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Electromyography and Kinematic Analysis of Lower Limb Function During Standardized Bodyweight Squats in Individuals with Flexible Flat Foot

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ABSTRACT

Aims: This study explores the effects of Flexible Flat Foot (FFF) on movement stability and muscle activation patterns during bodyweight squats. Since flat foot alters biomechanics and squats are essential for strength and injury prevention, their interaction is studied to improve rehabilitation and training interventions.

Method and Materials: In this study24 university amateur male athletes (12 with Flexible Flat Foot FFF, 12 healthy; age 18–28 years, ≥3 weekly strength training sessions) performed bodyweight squats to 90° knee flexion. Participants were classified into the FFF group based on a navicular drop of ≥10 mm during weight bearing, as measured by the Navicular Drop Test. Electromyography (EMG) of Tibialis Anterior (TA), GastrocNemius (GN), Vastus Medialis Oblique (VMO), Gluteus Maximus (Gmax), and Quadratus Lumborum (Ql), along with kinematic analysis of ankle, knee, hip, and pelvis, were measured within the 0–90° knee flexion range. Comparisons between groups were made for eccentric and concentric phases.

Findings: Compared with controls, the FFF group showed significantly reduced activation of the Vastus Medialis Oblique (VMO) (eccentric: $P = 0.023$; concentric: $P = 0.026$) and (TA (eccentric: $P = 0.001$). Conversely, Gmax activity was higher in both phases (eccentric: $P = 0.001$; concentric: $P = 0.041$). Kinematic analysis also showed reduced flexion angles at the hip, knee, and ankle joints during the eccentric phase ($P = 0.025$, $P = 0.055$, $P = 0.025$, respectively). Pelvic abduction–adduction range of motion increased significantly in the concentric phase ($P = 0.037$), while non-significant decreases were observed in hip, knee, and ankle extension ROM ($P = 0.055$).

Conclusion: : This study demonstrated that individuals with flexible flatfoot exhibit altered muscle activation patterns (reduced VMO and TA activity, elevated Gmax activity) and restricted joint kinematics (reduced flexion-extension at the femur, knee, and ankle) during bodyweight squats compared to individuals with normal arches. These findings highlight a distinct biomechanical profile associated with flexible flatfoot during a fundamental closed-kinetic-chain exercise. They underscore the importance of considering foot posture when assessing squatting mechanics. Future rehabilitation or training protocols for this population may benefit from addressing these specific neuromuscular and kinematic alterations. Further research is warranted to investigate the longitudinal development and potential clinical implications of these biomechanical differences.

Keywords: Kinematics, Squat, Electromyographic, Flexible Flat Foot, Lower limb

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Introduction

Pesplanus, commonly known as FlatFoot (FF), is a widespread lower limb ailment that accounts for nearly a quarter of the world population[1]. The term refers to the feature most of the time: the dropping of the medial longitudinal arch under pressure [2]. The most common type of flatfoot is Flexible Flat Foot (FFF), which is defined as the sinking of the arch under heavy load, and it returns when the foot is not bearing weight [3]. The malfunction of this organ is related to changes in the foot apparatus, e.g., inverted calcaneal and forefoot abduction[4], which can set off a sequence of proximal compensatory movements along the kinetic chain, such as increased tibial internal rotation and altered hip and pelvic mechanics [5, 6]. As known adaptations, these may affect lower-limb kinematics and muscle function, leading to joint stress and injuries during functional activities [7, 8].

Studies on FFF have mainly focused on gait and static postures [9], in which the biomechanical and neuromuscular effects of FFF have been investigated. However, a significant research gap still exists regarding dynamic, closed-chain functional movements. The bodyweight squat is a pivotal movement pattern necessary for daily activities (e.g., sitting, lifting) and rehabilitation protocols [10, 11]. Its nature as a closed-chain movement makes it an essential model for studying integrated lower-limb mechanics. There is some initial evidence that flatfoot can negatively affect squat mechanics and stability [12]; however, the question of how lower extremity muscle activation patterns (EMG) and joint kinematics change specifically during a standardized bodyweight squat in individuals with FFF remains unanswered. This research was conducted to fill a gap in the literature regarding the impact of flexible flatfoot on lower-limb muscle activity and three-dimensional joint kinematics during a controlled bodyweight squat. We predicted that people with FFF would show different EMG activation and kinematic patterns than those of healthy individuals. The results are supposed to serve as a stepping stone towards the practice of evidence-based exercise prescription and rehabilitation strategies tailored to this population.

Method and Materials

This study was a controlled laboratory design that was conducted non-interventionally. Using G*Power software, with a power of 0.95, an effect size of 0.45, and an alpha level of 0.05, 24 male university athletes (amateur-level, training ≥3 times.week in resistance and conditioning exercises), aged between 18 and 28 years, were recruited. They were divided into two groups: a healthy group (n=12) and an FFF group (n=12). Firstly, the participants were selected and divided into groups based on their condition using a standardized method that involved not only clinical inspection but also the navicular drop test (NDT, Brody method) [13]. To ensure the procedure was conducted consistently, the NDT was performed by a single, experienced corrective exercise specialist. To increase the precision of the NDT, the measurement was taken twice for each participant, and the average value was calculated and used for further analysis.

The participants were classified according to a very reliable cut-off value [14]: those with a navicular drop greater than 10 mm were considered the FFF group, indicating considerable collapse of the medial longitudinal arch during weight-bearing. Participants whose navicular drop was 10 mm or less and who showed no visual signs of arch collapse during clinical inspection formed the healthy control group. Recruitment was conducted through the university committee and sports clubs, with inclusion criteria of at least three regular training sessions per week, age between 18 and 28 years, and a Body Mass Index (BMI) of 18 to 24. Exclusion criteria were abnormalities in the lumbar and pelvic regions, recent musculoskeletal injuries, past six months lower limb injuries, recent fractures or surgeries, neurological or pathological conditions, and inability to perform the exercises due to a lack of proper understanding of the movement. Those with a history of inner ear problems that led to balance disturbances were also excluded. Informed written consent was obtained from all participants, and approval was granted by the Ethics Committee of Allamah Tabatabaei University (IR.ATU.REC.1402.067), which ensured that all procedures were in accordance with the relevant guidelines and regulations.

The subjects were observed once in the lab for 1 hour of testing, wearing loose-fitting athletic attire and no shoes to exclude foot-related influences. Body weight squats were performed within the 0–90° knee flexion range for all subjects. Three trials per subject were performed. To avoid fatigue, each set of squats was followed by a 1-minute rest for the participants. Participants used a platform with a maximum knee flexion of 90 degrees to maintain an optimal sitting posture and avoid miscalculations. During the eccentric phase, the participants were positioned such that the maximum knee flexion formed a 90-degree angle (0–90° knee flexion) at hip contact with the surface.

To acquire kinematic data, an Inertial Measurement Unit (IMU) system was used, with sensors placed on the dominant leg at four joints of pelvic, hip, knee and ankle (Figure 1).

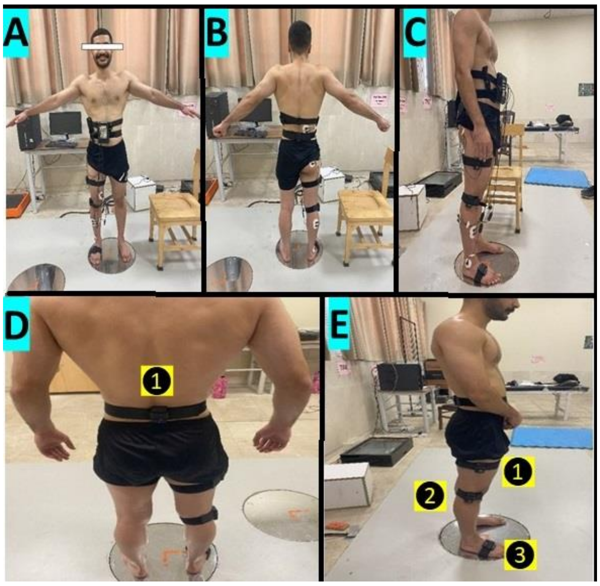

Fig 1) The EMG sensors (A: Anterior-view, B: Posterior-view, C: Lateral-view) and the IMU modules (D: Posterior-view {1= Pelvic joint}, E: Lateral-view {1= Hip joint, 2= Knee joint, 3= Ankle joint})

EMG signals were recorded from the dominant side (Tibialis Anterior (TA), GN, Vastus Medialis Oblique (VMO), Gluteus Maximus (Gmax), and Quadratus Lumborum (QL) muscles). The EMG and kinematic data devices were synchronized and recorded simultaneously. Participants performed squats with their backs to the indicated surface, beginning the movement upon hearing a beep and returning to the starting position upon their pelvis striking the surface. Height on the surface was also adjusted to each participant to have a knee flexion of 90 degrees with contact on the pelvis. Body weight alone was used to perform the squat, and participants attempted to balance and maintain technique. The motion was captured between two phases: the eccentric phase (descending) and concentric phase (ascending) as a single movement without interruption (Figure 2). During the trials, verbal cues were provided to prevent compensatory movement patterns such as lateral trunk bending or pelvic rotation. If such movements were observed, the trial was considered incorrect. Correct positioning and movement posture were taught to every participant a few times before proceeding further.

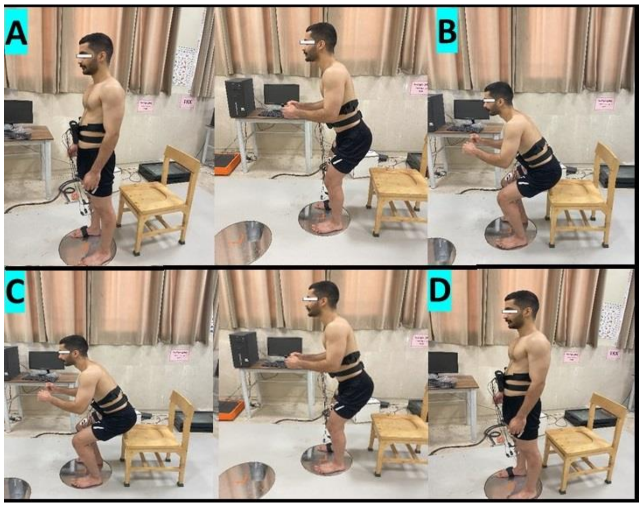

Fig 2) The squat movement was performed while the IMU modules and EMG sensors were attached to the subject (A and B: Eccentric phase, C and D: Concentric phase). A barrier was used to determine the maximum knee flexion angle.

A careful two-step quality-control protocol was implemented to maintain the integrity of the recorded data. Firstly, as part of data acquisition, compensatory movements (e.g., excessive lateral trunk flexion or pelvic rotation) that could be observed in the trials, and even after verbal cues, were immediately discarded, and the trials were repeated. Secondly, as part of post-processing, the device software and MATLAB were used to inspect all EMG and kinematic signals visually. Any EMG trial showing excessive baseline noise, an artifact (e.g., movement artifacts or powerline interference), or signal dropout was excluded from the data. IMU data were verified for sensor misalignment or drift by checking the consistency of the static calibration posture at the beginning and the end of each trial. Only trials that passed visual inspection were considered for inclusion in the analysis. The first two valid trials for each participant were averaged.

Also, the subsection on "Ethical Approval and Informed Consent" has been moved to the end of the methods section to improve clarity.

A 16-channel wireless EMG system (Live, Iran) measured TA, GastrocNemius (GN), VMO, Gmax, and QL muscle activity during eccentric and concentric phases of the squat. Signals were sampled at 1000 Hz, rectified, filtered (20–490 Hz), and smoothed using a symmetric RMS filter. EMG signals were normalized to maximum voluntary isometric contraction (MVIC), and the mean muscle activity was represented as a percentage of MVIC. The averages of the two trials were computed, and all data were processed using MATLAB (MathWorks, Natick, MA). According to SENIAM recommendations [15], the electrodes were placed along the direction of muscle fibers and on the dominant leg, defined as the leg selected for kicking a soccer ball [16]. Before electrode placement, the skin was shaved, abraded, and cleaned with 75% alcohol to minimize resistance. EMG data were normalized to Maximum Voluntary Isometric Contraction (MVIC) from standardized tests for each muscle (TA, GN, VMO, Gmax, QL). Each MVIC was performed three times for 6 seconds, with 1-minute rest intervals between repetitions. For determining the MVIC of the TA muscle, the subject sat on a chair with his dominant leg at a 90–100-degree angle. One hand holds the subject's distal calf as resistance is applied on the medial side and dorsal surface of the foot along the axis of eversion and plantar flexion.

For the GN muscle, Upright position: plantar flexion against external resistance in the neutral position of the ankle, e.g., attempting to lift an overloaded Olympic bar using the ankle joint only [17]. For the Gmax muscle, prone with 90° knee flexion and hip extension - Most frequently reported position, where the subjects are in the prone position with knee flexed to 90° and performing hip extension against resistance applied near the top of the knee [18]. To measure the MVIC of the QL muscle, the patient is side-lying with the supine position on an exam table or mat, which is the most commonly, used position. The patient lies with the side to be tested up, and the examiner provides manual resistance at the lateral border of the ribcage or upper trunk when the patient moves to maximum lateral trunk flexion toward the ceiling [19]. For the VMO muscle, testing always involves maximal effort isometric knee extension against anchored resistance while proper stabilization of the thorax, pelvis, and thigh is in place [20]. The highest mean peak of the three was recorded as the MVIC of the QL muscle. For all the muscles, one RMS value was obtained and partitioned by MVIC, multiplied by 100 for normalization [21].

For lower-limb kinematic assessment during squats, a wireless IMU-based motion analysis system (BSN Company, Iran) was employed. Two IMU modules with three-axis gyroscopes, accelerometers, and magnetometers were used by the system, which had no data volume limit other than wireless range, to measure angular velocity, acceleration, and magnetic field. The sensors were attached to the limbs using straps to record segmental movement. Before data collection, all the IMU modules were calibrated by the device software and manual instructions. This device has been used in previous studies to measure kinematic variables [22, 23]. Calibration was initiated by leaving the module static on a flat surface and selecting the "Accelerometer-Gyroscope" option in the software interface, triggering the process when the module LED flashed three times. Magnetometer calibration was performed by freehand spatial rotation of the module until a colored sphere appeared on the screen, continuing until a predetermined maximum number of points was reached. Calibration required 1200 points per IMU module to be accurate. Modules were synchronized to prevent timing problems. Four were attached to: trunk, pelvis, head (Y-axis along floor), thighs, calves (Y-axis along floor), and metatarsals (Y-axis along limb). Data on pelvic, femoral, knee, and ankle joint angles were recorded along the X, Y, and Z axes and exported to Excel (Figure 3). Movement phases were determined from the knee flexion graph: the eccentric phase comprised data before maximum knee flexion (descent), and the concentric phase (ascent) was defined by a 90-degree criterion.

Descriptive statistics (mean and standard deviation) were used to report participant demographics, and the Shapiro-Wilk test assessed data normality. Because of a small sample size and the data not being normally distributed (verified by the Shapiro-Wilk test),

Fig 3) IMU System (BSN Company, Iran) and Location of IMU device modules

non-parametric tests were used. The Mann-Whitney U test was applied to all comparisons between groups (FFF vs. healthy control). For all non-parametric tests, the Mann-Whitney U test and the Wilcoxon signed-rank test were used to compare the performances of variables between the FFF and healthy foot conditions. The data analysis was performed at a 95% confidence level with a significance level (alpha) of ≤ 0.05 using SPSS software (version 26).

Findings

The research involved 24 male recreational athletes (12 healthy, 12 FFF). EMG analysis revealed that the FFF subjects had lower TA (6.0% versus 15.0%, p = 0.001) and VMO (7.5% versus 13.5%, p = 0.023) activity but higher Gmax activity (15.2% versus 5.8%, p = 0.001) in the eccentric phase. The same trends were observed in the concentric phase, while differences in TA were more modest (8.3% versus 12.7%, p = 0.096). There was no difference in GN and QL activity between groups. Kinematic measurements revealed smaller flexion angles at the hip, knee, and ankle joints during the eccentric phase in FFF subjects, especially the hip (4.17° vs. 8.83°, p = 0.025) and ankle (4.17° vs. 8.83°, p = 0.025) during the eccentric phase. Concentric, abduction–adduction of the pelvis was larger for FFF (8.67° than 4.33°, p = 0.037, with the rest of the joint differences being more minor and nonsignificant.

These results show that FFF changes lower-limb muscle activation and sagittal-plane joint movement, and elicits compensatory increases in frontal-plane pelvic movement via controlled bodyweight squats.

Table 1- Muscle EMG activity was compared between eccentric and concentric phases of bodyweight squats in Flexible Flat Feet (FFF) and healthy controls.

Variables		Eccentric phase	Concentric phase
Muscle Activity (%MVIC)	Group		
Tibialis Anterior (TA)	Healthy	15.00	12.70
	Flexible Flat Foot	6.00	8.30
	Mann-Whitney U test, P Value	0.001*	0.096
Gastroc Nemius (GN)	Healthy	10.60	9.60
	Flexible Flat Foot	10.40	11.40
	Mann-Whitney U test, P Value	0.940	0.496
Vastus Medialis Oblique (VMO)	Healthy	13.50	13.45
	Flexible Flat Foot	7.50	7.55
	Mann-Whitney U test, P Value	0.023*	0.026*
Gluteus Maximus(Gmax)	Healthy	5.80	7.80
	Flexible Flat Foot	15.20	13.20
	Mann-Whitney U test, P Value	0.001*	0.041*
Quadratus Lumborum (OL)	Healthy	10.00	9.70
	Flexible Flat Foot	11.00	11.30
	Mann-Whitney U test, P Value	0.705	0.545

Table 2) IMU sensor kinematic data, separated by joint and angle of movement, were gathered for FFF and healthy populations during eccentric and concentric squat phases.

Variables		Eccentric phase			Concentric phase		
Kinematic Evaluation	Group	Abduction & Adduction	In-Rotation & Ex-Rotation	Flexion & Extension	Abduction & Adduction	In-Rotation & Ex-Rotation	Flexion & Extension
Pelvic (°)	Healthy	5.33	6.17	6.50	4.33	6.17	7.50
	*FFF	7.67	6.83	6.50	8.67	6.83	5.50
	Mann-Whitney U test, P Value	0.262	0.749	1	0.037*	0.749	0.337
Hip (°)	Healthy	6.67	6.17	8.83	8.17	6.17	8.50
	FFF	6.33	6.83	4.17	4.83	6.83	4.50
	Mann-Whitney U test, P Value	0.873	0.749	0.025*	0.109	0.749	0.055*
Knee (°)	Healthy	7.67	6.48	8.50	8.33	6.67	8.50
	FFF	5.33	6.56	4.50	4.67	6.33	4.50
	Mann-Whitney U test, P Value	0.262	1	0.055*	0.078	0.873	0.055*
Ankle (°)	Healthy	6.83	6.49	8.83	7.33	6.83	8.50
	FFF	6.17	6.50	4.17	5.67	6.17	4.50
	Mann-Whitney U test, P Value	0.49	1	0.025*	0.423	0.749	0.055*

*FFF: Flexible Flat Foot

Discussion

This research offers a direct comparison of lower limb electromyography and three-dimensional kinematics during a standardized bodyweight squat in individuals with flexible flatfoot (FFF) versus healthy controls. The significant discoveries were: 1-the activation of Vastus Medialis Oblique (VMO) and Tibialis Anterior (TA) muscles was significantly lessened, 2- the activation of Gluteus Maximus (Gmax) was increased, and 3-the FFF group exhibited a kinematic pattern with decreased sagittal plane range of motion (ROM) at the hip, knee, and ankle during the eccentric phase and increased pelvic frontal plane (abduction.adduction) ROM during the concentric phase. This study compared muscle activity and joint kinematics during bodyweight squats between healthy controls and individuals with FFF to identify key exercise and rehabilitation factors in the FFF group. The results showed significant differences: reduced VMO and TA activity, and greater Gmax activity in FFF subjects during both phases. There were kinematic reductions in hip, knee, and ankle flexion.extension during the eccentric phase, with an increase in pelvic abduction.adduction during the concentric phase in the FFF group. Some of these changes were less significant compared to the other findings, but we have reported them as well. Previous research clearly demonstrates the considerable role of muscles in enhancing and stabilizing joints (24). Evidence suggests that during the occurrence of a musculoskeletal abnormality in one of the joints, the muscles and ligaments on the concave side are shortened. In contrast, those on the convex side are stretched (25). Patients with musculoskeletal abnormalities most probably have deviated muscle activation patterns that trigger compensatory movement in other muscles and joints. Postural control and muscle function are affected by even slight deviations from normal biomechanics, corroborating the results of this study (24, 26).

The neuromuscular changes that were observed correspond to the changes that have been documented as the ones that happen in the compensation mechanism due to foot pronation and arch collapse. The decreased activation of the TA, which is a major ankle dorsiflexor and foot inverter, goes together with the publications that have been made about reduced dorsiflexor activity in FFF, which has been correlated most of the time with the changes in the foot posture and the limited dorsiflexion ROM that are the characteristics of this group of people [27, 28].

Similarly, the reduced VMO activation could indicate a suppression or change in the quadriceps recruitment pattern, the quadriceps being maybe a later branching effect of the quadriceps feed from ankle mechanics being compromised or from changes in the lower limb alignment [27, 29]

On the other hand, the higher Gmax activity in the FFF group probably reflects a proximal compensatory strategy. The excessive activation of one of the primary hip extensors could be a way for the stabilizing mechanism to control the pelvis and the trunk when the rest of the limb is impaired, and the hip abductors are possibly weaker, an idea that is in line with the findings of earlier studies [30, 31].

Moreover, our kinematic findings reveal in detail how the movement strategy was changed. The highly decreased range of motion for hip, knee, and ankle flexion.extension during squat descent in the FFF group is strong evidence for the association of limited ankle dorsiflexion with proximal joint mobility restriction in closed-chain tasks [32].

The limitation of the group in the sagittal plane seems to have been counteracted by a more significant use of the other planes, which is supported by the higher pelvic abduction/adduction ROM that was visible during the ascent phase. The present result is similar to the report of the other studies that have shown the increase of frontal plane pelvis motion to be used as a compensation for the reduction of distal mobility or stability [33].

Importantly, we did not observe an increase in knee valgus or internal rotation of the tibia, which are commonly reported in weighted studies of the flat foot [32]. This difference could be due to our study being a controlled, unweighted, barefoot protocol, in which we deliberately limited trunk and pelvic movement to isolate the mechanics of the lower limb. By restricting these movements, the methodological decision might have led to compensatory demands on the knee-to-pelvis system, thus indicating that the particular expressions of FFF adaptations depend heavily on the nature of the task. Hence, our findings describe the compensatory pattern under highly controlled squatting conditions and are consistent with previous literature. The combined neuromuscular and kinematic pattern that emerged—less activation of the stabilizers (TA, VMO), over activation of the proximal muscles through compensatory mechanisms (Gmax), limitation of sagittal plane movement, and enhanced frontal plane pelvic movement—indicates an inefficient squatting strategy in individuals with FFF. Such a changed strategy may result in reduced performance efficiency, increased joint loading in a non-optimal manner, and a higher risk of injury over time. We should also be aware of the study's limitations, including a small sample size of young male athletes, a cross-sectional design, and non-standard movement speeds. Moreover, although the controlled squat protocol has increased the experiment's rigor, it may not entirely mirror the more natural movement patterns of loaded squatting. The forthcoming studies need to focus on the presence of these adaptations in larger, more diverse populations during functional and loaded activities and to determine the effectiveness of targeted interventions for correcting these patterns. In summary, this study reveals that a flexible flatfoot influences the neuromuscular and kinematic groundwork of the bodyweight squat. The data obtained highlight that rehabilitation and training strategies for FFF individuals should not only focus on local foot function but also on the typical proximal compensatory patterns to improve movement quality and reduce the risk of injury.

Despite the noteworthy findings of this research, there are a few limitations to mention. First, the relatively small sample size of 24 young male recreational athletes limits generalizability to other ages, females, and individuals with different activity levels. Second, the cross-sectional design precludes causal inference and assessment of long-term adaptation. Third, the tests were performed barefoot in laboratory conditions without external load, which are not necessarily representative of authentic training or sporting conditions. Fourth, EMG recordings were limited to particular muscles and kinematic analysis to specific lower-limb joints, and hence other muscles and joints involved in compensatory strategies were not covered. Fifthly, isolated RMS-based muscle activity was investigated, rather than time-domain parameters such as activation onset.offset and coactivation indices, which are central to neuromuscular coordination, specifically in individuals with flexible flatfoot (FFF). Sixthly, participants were not excluded for structural lower-limb deformities, including genu varum, genu valgum, or hip morphological abnormalities, which could act as confounding variables. Seventh, kinetic variables and plantar pressure patterns were not assessed, which would have provided a more complete biomechanical description of the movement. Eighth, the speed of the movement during the squat was not regulated by a meter or any other objective tool (e.g., a metronome). Even though the participants were instructed to perform the movement in a controlled manner, the absence of a standardized tempo suggests that changes in movement speed may have affected EMG amplitude and kinematic variables. Lastly, the research protocol for standing squat, which accounted for compensatory movements such as lateral trunk flexion and pelvic rotation, may have restricted natural knee kinematics among participants with FFF. As such, compensatory mechanisms may have been diverted to the pelvis and ankle, and these effects may have been underestimated using this approach. Future studies should evaluate both standardized and unconstrained natural squats to quantify the full range of compensatory strategies.

To build upon and extend existing knowledge, future studies should overcome these limitations. First, larger and more diverse participant samples across age, sex, physical activity level, and clinical status would improve the external validity of the results. Second, longitudinal or interventional study designs that assess the impact of specific rehabilitation or training programs on muscle activation and kinematic patterns in individuals with FFF would be welcome. Third, the investigation of functional and sport-specific tasks—such as loaded squats, running, jumping, and change-of-direction tasks—would provide increased ecological validity. Fourth, considering a broader range of muscles (e.g., hip rotators, hamstrings, and trunk stabilizers) and upper-body contributions would present a more integrated view of whole-body compensation strategies. Fifth, future studies are encouraged to incorporate time-domain EMG analyses alongside amplitude-based comparisons to provide a more complete picture of motor control and joint stabilization during dynamic movements such as squatting. Lastly, the examination of sex-specific and age-related differences in biomechanical changes and compensation strategies may usefully inform the creation of more targeted intervention protocols.

Conclusion

The findings of this study demonstrate that flexible flatfoot (FFF) significantly alters the EMG activity of lower limb muscles and the dynamic joint kinematics during unweighted bodyweight squats. Specifically, compared to healthy controls, individuals with FFF exhibited reduced activation of the tibialis anterior and vastus medialis oblique, increased activation of the gluteus maximus, and restricted sagittal plane range of motion at the hip, knee, and ankle joints. These results demonstrate neuromuscular and kinematic alterations during squatting in individuals with FFF, characterized by reduced activation of key stabilizers, compensatory overactivation of proximal muscles, and restricted sagittal plane motion. These alterations are indicative of an adapted movement strategy that may affect squat performance.

Consequently, exercise interventions for FFF individuals should be directed at augmenting ankle dorsiflexor and quadriceps activation, improving hip and knee flexion mobility, and optimizing gluteal muscle recruitment to promote more efficient movement patterns during squatting tasks. Rehabilitation and strength training interventions need to account for these neuromuscular and kinematic alterations to improve squat performance and reduce potential joint stress. Increased heterogeneous samples, together with long-term intervention trials, are justified in future research to establish evidence-based exercise prescriptions for FFF individuals.

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Authors' Contribution
All authors contributed equally to the conception and design of the study, data collection and analysis, interpretation of the results, and drafting of the manuscript. Each author approved the final version of the manuscript for submission
Conflicts' of Interest
The authors declared no conflict of interest.
Ethical Permission
The approval was obtained by the Ethics Committee of Allamah Tabatabaei University (IR.ATU.REC.1402.067), which ensured that all procedures were in accordance with the relevant guidelines and regulations. Informed written consent was obtained from all participants, and
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References

1. Alsaidi FA, Moria KM. Flatfeet severity-level detection based on alignment measuring. Sensors. 2023;23(19):8219.
2. Wójcik G. The effect of high-heeled footwear on the induction of selected musculoskeletal conditions and potential beneficial uses in prophylaxis and management. Health Problems of Civilization. 2019;13(3):209-16.
3. Aphale S, Shinde S, Ambali MP, Mane M, Mane Sr M. The Effect of an Aquatic Exercise Program on Pain and Functional Performance in Overweight Adolescent Runners With Functional Flat Feet. Cureus. 2025;17(2).
4. Borges CdS, Fernandes LFRM, Bertoncello D. Relationship between lumbar changes and modifications in the plantar arch in women with low back pain. Acta ortopedica brasileira. 2013;21:135-8.
5. Piri E, Sobhani V, Jafarnezhadgero A, Arabzadeh E, Shamsoddini A, Zago M, et al. Effect of double-density foot orthoses on ground reaction forces and lower limb muscle activities during running in adults with and without pronated feet. BMC Sports Science, Medicine and Rehabilitation. 2025;17(1):54.
6. Aghakeshizadeh F, Letafatkar A, Ataabadi PA, Hosseinzadeh M. The effect of taping on maximum plantar pressure and ground reaction force in people with flat foot after applying a fatigue protocol. 2021.
7. Chen H, Sun D, Fang Y, Gao S, Zhang Q, Bíró I, et al. Effect of orthopedic insoles on lower limb motion kinematics and kinetics in adults with flat foot: a systematic review. Frontiers in Bioengineering and Biotechnology. 2024;12:1435554.
8. Kakavas G, Malliaropoulos NG, Forelli F, Mazeas J, Skarpas G, Maffuli N, et al. How Subtalar Kinematics Affects Knee Laxity in Soccer Players After Anterior Cruciate Ligament Injury? Cureus. 2023;15(10).
9. 9. Šćepanović T, Kojić M, Mikić M, Štajer V, Ödek U, Penjak A. Effects of an integrative warm-up method on the range of motion, core stability, and quality of squat performance of young adults. Frontiers in Sports and Active Living. 2024;6:1323515.
10. Sánchez-Medina L, Pallarés JG, Pérez CE, Morán-Navarro R, González-Badillo JJ. Estimation of relative load from bar velocity in the full back squat exercise. Sports medicine international open. 2017;1(02):E80-E8.
11. Kim S, Miller M, Tallarico A, Helder S, Liu Y, Lee S. Relationships between physical characteristics and biomechanics of lower extremity during the squat. Journal of Exercise Science & Fitness. 2021;19(4):269-77.
12. Kim JS, Lee MY. The effect of short foot exercise using visual feedback on the balance and accuracy of knee joint movement in subjects with flexible flatfoot. Medicine. 2020;99(13):e19260.
13. McPoil TG, Cornwall MW, Medoff L, Vicenzino B, Forsberg K, Hilz D. Arch height change during sit-to-stand: an alternative for the navicular drop test. Journal of foot and ankle research. 2008;1:1-11.
14. Magee DJ, Manske RC. Orthopedic physical assessment-E-Book. St Louis, MO: Elsevier Health Sciences. 2014.
15. 15. Hermens HJ, Freriks B, Disselhorst-Klug C, Rau G. Development of recommendations for SEMG sensors and sensor placement procedures. Journal of electromyography and Kinesiology. 2000;10(5):361-74.
16. Carcia CR, Cacolice PA, McGeary S. Defining lower extremity dominance: the relationship between preferred lower extremity and two functional tasks. International journal of sports physical therapy. 2019;14(2):188.
17. Lee J, Cynn H, Shin A, Kim B. Combined effects of

gastrocnemius stretch and tibialis anterior resistance exercise in subjects with limited ankle dorsiflexion. Physical Therapy Rehabilitation Science. 2021;10(1):10-5.

18. Bae C-H, Choe Y-W, Kim M-K. Effects of different external loads on the activities of the gluteus maximus and biceps femoris during prone hip extension in healthy young men. Korean Society of Physical Medicine. 2020;15(2):1-9.
19. Ducas J, Marineau E, Abboud J. Task-dependent neuromuscular adaptations in low back pain: a controlled experimental study. Frontiers in Human Neuroscience. 2024;18:1459711.
20. Escamilla RF, Thompson IS, Asuncion R, Bravo J, Chang T, Fournier T, et al. Effects of Step Length and Stride Variation During Forward Lunges on Lower-extremity Muscle Activity. Journal of Functional Morphology and Kinesiology. 2025;10(1):42.
21. Slater LV, Hart JM. Muscle activation patterns during different squat techniques. The Journal of Strength & Conditioning Research. 2017;31(3):667-76.
22. Heidarimoghaddam R, Hosseini MH, Ilbeigi S, Anbarian M, Tapak L. Kinematic analysis of head and trunk in individual and team one-handed carrying. International Journal of Industrial Ergonomics. 2023;94:103422.
23. Saemi E, Hasanvand A, Doustan M, Asadi A, Becker K. Standing long jump performance is enhanced when using an external as well as holistic focus of attention: A kinematic study. Sensors. 2024;24 (17):5602.
24. Williams GN, Barrance PJ, Snyder-Mackler L, Axe MJ, Buchanan TS. Specificity of muscle action after anterior cruciate ligament injury. Journal of orthopaedic research. 2003;21(6):1131-7.
25. Hunt AE, Smith RM. Mechanics and control of the flat versus normal foot during the stance phase of walking. Clinical biomechanics. 2004;19(4):391-7.
26. Rios JL, Gorges AL, dos Santos MJ. Individuals with chronic ankle instability compensate for their ankle deficits using proximal musculature to maintain reduced postural sway while kicking a ball. Human movement science. 2015;43:33-44.
27. Almansoof HS, Nuhmani S, Muaidi Q. Role of ankle dorsiflexion in sports performance and injury risk: A narrative review. Electronic Journal of General Medicine. 2023;20(5).
28. Macrum E, Bell DR, Boling M, Lewek M, Padua D. Effect of limiting ankle-dorsiflexion range of motion on lower extremity kinematics and muscle-activation patterns during a squat. Journal of sport rehabilitation. 2012;21(2):144-50.
29. Dill KE, Begalle RL, Frank BS, Zinder SM, Padua DA. Altered knee and ankle kinematics during squatting in those with limited weight-bearing–lunge ankle-dorsiflexion range of motion. Journal of athletic training. 2014;49(6):723-32.
30. Lee KY, Lee J-H, Im S-K. Effect of gluteal muscle strengthening exercise on sagittal balance and muscle volume in adult spinal deformity following long-segment fixation surgery. Scientific Reports. 2022;12(1):9063.
31. Vecchio L, Daewoud H, Green S. The health and performance benefits of the squat, deadlift. and bench press MOJ Yoga Phys Ther. 2018;3(40-47):105-6.
32. Park S-Y, Bang H-S, Park D-J. Potential for foot dysfunction and plantar fasciitis according to the shape of the foot arch in young adults. Journal of exercise rehabilitation. 2018;14(3):497.
33. Bent MA, Ciccodicola EM, Rethlefsen SA, Wren TA. Increased Asymmetry of Trunk, Pelvis, and Hip Motion during Gait in Ambulatory Children with Spina Bifida. Symmetry. 2021;13(9):1595.