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Article

Impact of Obesity on Foot Kinematics: Greater Arch Compression and Metatarsophalangeal Joint Dorsiflexion despite Similar Joint Coupling Ratios

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Abstract: This study investigates the sagittal plane dynamics of the foot, particularly the metatarsophalangeal (MTP) joint and medial longitudinal arch (MLA) movements, in relation to obesity and foot health. The kinematics of the MTP and arch joints were measured in 17 individuals with class 2–3 obesity (BMI > 35 kg/m²) and 10 normal-weight individuals (BMI $\leq$ 24.9 kg/m²) using marker-based tracking. Analysis was conducted during heel lifting while seated and during walking at self-selected speeds. The results indicated that obese participants exhibited 20.92% greater MTP joint dorsiflexion at the end of the push-off phase and 19.84% greater MLA compression during the stance phase compared to normal-weight controls. However, no significant differences were found in the kinematic joint coupling ratio. While these findings reveal the different biomechanical behaviors of the MTP joint and MLA in obese compared to normal-weight individuals, it is important to interpret the implications of these differences with caution. This study identifies specific biomechanical variations that could be further explored to understand their potential impact on foot health in obese populations.

Keywords: foot kinematics; obesity; arch compression; MTP joint dorsiflexion; plantar aponeurosis; midfoot mobility

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1. Introduction

The global increase in obesity prevalence necessitates a comprehensive understanding of its manifold implications, especially within the domain of musculoskeletal health (e.g., for a review, see [1,2]). With feet playing an important role in movement, balance and agility, foot health has become increasingly important. Several studies have linked obesity to declining foot health. For instance, Mickle and Steele [3] and Frey and Zamora [4] have identified that obese populations exhibit a higher incidence of foot pain and functional limitations. Expanding on this theme, Butterworth et al. [5] proposed that obesity increases stress on the foot, both directly through increased body weight and indirectly through changes in foot structure. These findings not only emphasize discomfort but also demonstrate that such restrictions can negatively impact one's quality of life.

The adverse effects on foot health can be linked to various underlying biomechanical factors. A recurring observation across studies is the alteration in plantar pressure distribution in obese individuals. There is a consensus showing higher plantar pressures in different foot regions, most prominently under the metatarsal heads and the midfoot area [6]. This redistribution of pressure is thought to be influenced by factors such as increased body mass [3], foot strength and laxity [7], and foot type, particularly pes planus [8].

It is worth noting that the analysis of foot function in obese individuals has predominantly been based on kinetic observations. Current kinematic studies have either concentrated on ankle joint kinematics [7,9–11] or frontal plane movement of the foot, such as pronation [9,11]. For example, research by Messier et al. [9] indicated that obese individuals, especially females, exhibit pronounced rearfoot motion and a tendency towards a pronated foot posture. However, a significant gap remains in the understanding of the sagittal plane dynamics of the foot, especially concerning the movement of the metatarsophalangeal (MTP) joint and the medial longitudinal arch (MLA).

An understanding of sagittal movement is essential as it is closely linked to the dynamics of the plantar aponeurosis during gait (e.g., [12–15]). Among others, McDonald et al. [16] state that arch compression during gait largely stresses the plantar aponeurosis. This fact becomes especially important when we consider the high frequency of cases of plantar fasciitis among obese individuals. During gait, the arch drops predominantly during the stance phase, only to rise rapidly towards the end [17,18]. The rise in the arch is closely coupled with MTP joint motion [17–19]. One of the fundamental principles underpinning the close kinematic coupling between MTP and MLA joint motion is the 'windlass mechanism', a concept pioneered by Hicks [20]. The interplay between the MTP joints and the plantar aponeurosis influences MLA dynamics, effectively altering push-off mechanics during walking and running.

Beyond the established windlass mechanism, there is emerging evidence that foot muscles and elastic energy in ligaments and tendons further contribute to the kinematic coupling between MTP and midfoot joints [19,21,22]. These findings, together with the reported changes in plantar pressure, foot strength, and pronation in overweight people, raise the need to investigate the likely changes in joint coupling and how the bodyweight could affect the down- and upward motion of the arch during gait.

In this study, we aimed to measure the kinematics of the MTP and arch joints in both individuals with normal weight and individuals who are obese with a BMI greater 30 kg/m^2 (class 2–3 obesity). Our initial assumption was that there would be a clear distinction in kinematic coupling between the two groups, mainly due to the impact of extra body mass. This could potentially manifest as increased MTP joint dorsiflexion in obese individuals during the propulsive phase and a decreased rise in the MLA at the end of the stance phase. Besides gait analysis, heel lifting while sitting was included as an additional task and used as a controlled mechanism for systematic manipulation of the MTP joint. The rationale for this choice was based on significant correlations observed between midfoot and ankle motion and ankle power during heel raises and the push-off phase of walking [23]. Such correlations suggest that heel raises may serve as a useful surrogate for the push-off phase of gait, independent of the influences of body weight. Consequently, we pursued a secondary assumption: that kinematic coupling during the heel-raise task would show negligible differences between obese and normal-weight participants. By examining these assumptions, we seek to unravel the intricate ways in which body weight can influence foot function and, by extension, overall biomechanical health.

2. Materials and Methods

2.1. Participants

Our sample consisted of 17 obese (BMI $\geq$ 35 kg/m^2) and 10 normal-weight (BMI $\leq$ 24.9 kg/m^2) individuals. Inclusion criteria for the obese sample was that participants had to have a BMI greater than 35 kg/m^2. The participant's height and body mass were measured to calculate body mass index (BMI) using the formula body mass/height^2. In addition, we applied a custom-made device to determine the total foot length, truncated foot length (from the heel to the first metatarsophalangeal joint), and dorsum height at 50% of foot length in seated position [24] (Figure 1A). The AHI was calculated by dividing the foot's dorsum height at 50% of foot length by the total length of the foot, excluding the toes, while the participants were sitting. This has been established as a reliable and consistent measure of MLA height [25]. In line with previous studies, participants with 'low' MLA values were identified when their (AHI) was below 0.297 [26,27], which is 1.5 standard deviations below the mean AHI of a large sample of adult males [25]. Table 1 summarizes all anthropometric data. Written informed consent was obtained from all participants. The study was conducted in accordance with the Declaration of Helsinki for studies involving humans. The ethics committee of our faculty granted ethical approval for this study (Approval Number: V-287-17-FE-Füße-02072018) after reviewing the non-invasive procedures and low-risk design of our research. The committee affirmed that the procedures are similar to those encountered in everyday life and do not increase risk levels for participants.

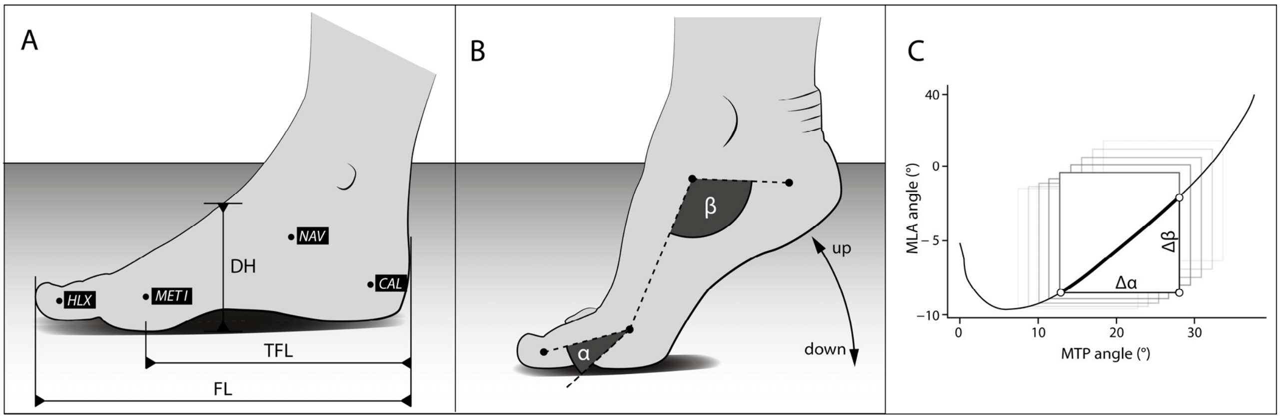

Figure 1. Kinematic analysis of the foot. Panel (**A**): Static measurements of the foot illustrating foot length (FL), truncated foot length (TFL), and dorsal height at 50% foot length (DH). Key markers (shown as black dots) are positioned at specific anatomical landmarks: the hallux (HLX), the first metatarsophalangeal joint (MET I), the navicular tuberosity (NAV), and the medial aspect of the calcaneus (CAL). Panel (**B**): Representation of the foot during the heel raising task in a seated position. Motion was captured in the sagittal plane using a high-speed camera, focusing on markers required to calculate the metatarsophalangeal (MTP) angle (shown as α) and the medial longitudinal arch (MLA) angle (shown as β). Panel (**C**): The plot represents the relationship between the MTP and MLA angles during the push-off phase of walking. The degree of coupling between the MTP and MLA joints during toe extension was determined using a sliding window analysis technique. Specifically, segments or 'windows' of 50 consecutive data points were examined. The correlation coefficient was calculated within each window to assess linearity and identify the segment that displayed the most linear relationship between the movements of the toe and the medial longitudinal arch. The slope of the line was then calculated using the 50 data points within this identified window (kinematic joint coupling ratio = $\Delta\beta/\Delta\alpha$).

Table 1. Summary of anthropometric data including foot dimensions of normal and overweight subjects (NW and OW, respectively).

	NW	**OW**
n (female/male)	10 (6/4)	17 (9/8)
age (yrs)	28.7 ± 9.1 (20–51)	44.1 ± 12.8 (25–65)
weight (kg)	68.1 ± 11.5 (53.0–88.6)	136.3 ± 27.5 (92.0–196.0)
height (cm)	175.6 ± 11.3 (160–193)	174.1 ± 12.9 (152–197)
BMI (kg/m²)	21.9 ± 2.1 (18.4–24.8)	44.8 ± 6.9 (37.2–61.3)
foot length (cm)	24.8 ± 1.8 (22.4–28.3)	26.2 ± 2.2 (21.2–29.5)
truncated foot length (cm)	18.3 ± 1.4 (16.0–20.8)	19.6 ± 1.8 (16.0–22.1)
arch height (cm)	6.5 ± 0.5 (5.7–7.5)	6.6 ± 0.9 (4.5–8.0)
AHI	0.38 ± 0.01 (0.35–0.40)	0.41 ± 0.7 (0.34–0.60)

2.2. Kinematic Measurements

Small LED markers were placed on the right lower limbs of participants before recording. The markers used to calculate MTP and midfoot motion included those placed on the medial aspect of the hallux at the joint between proximal and distal phalanges (HLX), the first metatarsal head (MET1), the navicular tuberosity (NAV), and the posterior calcaneus (CAL) (Figure 1A). We carefully considered the placement of foot markers to ensure alignment with underlying bony landmarks to mitigate the effects of soft tissue artifact, a critical factor in the accuracy of gait analysis. This approach to marker placement is particularly important in obese populations, where variations in soft tissue can affect kinematic data. However, Horsak et al. [28] highlighted that from an anatomical plane perspective, waveform similarity in marker-based recordings showed the highest values for the sagittal (and frontal) plane, justifying their use as a clinical outcome measure for obese populations. After marker placement, participants were instructed to practice two tasks: heel lifting while sitting and walking at a self-selected speed. We measured stance time—from heel strike to toe-off—as an estimate of walking speed, e.g., [29].

For the active heel raises, participants performed ten consecutive heel raises with their right leg to a metronome tempo of 45 beats per minute to control ankle angular velocity. After the static trial, participants were asked to walk barefoot across a 6 m track, maintaining a normal gait at a constant, comfortable speed, with all participants completing a minimum of three trials. The static and dynamic tasks were video-recorded using a GoPro Hero 10 camera (GoPro, Inc., San Mateo, CA, USA), with a 7.5 mm 3MP M12 lens (Back-Bone, Inc., Kanata, ON, Canada), that was positioned 0.5 m from the walking track to record a medial view of the individual's right foot at a frame capture rate of 200 Hz. To provide context on the accuracy of our camera-based 2D motion tracking system, Feng and Max [30] used a similar custom camera system operating at 240 Hz and reported an average root mean square error (RMSE) across dynamic tests of 0.18 ± 0.12 mm for dynamic tracking. For the dynamic walking task, we selected two to five trials for analysis per participant, based on an observation of the participant using consistent speed and walking kinematics, and all markers appearing unobstructed in the camera field of view during stance phase. The 2D marker positions were digitized using the DLTdv8 MATLAB package (MathWorks, Inc., Natick, MA, USA) [31], and marker positions were used to calculate the MTP joint angle and the angle of the medial longitudinal arch (MLA). The DLTdv8 algorithm is a MATLAB app that can operate independently of MATLAB or a MATLAB license. This algorithm enabled the direct reading of mp4 movie files, specifically recordings of the medial view of the foot during the stance phase of gait. It facilitated the tracking of markers placed on anatomical landmarks of the foot and the export of the 2D coordinates of these markers as a text file for further analysis. It is important to note that this method does not account for out-of-plane motions due to the absence of 3D tracking capabilities.

While 2D angles provide a simplified representation of foot kinematics, it is important to recognize the inherent limitations of this approach, particularly its inability to capture out-of-plane motions such as pronation/supination and eversion/inversion that occur at the midtarsal joints. Such motions could potentially affect the accuracy of MLA angle measurements. Caravaggi et al.'s comparative analysis between 3D and 2D measurements of MLA motion reported that the variability of the 3D measurements of MLA motion were smaller than the corresponding 2D measurements during walking and running [32].

The MTP joint angle was defined by vectors connecting the CAL, MET1, and HLX markers (Figure 1B). The MLA angle was defined by vectors connecting the MET1, NAV, and CAL markers (Figure 1B). From these angles, we determined maximum MTP joint angle (MTPmax), and minimum MLA angle (MLAdrop) for the walking task. Further, we calculated the rate of change in MLA angle per degree of MTP joint dorsiflexion as kinematic joint coupling ratio for both tasks: heel lifting and walking. To ensure the reliability of our measurements, our protocol follows established methods in the field [33,34]. Although direct assessment of inter-trial and intersession reliability was not performed in this study, Caravaggi et al. [32] reported inter-trial and intersession reliability values of 1.1° and 5.7°, respectively, for MLAdrop.

To measure the amount of coupling between the MTP and MLA joints during the toe extension phase, a sliding window analysis technique was used. This was based on observing a non-linear relationship between these joints as the MTP joint moves from neutral to peak MTP joint extension, which is consistent with the existing literature [17,18,35]. The methodology used involved examining segments, or 'windows', of 50 consecutive data points. Within each window, the correlation coefficient was calculated to assess linearity and identify the segment that displayed the most linear relationship between the movements of the toe and the arch. The linear slope of the 50 data points was then calculated within this identified window (kinematic joint coupling ratio = ΔMLA angle/ΔMTP angle) (Figure 1). This slope served as a metric to quantify the kinematic coupling between the MTP and MLA joints.

2.3. Statistical Analysis

Data were first summarized using descriptive statistics, including means and standard deviations between the obese and normal-weight individuals (OW versus NW). Before making group comparisons, normality of the distributions within each group was assessed using the Shapiro–Wilk test. The homogeneity of variances between groups was assessed using Levene's test. The choice of statistical tests to compare the groups was guided by the results of the normality and homogeneity of variances tests: If the data met the assumptions of normality and homogeneity of variances, a standard independent samples *t*-test was used. If the assumption of homogeneity of variances was violated, Welch's *t*-test, which does not require equal variances between groups, was used. All tests were two-tailed, and results were considered statistically significant at the 0.05 level.

3. Results

Descriptive statistics for heel lifting and walking are shown in Table 2, partially showing differences between OW and NW. During data collection and analysis of the kinematic measurements, we encountered instances of missing data attributed to various participant-specific circumstances. Specifically, three participants were unable to achieve MTP joint dorsiflexion greater than 10° during the heel-raising task. For another four participants, fewer than three valid walking trials were recorded. Additionally, one participant chose to discontinue the session before the kinematic measurements could be taken.

Table 2. Summary of kinematic measurements of normal and overweight subjects (NW and OW, respectively), including mean ± standard deviations and measured range of data.

	NW	OW	*p*-Values
Heel-raising task while sitting			
n (f/m)	10 (6/4)	13 (7/6)	
Joint coupling ratio	0.66 ± 0.26 (0.20–0.96]	0.64 ± 0.36 (0.19–1.60)	0.87
Walking talk			
n (f/m)	10 (6/4)	12 (6/6)	
Joint coupling ratio	0.48 ± 0.14 (0.22–0.64)	0.46 ± 0.15 (0.32–0.87)	0.75
MTP_max (°)	31.03 ± 5.80 (18.20–36.73)	37.52 ± 8.40 (21.05–49.84)	0.05
MLA drop (°)	−8.67 ± 2.26 (−4.94–−10.81)	−10.39 ± 2.50 (−6.71–−15.53)	0.11

No significant difference between groups was found for the kinematic joint coupling ratio, either for heel raising (p = 0.87, standard independent samples *t*-test) or for walking

(p = 0.75, Welch's t-test). Furthermore, for the walking task, the OW and NW groups were not significantly different for maximum MLA arch compression (MLAdrop: p = 0.11, standard independent samples t-test), although the direction of the means suggested a 19.84% greater arch drop in OW compared to NW (Figure 2B,D). A significant difference was observed for maximum MTP joint dorsiflexion (MTPmin), with obese individuals showing approximately 20.92% greater MTP joint dorsiflexion at the end of stance phase (Figure 2A,C) (p = 0.05, standard t-test). The comparison of stance time showed no statistically significant differences (p = 0.17, standard t-test), with a duration of 0.75 ± 0.06 s in NW compared to 0.80 ± 0.09 s in OW, indicating similar estimates of walking speed between the two groups.

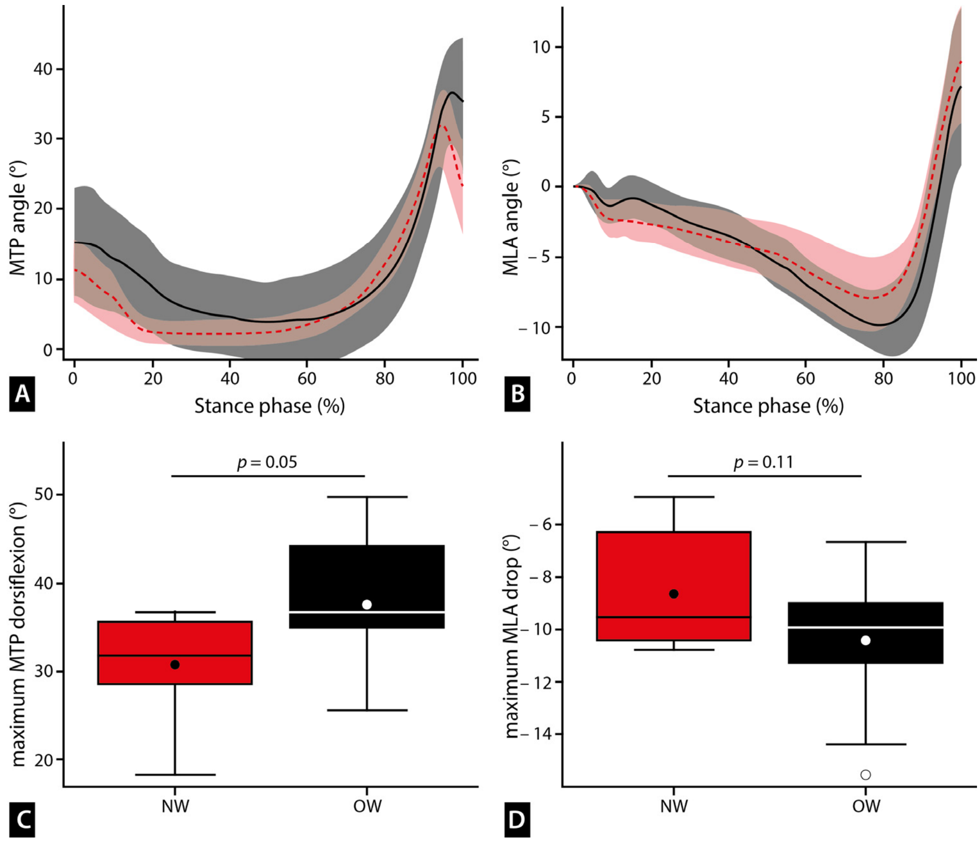

Figure 2. The panels provide a comparison of MTP joint motion and MLA angle in normal-weight (NW, red lines and bars) and obese (OW, black lines and bars) subjects during walking. Panels (**A**,**C**) show the MTP joint motions observed in each group, while panels (**B**,**D**) focus on the variations in MLA angles. To differentiate between the two groups of subjects, red dashed lines are used for NW subjects and black solid lines are used for OW subjects. The shaded bands indicate the standard deviations. For box plots, black and white dots added to boxes represent means. The results of the statistical comparison between groups (p-values) are included in panels (**C**,**D**).

4. Discussion

In recent years, a growing body of research has highlighted the relationship between obesity and declining foot health, with obese individuals consistently reporting higher incidences of foot pain and functional limitations [3,4]. The biomechanical factors underlying these observations have largely been anchored in kinetic observations (e.g., [9–11]), leaving gaps in our understanding of changes in foot kinematics in this population. In particular, the literature has been sparse in examining foot kinematics in the sagittal plane, encompassing MTP and MLA dynamics. Data clarifying the relationship between MTP joint dorsiflexion and arch dynamics in obese individuals are largely lacking. To address these missing data, our study sought to assess the kinematic joint coupling between the

MTP joint and the MLA in an obese cohort. Our main findings were that obese participants exhibited greater MTP joint dorsiflexion towards the end of the push-off phase compared to their normal-weight controls. In addition, pronounced MLA compression during the stance phase was observed in the obese group. Contrary to expectations, no differences were found in the kinematic joint coupling index.

For the static measurements, our study found an average AHI of 0.38 in the normal-weight group, consistent with the findings of Pohl and Farr [36]. Using the criteria of Holowka et al. [33], where an AHI below 0.297 indicates a low arch, none of our participants, including those in the obese group (average AHI of 0.41), fell into this 'low arch' category. This is consistent with previous research that has not found an association between a person's BMI and their arch height [11,37–39]. Interestingly, the data from our obese participants pointed to the presence of pes cavus, or a higher arch. This observation mirrors the work of Wozniacka et al. [40] who found similar foot structures in obese individuals. Zhao et al. [41] have suggested that a greater arch height in obese individuals may be due to increased fat padding on the sole of the foot, giving the impression of a higher arch. Unfortunately, we were not able to distinguish between the soft tissues of the foot and its bones (e.g., by using X-ray or ultrasound), suggesting the need for future research to take foot fat into account when assessing foot posture in obese individuals.

While obese and normal-weight individuals had comparable static foot postures, our kinematic analysis revealed differences between the two groups in terms of maximum arch compression and MTP joint dorsiflexion during the second half of the stance phase. In general, both populations exhibited longitudinal arch compression even with initial MTP joint dorsiflexion, a phenomenon termed 'inhibited windlass' by Welte et al. [18], indicating plantar aponeurosis elongation. Following an initial 5–10° of MTP dorsiflexion, there was dorsiflexion at the MTP joint and simultaneous arch rising (termed 'pure forward-windlass' by Welte et al. [18]). This showed a robust relationship up to the point of maximal MTP joint dorsiflexion. Peak MLA compression typically manifested around the 80% mark of the stance phase, echoing findings from previous studies such as Sichting and Ebrecht [17] and Welte et al. [18] (Figure 2B). It is during this phase that the highest arch compression force [17] and plantar aponeurosis stress [13] have been documented. Despite the similarities in the windlass behavior during gait (inhibited windlass followed by pure forward-windlass), obese participants had considerably greater maximum MLA compression, approximately 20% more. The increased arch compression observed in obese individuals may reflect kinetic studies that have documented increased plantar pressures, particularly under the metatarsal heads and midfoot region, during the final segment of the stance phase. Determining the exact reasons for this increased compression in obese individuals is challenging given our methodology, but it may be due to the foot muscles, both intrinsic and extrinsic, struggling to resist the increased arch compression forces due to the increased body mass in obese individuals [7,42,43]. Consequently, it is plausible that the strain on the plantar aponeurosis is more pronounced in obese individuals, adding weight to the observations that they are predisposed to conditions such as plantar fasciitis and heel pain, as reported by Irving et al. [44]. In light of these observations, there is a clear need for future research to further investigate the greater arch compression exhibited by obese individuals at the end of the stance phase, particularly its effect on the stress to the plantar aponeurosis.

While the differences in arch dynamics are certainly noteworthy, our findings also reveal another notable variation: the obese individuals demonstrated a 20.92% increase in total MTP dorsiflexion during the push-off phase. This increased MTP dorsiflexion is well in line with kinetic studies highlighting increased plantar pressures under the metatarsal heads towards the end of the stance phase [6]. Similar to our assumptions regarding increased MLA compression in the obese group, this pronounced MTP dorsiflexion may be due to both intrinsic and extrinsic foot muscles struggling with the task of stabilizing the MTP joint during push-off. In support of this, recent studies by Farris et al. [21,22] have shown that passive dorsiflexion of the MTP joint activates intrinsic foot muscles, potentially helping to stiffen the MTP joints during locomotion. However, more than just muscles are involved. The soft tissues surrounding the MTP joint, which include the joint capsules and the plantar aponeurosis, can also influence total dorsiflexion during push-off [45]. Unfortunately, our study lacks specific data on these tissue properties in our participants. Thus, while we acknowledge the potential impact of increased body mass on these soft tissue properties, our explanations for increased MTP joint dorsiflexion at the end of stance phase in obese individuals remain speculative.

In terms of the kinematic joint coupling ratio between normal-weight and obese individuals, our results showed an unexpected similarity in the kinematic joint coupling ratio between normal-weight and obese individuals. The mechanism underlying this similarity in joint coupling ratio, whether it be plantar aponeurosis loading, foot muscle activity, or a combination of both, remains puzzling. The similar rise of the arch during the pure forward-windlass phase in the obese group may be due to altered mechanical properties of the plantar aponeurosis, as indicated by Tas et al. [46], or even increased plantar aponeurosis stress and muscle activity. Although these theories tread on speculative ground, especially given our inability to validate such a hypothesis, it does point to the potential for increased stress on plantar structures, including muscles and connective tissues, in obese individuals. This, in turn, could potentially increase their susceptibility to foot pain and functional problems. Clearly, this is an area that warrants further research.

In this study, we encountered several limitations that warrant discussion. One of the primary challenges was recruiting subjects, resulting in a relatively small sample size. This limitation was due to poor compliance. Another limitation of this study is a mismatch in the age distribution between our obese and normal-weight groups. This raises the possibility of an effect of age on MTP and MLA motion. However, Allan et al. [47] examined the relationship between the maximum dorsiflexion of the first metatarsophalangeal joint (MTP) and age and found no significant relationship. This suggests that the age discrepancy in our study groups may not invalidate our findings. While acknowledging these limitations, our results provide novel insights into the biomechanics of the foot in obese individuals and emphasize the value of further research with larger, more demographically balanced cohorts to validate and extend our findings.

A further limitation of our study is that participants walked at a self-selected pace. To address this, we used stance time as an indirect indicator of walking speed, recognizing that walking speed can influence foot kinematics, including the movement of the MTP joint and the medial longitudinal arch. Our findings revealed that the obese group tended to show longer stance time compared to the normal-weight group, suggesting a slower walking pace among the obese participants. Although the difference was not statistically significant, it is consistent with the literature, which reported that individuals with obesity tend to walk at slower speeds, e.g., [48,49]. Our analysis involved individuals who were obese, with a BMI exceeding 35 kg/m^2, thereby increasing the probability of lower walking speeds in this category. However, the tendency towards slower walking speed in the obese group might also be explained by the age difference mentioned earlier. Walking speed tends to decrease with increasing age [29]. Considering the potential impact of walking speed, which may be mediated by factors such as body weight and age, on foot kinematics, a reduction in walking speed might typically be expected to result in decreased MTP joint dorsiflexion and MLA compression [50]. However, in contrast to these expectations, our study found that obese participants showed a tendency towards increased MTP dorsiflexion and MLA compression at the end of the stance phase. To yield more conclusive results, controlling for walking speed may be necessary and should be considered in future investigations.

Finally, our reliance on established, albeit likely less accurate, practices in 2D foot kinematic studies was a pragmatic decision based on the scope and resources of our study. Future investigations could certainly benefit from a more detailed investigation and documentation of camera calibration processes and their impact on the accuracy and reliability of 2D foot kinematic data and contribute valuable insights to the field.

Although this study involved a relatively small sample of participants, it provides valuable insight into the biomechanical behavior of the MTP joint and MLA during gait in both obese and normal-weight individuals. Specifically, we observed that despite the lack of significant differences in the kinematic joint coupling ratio between obese and normal-weight participants, there was a tendency for obese individuals to exhibit greater arch compression and MTP joint dorsiflexion during gait. This observation suggests that the biomechanical interaction between the MTP joint and the MLA during gait is consistent across individuals of varying body mass. However, the observed trends of increased arch compression and MTP joint dorsiflexion in obese individuals suggest the potential for greater mechanical stress on plantar foot structures, including the plantar aponeurosis and foot muscles.

It is important to emphasize that while these findings suggest biomechanical variations that may merit further investigation, they do not establish a direct causal relationship to foot health problems in obese individuals. The complexity of foot biomechanics and the multifactorial nature of foot health require a cautious interpretation of these findings. With the global increase in obesity, understanding its musculoskeletal implications remains a priority. This study contributes to the body of knowledge by highlighting biomechanical patterns that could inform future research aimed at elucidating the relationship between obesity, foot biomechanics, and health outcomes. Such research is essential to identify causative biomechanical risk factors and develop targeted preventive and therapeutic strategies to address the increased incidence of foot pain and functional limitations observed in obese populations.

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Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: The data presented in this study are available on request from the corresponding author.

Conflicts of Interest: The authors declare no conflicts of interest. The funders had no role in the design of the study; in the collection, analyses, or interpretation of data; in the writing of the manuscript; or in the decision to publish the results.

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