneficial effects of orthoses [21,22]. However, other studies reported that the FO is an effective orthotic device for rearfoot motion control, plantar pressure reduction and re-distribution in patients with flexible FF [19,20]. These changes in the components of plantar pressure and rearfoot motion could affect load distribution at the other joints. FO have been shown to be clinically effective in reducing symptoms in the lower limbs; nevertheless, their reported effectiveness has varied [21–24] and it is still a controversial issue. Yet, few studies have provided scientific evidence of these effects, specially applying corrective insoles for children with flexible FF [19]. Furthermore, there is a dearth of information regarding how FO can alter principal components of gait kinetics parameters such as net moments at lower limb joints.

Analyses of joint moments are at the core of mechanical analysis of human movement, and it is helpful to understand joint loading and muscle function. The use of clinical gait analysis to calculate joint moments is growing and from biomechanical perspective, many of lower limb diseases such as hip and knee osteoarthritis are being assessed by analysis of joint moments. At the knee, adduction moment can be used as a surrogate measure of medial–lateral load distribution within the joint [25], while knee flexion moments are related to quadriceps demand [26]. High joint moments have been associated with the severity and progression of osteoarthritis [27]. It is suggested that coexistence of an increased adduction moment at the hip and knee joints with an excessive hip adduction angle leads to an increase in iliotibial band hardness [28]. In fact, changes in joint moments have different influences on improvement or injuries of joints. For example, knee adduction moment is thought to increase load in the medial aspect of the tibial plateau and femoral condyle, thereby causing knee pain in runners [29]. However, as far as a review of the related literature shows, no research has yet been performed so far on the effects of orthoses on FF children through a three dimensional moment analysis of ankle, knee and hip joints.

Therefore, this study aimed to investigating how FO can modify the magnitude of three dimensional moments of ankle, knee, and hip joints during a stride of gait cycle in children with flexible FF.

2. Methods

2.1. Subjects

In this study, fourteen male children were selected as experimental group. The averages of subjects, age, weight, height and body mass index were 10.2 ± 1.4 years (age range between 8 and 12 years); 42.6 ± 7.5 kg; 150.6 ± 10.2 cm, 19.1 ± 3.5 kg/m^2, respectively. Based on previously established criterion for diagnosing FF [30], patients with a navicular drop greater than 10 mm were selected. A subject was excluded from the study if he had any previous history of bone fractures, surgery, orthopedic disease, neuromuscular problems, limb length discrepancies of greater than 5 mm, or if he was feeling fatigue, and had heavy physical tasks or exercise during the past two days. The subjects were all right foot dominant determined through kicking ball test. Ethics permits were obtained and all participants' parents provided written informed consent.

2.2. Apparatus

Kinematics data were collected at 100 Hz using a 6-camera VICON motion capture system (Vicon system, Oxford Metrics, Oxford, UK) and 16 spherical reflective markers with a diameter of 15 mm. The Plug in Gait marker set was used to identify the bilateral pelvis, thighs, legs, and feet provided in Nexus software.

Two force plates (Kistler, type 9281, Kistler Instruent AG, Winterthur, Switzerland) were used to record the kinetics data under each leg at 1000 Hz that were synchronized with the Vicon system. The force plates were located at the center of the calibrated volume.

2.3. Kinematic and kinetic data collection and analysis

Reflective markers were attached bilaterally to subjects on the following landmarks: anterior superior iliac spine, posterior superior iliac spine, lateral mid-thigh, lateral femoral epicondyle, mid shank, lateral malleoli, heel and toe between second and third metatarsal heads, although it should be noted that in the shod test conditions, the bilateral heels and toes markers could not be attached directly to the participants' skin as it was covered with shoes. In consequence, instead of anatomic landmarks, markers were attached to appropriate positions on the vamp of the shoe.

All kinetics data were filtered using a fourth-order low-pass Butterworth filter with a 20 Hz cutoff frequency and normalization was performed based on child's body weight (BW). The Kinematics data were filtered by zero lag fourth order Butterworth filter with cut off frequency of 6 Hz. Afterward, trials were interpolated to 100 points which were normalized in time domain within a stride cycle (0–100%). Plug-in-Gait lower body model (Vicon Motion Systems) [31] was used for data processing, and graphical reports were created in Polygon Authoring Tool.

2.4. Task and procedure

Firstly, subjects were asked to walk a number (three) of trials along the test walkway so as to familiarize themselves with the experimental surroundings. In addition, prior to each experiment condition, a static trial was captured to identify joint center locations and calculate the segment coordinate systems. Data were then collected under two specific test conditions including walking in shoes, and shoes which were embedded with FO. The order of conditions was randomised across subjects. Six walking trials were performed under each condition at self-selected walking velocity. A 2-min rest was allowed between conditions. A trial was accepted when subjects walked with no visible alteration in gait mechanics; whereas, it was excluded in case the foot was placed on the edge of the force plate.

2.5. Application of orthoses

Since shoe type has been thought to affect the walking style in children [32,33], in the present study all participants wore same shoes (New Balance 759, USA) during both conditions.

Arc support FO (Longxin, Industrial Ltd., LX-0701-1) were used as an intervention. FO peak longitudinal height of the mid-foot arc was 25 mm. One of the researchers carried out all the orthotics procedures with consistency in replication of experiments was achieving. The orthoses were placed in both left and right shoes (Fig. 1).

2.6. Statistical analysis

The normality of the distributions, of obtained data was verified using Shapiro-wilk test. Paired samples *T*-test was used to compare amounts of joint moment before and after wearing the FO. In addition, the effects sizes of paired samples *T*-test were calculated as r-squared formulae: $t^2/(t^2 + df)$. This parameter classifies the effect size as small if $r^2 \leq 0.01$, medium if $r^2 \leq 0.1$, and large if $r^2 \geq 0.25$ [34]. The level of significance was set as $p < 0.05$. Statistical analysis was conducted in SPSS (Version 22).

3. Results

The average walking speed was not significantly different between no intervention (1.21 ± 0.14 m/s) and FO (1.22 ± 0.16 m/s) ($P > 0.05$). The maximum amplitude of the net hip, knee and ankle moments during a stride of gait cycle for dominant lower limb are



Fig. 1. The view of shoe (A), and foot orthoses (B).

Table 1
Statistical comparison of maximum amplitude of Net Joint moments (ankle, knee and hip) between two different conditions (wearing 1. normal shoes, 2. shoes embedded with foot orthoses) during a stride cycle of walking in dominant lower limb.

Joint	Moment (N m/kg)	Mean ± SD Shoe	Mean ± SD Shoe + Orthoses	t	*p*-Value	r^2
Ankle	Dorsiflexor	−0.11 ± 0.03	−0.11 ± 0.02	−0.46	0.65	0.01
	Plantarflexor	1.47 ± 0.31	1.49 ± 0.34	0.55	0.59	0.02
	Invertor	−0.22 ± 0.13	−0.23 ± 0.13	−0.55	0.58	0.02
	Evertor	0.03 ± 0.02	0.02 ± 0.01	−4.14	0.001*	0.57
	Internal rotator	−0.04 ± 0.03	−0.05 ± 0.04	−0.63	0.54	0.03
	External rotator	0.23 ± 0.11	0.22 ± 0.10	−0.36	0.72	0.01
Knee	Flexor	−0.52 ± 0.10	−0.53 ± 0.12	−0.19	0.85	0.00
	Extensor	1.12 ± 0.52	1.01 ± 0.50	−1.84	0.09	0.21
	Adductor	−0.26 ± 0.09	−0.27 ± 0.10	−0.67	0.51	0.03
	Abductor	0.49 ± 0.13	0.44 ± 0.10	−4.07	0.001*	0.47
	Internal rotator	−0.06 ± 0.03	−0.06 ± 0.03	−0.40	0.69	0.01
	External rotator	0.20 ± 0.06	0.19 ± 0.07	−0.85	0.41	0.05
Hip	Flexor	−0.83 ± 0.31	−0.71 ± 0.31	2.78	0.02*	0.37
	Extensor	0.87 ± 0.13	0.82 ± 0.14	−1.63	0.13	0.00
	Adductor	−0.22 ± 0.08	−0.25 ± 0.12	−1.66	0.12	0.17
	Abductor	0.94 ± 0.14	0.86 ± 0.20	−2.21	0.04*	0.27
	Internal rotator	−0.15 ± 0.06	−0.14 ± 0.06	0.69	0.50	0.03
	External rotator	0.15 ± 0.06	0.14 ± 0.06	−2.09	0.06	0.25

SD Indicates standard deviation.

illustrated in Table 1. As seen, there are significant differences between the peak ankle evertor moment, knee and hip abductor moments and hip flexor moment when subjects wear normal shoes as compared with shoes embedded with FO. As the table shows, the use of orthoses decreased the peak moments of four variables. The amounts of decrease were 33% for ankle evertor moment ($p = 0.001$, $r^2 = 0.57$), 10% for knee abductor moment ($p = 0.001$, $r^2 = 0.47$), 14% for hip flexor moment ($p = 0.02$, $r^2 = 0.37$) and 8% for hip abductor moment ($p = 0.04$, $r^2 = 0.27$).

For the non-dominant lower limb, the maximum amplitudes of the net hip, knee and ankle moments during a stride of gait cycle are shown in Table 2. The subjects experienced greater peak dorsiflexor moment (17% increase) at the ankle when they wore FO ($p = 0.004$, $r^2 = 0.47$). Apart from that, using orthoses decreased the peak evertor and internal rotator moments at the ankle, flexor and internal rotator moments at the knee and finally extensor moment at the hip. The decrease was 80% for ankle evertor moment ($p = 0.01$, $r^2 = 0.38$), 28% for ankle internal rotator moment ($p = 0.003$, $r^2 = 0.50$), 13% for knee flexor moment ($p = 0.003$, $r^2 = 0.51$), 25% for knee internal rotator moment ($p = 0.000$, $r^2 = 0.74$) and 10% for hip extensor moment ($p = 0.02$, $r^2 = 0.34$).

Each participant's change in moment between two conditions for the dominant and non-dominant lower limbs was reported in Tables 3 and 4, respectively. In these two Tables, only those moments which significantly showed a change by the FO (Tables 1 and 2) are expressed.

As it is seen in Table 3, in dominant side, all subjects experienced a reduction in ankle evertor moment in response to the FO condition. Similarly, all children (except subject 2) had a lower knee abductor moment after using the FO. However, although, the total hip flexor and abductor moments decreased in the FO condition (Table 1), the responses of the subjects were quite different.

In non-dominant side (Table 4), in response to the FO, all children (except subject 8 with no response) experienced an increased ankle dorsiflexor moment. The FO resulted in a decreased ankle evertor moment (for 10 subjects), decreased ankle internal rotator moment (for 13 subjects), decreased knee flexor moment (for 12 subjects), and decreased knee internal rotator moment (for all 14 subjects), as well as a decreased hip extensor moment (for 10 subjects) in non-dominant lower limb.

4. Discussion

This study was conducted to examine how FO can modify the magnitude of three dimensional moments of ankle, knee, and hip joints during a stride of gait cycle in children with flexible FF. Based on the results of data analysis and statistical tests given above, the use of orthoses can decrease the peak ankle evertor moment, knee and hip abductor moments and hip flexor moment in the dominant lower limb. Although the magnitudes of differences were small in many of the variables, large effect size values were observed for all significant differences ($r^2 \geq 0.25$) [35].

Previous studies demonstrated that there was a negligible invertor ankle moment in the FF group compared to the normal group; and that rearfoot eversion could occur without muscle input in FF group [10]. In

Table 2
Statistical comparison of maximum amplitude of Net Joint moments (ankle, knee and hip) between two different conditions (wearing 1. Normal shoes, 2. shoes embedded with foot orthoses) during a stride cycle of walking in non-dominant lower limb.

Joint	Moment (N m/kg)	Mean ± SD Shoe	Mean ± SD Shoe + Orthoses	t	*p*-Value	r^2
Ankle	Dorsiflexor	−0.12 ± 0.02	−0.14 ± 0.03	−3.44	0.004*	0.47
	Plantarflexor	1.45 ± 0.17	1.41 ± 0.26	−1.11	0.28	0.08
	Invertor	−0.33 ± 0.05	−0.36 ± 0.05	−2.05	0.06	0.24
	Evertor	0.05 ± 0.06	0.01 ± 0.01	−2.85	0.01*	0.38
	Internal rotator	−0.07 ± 0.04	−0.05 ± 0.03	3.65	0.003*	0.50
	External rotator	0.17 ± 0.03	0.18 ± 0.02	0.70	0.49	0.04
Knee	Flexor	−0.59 ± 0.07	−0.51 ± 0.07	3.66	0.003*	0.51
	Extensor	0.90 ± 0.25	0.90 ± 0.26	−0.03	0.97	0.00
	Adductor	−0.09 ± 0.04	−0.08 ± 0.03	0.99	0.34	0.07
	Abductor	0.95 ± 0.12	0.90 ± 0.11	−1.49	0.16	0.14
	Internal rotator	−0.04 ± 0.01	−0.03 ± 0.01	6.09	0.000*	0.74
	External rotator	0.29 ± 0.04	0.28 ± 0.06	−0.42	0.68	0.01
Hip	Flexor	−0.47 ± 0.12	−0.48 ± 0.13	−0.26	0.80	0.00
	Extensor	0.93 ± 0.16	0.84 ± 0.19	−2.57	0.02*	0.34
	Adductor	−0.44 ± 0.22	−0.40 ± 0.19	1.49	0.16	0.14
	Abductor	1.45 ± 0.20	1.45 ± 0.19	−0.08	0.94	0.00
	Internal rotator	−0.10 ± 0.03	−0.11 ± 0.06	−1.01	0.33	0.07
	External rotator	0.19 ± 0.05	0.19 ± 0.04	0.28	0.78	0.01

SD Indicates standard deviation.
* Indicates significant difference ($P < 0.05$).

the present study, the ankle moment shows a decreased evertor moment when subjects wear orthoses. Therefore, the case can be made that FO will reduce rearfoot evertor moment in the dominant lower limb, although this change is not accompanied with a reduction of knee adduction moment during walking. These findings are in line with those obtained by Maly et al. [35] and Eslami et al. [36] but are in contrast with those of Mündermann et al. [37] and Nester et al. [38] who reported a positive and significant correlation between peak knee adduction moment and the magnitude of rearfoot eversion during shod and shod/orthoses running. Individual differences in response to orthoses, type of FO utilized and difference in motion patterns could account for the discrepancy in these findings. It seems that a large knee adduction moment could increase the risk of overloading the medial structures of the knee contributing to the iliotibial band friction [39] and patellofemoral pain syndromes [40] in runners. In the present study, an arch support orthoses did not change the peak knee adduction moment during the whole percent of stride.

The present study showed that compared with walking in shoes, using the orthoses is associated with a significant increase in the peak ankle dorsiflexion moment in non-dominant lower limb. Previous studies showed that, although not statistically significant dorsiflexion at pushoff tended to be less in the FF group for both the rearfoot and forefoot [41,42]. The only other significant kinematic difference between the two groups occurred at foot flat, whereby the rearfoot was in slight plantarflexion in the FF group rather than in slight dorsiflexion. This finding was similarly more indicative of restraint or stability than of excessive mobility [10], while our results demonstrated that arch support insole could increase dorsiflexion moment in non-dominant lower limb. In the sagittal plane, the greater dorsiflexor moment in the FO condition could be reflected in increased rearfoot dorsiflexion motion.

This study is the first to identify that using orthoses in non-dominant lower limb of children with flexible FF decreases the peak evertor and internal rotator moments at the ankle, flexor and internal rotator moments at the knee and finally extensor moment at the hip. Consequently, similar study assessing the effect of orthoses on three-dimensional joint moments of non-dominant lower limb was not found in literature, to compare the results of the present study. However, a recent meta-analysis has demonstrated that using FO is an effective intervention [43]. The outcome is a kinematic measurement and mechanical or therapeutic targets included reduced peak eversion [36,37,44,45], and eversion/inversion moment [37,46,47]. Further research is required to determine if the relative reductions observed in eversion moments are clinically relevant. In contrast to our results, Telfer et al. hold that the lack of a systematic main effect of posting level on internal tibial rotation [48]. Nester et al. investigated the effects of FO on kinetics of proximal joints in walking and showed no effects on the knee and hip [38]. Hsu et al. argued that orthoses increase knee internal abduction moment during late stance, and reduce the ankle inversion moment, hip adduction angle, hip abduction moment, and hip external rotation moment in patients with forefoot varus abnormality [49].

In this study, using orthoses did not show any significant effect on internal and external rotator moments at the hip joints. Twomey and McIntosh [14] like Schutte et al. [50] also reported that participants in the FF group were significantly more externally rotated at the hip, particularly during stance and again just prior to second heel strike. These findings were contrary to the findings of Lin et al. [15] who reported no differences in hip kinematics. This extra external hip rotation and foot progression angle combined with a less pronated

Table 3
Changes in maximum amplitude of Net Joint moments (% difference from orthoses to control shod by each participant) in dominant lower limb during a stride cycle of walking.

Joint	Moment	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	S12	S13	S14
Ankle	Evertor	−37	−51	−33	−38	−46	−41	−16	−13	−41	−64	−40	−54	−58	−32
Knee	Abductor	−8	+1	−3	−6	−10	−17	−7	−1	−14	−22	−18	−1	−11	−11
Hip	Flexor	−4	+6	−13	−4	−27	−52	−25	−8	+1	−16	−32	−22	−12	+15
	Abductor	−1	−14	+5	−3	−2	−37	+2	+1	+3	−24	−32	0	−14	−1

Note: "S" stands for subject.
Positive sign shows an enhancement in moment due to foot orthoses. Negative sign shows a reduction in moment due to foot orthoses.

Table 4
Changes in maximum amplitude of Net Joint moments (% difference from orthoses to control shod by each participant) in non-dominant lower limb during a stride cycle of walking.

Joint	Moment	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	S12	S13	S14
Ankle	Dorsiflexor	+7	+19	+8	+37	+15	+17	+17	0	−10	+78	+9	+21	+6	+25
	Evertor	−15	+5	−42	−15	+6	−8	−7	−24	−12	−37	+6	−24	+10	−10
	Internal rotator	−10	−31	−32	−59	−18	−18	+17	−16	−25	−60	−56	−87	−14	−22
Knee	Flexor	−25	−8	−9	−20	−2	−25	+13	−15	−24	+4	−33	−5	−8	−7
	Internal rotator	−45	−33	−20	−11	−25	−17	−21	−12	−26	−39	−58	−70	−6	−25
Hip	Extensor	−27	+2	−9	−18	−20	−14	+16	−2	−28	+1	−30	−1	0	−4

Note: "S" stands for subject.
Positive sign shows an enhancement in moment due to foot orthoses. Negative sign shows a reduction in moment due to foot orthoses.

forefoot suggests that the combination of these may compensate for the normal rotary dynamics in the low arched foot during gait [14]; however, the orthoses used in the present study did not change this parameter at the hip joint.

Based on our results about each participant's change in moment between two conditions, the responses of the some subjects were quite varied. This would allow the researchers to start to identify the characteristics of children who positively benefited from the orthoses versus the ones who didn't. Essentially (in adults) it is believed that there are responders and non-responders to orthoses [51], so this could provide evidence regarding the same phenomena in children.

Our findings demonstrated that the effects of orthoses on three dimensional moments differ in dominant and non-dominant lower limbs. According to previous studies, it is reported that there are significant bilateral differences in the cartilage volume [52], bone density [53], and strength and coordination abilities [54,55]. These asymmetries in lower extremity anatomy may lead to different joint moment production in response to same FO in dominant and non-dominant lower limbs. Newton et al. suggesting that responses to eccentric exercise were not necessarily the same between dominant and non-dominant limbs [56]. However, there is a dearth of research investigating whether dominant and non-dominant lower limbs respond similarly to FO during walking in FF children. Finding of the present study justifies the continued use of the dominant limb in clinical settings.

A number of limitations of the current study should be acknowledged. The use of net joint moments to determine the mechanical output of the controlling muscles meant that the contraction of an agonist could be counteracted by the contraction of the antagonist muscles, without detection [57]. Furthermore, the use of skin-mounted markers could lead to erroneous results due to the potential skin motion artifact [58].

5. Conclusion

Our findings imply that effects of orthoses on three dimensional moments differ in dominant and non-dominant lower limbs. Furthermore, our results demonstrated that dominant and non-dominant lower limb will also show different responses to the same intervention. While it would be difficult to conclude that there is direct relationship between these variables and injury, our findings suggest that the type of orthoses tested in this study can be effective in treating lower extremity injuries. Although three dimensional joint moment differences between the conditions have been identified, further research would need to be conducted including muscle activation, ankle and foot muscular strength, and hyper-mobility.

Conflict of interest

None of the authors have any conflict of interest associated with this study.

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