Characterizing the Effect of Lower Limb Dominance on Foot Kinematics During Walking

By

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Chapter 1

INTRODUCTION

1.1 Classification of Foot Dominance

Similar to handedness, individuals also present with footedness. The terms "footedness," "foot dominance," and "lower limb dominance" are used interchangeably to describe the stratification of the lower limbs according to their primary functional roles in bilateral tasks. The dominant limb is identified as that which is preferred for performing a mobilizing or manipulating behavior during a bilateral task, while the non-dominant limb is identified as providing support and/or stability [1]. For example, when kicking a ball, the dominant limb strikes the object, while the non-dominant limb offers support [2]. Footedness presents on a graduated scale, with some individuals exhibiting strong right- or left-footedness, some exhibiting weak footedness, and others presenting as ambipedal (no clear dominance) [3]. Moreover, footedness does not always correspond to handedness, in that an individual may be right-handed but left-footed [4]. The effects of foot dominance are most evident in bilateral tasks during which the limbs are performing drastically different roles (e.g., kicking a soccer ball, lunging, baseball batting etc.); however, there is evidence to suggest that limb dominance is also present in more symmetric bilateral tasks, such as cycling [5]–[7].

1.2 Standard Practices in Human Gait Analyses

Despite the different functional roles demonstrated by the lower limbs in bilateral tasks, biomechanical analyses of human locomotion do not typically consider the effects of lower limb dominance. It is common practice in studies of able-bodied gait to use the data collected from one limb (e.g., data from only right legs) to represent the function of both limbs [1], [8]–[11], or to pool data from both limbs to create a representative average [12]. Data from groups of unilateral prosthesis users [13], [14], hemiparetic stroke patients [15], and others with unilateral limb differences (e.g., anterior cruciate ligament rupture, plantar fasciitis, etc.) [16], [17] are similarly pooled into “affected” and “unaffected” limb categories, with lower limb dominance remaining unaccounted for. Although these practices simplify data collection and analysis, they rely on the assumptions that able-bodied gait is symmetric, that the cause of any asymmetries demonstrated in pathological gait are purely a result of the clinical impairment, and that any effects of limb dominance are negligible and will not confound data interpretation.

1.3 Able-Bodied Gait May Not Be Symmetric

Contrary to the assumptions present in many gait analyses, there is evidence which suggests that human locomotion does demonstrate intrinsic asymmetry, and that it may be related to the effects of lower limb dominance. Upon examining the shoe soles of healthy, young adults (students), [18] observed asymmetric wear patterns, indicating asymmetric use of the limbs in activities of daily living. More recently, [19] evaluated hip, knee, and ankle kinematics during overground walking, and found significant asymmetry of ankle sagittal plane motion throughout the gait cycle. Studies have also identified between-limb differences in EMG activity [20] and ground reaction forces during walking [21], [22]. However, there has been little previous exploration of between-limb differences in foot and toe dynamics, and we are aware of no studies that consider the effects of foot dominance.

When evaluating handedness, functional differences in the upper extremities are more pronounced at the distal joints [23], indicating that the effects of footedness may likewise be most apparent at the distal joints of the lower limb (i.e., the foot joints). This hypothesis of exaggerated asymmetry at the distal joints is also supported by the findings in [19], in that the only joint (among the hip, knee, and ankle) that demonstrated significant asymmetry was the ankle. Therefore, the objective of this work was to characterize the effect of lower limb dominance on foot kinematics during human locomotion. Here, we specifically focus on metatarsophalangeal joint (MPJ) and medial longitudinal arch (MLA) dynamics during walking.

1.4 Metatarsophalangeal Joint Dynamics

Motion of the MPJ has been identified to be important in both clinical [17], [24]–[26] and sporting [27]–[29] contexts. Furthermore, restriction of MPJ motion during walking in healthy individuals has been shown to result in compensatory mechanisms such as increased ankle, knee, and hip joint moments [30]. Subjective feedback indicated that adding a toe joint to a prosthetic foot was predominantly favored by a sample of able-bodied users wearing prosthetic adapters [31], was reported to be more comfortable in a case study with a unilateral transtibial prosthesis user [32], and was preferred in a larger sample of unilateral transtibial prosthesis users only by those who identified as having lost their dominant limb [33].

Given the importance of the MPJ, we sought to investigate the effect of lower limb dominance on MPJ kinematics during locomotion. Based on the asymmetric shoe wear patterns reported in [18], and the asymmetry in ankle kinematics reported by [19], we hypothesized that between-limb MPJ kinematic analysis would uncover small but meaningful differences (>4° as relevant in clinical and sporting contexts [17], [25], [27]), and that these would be associated with limb dominance. The preferences noted in [33] suggest that the dominant-side MPJ may adopt a more dynamic role than the non-dominant MPJ during locomotion. Therefore, we hypothesized that the dominant-side MPJ would exhibit a greater range of motion than the non-dominant MPJ, reflecting its role of manipulation and mobilization during bilateral tasks [1].

1.5 Medial Longitudinal Arch Dynamics

The medial longitudinal arch is comprised of several bones, and its motion is modulated by intrinsic foot musculature and ligamentous tissues, including the plantar fascia which runs along the bottom of the foot, connecting the calcaneus to the metatarsal heads and phalanges. Through the actions of both the arch-spring and windlass mechanisms, the MLA is thought to stabilize and support the foot during push-off. It achieves this by performing as a rigid lever, whereby the plantar fascia is tensioned and the bones of the arch are in a tightly-packed arrangement [34]. The arch-spring mechanism describes the motion of the MLA and the plantar fascia during vertical loading and unloading of the foot. When the foot is loaded (e.g., during stance), the arch flattens and horizontally elongates, thus stretching the plantar fascia. When the foot is offloaded, the MLA returns to its original conformation, and the plantar fascial spring recoils [35].

The windlass mechanism describes the response of the MLA and the plantar fascia to metatarsal dorsiflexion: the plantar fascia wraps around the metatarsal head and draws the calcaneus closer to the metatarsals. This results in the plantar fascia becoming tensioned and the arch subsequently becoming taller and horizontally compressed [36]. Bojsen-Møller (1979) describes two classes of the windlass mechanism: high- and low-gear windlass. In high-gear windlass, the plantar fascia wraps around the head of the first metatarsal. Due to the large radius of the first metatarsal head, high-gear windlass is associated with significant plantar fascial tensioning [34]. Without the high-gear windlass mechanism, decreased vertical loading on the MLA during push-off would result in plantar fascial recoil (due to the arch-spring mechanism). However, activation of high-gear windlass maintains tension in the plantar fascia, resulting in a slower recoil and prolonged support [37]. Furthermore, high-gear windlass is often associated with foot pronation and tight packing of the bones of the arch. This combination of the tensioned plantar fascia and the tightly packed bone structure results in a more rigid and supported foot. Low-gear windlass, on the other hand, occurs when the foot is more supinated and rolls over metatarsals three to five. Since these metatarsal heads are much smaller than that of the first metatarsal, the plantar fascia is less tightly wound and therefore remains slack [34]. Additionally, the bones of the arch are more loosely packed in this position, leading to less stability and increased foot mobility [38]. The mechanism described by [36] more closely reflects high-gear windlass; therefore the term "windlass" will henceforth refer to high-gear windlass, unless otherwise specified.

Given its role in support and stabilization, we hypothesized that the non-dominant foot would demonstrate high-gear windlass to provide support during push-off, and would therefore exhibit greater MLA range of motion throughout a stride. Furthermore, we hypothesized that the non-dominant foot would invoke the windlass mechanism for a longer time period. This would likely be evidenced by an increased time duration between heel-rise and toe-off, facilitating controlled recoil of the plantar fascia, and thus delivering maximal support throughout push-off.

Chapter 2

METHODS

2.1 Participant Recruitment

Five healthy adults (4 female, 1 male) were recruited for this study. Participant anthropometric characteristics are shown in Table 1. All participants provided informed consent, and the study was approved by Vanderbilt University's Human Subjects Review Board (IRB 141697) prior to data collection. Participants were required to be free from recent bone fractures and acute injuries, and have no known locomotor or neuromotor disorders or conditions that would affect their ability to safely complete the protocol.

Table 1. Summary of participant anthropometric characteristics.

Sex	4 Female; 1 Male
Age (years)	22.0 ± 0.65
Height (m)	1.68 ± 0.03
Mass (kg)	63.5 ± 4.15
Dominant Leg Length (m)	0.87 ± 0.038
Non-Dominant Leg Length (m)	0.87 ± 0.041
Dominant Foot Length (m)	0.24 ± 0.011
Non-Dominant Foot Length (m)	0.241 ± 0.010

2.2 Data Collection

A total of 52 retro-reflective markers were placed bilaterally on the lower body and feet. The lower body marker set included the pelvis (6), thighs (8), femoral epicondyles (4), shanks (8), and malleoli (4). The foot markers were placed in accordance with [39] and included the calcanei

(3 per foot; 6 total), and one on each of the first, second and fifth metatarsal heads and bases (6 per foot; 12 total), the talonavicular joint (1 per foot; 2 total), and the hallux (1 per foot; 2 total) (Figure 1).

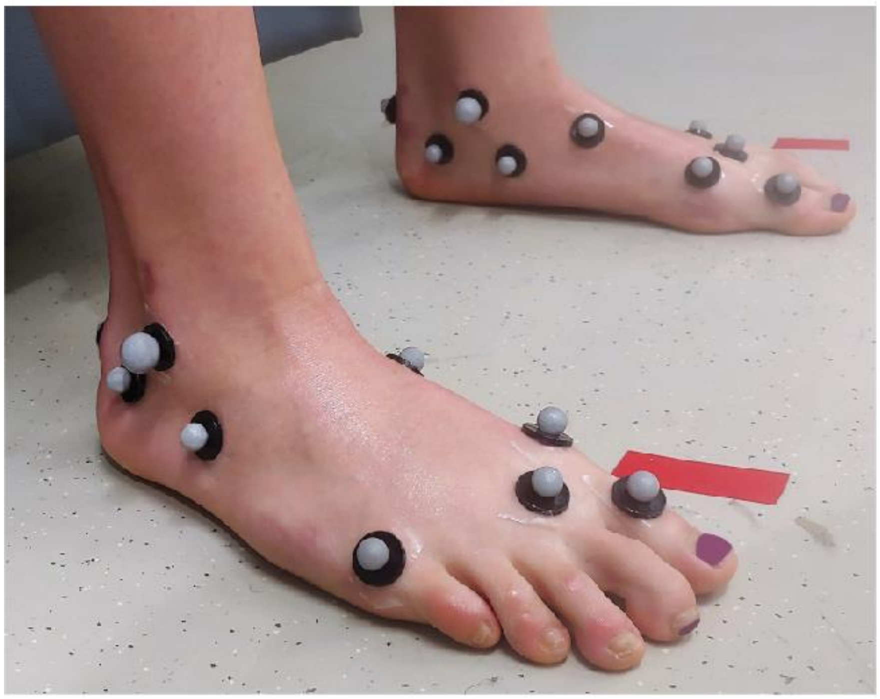

Figure 1. Placement of foot markers.

A trial of the participant in the anatomical position (quiet standing) was captured. This was followed by functional hip, knee, and ankle calibration trials, which were collected in accordance with [40]. An additional "unweighted" normalization trial was also collected for each foot in accordance with [41], which required the participant to offload the foot of interest (<20 N permitted) and gently rest it on the level treadmill surface in a near-perfect 0° ankle angle, as visually determined by the investigator. All participants then completed six randomized barefoot walking and running trials at speeds of 0.8-4.0 ms^{-1}. Synchronized ground reaction forces (1000

Hz; split-belt instrumented treadmill, Bertec, Columbus, USA) and three-dimensional motion capture (200 Hz; 10-camera system, Vicon, Oxford, UK) were recorded for all trials.

Following the completion of their walking and running trials, footedness was determined for each participant. Two measures of footedness were taken; first, a binary measurement determined foot dominance based on the answer to the question, "Which foot would you use to kick a stationary ball at a target straight in front of you?" As recommended by [2], the dominant limb was taken to be that which the participant indicated would perform the kicking action. A second measure of footedness was taken via the revised Waterloo Footedness Questionnaire (WFQ-R) [42] in order to ascertain the relative strength of foot dominance. The WFQ-R asks the participant to rate foot preference for a total of 10 tasks—five of which concern object manipulation (e.g., picking up a marble), and five of which target support (e.g., balancing on one foot). Strength of foot dominance, as measured by the WFQ-R, is reported on a scale from (-20) to (+20), where a negative score indicates left dominance and a positive score indicates right dominance. The binary metric was used for primary analysis, but the WFQ-R was used to more closely evaluate inter-participant variance. Both the binary test and the WFQ-R were administered following the completion of the walking and running trials to ensure that the captured kinematic and ground reaction force data were reflective of the participant's natural gait, and not affected by their perceptions of, or responses to, the questionnaire.

2.3 Data Processing

Marker trajectories and ground reaction force data were low-pass filtered at 8 Hz and 15 Hz, respectively, using a fourth order Butterworth filter. Joint angles were computed in Visual 3D (C-motion, Germantown, MD, USA), with foot joint angles calculated in accordance with [39].

The MPJ angle (named F2Ps in [39]) was calculated as the sagittal plane projection of the angle created by the markers on the hallux, the first metatarsal head, and the first metatarsal base, with a positive angle indicating dorsiflexion (Figure 2). The MLA angle was calculated as the sagittal plane projection of the angle created by the markers on the first metatarsal head, the sustentaculum tali, and the posterior calcaneus, with a greater positive angle indicating flattening and vertical elongation of the arch (Figure 2).

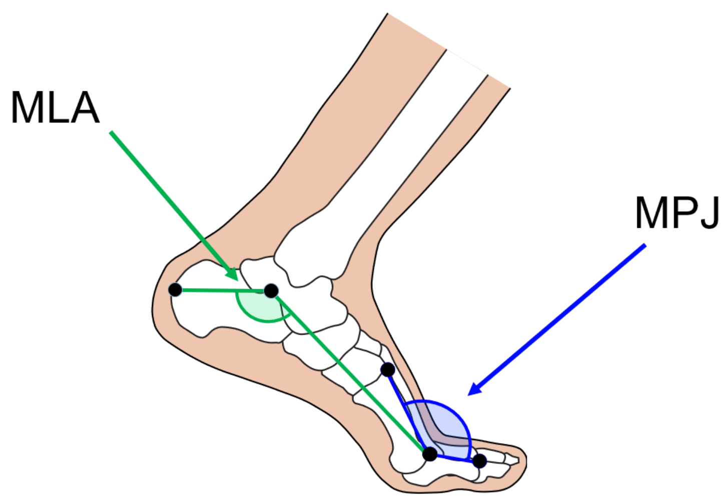

Figure 2. Determination of the foot joint angles.

A non-normalized dataset of the foot joint angles was outputted from Visual 3D. Two additional datasets of the foot joint angles were then generated in post-processing by a custom-built MATLAB (MathWorks, Natick, USA) function that subtracted reference angle values from the non-normalized walking and running time-series data (Figure 3). One dataset was normalized to the foot joint angles captured during quiet stance, and the other dataset was normalized to those captured during the unweighted normalization trials (Figure 4). The motivation for applying three unique normalization methods was due to the lack of apparent consensus across the literature. Our goal was to assess the time-series data produced via each method to ensure an objective evaluation of the peak angles.

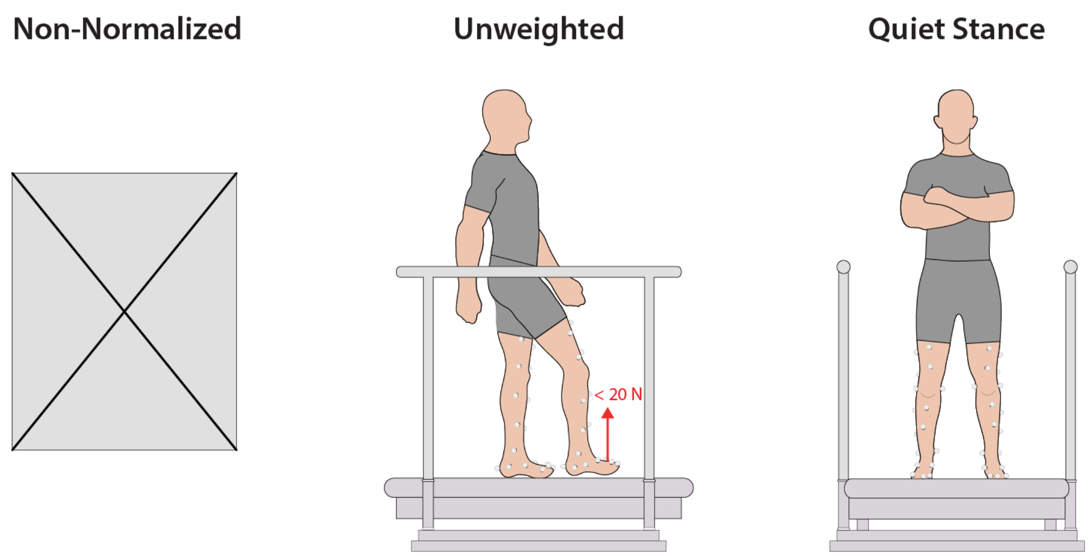

Figure 3. Reference angles used to normalize the foot joint angles were captured in three conditions.

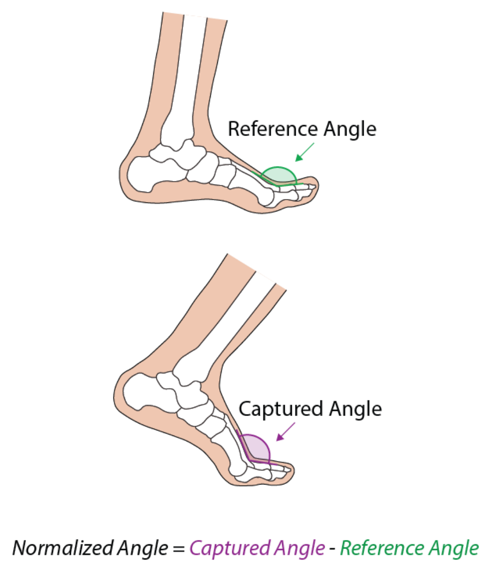

Figure 4. Calculation of normalized foot joint angles.

Spatiotemporal variables (stride length; stance and swing time of each limb) were computed using the ground reaction force and foot marker data. These parameters were assessed to determine if any notable differences were present that may affect the interpretation of the kinematic data. Kinematic data were divided into individual strides based on the vertical ground reaction forces. Range of motion for the MPJ and MLA were calculated as the difference between the absolute maximum and absolute minimum values of the respective joint angle throughout a stride. MPJ peak dorsiflexion angle was calculated as the absolute maximum MPJ angle throughout a stride. MLA peak recoil angle (i.e., the most acute MLA angle, during which the arch is vertically heightened and horizontally compressed) was calculated as the absolute minimum MLA angle throughout a stride. MPJ peak dorsiflexion angle and MLA peak recoil angle were each evaluated using the non-normalized dataset, as well as the two additional normalized datasets. Time between heel-rise and toe-off was calculated as the time period beginning with a 50% increase in posterior calcaneus marker vertical position relative to its position during mid-stance (calculated as the average position from 10-30% gait cycle [43]) and ending with toe-off.

2.4 Statistical Analysis

Due to the limited sample size and the gradated strength of footedness, we approached statistical analysis mainly at the individual participant level. However, descriptive statistics such as the mean, standard deviation, and range are included in the Results. This choice was further supported by findings from [20], [44]–[46], in which evaluating averaged group data occluded significant individual-level trends.

Chapter 3

RESULTS

3.1 Footedness Scores

All participants were identified as having right foot dominance, with scores ranging between +6 to +10, with an average score of +7.4/20 (Table 2).

Table 2. Footedness scores as determined by the binary test and the WFQ-R.

	Dominant Foot (binary test [2]**)**	**Dominance Score (/20; WFQ-R** [42]**)**	**Strength of Dominance (as interpreted by** [47]**;** [48]**)**
Participant 1	Right	+10	Weak Right; Right
Participant 2	Right	+6	Weak Right; Mixed
Participant 3	Right	+6	Weak Right; Mixed
Participant 4	Right	+8	Weak Right; Right
Participant 5	Right	+7	Weak Right; Right

3.2 Spatiotemporal Metrics

There were no appreciable differences in stride length between the two limbs across all participants and speeds (Figure 5). Stride length increased with speed; however, the between-limb difference in length was less than 0.4 cm at any given speed (Table 3).

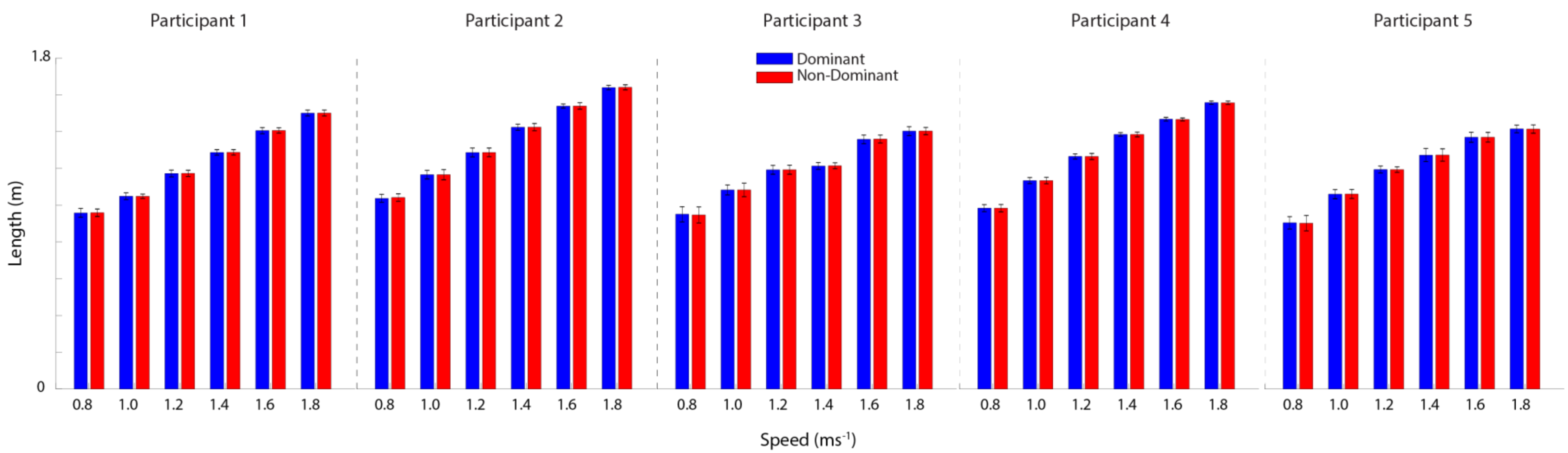

Figure 5. Mean stride length of the dominant (blue) and non-dominant (red) limb for each participant at all speeds. Error bars represent standard deviation.

Table 3. Absolute differences in stride length between the dominant and non-dominant limbs. Average row and column display group mean ± standard deviation.

	Absolute Difference in Stride Length (x10^{-4} m)						
	0.8 ms^{-1}	**1.0 ms^{-1}**	**1.2 ms^{-1}**	**1.4 ms^{-1}**	**1.6 ms^{-1}**	**1.8 ms^{-1}**	**Average**
Participant 1	9.58	4.28	3.82	2.74	7.57	3.92	5.32 ± 2.65
Participant 2	42.91	1.04	2.16	9.79	3.13	11.85	11.81 ± 15.85
Participant 3	36.56	0.638	8.0	7.45	14.61	2.58	11.64 ± 13.14
Participant 4	2.96	0.50	2.44	1.78	12.02	7.51	4.53 ± 4.38
Participant 5	15.48	13.84	1.23	2.21	0.32	9.73	7.14 ± 6.74
Average	21.5 ±17.37	4.06 ± 5.68	3.53 ± 2.67	4.79 ± 3.60	7.53 ± 5.95	7.12 ± 3.88	

Similarly, there were no clear trends regarding differences in stance/swing time ratio between the two limbs (Figure 6). One trial (Participant 3; 1.4 ms^{-1}) showed a distinct difference in stance/swing time ratio (Figure 6), but for all other trials (across participants and speeds), between-limb differences in both stance and swing times were less than 0.02 seconds (Tables 4 & 5), with no clear directionality trend (Figure 6).

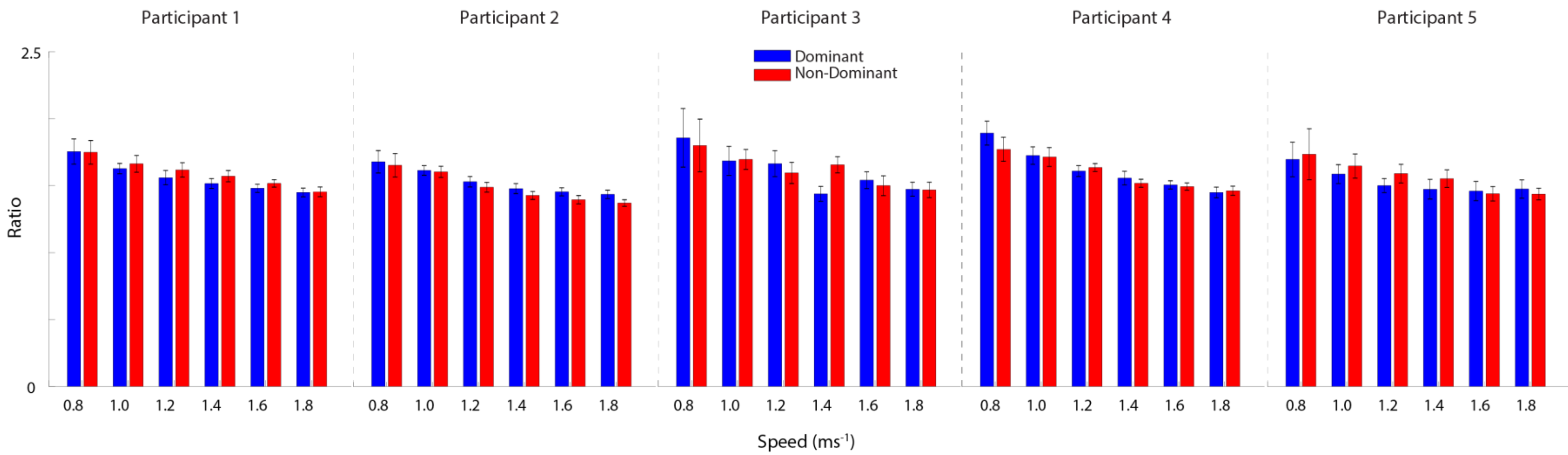

Figure 6. Mean stance/swing time ratio of the dominant (blue) and non-dominant (red) limb for each participant at all speeds. Error bars represent standard deviation.

Table 4. Absolute differences in stance time between the dominant and non-dominant limbs. Average row and column display group mean ± standard deviation.

	Absolute Difference in Stance Time (x10^{-3} s)						
	0.8 ms^{-1}	**1.0 ms^{-1}**	**1.2 ms^{-1}**	**1.4 ms^{-1}**	**1.6 ms^{-1}**	**1.8 ms^{-1}**	**Average**
Participant 1	0.27	5.0	8.52	7.86	5.667	0.78	4.8 ± 3.49
Participant 2	3.64	1.818	7.6	8.46	9.815	10.17	6.92 ± 3.42
Participant 3	11.19	1.67	10. 2	29.52	5.469	0.429	9.74 ± 10.62
Participant 4	18.1	1.74	4.2	6.111	1.786	2.1	5.67 ± 6.33
Participant 5	3.7	09.2	13.85	11.9	2.66	4.85	7.69 ± 4.63
Average	7.7 ± 7.21	3.9 ± 6.94	8.87 ± 7.97	12.77 ± 13.52	5.8 ± 15.91	3.67 ± 15.16	

Table 5. Absolute differences in swing time between the dominant and non-dominant limbs. Average row and column display group mean ± standard deviation.

	Absolute Difference in Swing Time (x10-3 s)						
	0.8 ms^{-1}	**1.0 ms^{-1}**	**1.2 ms^{-1}**	**1.4 ms^{-1}**	**1.6 ms^{-1}**	**1.8 ms^{-1}**	**Average**
Participant 1	1.36	5.2	8.33	7.86	5.0	0.47	4.70 ± 3.24
Participant 2	5.70	1.82	06.6	8.85	9.63	10.35	7.16 ± 3.16
Participant 3	8.10	2.08	9.62	29.19	5.63	1.0	9.27 ± 10.31
Participant 4	19.05	1.52	4.0	5.74	1.97	1.77	5.68 ± 6.75
Participant 5	4.57	9.4	13.46	11.38	2.66	4.71	7.70 ± 4.33
Average	7.75 ± 6.76	4.01 ± 6.63	8.40 ± 7.97	12.60 ± 13.25	4.98 ± 15.36	3.66 ± 14.82	

3.3 Metatarsophalangeal Joint Range of Motion

All participants exceeded the predetermined 4° threshold for a meaningful difference in MPJ range of motion between the two limbs (Figure 7), with average differences ranging between 3.4°-8.0° (Table 6). Three of the five participants demonstrated increased MPJ range of motion on their dominant side for all speeds. One participant (Participant 4) exhibited increased MPJ range of motion on their dominant side for all but one speed, and one participant (Participant 3) demonstrated increased MPJ range of motion on their non-dominant side for all speeds (Figure 7).

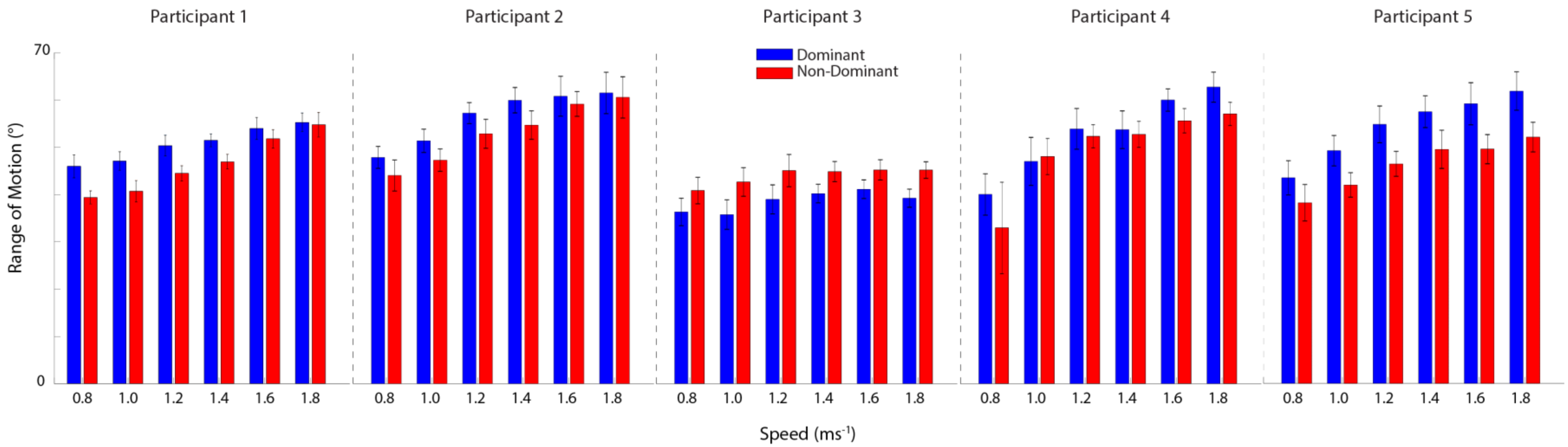

Figure 7. Mean metatarsophalangeal joint (MPJ) range of motion for the dominant (blue) and non-dominant (red) limb for each participant at all speeds. Error bars represent standard deviation.

Table 6. Absolute differences in metatarsophalangeal joint (MPJ) range of motion between the dominant and non-dominant limbs. Average row and column display group mean ± standard deviation.

	Absolute Difference in MPJ Range of Motion (°)						
	0.8 ms^{-1}	**1.0 ms^{-1}**	**1.2 ms^{-1}**	**1.4 ms^{-1}**	**1.6 ms^{-1}**	**1.8 ms^{-1}**	**Average**
Participant 1	6.55	6.36	5.85	4.51	2.19	0.50	4.33 ±2.48
Participant 2	4.19	4.10	4.38	5.25	1.64	0.94	3.412 ±1.71
Participant 3	4.46	6.80	6.01	4.59	4.03	5.88	5.30 ±1.0
Participant 4	7.06	1.03	1.57	1.00	4.38	5.65	3.45 ±2.61
Participant 5	5.27	7.25	8.35	7.95	9.61	9.74	8.03 ±1.66
Average	5.51 ±1.27	5.11 ±2.59	5.23 ±2.50	4.66 ±2.48	4.37 ±3.15	4.54 ±3.85	

3.4 Medial Longitudinal Arch Range of Motion

Between-limb differences in MLA range of motion (Figure 8) were less pronounced than the MPJ. However, small but consistent differences (average magnitudes ranging from 0.91°-6.9°) were present for all participants (Table 7). Three of the five participants demonstrated increased MLA range of motion on their non-dominant side for all speeds. One participant (Participant 1) exhibited increased MLA range of motion on their non-dominant side for all but two speeds, and one participant (Participant 2) demonstrated increased MLA range of motion on their dominant side for all speeds (Figure 8).

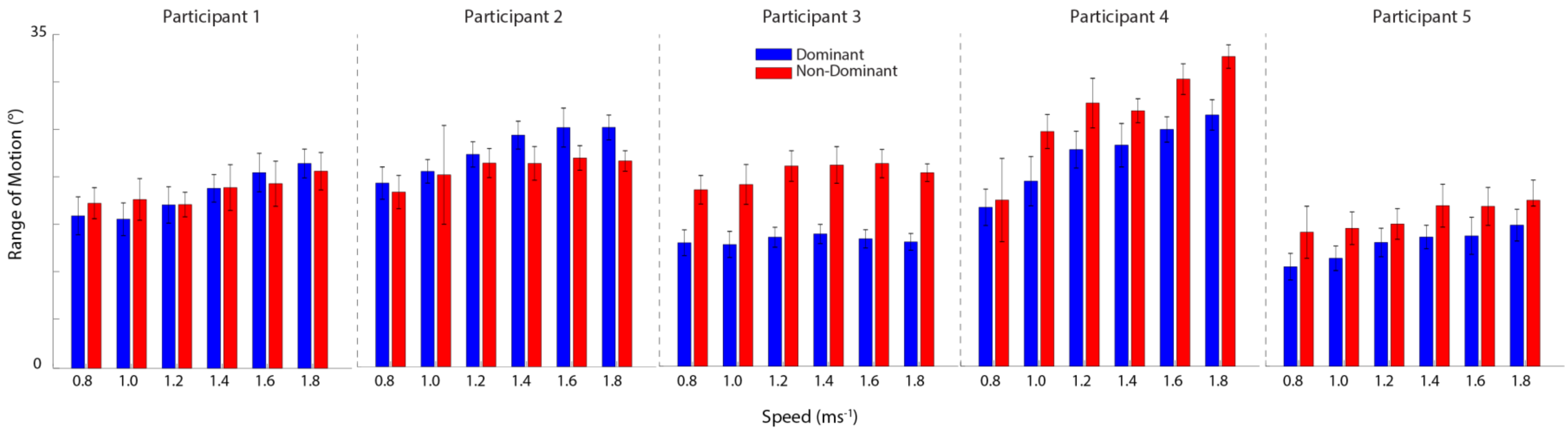

Figure 8. Mean medial longitudinal arch (MLA) range of motion for the dominant (blue) and non-dominant (red) limb for each participant at all speeds. Error bars represent standard deviation.

Table 7. Absolute differences in medial longitudinal arch (MLA) range of motion between the dominant and non-dominant limbs. Average row and column display group mean ± standard deviation.

	Absolute Difference in MLA Range of Motion (°)						
	0.8 ms^{-1}	**1.0 ms^{-1}**	**1.2 ms^{-1}**	**1.4 ms^{-1}**	**1.6 ms^{-1}**	**1.8 ms^{-1}**	**Average**
Participant 1	1.32	2.07	0.04	0.08	1.17	0.81	0.91 ± 0.48
Participant 2	0.98	0.37	0.91	2.98	3.20	3.53	1.99 ±1.39
Participant 3	5.55	6.28	7.46	7.23	7.88	7.25	6.94 ±0.86
Participant 4	0.77	5.24	4.90	3.64	5.30	6.16	4.34 ±1.93
Participant 5	3.58	3.12	1.93	3.26	3.06	2.59	2.92 ±0.58
Average	2.44 ±2.07	3.42 ±2.38	3.05 ±3.08	3.44 ±2.55	4.12 ±2.56	4.07 ±2.63	

3.6 Heel-Rise to Toe-Off Time

All participants demonstrated a greater time between heel-rise and toe-off on the non-dominant side (Figure 9). Average differences ranged between 56.6 ms-116.4 ms, which reflects a 28%-50% change in duration (Table 8). There was one trial where this difference was not noteworthy (Participant 4; 0.8 ms^{-1}).

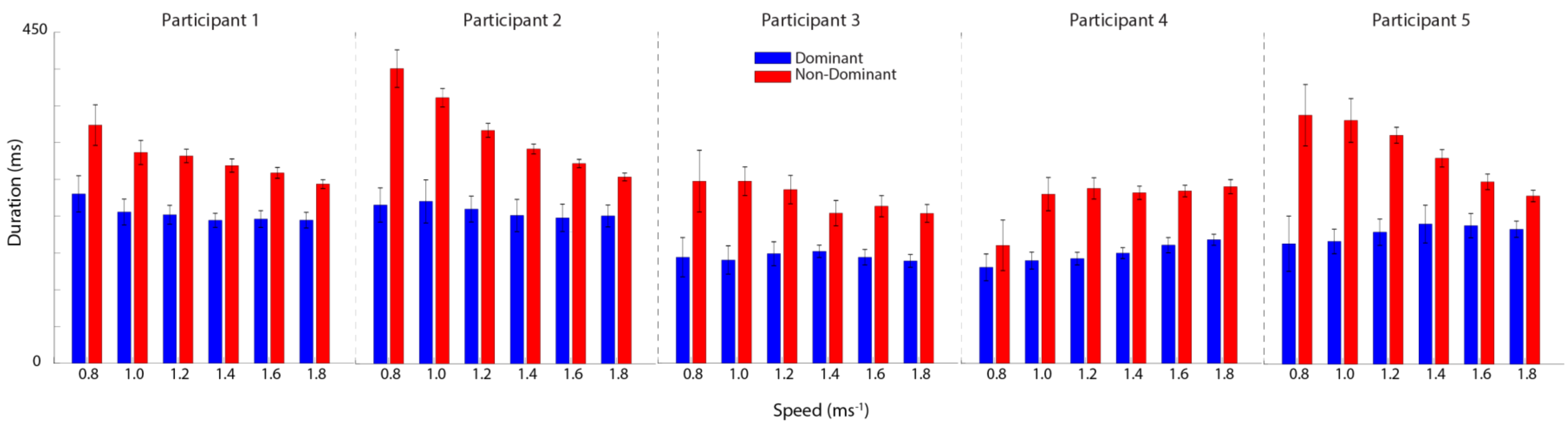

Figure 9. Mean time between heel-rise and toe-off for the dominant (blue) and non-dominant (red) limb for each participant at all speeds. Error bars represent standard deviation.

Table 8. Time between heel-rise and toe-off for the dominant and non-dominant limb for each participant. Average row and column display group mean ± standard deviation.

	Percent Difference in Time Between Heel-Rise to Toe-Off						
	0.8 ms^{-1}	**1.0 ms^{-1}**	**1.2 ms^{-1}**	**1.4 ms^{-1}**	**1.6 ms^{-1}**	**1.8 ms^{-1}**	**Average**
Participant 1	33.8	32.9	33.1	32.1	27.6	22.4	30.3 ±4.5
Participant 2	60.1	48.4	40.6	36.7	31.4	23.3	40.1 ±13.0
Participant 3	52.8	55.3	45.3	29.1	38.8	37.7	43.2 ±9.9
Participant 4	20.6	48.8	50.23	42.9	37.3	35.4	39.2 ±10.9
Participant 5	69.6	66.0	53.8	38.1	27.3	21.9	46.1 ±20.1
Average	47.44 ±19.9	50.3 ±12.1	44. 6 ±8.1	35.8 ±5.3	32.5 ±5.4	28.1 ±7.7	

3.5 Effect of Normalization Method on Peak Angles

The magnitude and direction of the between-limb differences for peak angles were heavily influenced by the chosen normalization method. The effects of normalization method on the peak MPJ dorsiflexion angle and peak MLA recoil angle at a moderate speed (1.2 ms^{-1}) are shown for Participant 2 (Figure 10). The range of reported angle differences, as well as the maximum effect change due to normalization method, for all participants at 1.2 ms^{-1}, are shown for peak MPJ dorsiflexion angle (Table 9) and peak MLA recoil angle (Table 10). For multiple participants, we see the between-limb differences shift from the non-dominant side demonstrating a peak angle with a greater magnitude (negative values in Tables 9 & 10; second column) to the dominant side demonstrating a greater magnitude (positive values in Tables 9 & 10; second column), solely based on the change in chosen normalization method.

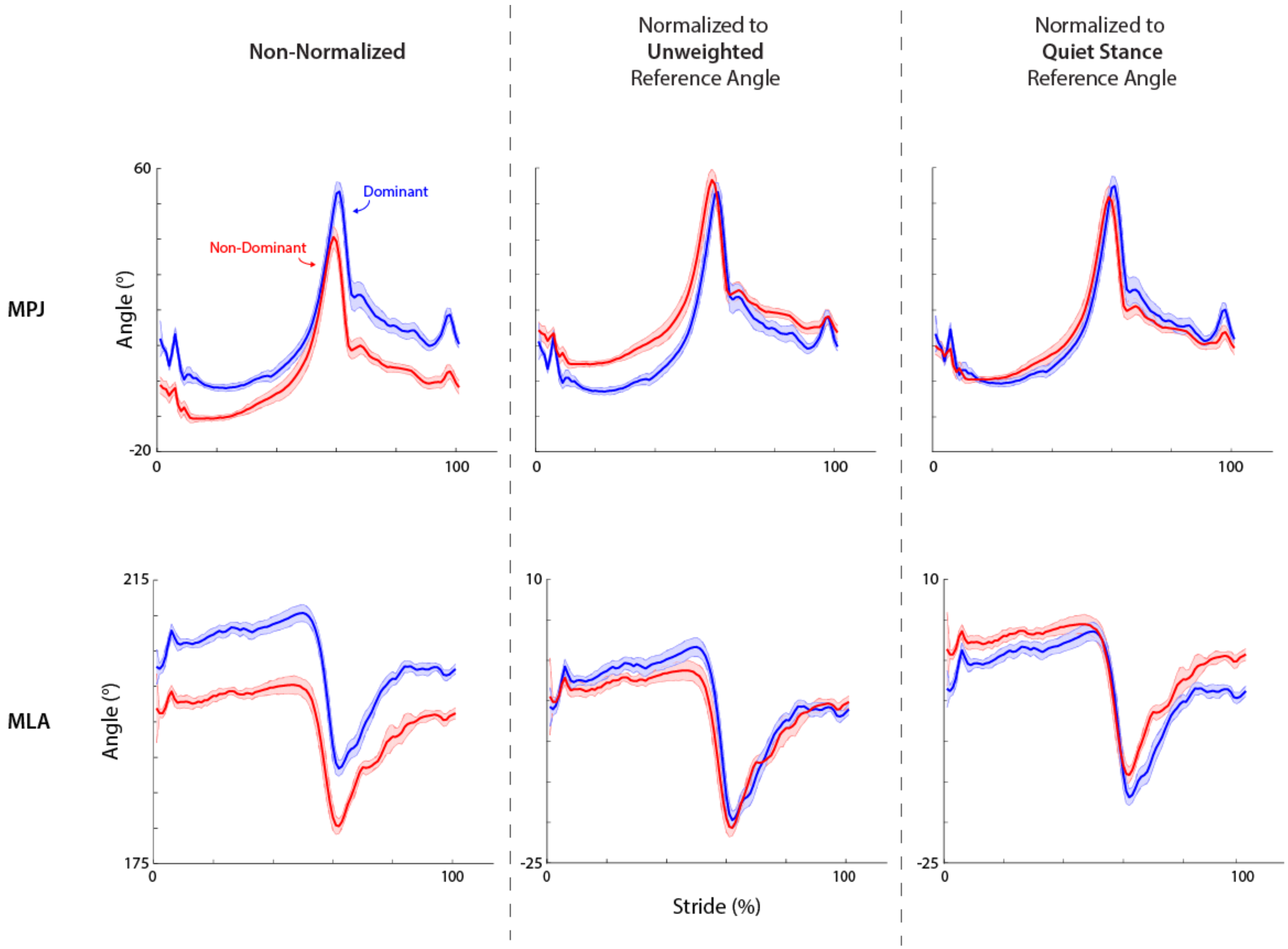

Figure 10. Effect of normalization method on dominant (blue) vs. non-dominant (red) limb comparison of peak metatarsophalangeal (MPJ) dorsiflexion angle and peak medial longitudinal arch (MLA) recoil angle for Participant 2 at 1.2 ms^{-1}. Shaded regions represent standard deviations.

Table 9. Effect of normalization method on between-limb differences in peak MPJ dorsiflexion angle.

	Range of Between-Limb Differences in Peak MPJ Angle due to Normalization Method (°) (+) indicates dominant side had a higher magnitude peak	**Maximum Effect Change due to Normalization Method (°)**
Participant 1	0.63 to 7.06	6.43
Participant 2	-3.08 to 13.12	16.21
Participant 3	-16.72 to -4.45	12.26
Participant 4	-0.14 to 5.93	6.06
Participant 5	7.00 to 22.34	15.34

Table 10. Effect of normalization method on between-limb differences in peak MLA recoil angle.

	Range of Between-Limb Differences in Peak MLA Angle due to Normalization Method (°) (+) indicates dominant side had a higher magnitude peak	**Maximum Effect Change due to Normalization Method (°)**
Participant 1	-1.058 to 6.32	7.38
Participant 2	-8.18 to 2.93	11.11
Participant 3	-7.00 to -2.62	4.38
Participant 4	-13.74 to -3.82	9.92
Participant 5	-5.39 to 8.94	14.34

Chapter 4

DISCUSSION

4.1 Overview

The existing literature has shown that asymmetries in ground reaction forces, electromyography profiles, and lower limb kinematics are present during able-bodied walking [19]–[22]. However, these asymmetries are rarely evaluated in the context of lower limb dominance. Furthermore, the effect of lower limb dominance on foot mechanics has not previously been explored, despite evidence to suggest that asymmetries are likely to be accentuated at the most distal joints during able-bodied walking [19], [23]. To address this knowledge gap, the objective of this study was to characterize MPJ and MLA kinematics with regards to footedness during walking. All five participants demonstrated notable asymmetry, with most exhibiting a higher MPJ range of motion on the dominant side. While MLA range of motion responses were varied, a greater time period between heel-rise and toe-off was observed on the non-dominant side. Taken together, these findings indicate that the functional roles of the feet seen in more overtly asymmetric bilateral tasks may also be present in walking. Furthermore, these findings highlight the importance of evaluating the behavior of both limbs during studies of human gait that are directly interested in, or indirectly concerned with, foot mechanics.

4.2 Between-Limb Differences in Kinematics

Participants generally demonstrated higher MPJ range of motion on their dominant side, aligning with the purported dynamic role of the dominant limb. The most widely accepted classification of footedness identifies the dominant limb as that which is preferred for mobilizing or manipulating behaviors, while the non-dominant limb provides support during bilateral tasks [1]. Based on this definition, it follows that the dominant foot would exhibit greater flexibility and dexterity during a bilateral task, when compared to the non-dominant foot. The hallux (via movement at the MPJ) is responsible for much of the dexterity seen at the foot [49]. Therefore, our finding that the dominant-side MPJ exhibited a greater range of motion (~4°-8°) for most participants aligns with the proposed functional role of the dominant foot.

Unlike the MPJ, increased range of motion at the MLA is associated with greater support and rigidity, rather than dexterity, and a slower release of the arch likely aids in stability. This is largely due to the action of the windlass mechanism, whereby the plantar fascia tensions during push-off as a result of its insertion at the head of the first metatarsal [36]. This, in turn, shifts the bones of the arch into a tightly packed conformation, decreasing the MLA angle and providing increased support for the foot [34]. Three of the five participants demonstrated a higher MLA range of motion (~1°-7°) on their non-dominant side, which suggests that, for them, the non-dominant foot invokes more extreme (high-gear) windlass function to provide stability and support during walking. The non-dominant limb for all participants also demonstrated a 28-50% increase in time duration between heel-rise and toe-off (with no notable change in stance time), indicating that the windlass mechanism may be utilized for a longer period of time as a means to stabilize the foot throughout push-off via a prolonged, less vigorous arch-spring recoil. There were no observable between-limb differences in the spatiotemporal metrics (i.e., stride length or stance/swing time ratio), thus indicating that the aforementioned differences seen in the foot kinematic data are not merely an accommodation to longer/shorter strides, but may reflect the proposed functional differences between the dominant and non-dominant feet.

Considering that the windlass mechanism is tied to metatarsal dorsiflexion, it may be anticipated that increased MPJ and MLA ranges of motion should occur on the same side. This is not reflected in the results, and may be explained, in part, by the differences in high- and low-gear windlass. During low-gear windlass, [34] notes that the plantar fascia remains relatively slack since it wraps around the small heads of metatarsals three to five. In this condition, MPJ dorsiflexion and plantar fascial tensioning are less tightly coupled, and thus increased MPJ dorsiflexion does not necessarily correspond to increased arch height. Conversely, during high-gear windlass, the plantar fascia wraps around the large head of the first metatarsal, and less MPJ dorsiflexion is required before the plantar fascia tensions and arch height increases [34]. Additionally, [37] notes that the passive windlass mechanism is not the only factor that determines MLA stiffness, but that it is also actively modulated by the intrinsic foot musculature.

Two participants demonstrated reversed trends from the rest of the group for MPJ and MLA range of motion, highlighting the importance of subject-specific analyses. Across all speeds, one participant (Participant 3) displayed a higher MPJ range of motion on their non-dominant side, and a different participant (Participant 2) displayed a higher MLA range of motion on their dominant side. Interestingly, these two participants were the only ones whose scores from the WFQ-R fell in the "mixed-footed" range, as outlined in [48]. For the participant who exhibited higher MPJ range of motion on the non-dominant side (Participant 3), it was noted that they demonstrated ~10° lower MPJ range of motion overall, when compared to the other participants (Figure 7). Upon evaluation of their time-series data, this participant also demonstrated a unique MPJ angle profile, in that their MPJ remained substantially dorsiflexed throughout the swing phase, whereas all other participants seemed to relax their toes following push-off (Appendix). Though this may accurately reflect their natural gait pattern, it is also possible that this participant altered their gait—potentially due to the barefoot and/or treadmill walking conditions. For the participant who demonstrated higher MLA range of motion on their dominant side (Participant 2), there were no marked abnormalities in their time-series data. However, further evaluation of their responses to the WFQ-R revealed that they preferred to use the same foot for both the object manipulation tasks and the support tasks, whereas this was not true for any other participant. If this participant prefers to use their dominant foot for manipulation as well as support, then this may explain why they have increased range of motion for both the MPJ and the MLA on their dominant side. The inter-subject variability demonstrated in these results highlights the importance of subject-specific analyses. If data were averaged across all participants, the reversed directionality of the differences discussed here may have obscured the overarching trends. Similarly, the unique findings for these two participants indicate that the between-limb differences cannot fully be explained by binary footedness. In an analysis of data pooled from all participants, this may have gone undetected.

4.3 Clinical and Research Implications

The between-limb differences in foot kinematics may be relevant in both clinical and research applications. Compared to healthy controls, plantar fasciitis patients have been shown to exhibit increased MPJ dorsiflexion (~4 deg) throughout a walking stride [17], [25]. Although it is unknown whether the increased dorsiflexion is a causative factor of plantar fasciitis or a compensatory mechanism to alleviate pain, it indicates that the magnitude of differences between the dominant and non-dominant MPJ ranges of motion expressed in the current study (3.4°-8.0°) are clinically relevant. A search of the existing literature did not identify any previous work that investigated prevalence of plantar fasciitis in dominant vs. non-dominant feet; however, [24] notes anecdotally that a higher fraction of unilateral plantar fasciitis patients recruited in their study presented with their dominant foot being affected. The between-limb differences in MPJ and MLA ranges of motion may also help to guide future development of prosthetic feet. When combined with preference findings in [33], the results from the present study suggest that the development of prosthetic feet specifically designed for dominant-side or non-dominant-side limb loss may be beneficial to the user in restoring a more comfortable and natural gait. In a case study by [32], a standard passive prosthetic foot (1S90, Ottobock) was found to deflect 8.9° at the toe during walking. If a set of dominant- and non-dominant feet were to be made based off of this value, then the complementary prosthesis (accounting for a between-limb difference in MPJ flexion of 3.4°-8.0°) may allow for approximately 1.5-2x more toe dorsiflexion.

Current biomechanical analyses of gait in clinical populations typically group data by "affected" vs. "unaffected" limb, without accounting for foot dominance. Based on the results from this study, we recommend that researchers consider the relevance of foot dominance when designing experimental protocols. For some studies, the effect of the clinical impairment may vastly outweigh the magnitude of asymmetries arising from foot dominance. Likewise, studies that are directed at more proximal joints (e.g., hip) may not be significantly affected by differences at the distal foot joints. In studies that evaluate foot mechanics or other distal joint parameters that may be closely coupled with foot mechanics, however, intrinsic asymmetries of the magnitudes determined in this study (~1°-8°) may have a nontrivial impact on the interpretation of results.

4.4 Limitations

The generalizability of these results to a broader population is limited by the small sample size of this study, which consisted of college-aged adults, included more female than male participants, and included two athletes competing at the varsity level. Furthermore, all participants were identified to be right-footed. Therefore, it cannot be guaranteed that any trends specifically resulted from limb dominance rather than simply left vs. right, especially given that societal pressures during certain activities (e.g., the location of foot pedals in cars) may influence the behavior of the left and right feet. Moreover, anatomical morphology was not explored in this study. It is possible that intrinsic inter- and intra-subject differences in foot anatomy (e.g., flat or high arches) may affect the functional roles of the feet, as [50] shows that people with flat arches tend to display more pronation during gait (which may, in turn, affect utilization of high- vs. low-gear windlass). Finally, this study evaluated foot joint mechanics during barefoot treadmill walking, and may not fully reflect walking in daily living. Although participants were given time to acclimate to every speed prior to data collection, it is possible that participants may have walked with a more cautious gait, as to protect against scraping or jamming their toes on the treadmill. Furthermore, we cannot speculate if/how the observed effects of foot dominance may change due to movement restriction or additional support when wearing shoes.

4.5 Recommendations for Future Work

Interpretations of between-limb differences in peak MPJ dorsiflexion angle and MLA recoil angle were occluded by the confounds of normalization method. There is currently no standard reference position of the foot joints to be used for normalization of kinematic data. Some studies utilize unnormalized foot joint angles [27], others normalize to the position of the foot during quiet stance [17], [50], and others use different methodologies [41], [51]. Upon comparing the results we obtained using three normalization methods, we found three different outcomes regarding peak joint angles. At a moderate walking speed (1.2 ms^{-1}), the selection of a different normalization method changed the reported between-limb difference in joint angle by up to 25°, and in some cases reversed the direction of the difference. Considering that 1.2 ms^{-1} is a standard comfortable walking speed, we predict that the effects of normalization method may be even more exacerbated at more extreme walking speeds (e.g., 0.8 ms^{-1} or 1.8 ms^{-1}). Similar to [11], who notes discrepancies in the range of values reported across the field for MPJ angle and attributes these discrepancies to the utilization of different reference angles during calculation, the results from this study highlight the importance of establishing a standard normalization method for calculating foot joint angles. Future work should be directed towards establishing a standard reference position for use in multi-segment foot models.

Another challenge noted in this study is the lack of consensus for measuring the strength of foot dominance. Both the WFQ-R and the binary (ball-kicking) evaluation have been widely accepted for accurately identifying the dominant limb [2], [52], [53]; however, there is no standard metric for quantifying strength of footedness. The original paper describing the WFQ-R [42] outlined a scoring method, which rates each response of (i) left-always, (ii) left-usually, (iii) equal, (iv) right-usually, and (v) right-always on a scale from -2 to 2. The degree of lateralization was then calculated using a log-odds ratio (λ) of the total left- and right- scores, in which $\lambda = \ln(right\ limb\ score/left\ limb\ score)$ as outlined in [54]. Using this method, negative λ values indicate left-dominance and positive λ values indicate right dominance. However, due to the logarithmic nature of this calculation, strength of foot dominance can range from -∞ to +∞ and thus, no method for further stratification was provided. Some studies that have used the WFQ-R have developed their own methods for evaluating footedness based on the sum of the total left- and right scores. On a scale from (-20) to (+20), [48] outlines scores less than (-6) as left-footed, scores from (-6) to (+6) as mixed-footed, and scores greater than (+6) as right-footed; [47] outlines scores less than (-11) as strong left-footed, from (-10) to (-1) as weak left-footed, (0) as mixed-footed, from (+1) to (+10) as weak right-footed, and greater than (+11) as strong right-footed. Although we are confident in the binary classification of participants being generally left- or right-footed, we are unable to speak to the accuracy of further stratification by strength of footedness. Therefore, future work is needed to establish a standard metric for evaluating strength of foot dominance.

Although the results from this study indicate that able-bodied walking does involve functional differences between the dominant and non-dominant feet, further work is necessary to confirm these findings in a larger population. Additionally, this study focused on level walking gait; however, future work directed at evaluating the role of foot dominance during uneven terrain locomotion or during running is also of interest.

4.6 Conclusion

In summary, this study evaluated foot mechanics during able-bodied walking, and identified between-limb differences associated with lower limb dominance. In general, participants demonstrated higher MPJ range of motion on the dominant side, varied MLA range of motion, and increased time between heel-rise and toe-off on the non-dominant side. These findings suggest that the conventional dynamic and support roles of the dominant and non-dominant limb, respectively, may also be present in walking. In addition to the differences noted between the two limbs, and their implications for future gait analyses, the results of this study bring to light two broad considerations for the field: 1) the importance of participant-level data analysis, and 2) the influence of normalization method on the interpretation of kinematic data, especially when using a multi-segment foot model.

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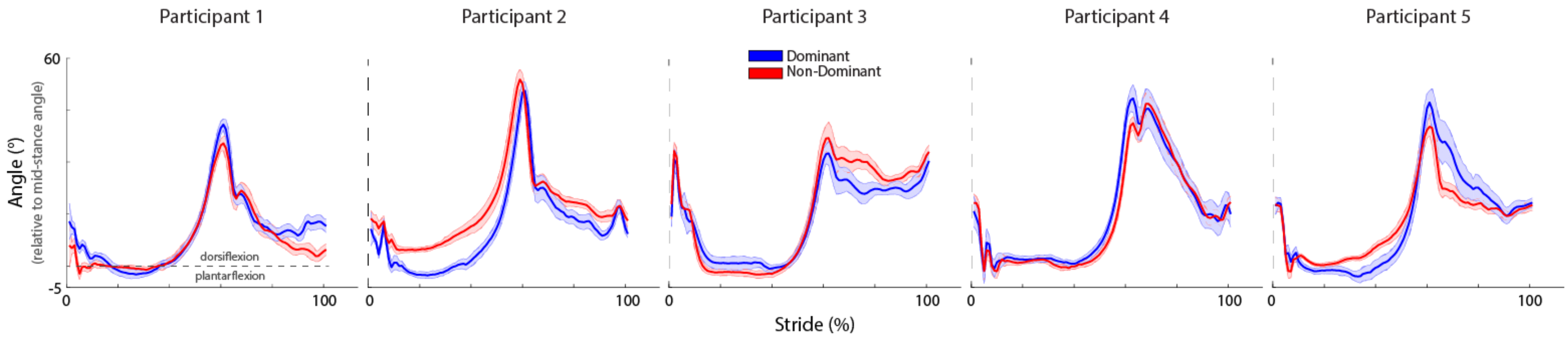
APPENDIX



Participant 1
Participant 2
Participant 3
Participant 4
Participant 5
Dominant
Non-Dominant
Angle (°)
(relative to mid-stance angle)
60
-5
dorsiflexion
plantarflexion
0
100
Stride (%)