Kinetic analysis of gait in adults
with asymptomatic flatfoot

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Purpose: The aim of our study was to assess the influence of asymptomatic flatfoot on the kinetic parameters of the lower limb during gait. *Methods*: Individuals of both sexes were studied: 15 women [age 25 ± 5 years] and 19 men [age 25 ± 4 years] with bilateral asymptomatic flatfoot, as well as 16 women [age 26 ± 4 years] and 14 men [age 24 ± 3 years] with normal feet on both sides. A three-dimensional VICON motion analysis system coupled with KISTLER dynamometric platforms was used to perform kinetic gait analysis. *Results*: Women with flatfoot showed significantly lower maximal relative moments in the ankle in the sagittal plane ($p < 0.05$) and significantly lower maximal relative moments in the knee in the sagittal plane in the Terminal Stance ($p < 0.001$). In men, a significant difference was found in terms of hip rotation moment in the transverse plane in the Mid Stance ($p < 0.01$): men with normal feet showed moments of external rotation, while men with flatfoot generated internal rotational moments. Moreover, men with flatfoot showed significantly lower ($p < 0.01$) maximal relative moments in the knee in the transverse plane in the Mid Stance. *Conclusions*: Women with flatfoot have a weakened lower limb propulsion mechanism, whereas, in men with flatfoot, there is a change in the mechanics of the lower limb in the transverse plane. Our findings cast some doubt on flatfoot as a putative risk factor for stress injuries and degenerative changes in lower-limb structures, and suggest that gender differentiation should be taken into account in the analysis and therapy of flatfoot.

Key words: *adults; joint moments; asymptomatic flatfoot; stress injuries; gait kinetics*

1. Introduction

As the foot forms part of a closed biokinetic chain, pathological foot alignment gets transferred upwards to higher segments of the lower limb. Because the talus bone is located between the tibia and fibula, its excessive displacement in the medial-plantar-posterior direction is transferred to the shank, causing its excessive internal rotation and adduction, which, in turn, causes motion of the femur in the same directions. Such an alignment of the lower limb, known as functional valgus or flatfoot, is considered detrimental for the entire motor system [18], [28]. In the literature, it is very often indicated as one of the etiological factors underlying stress-related and degenerative changes in the individual structures of the lower limb [28]. At the foot level, these changes affect the soft tissue structures that act as passive or active stabilizers of the subtalar joint [7] and work to counteract its excessive mobility – mainly the posterior tibial tendon [38], the Achilles tendon [17], the aponeurosis plantaris [38]. As a result of improper rolling of the foot, the first metatarsophalangeal joint is also subluxed, thus giving rise to hallux valgus [5], [9]. Within the lower limb segments above the foot, in turn, the changes affect those soft tissue structures which limit excessive functional valgus of the lower limb – the tendons

involved in goose foot [10]. Incorrect alignment of the axis of the lower limb may also lead to dysfunction of the patellar-pulmonary joint [32] and to anterior cruciate ligament injuries [11]. Overall, the literature includes numerous studies on the kinematic analysis of gait in adults with flatfoot [3], [19], [27]. However, although a relationship between foot alignment and stress injuries appears biomechanically and physiologically plausible, such studies have failed to provide conclusive evidence in support of such a relationship [23].

The aim of our study, therefore, was to assess the influence of flatfoot on the kinetic parameters of the lower limb during free gait. Specifically, we sought to better understand how flatfoot affects the maximal values of force moments occurring in the lower limb joints during gait and thus greater recognition of mechanisms whereby flatfoot possibly contributes to damage of the lower limb joint structures.

2. Materials and methods

2.1. Participants

Purposive sample selection was performed based on the following inclusion and exclusion criteria. The criteria for inclusion into the test group with flatfoot were: age of 18–35 years, BMI within normal limits, bilateral occurrence of flatfoot (normalized navicular height <0.21 [24]; Clarke's angle <41° [31]; forefoot-shank angle >5° [37]). The criteria for inclusion into the test group with healthy feet: age of 18–35 years, BMI within normal limits, bilateral occurrence of correctly shaped feet (normalized navicular height 0.24–0.30 [24]; Clarke's angle 42–54° [31]; forefoot-shank angle <5° [37]).

To exclude the influence of other factors that may be significant for the type of gait, the following exclusion criteria were applied: symptomatic flatfoot, any

past surgical procedures or injuries in the lower limb, dysfunctions of the central or peripheral nervous system, rheumatological diseases, dysfunctions of the vestibular system, general joint hypermobility, connective tissue diseases, pregnancy, the practice of competitive sports, lack of laterality in the lower limbs.

The demographics for the experimental and control groups are presented in Table 1.

The procedure of qualifying the participants and the course of individual stages of the study are presented in Fig. 1.

The subjects were informed about the examination and about the aim of the study. Written consent for participation was expressed by the subjects. The study was approved by the Ethics Committee of the Józef Piłsudski University of Physical Education, Warsaw (decision no. SKE 01-49/2014).

2.2. Apparatus

Each subject was tested using a VICON three-dimensional motion analysis system, which consisted of a computer equipped with the NEXUS software and a set of 9 VICON cameras with a recording frequency of 100 frames per second. The cameras recorded the gait of each participant as they walked along a 10-meter pathway. Along the path, 3 KISTLER dynamometric platforms were placed, measuring 90×60 cm, used to register the ground reaction forces at a frequency of 100 Hz, synchronized with the recording of the participants' movement using the cameras. 34 markers were applied on the body of each participant, according to the Plug in Gait Full Body (SACRUM) model.

2.3. Procedure

At the beginning, each participant adopted a free standing position on two dynamometric platforms (each

Table 1. The demographics for the study groups

| Parameters	Flatfoot Group (n = 34 participants)		Normal Foot Group (n = 30 participants)	
	Female = 15 participants (30 feet) (Average ± SD)	Male = 19 participants (38 feet) (Average ± SD)	Female = 16 participants (32 feet) (Average ± SD)	Male = 14 participants (28 feet) (Average ± SD)
Age [years]	25.0 ±5.0	25.0 ± 4.0	26.0 ± 4.0	24.0 ± 3.0
Body weight [kg]	56.4 ± 5.1	75.3 ± 5.5	58.1 ± 6.7	75.5 ± 9.8
Height [cm]	165.3 ± 5.7	179.3 ± 6.0	166.8 ± 4.6	180.1 ± 8.1
Body mass index [kg/m^2]	20.6 ± 1.5	23.4 ± 1.1	20.8 ± 2.1	23.2 ± 1.3
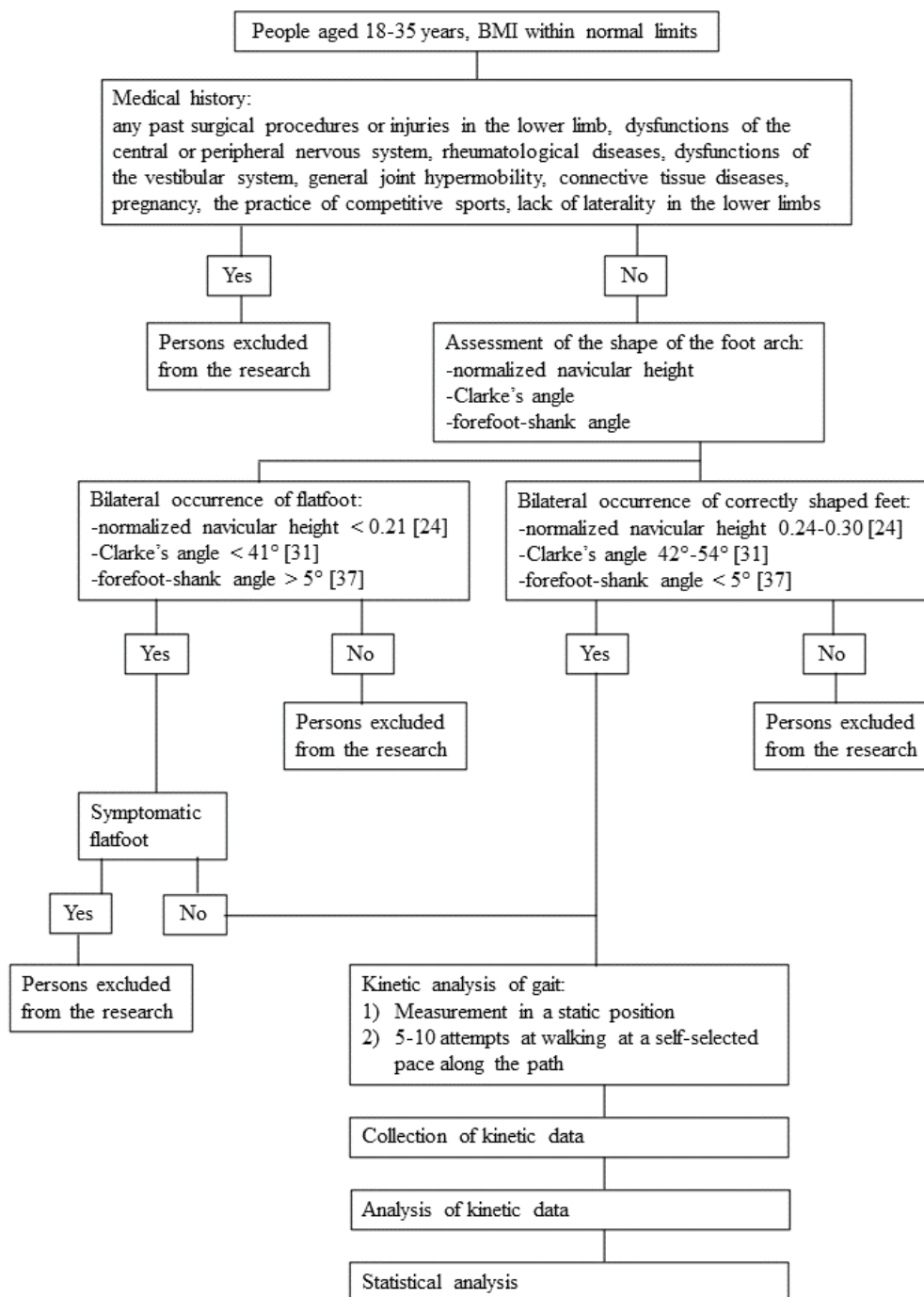
Fig. 1. Diagram presenting the procedure of qualifying the participants
and the course of individual stages of the study

lower limb was placed on a different platform), during which measurements were taken in static position. Next, the participant made 5–10 attempts at walking at a self-selected pace along the path. An attempt was considered correctly performed if the participant stepped with one lower limb on one dynamometric platform and with the other on the other (although participants were not informed of this, in order to preserve naturalness of gait during the attempts).

2.4. Collection and analysis of kinetic data

All kinetic data were normalized to the body weight of the subjects. They were then interpolated to 100 points, which were normalized to the time of the gait cycle (0–100%). Force moments in the ankle, knee and hip joints in the sagittal, frontal and trans-

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verse planes were plotted during the gait cycle using Microsoft Office Excel 2007. From these graphs, the maximal values of relative force moments in the examined joints were identified in each of the three planes. Then the average value of maximal relative moment in the tested joints in three planes of motion in the first half (Loading Response and Mid Stance sub-phases) and second half (Terminal Stance sub-phase) of the stance phase was calculated over 5 gait cycles (after rejecting outliers).

2.5. Statistical analysis

Statistical analysis was performed using STATISTICA 13.0 (StatSoft). Normality of the data was tested using the Shapiro–Wilk test. Because distribution of some variables differed from normal or the assumption of equal variances was not met (Levene's test), the Mann–Whitney U-test was performed for group comparison. The level of significance was set at $\alpha = 0.05$.

3. Results

The results of maximal relative moments measured in the ankle, knee and hip joint in the stance phase of free gait are shown in Tables 2, 3.

The group of women with flatfoot showed significantly lower maximal relative moments in the ankle joint in the sagittal plane ($p < 0.05$) and significantly lower values ($p < 0.001$) in the knee joint in the sagittal plane in the Terminal Stance from the ground, compared to the group of women with healthy feet.

In the group of men with flatfoot, significantly lower maximal relative moments ($p < 0.01$) were found in the knee joint in the transverse plane in the Mid Stance, compared to the men with healthy feet. A significant difference was also found between the hip rotation moment in the transverse plane at the Mid Stance ($p < 0.01$), with men with healthy feet generating external rotation and men with flatfoot showing internal rotation.

4. Discussion

This study sought to clarify the relationship between flatfoot and dysfunctions within the structures of the lower limb.

The results obtained and their statistical analyses showed that women with flatfoot exhibit significantly lower maximal values of flexion moment in the knee joint and plantar flexion moment in the ankle joint during the Terminal Stance of gait, compared to women with healthy feet. The Terminal Stance involves the maximal dorsal flexion of the upper ankle joint, stabi-

Table 2. Maximal relative moments in the stance phase during free gait in female participants

Joint	The sub-phases of the stance phase	Movement plane	Normal Foot Group (Average ± SD)	Flatfoot Group (Average ± SD)	p
Ankle	Loading Response	F	0.02 ± 0.09	0.03 ± 0.09	0.994
	Mid Stance	F	−0.01 ± 0.07	−0.04 ± 0.07	0.819
	Terminal Stance	S	1.59 ± 0.17	1.54 ± 0.16*	0.020
		F	0.13 ± 0.09	0.15 ± 0.07	0.973
Knee	Mid Stance	S	0.05 ± 0.45	0.38 ± 0.49	0.313
		F	0.57 ± 0.18	0.58 ± 0.18	0.837
		T	0.12 ± 0.20	0.03 ± 0.17	0.486
	Terminal Stance	S	−0.44 ± 0.25	−0.30 ± 0.20***	0.001
		F	0.42 ± 0.15	0.43 ± 0.12	0.597
		T	0.20 ± 0.16	0.15 ± 0.18	0.508
Hip	Mid Stance	S	0.88 ± 0.55	0.92 ± 0.41	0.788
		F	0.76 ± 0.23	0.72 ± 0.29	0.133
		T	−0.02 ± 0.20	0.03 ± 0.17	0.349
	Terminal Stance	S	−1.14 ± 0.41	−1.18 ± 0.52	0.544
		F	0.89 ± 0.24	0.89 ± 0.32	0.792
		T	0.20 ± 0.13	0.15 ± 0.12	0.459

S – sagittal plane, F – frontal plane, T – transverse plane; * indicates significant difference ($p < 0.05$); *** indicates significant difference ($p < 0.001$).

Table 3. Maximal relative moments in the stance phase during free gait in male participants

Joint	The sub-phases of the stance phase	Movement plane	Normal Foot Group (Average ± SD)	Flatfoot Group (Average ± SD)	p
Ankle	Loading Response	F	0.05 ± 0.10	0.03 ± 0.11	0.512
	Mid Stance	F	−0.03 ± 0.07	−0.04 ± 0.11	0.987
	Terminal Stance	S	1.74 ± 0.19	1.66 ± 0.16	0.075
		F	0.18 ± 0.08	0.15 ± 0.10	0.297
Knee	Mid Stance	S	0.04 ± 0.46	0.24 ± 0.42	0.202
		F	0.54 ± 0.13	0.59 ± 0.24	0.505
		T	0.17 ± 0.20	0.03 ± 0.10**	0.006
	Terminal Stance	S	−0.59 ± 0.18)	−0.49 ± 0.21	0.125
		F	0.37 ± 0.07	0.40 ± 0.14	0.159
		T	0.15 ± 0.19	0.20 ± 0.13	0.625
Hip	Mid Stance	S	0.98 ± 0.47	0.91 ± 0.42	0.507
		F	0.55 ± 0.25	0.62 ± 0.18	0.286
		T	0.03 ± 0.21	−0.10 ± 0.18**	0.007
	Terminal Stance	S	−0.96 ± 0.42	−1.09 ± 0.44	0.339
		F	1.02 ± 0.19	0.94 ± 0.25	0.219
		T	0.12 ± 0.18	0.19 ± 0.12	0.613

S – sagittal plane, F – frontal plane, T – transverse plane; ** Indicates significant difference ($p < 0.01$).

lized by the triceps surae muscle of the calf, which enables the foot to form a rigid lever and to effectively lift the heel from the ground [5], [13], [30]. This is accompanied by a shifting of the floor reaction force vector to the level of the forefoot. This results in an extension moment in the knee joint, striving towards its hyperextension, which is opposed by a flexion moment in the knee joint, generated by the gastrocnemius muscle of the calf. This phenomenon is important for effective lower limb propulsion and protects the knee joint from excessive overextension in this phase [5], [15].

The lower maximal flexion moments in the upper ankle seen in the group of women with flatfoot result, it seems, from a weakening in the mechanism blocking the transverse tarsal joint before the Terminal Stance begins. Chimenti et al. [29], Hunt et al. [13], Okita et al. [30] showed that this weakness makes it difficult to turn the foot into a rigid lever, which creates more difficult conditions for the effective contraction of the triceps surae in the calf. Prachgosin et al. [33] confirmed this, reporting significantly lower values of the floor reaction forces in subjects with flatfoot in the second half of the stance phase as compared to subjects with healthy feet. In turn, Buldt et al. [2], Hillstrom et al. [12], Rao et al. [34] showed that the distribution of foot pressure on the ground during free gait in subjects with flatfoot indicates lower maximal pressure around the first and the fifth metatarsophalangeal joint, and higher pressure at the first and second toe, with similar results also observed in terms of maximal force. Myoung-Kwon [25] reported increased pressure around the second and third metatarsophalangeal joint during the Terminal Stance in people with flatfoot.

The ground pressure distribution results indicate that the location of the ground reaction force vector changes, and thus may affect the moment arm length of internal flexion in the ankle joint. Wolfram et al. [40], in turn, showed that the position of the subtalar joint during plantar ankle flexion does not affect the length of the Achilles tendon moment arm. It should be noted, however, that these tests were conducted under static conditions. Based on the literature review presented above, it seems that the maximal values of the plantar flexion moment in the ankle joint and the flexion moment in the knee joint, which is also generated by the gastrocnemius muscle of the calf [5], [15], are caused by more difficult conditions for the triceps surae or by a change in the location of the ground reaction force vector.

Because our results show that the maximal values of force moments within the hip joint do not differ significantly between the tested groups, it appears that the difference we observed between the tested force moments in the knee and ankle joint in the sagittal plane in the Terminal Stance is indeed the result of the flatfoot pathology. Unfortunately the literature offers no results of kinetic analysis of flatfoot gait in women, making it impossible to relate these obtained results to other studies. It seems that the observed lowering of the maximal values of the plantar flexion moments

in the ankle and knee flexion moments in the Terminal Stance may weaken the effectiveness of the lower limb's propulsion function. In the literature, the plantar flexion moment of the ankle is considered to be the most important lower limb propulsion element [36]. The results obtained do not, therefore, bear out the assumption that flatfoot is a risk factor for Achilles tendon stress injuries in women, given that the plantar flexion moment was actually found to be lower in our group of women with flatfoot compared to those with healthy feet.

The group of men with flatfoot, in turn, exhibited an internal rotation moment in the hip joint during the Mid Stance, whereas the group of men with healthy feet showed an external rotation moment in the same joint and phrase. In the knee joint, on the other hand, the men with flatfoot were found to have a significantly lower maximal external rotational moment during the Mid Stance in comparison with men with healthy feet.

According to the standards posited by Winter [39], under conditions of proper functioning, there should be an external rotation moment in the hip joint in the Mid Stance, in order to release the pelvis rotation during free gait. Such internal rotation moment in our group of men with flatfoot (but not in women with flatfoot) presumably results from the increased tissue stiffness of the hip joint, which is characteristic for the male population [37]. Carvalhais et al. [4] demonstrated the relationship between the stiffness of the hip joint tissues and the mobility of individual lower limb segments in the transverse plane. During gait, the stiffness of the hip joint is increased by stretching the ligaments that control its rotation, as reported by Fonseca et al. [20] and Souze et al. [37]. Souze et al. [37] also demonstrated that the passive interaction of the hip joint soft tissue can produce an internal moment in the hip joint during the Mid Stance. The internal moment of rotation in the hip joint we observed in in men with flatfoot may result from stress on the frontal ligament of the hip joint (i.e., the upper strand of the iliofemoral and pubofemoral ligament) during the Mid Stance.

The difference found in the knee joint among the men, on the other hand, may result from reduced mobility of the hindfoot due to increased stiffness of the hip joint. Such a relationship is confirmed by the results of Carvalhais et al. [4], Souze et al. [37]. Dicharry [8] also presents the effect of pronation and supination of the ankle joint on the whole kinetic chain of the lower limb. Reishl et al. [35] reported that there is no connection between hindfoot pronation and the internal rotation of the shank and thigh. There is one report in the literature on the kinetics of gait in men with flatfoot, which shows a significantly higher maximal value of the ankle plantar flexion moment during the Terminal Stance, although it concerns symptomatic patients [13].

Overall, it is notable that the results we obtained do not correspond to the kinetic pattern of any of the disorders that are commonly mentioned as putative consequences of flatfoot, i.e., patellofemoral joint stress [14], degenerative changes in the knee joint [26], tibial stress injuries [1], predisposition to lower limb injuries [22].

Our study also found significant differences between the groups of women and men with flatfoot in terms of the maximal moments of force in the hip joint in the frontal plane during the Mid Stance, and in the knee and ankle joint in the sagittal plane during the Terminal Stance. Lee et al. [16] and Nguyen [29] reported differences in the gait patterns of women and men, explainable in terms of differences in the anthropometric skeletal dimensions of the two genders. In the group of women, a significantly higher maximal adductive moment was observed in the hip joint in the Mid Stance compared to men, which is probably related to a wider pelvic structure, and thus a larger external abductive moment in the indicated gait phase. This was also confirmed by Cho et al. [6]. In the group of women with flatfoot, the maximal internal flexion moment in the knee joint and the internal flexion moment in the ankle joint in the Terminal Stance were significantly lower than in the group of men with flatfoot. This seems to be due to the development of a lower value of force by the triceps surae when lifting the heel off the ground, which affects the above mentioned moments in the ankle and knee joint in the sagittal plane. Identical differences in the maximal values of the internal ankle moments were found by Cho et al. [6], who explained the indicated phenomenon in terms of the shorter lever arm in women during heel-off. These results suggest that proper motor analysis and physiotherapy planning for individuals with flatfoot should take sex differentiation into account. Many kinetic studies of gait, however, have not analyzed results based on the division between sexes [3], [19], [25].

In terms of limitations of the present study, we should note that the Oxford Foot Model was not used, which would allow for detailed kinetic analysis of the foot [19]. Moreover, there is a potential risk of results being affected by displacement of the skin under the markers [21].

The clinical implications of the present study are as follows: in the group of women with flatfeet, func-

tional exercises should be used to improve the Terminal Stance, aimed at greater activation of the gastrocnemius muscle. In the group of men with flatfeet, functional exercises of the Mid Stance should be performed, which should be aimed at improving the control of the hip joint in the transverse plane.

Moreover, In the group of females with a flatfeet, orthoses should be aimed at supporting hindfoot supination towards the end of the Mid Stance. In the group of males with a flatfeet, orthoses should primarily affect the work of the joints of the foot and lower limb in the transverse plane in the Mid Stance.

5. Conclusions

Our results suggest that women with flatfoot have a weakened lower-limb propulsion mechanism. In men with a flatfoot, in turn, the lower limb exhibits altered mechanics in the transverse plane. Overall, our results do not correspond to the kinetic pattern of any of the disorders that are commonly mentioned as putative consequences of flatfoot. Moreover, our results suggest that motor analysis and physiotherapy planning for individuals with flatfoot need to take sex differentiation into account.

Future research will focus on performing a kinetic, kinematic and functional EMG analysis during gait in people with flatfeet before and after physiotherapy based on:
- kinesiotherapy, which will be appropriately adjusted to the group of women and men,
- orthotic treatment, which will be individually selected for each research participant.

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