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Study of Force Changes Based on Orthotic Elements Under the First Ray

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Abstract

The first ray plays a fundamental role in foot biomechanics, particularly in stabilizing the medial longitudinal arch and enabling efficient weight transfer during the mid-stance and propulsion phases of gait. When dorsiflexed—a condition known as metatarsus primus elevatus—especially in its flexible form, this structure disrupts load distribution, impairs propulsion, and contributes to various clinical symptoms. Despite its clinical importance, the biomechanical impact of orthotic elements placed beneath the first ray remains underexplored. This study aimed to quantify the variations in medio-lateral (Fx), antero-posterior (Fy), and vertical (Fz) force vectors generated during gait in response to different orthotic elements positioned under the first ray. A quasi-experimental, post-test design was conducted involving 22 participants (10 men and 12 women) diagnosed with flexible metatarsus primus elevatus. Each participant was evaluated using custom-made insoles incorporating various orthotic elements, while gait data were collected using a dynamometric platform during the mid-stance and propulsion phases. Significant gait-phase-dependent force alterations were observed. A cut-out (E) reduced medio-lateral forces during propulsion ($p < 0.05$), while a kinetic wedge (F) was correlated with late-stance stability ($r = -0.526$). The foot posture index (FPI)/body mass index (BMI) mediated the vertical forces. The effect sizes reached 0.45–0.42 for antero-posterior force modulation. Phase-targeted orthoses (a cut-out for propulsion, a kinetic wedge for late stance) and patient factors (FPI/BMI) appear to promote biomechanical efficacy in metatarsus primus elevatus, enabling personalized therapeutic strategies.

Keywords: foot; first ray; foot orthoses; forces

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1. Introduction

The first ray plays a crucial role in the biomechanics of gait, particularly during mid-stance and propulsion in the support phase. It contributes to the formation of the medial longitudinal arch of the foot, adapting to terrain irregularities and acting as a rigid segment that propels the body weight forward [1].

The dorsiflexed first ray, initially described by Lambrinudi [2] in 1938 as metatarsus primus elevatus, is characterized by a position in which the head of the first metatarsal lies above the plane of the other metatarsal heads. Today, the definition of metatarsus primus elevatus places greater emphasis on movement, affecting the normal function of the foot

and leading to a variety of symptoms [3]. A first ray is considered to exhibit this morpho-functional alteration when it shows a range of dorsiflexion greater than plantarflexion from its neutral position. From a biomechanical perspective, the first ray plays an essential role in the propulsion process of the foot during gait, assisting in the transfer of weight to the hallux and stabilizing the foot during push-off. The dorsiflexion of the first ray appears to disrupt this process by preventing the first metatarsal from aligning properly with the rest of the metatarsals during stance, resulting in uneven force distribution and potentially inducing pain in the area of the first metatarsal head, as well as in other parts of the foot, such as the second metatarsal. This condition can lead to dysfunction in the function of the medial longitudinal arch and altered force distribution during the gait cycle, especially during the mid-stance and propulsion phases [2,4]. The clinical consequences include pain in the first metatarsal area, difficulty with propulsion, and possible problems in other structures of the foot, such as the hallux valgus or metatarsalgia [5–9].

Munuera [5], building on Michaud's [6] classification for the plantarflexed first ray, developed a new categorization for the metatarsus primus elevatus. This classification distinguishes three types: flexible, when the first ray can plantarflex and lie below the other metatarsal heads; semi-flexible, when it reaches the level of the other metatarsal heads; and rigid, when its plantarflexion capacity is insufficient to align with the other metatarsals. This alteration can have either a congenital or acquired origin [10]. Several studies have evaluated the validity and reliability of the techniques and methods used to measure the mobility of the first ray [11–14].

The treatment of the first metatarsophalangeal segment focuses on reducing symptoms without restricting the plantarflexion of the first ray [15], with podiatric interventions being the primary option. Foot orthoses aim to improve foot functionality, with specific adaptations based on the classification of the deformity. The pathological pronation of the hindfoot is one of the main factors contributing to the misalignment and progression of various pathologies of the first ray, such as hallux valgus. This biomechanical alteration causes an abnormal distribution of forces during gait, which generates additional stress on the metatarsophalangeal joint and promotes a progressive deformity. Orthotic insoles play a crucial role in the management of these pathologies because, by correcting excessive pronation, they allow for a more equitable distribution of loads and improve the foot alignment, which, in turn, optimizes gait and delays the progression of the deformity [5].

Despite the clinical use of orthoses for first-ray pathologies, the evidence quantifying their biomechanical impact on three-dimensional force vectors remains limited. Previous studies [16–18] have primarily focused on mobility assessments or plantar pressure outcomes, with scarce data derived from dynamometric platforms analyzing the medio-lateral (X), antero-posterior (Y), and vertical (Z) forces. Importantly, no previous research has systematically examined how orthotic elements placed specifically under the first ray modify ground reaction forces in cases of flexible metatarsus primus elevatus, a gap that hampers the development of targeted orthotic designs. This knowledge deficit motivates our study, as understanding the directional reorientation of forces is essential to optimize the orthotic efficacy and reduce the risk of compensatory biomechanical adaptations. This study aimed to quantify variations in the medio-lateral (X-axis), antero-posterior (Y-axis), and vertical (Z-axis) force vectors, based on orthotic elements positioned under the first ray, using a dynamometric platform in individuals with flexible metatarsus primus elevatus.

2. Materials and Methods

2.1. Study Design

This was a crossover trial with repeated measures under randomized conditions [19]. The study received approval from the Ethics Committee of Junta of Andalucía, in accor-

dance with the Declaration of Helsinki and the national legislation on biomedical research involving human participants. All the participants provided signed informed consent prior to their participation.

2.2. Selection of Participants

The sample consisted of 22 feet from 22 individuals (10 men and 12 women) who were patients from the Podiatric Clinical Area of the Faculty of Nursing, Physiotherapy and Podiatry at the University of Seville. The inclusion criteria were established to ensure sample homogeneity and avoid the influence of external factors that could bias the results. The participants had to be over 20 years of age, regardless of gender, as the musculoskeletal system of the lower limb is considered to reach maturity by this age. This ensured that the joint mobility and biomechanics of the first ray were representative of an adult population, without the developmental alterations in bone and muscle structure that occur at younger ages. Additionally, the selected participants had a biomechanical alteration in the first ray of the foot, characterized by dorsiflexion, but one that was correctable, meaning that the joint could move into a plantarflexed position.

As for the exclusion criteria, these were established to identify individuals whose conditions could interfere with the study's objectives. Individuals were excluded if they had previously undergone osteoarticular surgery on the foot, had experienced severe foot trauma leading to significant alterations in bone morphology, suffered from degenerative osteoarticular disorders or neuromuscular conditions, or showed visible deformities in the forefoot that could influence the study's results.

The mobility of the first ray was assessed with the individual on a stretcher in the supine position using the exploration technique described by Root and colleagues, a classic maneuver widely accepted and used in clinical settings [5,20]. However, due to the reduced reliability of this technique, it was supplemented with a quantitative assessment to ensure more objective and reliable data. This involved quantifying the mobility of the first ray using an innovative and recently patented device capable of measuring the range of motion in millimeters, both at maximum dorsiflexion and maximum plantarflexion [4]. Prior to the study's commencement, we conducted extensive reliability testing involving twelve randomly selected participants who underwent three separate measurement sessions at one-week intervals. This protocol yielded excellent reliability coefficients, with intra-rater ICCs of 0.981 (95% CI: 0.935–0.995) for Investigator 1 and 0.936 (95% CI: 0.776–0.981) for Investigator 2, while the inter-rater reliability between both investigators reached an ICC of 0.978 (95% CI: 0.925–0.994). The examiner used one hand to hold the long arm of the instrument over the heads of the second to fifth metatarsals, while the other hand held the short arm over the head of the first metatarsal. From this position, the head of the first metatarsal was moved upwards until the limit of its dorsiflexion range was reached and then downwards to its maximum plantarflexion range. The range of motion was established by observing how many millimeters the instrument displaced in both positions. The evaluator repeated the measurements three times for each subject. This combination of qualitative and quantitative assessments allowed for a more comprehensive evaluation of the first-ray mobility, ensuring that both subjective observations and objective data were collected for further analysis.

2.3. Orthotic Interventions

A custom plantar support was created for each participant, molded with the subtalar joint in a neutral position and ensuring proper load on the first metatarsal head [21]. The base was fabricated using 3 mm retrocapital polypropylene, covered with 3 mm Nora®-EVA Lunasoft (48 Shore A, Nora systems GmbH, Weinheim, Germany). The par-

ticipants were provided with standardized Kalenji® (Villeneuve-d'Ascq, France) sports footwear to eliminate variability from footwear differences.

Seven orthotic interventions were evaluated: without a foot orthosis (WFO), a foot orthosis without elements (A), metatarsal elevation plus hallux (B), Morton's extension (C), proprioceptive stimulation (D), a cut-out (E), and a kinetic wedge (F) (Figure 1). The orthotic elements were added to the custom plantar supports with double-sided adhesive tape to analyze their individual impact on the force distribution. For each intervention, five measurements were taken, and the averages were calculated to obtain representative data. The tests were performed in random order to minimize possible biases in the results.



Figure 1. Orthotic elements added to the custom plantar supports: (**B**) metatarsal elevation plus hallux, (**C**) Morton's extension, (**D**) proprioceptive stimulation, (**E**) cut-out, and (**F**) kinetic wedge.

2.4. Gait Data Collection

Each participant walked in a straight line on a dynamometric platform (FP4060-07, Bertec, Columbus, OH, USA), which recorded the forces in all three spatial axes during the gait of the evaluated foot. The data were sampled at 500 Hz and filtered using a fourth-order low-pass Butterworth filter with a cutoff frequency of 20 Hz, following the recommendations for biomechanical gait data processing [22]. The Bertec FP4060-07 force platform is widely used in gait-analysis studies due to its high precision (sampling at 500 Hz) and low noise levels [22]. Although test–retest reliability data specific to this model are not publicly available in the peer-reviewed literature, force platforms with analogous designs have reported intraclass correlation coefficients (ICCs) > 0.90 for repeated gait measurements [23,24]. In our study, we followed strict protocols: calibration before each session, three practice trials per participant, and the averaging of five measurements per condition to ensure data consistency.

The stance phase of the gait was defined according to standard biomechanical descriptions found in the literature, which divide the gait cycle into distinct phases based on functional and kinematic events [4,25]. Specifically, the stance phase encompasses approximately 60% of the gait cycle, beginning at the initial contact (heel strike) and ending at toe-off. Within this phase, we focused on two sub phases: mid-stance (28% to 66% of the stance phase) and propulsion (67% to 100% of the stance phase), as these reflect the involvement of the first ray during the foot support phase. To evaluate the impact of orthotic interventions on the force distribution, for each spatial axis (medio-lateral, antero-posterior, and vertical), the greatest (G) and lowest (L) ground reaction force values were extracted

during these two periods of the stance phase: mid-stance (period 1) and propulsion (period 2). Accordingly, the variables were labeled as FxG1, FxL1, FyG1, FyL1, FzG1, FzL1, FxG2, FxL2, FyG2, FyL2, FzG2, and FzL2, where "Fx", "Fy", and "Fz" refer to the spatial axes; "G" and "L" indicate the greatest and lowest recorded forces; and "1" and "2" correspond to the mid-stance and propulsion periods, respectively.

The raw force data collected from the platform were processed using a custom Excel macro, in which each participant's body weight (in kilograms) was entered manually. The macro then converted the weight to Newtons (kg × 9.8) and applied the following normalization formula: (force in N/body weight in N) × 100. This allowed all ground reaction force values to be expressed as a percentage of each participant's body weight.

2.5. Descriptive and Statistical Analysis

The descriptive variables included sex, age, body mass (kg), height (m), body mass index (BMI), shoe size, and the foot posture index (FPI) [26,27]. The independent variable was the intervention in the first ray, evaluated under seven orthopedic conditions: no treatment, plantar support, metatarsal elevation plus hallux, Morton's extension, proprioceptive stimulation, a cut-out, and a kinetic wedge. As dependent variables, the greatest and lowest values of forces in the three spatial directions (medio-lateral, antero-posterior, and vertical) were analyzed.

The statistical data were analyzed using SPSS® version 27 for Windows, and significance was accepted at $p < 0.05$. The Shapiro–Wilk normality test was applied. Descriptive analyses of the age, body mass (kg), height (m), BMI, shoe size, and FPI were performed to describe the sample characteristics. The mean with standard deviation, median, and interquartile range values were calculated for each variable. A multivariate linear regression analysis was performed to assess whether the descriptive variables (age, sex, weight, height, BMI, shoe size, and foot posture index) acted as potential confounding factors in relation to the ground reaction force outcomes.

Correlation analyses were performed to explore potential associations between the descriptive variables (age, BMI, shoe size, and FPI) and the ground reaction force components. Given the non-normal distribution of several variables, Spearman's rho was used. These correlations were interpreted as indicators of a general association rather than linear relationships or adjustments for confounding factors. To provide clinical relevance, the magnitude of r or ρ was interpreted according to the following scale: 0.00–0.25 (poor), 0.26–0.50 (fair), 0.51–0.75 (moderate), and >0.75 (strong), as suggested by Cohen [28].

For related samples, non-parametric Wilcoxon and Friedman tests were applied, with post hoc tests conducted when the Friedman test detected significant differences. Rosenthal's indices were used for non-parametric tests, classifying them into four ranges: no significant effect (<0.2), small effect (0.2–0.5), medium effect (0.5–0.8), and large effect (>0.8) [29,30]. Related sample tests were used to compare the effects of different orthotic interventions on the dependent variables. These interventions included (1) no orthotic treatment (control group), (2) orthotic treatment (with custom plantar support), (3) first-ray elongation, (4) Morton's extension, (5) proprioceptive stimulation, (6) a first-ray cut-out, and (7) a kinetic wedge. Each of these conditions was compared to evaluate significant differences in the force distribution and first-ray mobility during gait.

3. Results

A total of 22 feet (N = 22) from the dominant lower limbs of 22 participants were analyzed, comprising 12 women and 10 men, with a mean age of 28.9 years and a mean body mass index (BMI) of 24.1 (Table 1).

Table 1. Main characteristics of total sample.

	Mean ± SD	Median	RIQ
Age (years)	28.9 ± 10.6	24.5	22.8–28.5
Body mass (kg)	70.5 ± 16.2	67.0	59.5–82.3
Height (m)	1.70 ± 0.09	1.70	1.63–1.78
BMI (kg/m^2)	24.1 ± 4.1	23.3	21.8–25.9
Shoe size	40.5 ± 3.0	40.5	38–43
FPI	3.6 ± 3.1	4	2–6

Abbreviations: kg, kilogram; m, meter; m^2, square meter; SD, standard deviation; RIQ, interquartile range.

A multivariate linear regression analysis was performed to evaluate whether the descriptive variables (age, sex, weight, height, BMI, shoe size, and foot posture index) could act as potential confounding factors affecting the ground reaction force components. This analysis was conducted across all orthotic conditions and for each force direction (Fx, Fy, and Fz), during both the mid-stance and propulsion phases.

As shown in Table 2, most of the coefficients of determination (R^2) ranged from 0 to below 0.5. Considering practical thresholds for identifying confounding factors (e.g., $r \geq 0.5$), the observed values generally fell short of this criterion, except for FxG2 in condition C ($R^2 = 0.507$) and FyG1 in condition F ($R^2 = 0.515$). The majority of the R^2 values were closer to 0.2–0.3, implying correlation coefficients well below 0.5 (e.g., $r \approx 0.45$ or lower). Although these two specific variables showed R^2 values slightly above 0.5, they did not exhibit statistically significant correlations with any of the descriptive variables in any condition. As such, their potential as confounding factors was considered negligible and does not affect the main findings of the study. We therefore found no statistical evidence to support the presence of confounding variables.

Table 2. Coefficients of determination (R^2) obtained from multivariate linear regression models evaluating the relationship between descriptive variables (age, sex, weight, height, BMI, shoe size, and foot posture index) and ground reaction force components (Fx, Fy, and Fz) across all orthotic conditions and gait phases.

		R^2		
		Fx	Fy	Fz
WFO	G1	0.312	0.117	0
	L1	0	0	0
	G2	0	0.204	0.172
	L2	0.110	0.095	0.090
A	G1	0	0.420	0.185
	L1	0	0.081	0.304
	G2	0.493	0.190	0
	L2	0	0	0.425
B	G1	0	0.390	0.304
	L1	0.394	0.126	0.424
	G2	0	0.372	0.065
	L2	0.401	0.328	0.108

Table 2. *Cont.*

		R^2		
		Fx	**Fy**	**Fz**
C	G1	0	0.108	0
	L1	0	0	0
	G2	0.507	0	0
	L2	0.219	0	0.169
D	G1	0	0.090	0
	L1	0	0.081	0
	G2	0.132	0	0.082
	L2	0.394	0.257	0.173
E	G1	0.259	0.113	0.096
	L1	0.275	0.131	0.141
	G2	0	0.211	0
	L2	0.254	0.113	0
F	G1	0	0.515	0
	L1	0	0	0
	G2	0.470	0.104	0
	L2	0.392	0	0.172

R^2 values represent the proportion of variance in each ground reaction force component explained by the combined effect of the descriptive variables. Values are shown separately for each orthotic condition and for the mid-stance and propulsion phases. Abbreviations: WFO, without a foot orthosis; A, a foot orthosis without elements; B, metatarsal elevation plus hallux, C, Morton's extension; D, proprioceptive stimulation, E, a cut-out; F, a kinetic wedge; Fx, medio-lateral force; Fy, antero-posterior force; Fz, vertical force; G1: greatest values of forces during mid-stance phase; L1: lowest values of forces during mid-stance phase; G2: greatest values of forces during propulsion phase; L2: least values of forces during propulsion phase.

As shown in Table 3, the greatest vertical forces during propulsion (FzG2) were observed under the no-orthosis (WFO) condition (109.3%), followed by orthoses F (105.6%) and B (104.6%), reflecting greater loading during push-off. For antero-posterior forces (Fy) in mid-stance (FyG1), the greatest propulsive forces were registered with orthoses E (9.2%) and B (8.9%). Regarding medio-lateral forces (Fx) during propulsion (FxL2), condition B showed the most lateral deviation (–2.9%), while condition A exhibited the least (–1.7%), suggesting better medial stabilization.

Table 4 displays the correlations between the ground reaction force components (Fx, Fy, and Fz) and the participant variables (age, BMI, shoe size, and foot posture index, FPI) across various orthotic conditions and gait phases. Significant negative correlations were found between the FPI and the medio-lateral force (Fx) during late stance (L2) for orthoses B, C, D, E, and F (r values between -0.436 and -0.608; $p < 0.05$), indicating increased lateral shifts in more pronated feet. The FPI also showed significant positive correlations with the vertical force (Fz) for orthosis B during mid-stance and propulsion, and for orthosis E during mid-stance. Fy was positively correlated with the FPI only in orthosis B during mid-stance. The BMI was negatively associated with Fz for WFO (G2) and orthosis F (L2).

Table 3. Recorded variables of Fx, Fy, and Fz using the dynamometric platform.

	FORCE	**Fx**				**Fy**				**Fz**			
		MEAN	**SD**	**MED**	**IQR**	**MEAN**	**SD**	**MED**	**IQR**	**MEAN**	**SD**	**MED**	**IQR**
WFO	G1	3.9	6.2	6.1	−3.2–9.4	7.0	5.4	9	1.5–10.5	82.0	31.1	91.1	58.7–105.4
	L1	0.8	6.3	2.7	−5.8–6.6	−5.0	3.1	−4.8	−6.4; −3.6	55.5	27.6	56.4	32.2–83.9
	G2	4.7	4.1	4.9	0.8–8.5	3.5	4.6	1.1	0.3–2.8	109.3	33.1	109.9	98.0–138.1
	L2	−2.5	4.2	−1.9	−5.7–1.0	−16.1	7.0	−16.3	−20.8; −13.0	15.5	17.4	1.1	0.7–33.3
A	G1	3.9	6.3	6	−8–9.0	7.0	5.7	7.5	3.5–11.1	92.0	29.0	98.4	74.1–105.1
	L1	0.3	5.9	1.2	−5.4–4.2	−7.1	7.5	−5	−6.4; −3.2	59.5	26.4	70.4	32.9–80.6
	G2	4.8	4.3	5.2	0.3–9.0	2.6	5.3	0.4	−0.01–4.0	106.1	21.9	105.2	98.9–112.9
	L2	−1.7	5.4	−1.6	−6.5–1.5	−14.2	6.9	−14.4	−20.2; −7.2	15.2	17.0	1.0	0.7–32.8
B	G1	2.8	5.9	2.6	−3.4–8.1	8.9	6.0	10.8	4.6–12.7	89.0	36.3	98.2	75.5–106.6
	L1	−0.7	5.1	−0.8	−5.1–3.3	−4.6	2.2	−4.5	−6.5; −2.9	54.9	27.4	56.7	31.7–79.6
	G2	4.1	4.2	2.9	0.3–8.4	2.6	3.4	0.7	−0.1–5.6	104.6	25.3	107.4	99–113
	L2	−2.9	4.3	−2.4	−5.8–0.3	−14.9	6.5	−14.9	−20.2; −9.5	14.1	16.4	1	0.8–29.6
C	G1	3.7	6.2	5.1	−3.2–9.5	8.5	7.4	11.2	2.5–12.7	82.9	36.2	94.9	38.8–104.5
	L1	0.3	6.0	0.1	−4.8–5.5	−5.4	5.4	−3.9	−5.2; −2.9	49.5	23.4	39	29.7–71.4
	G2	4.8	4.4	4.1	0.5–9.0	3.4	5.7	0.6	−0.2–5.6	99.3	35.4	105.4	83.3–114.6
	L2	−2.6	5.5	−2.6	−5.8–0.5	−15.5	6.6	−15.6	−20.6; −10.4	16.8	17.3	10.6	0.7–33.2
D	G1	3.5	6.2	5	−3.9–9.5	8.5	6.7	10.2	4.3–14.2	83.9	30.2	92.4	53.1–105.5
	L1	0.3	6.2	1.7	−5.7–5.6	−4.8	1.8	−4.6	−6.1; −3.5	51.6	24.2	42	32.1–77.0
	G2	5.1	4.6	5.4	0.1–9.0	3.0	4.6	1.3	−0.2–5.9	102.7	34.1	106.5	88.5–112.8
	L2	−2.4	5.0	−0.6	−7.3–1.8	−15.6	6.7	−17.3	−19.7; −8.9	16.8	17.3	10.6	0.7–33.2

Table 3. *Cont.*

	FORCE	Fx				Fy				Fz			
		MEAN	**SD**	**MED**	**IQR**	**MEAN**	**SD**	**MED**	**IQR**	**MEAN**	**SD**	**MED**	**IQR**
E	G1	3.4	6.5	5.4	−3.9–8.7	9.2	5.3	10.1	4.9–13.0	91.5	27.3	99	84.4–106.3
	L1	−0.03	5.9	0.7	−6.2–4.2	−4.4	1.9	−4.1	−5.8; −3.1	55.7	26.3	57.2	32.2–80.0
	G2	4.3	4.3	4.8	−0.03–8.4	2.1	3.8	0.7	−0.04–3.6	98.9	23.5	104.5	91.0–109.8
	L2	−2.4	5.1	−0.6	−7.6–2.5	−15.1	6.3	−16.7	−20.3; −8.1	13.8	15.4	1.1	0.7–30.7
F	G1	3.3	6.2	5.3	−3.4–9.0	8.8	6.2	10.3	2.7–14.1	92.5	25.0	97.2	70.9–107.0
	L1	−0.1	6.5	−0.1	−5.9–6.2	−4.6	3.3	−4.3	−5.3; −2.8	56.1	24.2	63.6	32.8–77.3
	G2	4.7	4.4	4.4	0.1–9.0	3.2	5.8	0.8	−0.1–5.9	105.6	21.9	105.4	95.1–111.3
	L2	−2.6	5.6	−2.4	−7.8–1.9	−14.6	6.1	−15.5	−18.9; −8.7	16.9	17.2	10.6	0.8–33.2

Abbreviations: SD, standard deviation; MED, median; IQR, interquartile range; WFO, without a foot orthosis; A, a foot orthosis without elements; B, a metatarsal elevation plus hallux; C, Morton's extension; D, proprioceptive stimulation; E, a cut-out; F, a kinetic wedge; Fx, medio-lateral force; Fy, antero-posterior force; Fz, vertical force; G1: greatest values of forces during mid-stance phase; L1: lowest values of forces during mid-stance phase; G2: greatest values of forces during propulsion phase; L2: lowest values of forces during propulsion phase.

Table 4. Correlations between medio-lateral (Fx), vertical (Fz), and antero-posterior (Fy) axis ground reaction forces and descriptive variables across different orthotic conditions and gait phases.

Fx Correlations		**Age**		**BMI**		**Shoe Size**		**FPI**	
		r	***p***	**r**	***p***	**r**	***p***	**r**	***p***
B	L1	0.248	0.265 [2]	0.162	0.470 [1]	−0.028	0.900 [1]	−0.563	0.006 [1]
	L2	0.086	0.704 [2]	0.136	0.547 [2]	0.056	0.806 [2]	−0.608	0.003 [1]
C	L2	−0.028	0.902 [2]	0.029	0.897 [2]	0.108	0.633 [1]	−0.436	0.043 [1]
D	L2	0.108	0.633 [2]	−0.022	0.924 [2]	−0.072	0.751 [1]	−0.489	0.021 [1]
E	L1	0.146	0.516 [2]	−0.022	0.921 [2]	−0.049	0.829 [1]	−0.444	0.039 [1]
	L2	0.130	0.565 [2]	0.084	0.709 [2]	0.155	0.491 [2]	−0.475	0.026 [2]
F	L2	0.024	0.914 [2]	0.029	0.897 [2]	−0.018	0.936 [1]	−0.526	0.012 [1]
Fy Correlations		**Age**		**BMI**		**Shoe Size**		**FPI**	
		r	***p***	**r**	***p***	**r**	***p***	**r**	***p***
B	G1	−0.093	0.679 [2]	−0.380	0.081 [1]	−0.024	0.914 [1]	0.570	0.006 [1]
E	L2	0.463	0.030 [2]	0.223	0.318 [2]	−0.265	0.234 [2]	−0.346	0.115 [2]
Fz Correlations		**Age**		**BMI**		**Shoe Size**		**FPI**	
		r	***p***	**r**	***p***	**r**	***p***	**r**	***p***
WFO	G2	−0.383	0.078 [2]	−0.523	0.012 [2]	−0.239	0.284 [2]	0.178	0.428 [2]
	L2	−0.157	0.485 [2]	−0.492	0.020 [2]	−0.323	0.143 [2]	−0.065	0.775 [2]
B	G1	−0.034	0.880 [2]	−0.477	0.025 [1]	0.007	0.974 [1]	0.580	0.005 [1]
	G2	0.112	0.619 [2]	−0.295	0.183 [2]	−0.179	0.425 [2]	0.426	0.048 [2]
	L2	0.259	0.245 [2]	−0.190	0.397 [2]	−0.458	0.032 [2]	0.003	0.991 [2]
E	L1	−0.285	0.199 [2]	−0.086	0.7042	−0.123	0.586 [2]	0.436	0.043 [2]
F	L2	−0.117	0.603 [2]	−0.508	0.016 [2]	−0.457	0.033 [2]	0.053	0.816 [2]

Abbreviations: [1] Pearson's correlation; [2] Spearman's rho; BMI, body mass index; FPI, foot posture index; WFO, without foot orthosis; A, foot orthosis without elements; B, metatarsal elevation plus hallux; C, Morton's extension; D, proprioceptive stimulation; E, a cut-out; F, a kinetic wedge; Fx, medio-lateral force; Fz, vertical force; G1: greatest values of forces during mid-stance phase; L1: lowest values of forces during mid-stance phase; G2: greatest values of forces during propulsion phase; L2: lowest values of forces during propulsion phase.

On the other hand, the Wilcoxon test compared the force vectors between the no-insole condition (WFO-A) and each orthotic intervention. Table 5 presents the *p*-values obtained from these comparisons. For FY, only one statistically significant difference was found: the maximum antero-posterior force during mid-stance showed significance when comparing WFO-A with WFO-D ($p = 0.0332$). Regarding FZ, a statistically significant difference in the maximum vertical force during propulsion was found between WFO-A and WFO-C ($p = 0.0492$).

The Friedman test was applied to compare the medio-lateral (Fx), antero-posterior (Fy), and vertical (Fz) forces across the seven orthotic interventions, assessing the variations during the mid-stance and propulsion phases. The results showed significant differences in Fx during propulsion ($p = 0.032$) and in Fy during mid-stance ($p = 0.004$ and 0.015 for G1 and L1, respectively) (Table 6).

Table 5. Comparative analysis (*p*-values) of Fx, Fy, and Fz between the no-insole (WFO) condition and the orthotic intervention.

	FX						**FY**						**FZ**					
	WFO-A p1	**WFO-B p1**	**WFO-C p1**	**WFO-D p1**	**WFO-E p1**	**WFO-F p1**	**WFO-A p1**	**WFO-B p1**	**WFO-C p1**	**WFO-D p1**	**WFO-E p1**	**WFO-F p1**	**WFO-A p1**	**WFO-B p1**	**WFO-C p1**	**WFO-D p1**	**WFO-E p1**	**WFO-F p1**
G1	0.897 [2]	0.485 [2]	0.455 [2]	0.110 [2]	0.758 [2]	0.173 [2]	0.958 [1]	0.201 [1]	0.308 [1]	0.033 [2]	0.093 [1]	0.089 [1]	0.116 [1]	0.508 [1]	0.917 [1]	0.465 [2]	0.381 [2]	0.061 [1]
L1	0.429 [1]	0.072 [1]	0.545 [1]	0.543 [1]	0.523 [1]	0.114 [1]	0.299 [2]	0.745 [2]	0.230 [2]	0.795 [2]	0.322 [2]	0.148 [2]	0.211 [2]	0.811 [1]	0.188 [2]	0.167 [2]	0.931 [2]	0.758 [2]
G2	0.456 [2]	0.236 [2]	0.404 [2]	0.302 [1]	0.052 [2]	0.258 [2]	0.363 [2]	0.338 [2]	0.709 [2]	0.426 [2]	0.049 [2]	0.602 [2]	0.527 [2]	0.426 [2]	0.527 [2]	0.506 [2]	0.101 [2]	0.306 [2]
L2	0.370 [1]	0.492 [1]	0.872 [1]	0.931 [1]	0.661 [2]	0.890 [1]	0.196 [1]	0.433 [1]	0.693 [1]	0.600 [1]	0.223 [2]	0.426 [2]	0.486 [2]	0.794 [2]	0.614 [2]	0.922 [2]	0.672 [2]	0.266 [2]

Abbreviations: WFO, without foot orthosis; A, foot orthosis without elements; B, metatarsal elevation plus hallux; C, Morton's extension; D, proprioceptive stimulation; E, a cut-out; F, a kinetic wedge; Fx, medio-lateral force; Fy, antero-posterior force; Fz, vertical force; G1: greatest values of forces during mid-stance phase; L1: lowest values of forces during mid-stance phase; G2: greatest values of forces during propulsion phase; L2: lowest values of forces during propulsion phase; [1] related-samples *t*-test; [2] Wilcoxon signed-rank test for related samples.

Table 6. Comparative analysis (*p*-values) of Fx, Fy, and Fz under different orthotic interventions.

Force Measurement	Fx	Fy	Fz
G1 (Mid-Stance)	0.140 [2]	0.004 [2]	0.140 [2]
L1 (Mid-Stance)	0.480 [1]	0.015 [2]	0.107 [2]
G2 (Propulsion)	0.032 [2]	0.640 [2]	0.620 [2]
L2 (Propulsion)	0.436 [2]	0.930 [2]	0.411 [2]

Abbreviations: Fx, medio-lateral force; Fy, antero-posterior force; Fz, vertical force; G1: greatest values of forces during mid-stance phase; L1: lowest values of forces during mid-stance phase; G2: greatest values of forces during propulsion phase; L2: lowest values of forces during propulsion phase; [1] repeated-measures ANOVA; [2] Friedman's test for related samples for a two-way analysis of variance by rank.

The post hoc comparisons (Table 7) showed significant differences in the medio-lateral forces during propulsion between WFO-E and WFO-F (p = 0.031), and in the antero-posterior forces during mid-stance between A–E and A–F (p = 0.031) and A–D and A–F (p = 0.025). A trend toward significance was observed between WFO-F and A–F (p = 0.057) in the antero-posterior direction.

Table 7. Post hoc tests; pairwise comparisons.

FX	WFO-E	WFO-F	A–D	A–F
G2	0.031	>0.999	>0.999	>0.999
FY	**WFO-F**	**A–E**	**A–F**	
G1	0.040	0.490	0.057	
L1	>0.999	0.031	0.025	

Abbreviations: Fx, medio-lateral force; Fy, antero-posterior force; G1: greatest values of forces during mid-stance phase; L1: lowest values of forces during mid-stance phase; G2: greatest values of forces during propulsion phase; L2: lowest values of forces during propulsion phase; WFO, without foot orthosis; A, foot orthosis without elements; B, metatarsal elevation plus hallux; C, Morton's extension; D, proprioceptive stimulation; E, a cut-out; F, a kinetic wedge.

Table 8 shows the effect sizes for significant comparisons in the antero-posterior force (Fy). Moderate effect sizes were observed for WFO-D during mid-stance (p = 0.033, d = 0.453) and for WFO-E during propulsion (p = 0.049, d = 0.419), indicating meaningful differences between the conditions.

Table 8. Effect size of Fy.

	FY	
	p1	**Effect Size**
WFO-D G1	0.033	0.453
WFO-E G2	0.049	0.419

Abbreviations: Fy, antero-posterior force; WFO, without foot orthosis; G1: greatest values of forces during mid-stance phase; G2: greatest values of forces during propulsion phase; D, proprioceptive stimulation; E, a cut-out.

4. Discussion

Although the first ray plays a fundamental role in the biomechanical stability and propulsion of the foot, especially during mid-stance and push-off, there is a lack of studies specifically addressing how different orthotic elements placed beneath this structure affect the ground reaction forces during gait. Most existing research has focused on the kinematic aspects or general effects of foot orthoses, without analyzing how targeted modifications under the first ray influence force vectors in the sagittal, frontal, and transverse planes. Understanding these effects is particularly relevant in individuals with flexible metatarsus

primus elevatus, where the mechanical dysfunction of the first ray can lead to altered load distribution and an increased risk of a forefoot pathology. Therefore, this study sought to address this gap by quantifying changes in the medio-lateral (X-axis), antero-posterior (Y-axis), and vertical (Z-axis) ground reaction forces in response to different orthotic interventions applied under the first ray.

The literature review [31–34] highlighted the lack of podiatry studies using dynamometric platforms to analyze force vector variations in the three spatial directions with orthotics under the first ray. While some studies [35–38] on human gait exist, they are not directly comparable to this research. Despite various treatments [39–46] aimed at optimizing first-ray functionality and improving gait biomechanics, no classification based on force control during the gait cycle has been established, emphasizing the need for personalized interventions for metatarsus primus elevatus.

The sample used in this study consisted of young individuals with a normal weight, height, and BMI, ensuring uniformity in foot size and minimizing gait-influencing factors. The foot posture index (FPI) showed a normal posture with slight pronation [26,27], associated with limitations in first metatarsophalangeal joint dorsiflexion [47].

The correlation analysis revealed that greater foot pronation (a higher FPI) resulted in significantly lower medio-lateral forces (Fx) during the propulsion phase (L2) with the following orthoses: metatarsal elevation plus hallux, Morton's extension, proprioceptive stimulus, a cut-out, or a kinetic wedge. No significant FPI-Fx correlations were observed during mid-stance (G1/L1). This suggests that these devices modulate the pronatory forces in pronated feet. Regarding the vertical forces (Fz), significant positive correlations with the FPI occurred specifically for the orthosis with metatarsal elevation plus hallux during mid-stance (G1: $r = 0.580$) and propulsion (G2: $r = 0.426$), and for the orthosis with a cut-out during mid-stance (G1: $r = 0.436$). This implies that more pronated feet leverage these specific orthoses to increase the vertical loading, potentially through force redistribution from the medial to the lateral forefoot.

Significant differences in the medio-lateral forces were observed, specifically during propulsion (G2 phase) between the no-orthosis (WFO) and the cut-out (E) conditions, which is consistent with the cut-out's biomechanical role in modulating supination. The kinetic wedge (F) showed no statistically supported effects during mid-stance or propulsion in the pairwise analyses. This phase-dependent behavior, with the cut-out modulating forces during propulsion and the kinetic wedge during late stance, aligns with their shared mechanism of restricting first-ray dorsiflexion, though the recorded force magnitudes were consistent with typical gait patterns. This phase-dependent control, which enhances the medial stability during late stance and the lateral stability during propulsion, suggests how targeted orthotic designs can address gait-phase-specific instability in metatarsus primus elevatus.

The consistency in the antero-posterior forces (Fy) across all conditions implies neuromuscular adaptations in the sagittal plane's preserved propulsive mechanics, despite orthotic interventions. This suggests that first-ray modifications predominantly alter the frontal (Fx) and vertical (Fz) plane control. The vertical forces (Fz) showed no phase-wide orthotic effects during propulsion, though individual correlations revealed BMI- and FPI-dependent modulation. The absence of Fz changes aligns with the results of Scherer et al. [35], where insoles redistributed local pressures without modifying the net vertical impulse. While proprioceptive mechanisms [48] offer a theoretical rationale for force modulation, their role remains unexplored in this dataset due to non-significant D-specific outcomes.

The Friedman test results showed significant differences in the medio-lateral forces during propulsion (G2), attributable to the cut-out orthosis (E) versus no orthosis. This

suggests the kinetic wedge's effectiveness in modulating pronatory forces, particularly during propulsion. This may relate to the enhanced activation of the flexor hallucis longus, improving first-ray stability [47]. The effect size quantification was limited to the antero-posterior forces (Fy), revealing moderate impacts for the proprioceptive stimulus (D) during mid-stance and the cut-out (E) during propulsion. The pronatory force (Fx) and vertical loading (Fz) effects remain unquantified in this dataset.

This study has certain limitations that should be acknowledged. One of them is that it should be interpreted primarily as a methodological contribution rather than a definitive evaluation of the clinical outcomes. The sample size was limited, which reduces the generalizability of the statistical findings and precludes subgroup analyses (e.g., by sex or first-ray mobility). Although it provided sufficient power to detect global orthotic effects, larger and more diverse samples are needed to confirm these results in clinical populations.

The sample consisted of asymptomatic adults with correctable dorsiflexion of the first ray, minimizing confounders while limiting the applicability to patients with more severe structural foot conditions. The design was cross-sectional and focused on immediate biomechanical responses without assessing long-term adaptations or clinical outcomes.

Also, this study relied solely on tri-axial ground reaction force data without including kinematic or electromyographic measures, preventing the identification of the precise mechanisms driving force redistribution. Another limitation is that this study did not evaluate energy absorption, power generation, or gait efficiency, which should be addressed in future research.

And finally, although more advanced statistical models could potentially enhance the analysis of repeated-measures data, such as mixed-effects models, their application in small-sample exploratory studies may lead to overfitting or unstable estimates. For this reason, we adopted a conservative and methodologically sound strategy using non-parametric tests, which were deemed more appropriate for the characteristics and objectives of the present work.

5. Final Considerations

The purpose of this research was to analyze the impact of orthotic elements placed under the first ray on the variations in the force vectors (Fx, Fy, and Fz) during the support and propulsion phases in individuals with flexible metatarsus primus elevatus. The study aimed to provide relevant information on how these interventions can influence gait biomechanics in this patient population.

The main results demonstrated that orthotic elements significantly modulate forces during the critical phases of gait, with the kinetic wedge standing out as the most effective intervention for controlling pronatory forces during propulsion. Additionally, both the proprioceptive stimulus and the kinetic wedge had a considerable effect on propulsive forces during the mid-stance phase. These findings support the use of this experimental setup as a replicable protocol for future research, emphasizing the importance of selecting the most appropriate orthotic interventions to optimize gait biomechanics in patients with flexible metatarsus primus elevatus.

In conclusion, although the limited sample of this study restricts its generalizability, the results highlight the potential of a standardized methodological protocol to improve the understanding of first-ray functionality and gait mechanics. These interventions offer a personalized and potentially effective therapeutic approach for patients with this condition.

Future studies should build upon this framework, incorporating larger and longitudinal samples to validate the long-term biomechanical and clinical effects of orthotic modifications. Furthermore, the development of more targeted orthotic designs for pa-

tients with osteoarticular and neuromuscular conditions is recommended to improve the therapeutic outcomes in podiatric practice.

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