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Original Article

Effects of Arm Swing on Plantar Pressure Behavior During Walking

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ABSTRACT

Objective: This study aimed to investigate the influence of different arm swing conditions on plantar pressure behavior during walking in healthy individuals.

Methods: The study included 29 healthy (22.55 ± 1.02 years) volunteers. The foot pressure was analyzed under 3 conditions: both arms should be freely swinging and the dominant arm should be restricted and should be held. Time and magnitudes of peak forces, gait velocity, duration of stance subphases, peak forces for 5 different areas in foot-sole, accelerations of the center of pressure, and mediolateral displacements of center of pressure were the interesting parameters.

Results: When the arm swing was held, the onset of terminal stance was earlier and the anterior-posterior center of pressure acceleration decreased at the midfoot on the affected side (0.32 ± 0.04 seconds, 2.96 ± 0.27 m/ms^2) than on the contralateral side (0.34 ± 0.05 seconds, 3.12 ± 0.28 m/ms^2) (P = .04, P = .02). The differences in anterior-posterior center of pressure acceleration between heel and forefoot and the mediolateral displacements of center of pressure were lower on the affected side at held (3.75 ± 0.31 m/ms^2, 0.06 ± 0.02 m, respectively) compared to the free swing (3.82 ± 0.30 m/ms^2, 0.07 ± 0.02 m) (P = .02, P = .01), while the peak force at the medial forefoot was lower on the contralateral side when the arm was held (28.87 ± 6.22 N) compared to the free swing (30.54 ± 5.86 N) (P = .01).

Conclusion: The lack of arm swing may interact with ipsilateral early onset and longer late stance phase during walking in healthy individuals. The foot pressure behaviors during walking should be investigated for unilaterally affected patients.

Keywords: Arm swing, foot pressure, gait

Introduction

Walking dynamics have been shown to be altered by neurologic diseases (hemiplegia, parkinsonism, cerebral palsy, spinal cord injuries, etc.), in which the arm swing is directly affected.[1-3] The changes in gait dynamics also alter foot pressure parameters such as force, pressure, and stance time in hemiplegics on both sides.[4-7] A recent study investigated the hemiplegic plantar pressure behavior and showed that reduced arm swing might change the plantar force behavior by reducing the first and the second maximum forces on both affected and contralateral sides compared to healthy individuals.[6] Additionally, stance phase durations were found to be longer in hemiplegics than in healthy peers, not only for the affected extremity but for both extremities. Limited arm swing range, as one of the reasons influencing the weight transfer from one side to the other during walking, has not been clearly defined for either patients with hemiplegics or healthy individuals.

Bruijn et al[8] found that arm swing contributed to the overall human gait stability for normal walking in healthy male individuals. Besides, Grodner et al[9] found that the time of peaks in ground reaction force graphs and the durations of walking subphases were different on the affected side than on the contralateral side of the lower extremity in patients with obstetric brachial plexus injuries who had limited arm swing during walking. Also, Baron et al[10] suggested that arm swing provided a sensitive measure of declines in gait function in Parkinson's disease under dual-task conditions. Altering arm swing might affect ground reaction forces and gait stability. The role of arm swing asymmetries on gait kinematics is usually ignored, and gait treatments focus mainly on lower extremities for patients whose upper and lower extremities are affected. Although there are studies conducted on the effects of arm swing during walking in healthy and pathologic populations,[3,11-15] pure effects of arm swing alterations on

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plantar pressure behavior during walking are still not clear for healthy individuals and this may help understand the fundamentals of biomechanical alterations in neurologically affected populations. With the help of this study, we will be able to determine the direct impacts of arm swing during walking on ground reaction forces and foot pressure behavior.

Therefore, the present study aimed to investigate the effects of different arm swing conditions on plantar pressure behavior during walking in healthy individuals.

Methods

Subjects

A total of 29 healthy participants, aged 18-30 years (22.55 ± 1.02 years, 15 females and 14 males, body mass index: 23.17 ± 3.75), were included in the study. Participants with any upper extremity/trunk pathology that would prevent arm swing during walking, any neuromuscular or musculoskeletal system pathologies, or past lower/upper extremity surgery in the last 2 years were excluded from the study.

The study was carried out at Gait Analysis Laboratory, Department of Orthopedics and Traumatology, Istanbul Faculty of Medicine, Istanbul University. The study was approved by the Istanbul University Faculty of Medicine Clinical Research Ethics Committee and was conducted following the Helsinki Declaration (Date: May 13, 2016, No: 09).

Research Design

The plantar pressure behavior was analyzed using pressure sensors (Matscan, Tekscan Inc., Mass, USA) at the self-selected speed of walking and it was analyzed under 3 different conditions: (1) both arms should be in a free swing (FS), (2) the dominant arm swing should be restricted by an elastic band at the level of the umbilicus to reduce the swing range (RS), and (3) the dominant arm should be held to the trunk with an arm sling, which kept the arm across the opposite shoulder and did not allow arm swing (H). Video recording was performed using a camera and a tripod on the lateral side of the walkway (Panasonic Lumix DMC-FZ300) to check the restriction status of different arm swing conditions. Each participant walked for 3 minutes before recording to set the self-selected speed gait. During each trial, gait speed was calculated using Kinovea motion analysis software (Kinovea 0.8.15).

In the pedobarographic analysis, the time and the magnitude of the first and second peak forces (t_1, t_2 (s), F_1, F_2 (Newton (N)) respectively), the time and the magnitude of the lowest force between the first and second peak forces [t_3 (s), F_3 (N), respectively], stance time (t_{stance}), maximum force ($F_{max,}$ N), normalized duration of terminal stance ($[t_2 - t_3]/t_{stance}$), pre-swing ($[t_{stance} - t_2]/t_{stance}$), late stance (terminal stance and pre-swing times ($[t_s - t_3]/t_s$)), midstance ($[t_3 - t_1]/t_s$), loading response (t_1/t_{stance}), and t2 (t_2/t_{stance}) by stance time (t_{stance}), and gait speed were the parameters of interest (Figure 1).[16]

Additionally, the following parameters were analyzed: acceleration of the center of pressure for anterior–posterior direction [CoP_{AP} (m/ms^2)] in the 3 different foot areas [heel, midfoot (MF), and forefoot] and mediolateral (ML) direction [CoP_{ML} (m/ms^2)],[17,18] CoP_{AP} acceleration difference between heel and forefoot, mean deviation of the CoP_{ML} displacement (m), peak force values in the 5 different foot areas [heel, medial MF, lateral MF, medial forefoot (MFF), and lateral forefoot] (N).

Statistical Analysis

In statistics, the Kolmogorov–Smirnov test was used for the analysis before the normal distribution of data was determined. The independent-samples *t* test was used to compare the parameters of interest between the affected and contralateral sides. Repeated-measures analysis of variance was used to compare the parameters of interest on the affected and contralateral sides between different arm swing conditions and was conducted using Statistical Package for the Social Sciences 21.0 (IBM SPSS Corp., Armonk, NY, USA).

Results

Alterations Between Affected and Contralateral Sides Under Different Arm Swing Conditions

For the held arm swing condition, t_3 and t_3/t_{stance} were shorter on the affected side (0.32 ± 0.04 seconds and 0.49 ± 0.04 seconds, respectively) than on the contralateral side (0.34 ± 0.05 seconds and 0.51 ± 0.05 seconds, respectively) (P = .04 and .02, respectively) (Figure 1). Late stance durations ($[t_s - t_3]/t_s$) was longer on the affected side (0.52 ± 0.04 seconds) than on the contralateral side (0.49 ± 0.05 seconds) (P = .02) (Table 1). Also, mean CoP_{AP} acceleration at the MF was lower on the affected side (2.96 ± 0.27 m/ms^2) than on the contralateral side (3.12 ± 0.28 m/ms^2) (P = .02) (Table 2).

For all arm swing conditions, the terminal stance duration ($[t_2 - t_3]/t_{stance}$) was significantly longer on the affected side than on the contralateral side (Table 1). The time and the magnitude of the first and the second peak forces (t_1, t_2, F_1, F_2, respectively) were not significantly different between the affected and contralateral sides. Also, no significant differences were found in stance time (t_{stance}), maximum force (F_{max}), normalized pre-swing duration ($[t_{stance} - t_2]/t_{stance}$), midstance ($[t_3 - t_1]/t_s$), and loading response (t_1/t_{stance}) between affected and contralateral sides (Table 1).

Alterations Between Different Arm Swing Conditions on the Same Side

The peak force at MFF was lower (28.87 ± 6.22 N) for the held arm swing condition than for the FS arm condition (30.54 ± 5.86 N) on the contralateral side (P = .01). The mean CoP_{AP} acceleration difference between heel and forefoot on the affected side was lower for the held arm swing condition (3.75 ± 0.31 m/ms^2) than for the FS arm condition (3.82 ± 0.30 m/ms^2) (P = .02) (Table 2). The ML displacements of CoP on the affected side were lower for the held arm swing condition (0.06 ± 0.02 m) than for the free arm swing (0.07 ± 0.02 m) (P = .03) and restricted arm swing condition (0.07 ± 0.02 m) (P = .01) (Table 2).

No significant change was found in other parameters of interest between different arm swing conditions for both affected and contralateral sides.

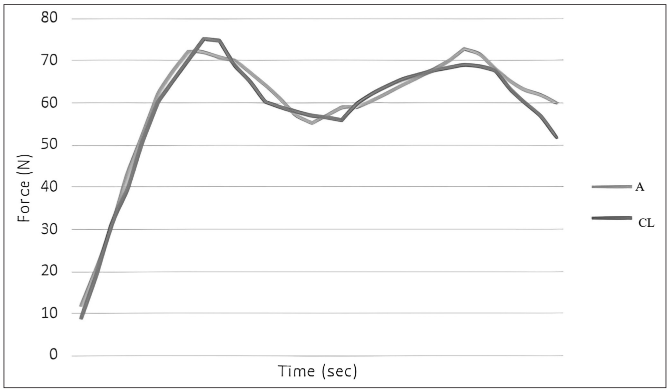

Figure 1. The behavior of ground reaction forces for affected and contralateral sides in held arm swing condition. A, affected; CL, contralateral; N, newton; sec, second.

Table 1. Comparison of the Parameters of Interest Between Affected and Contralateral Sides

	Free Swing			Restricted Swing			Held		
	A	CL	*P*	A	CL	*P*	A	CL	*P*
F_{max} (N)	75.74 ± 13.70	76.70 ± 15.64	.801	75.40 ± 14.81	76.50 ± 15.45	.781	75.03 ± 14.08	75.61 ± 14.92	.803
F_1 (N)	72.31 ± 14.04	76.45 ± 15.84	.292	72.87 ± 15.61	76.25 ± 15.72	.412	72.09 ± 14.95	75.13 ± 15.31	.442
F_2 (N)	73.86 ± 12.67	69.33 ± 12.17	.175	73.13 ± 14.19	69.32 ± 12.66	.284	72.82 ± 13.06	69.5 ± 12.5	.322
F_3 (N)	54.73 ± 10.98	56.08 ± 12.46	.665	53.69 ± 11.32	55.32 ± 12.14	.601	55.21 ± 11.65	55.94 ± 12.48	.826
F_{MAX_MFF} (N)	31.75 ± 7.29	30.54 ± 5.86	.484	31.95 ± 6.70	29.45 ± 5.31	.127	31.36 ± 6.11	28.87 ± 6.22	.137
F_{MAX_MMF} (N)	12.73 ± 4.34	9.51 ± 3.72	**< .01**	11.73 ± 4.35	9.30 ± 2.99	**.010***	12.26 ± 5.07	9.42 ± 3.27	**.011***
F_{MAX_LFF} (N)	28.24 ± 6.08	26.74 ± 6.88	.385	27.95 ± 6.19	26.89 ± 6.88	.545	27.90 ± 6.09	26.89 ± 6.11	.535
F_{MAX_LMF} (N)	17.54 ± 4.87	15.71 ± 7.79	.285	18.20 ± 6.34	15.20 ± 6.68	.089	19.01 ± 6.72	16.61 ± 6.69	.172
F_{MAX_HEEL} (N)	28.20 ± 6.13	30.16 ± 6.29	.232	27.49 ± 5.67	29.69 ± 5.55	.146	27.01 ± 5.57	29.29 ± 5.18	.118
t_{stance} (seconds)	0.66 ± 0.04	0.66 ± 0.05	.861	0.65 ± 0.06	0.66 ± 0.06	.557	0.67 ± 0.05	0.67 ± 0.05	.722
t_1 (seconds)	0.17 ± 0.03	0.17 ± 0.04	.802	0.16 ± 0.03	0.17 ± 0.04	.372	0.17 ± 0.03	0.17 ± 0.03	.686
t_2 (seconds)	0.51 ± 0.04	0.50 ± 0.04	.482	0.50 ± 0.05	0.50 ± 0.04	.932	0.51 ± 0.03	0.51 ± 0.04	.593
t_3 (seconds)	0.32 ± 0.04	0.33 ± 0.04	.368	0.31 ± 0.04	0.33 ± 0.05	.123	0.32 ± 0.04	0.34 ± 0.05	**.041***
t_1/t_s	0.25 ± 0.03	0.25 ± 0.04	1.000	0.25 ± 0.03	0.26 ± 0.04	.552	0.25 ± 0.03	0.26 ± 0.03	.612
t_2/t_s	0.77 ± 0.03	0.75 ± 0.03	.081	0.77 ± 0.03	0.75 ± 0.03	.291	0.76 ± 0.03	0.76 ± 0.04	.900
t_3/t_s	0.48 ± 0.05	0.49 ± 0.05	.288	0.48 ± 0.05	0.50 ± 0.04	.092	0.49 ± 0.04	0.51 ± 0.05	**.021***
$[t_2 - t_3]/t_s$	0.28 ± 0.04	0.26 ± 0.04	**.032***	0.29 ± 0.04	0.26 ± 0.04	**.010***	0.28 ± 0.04	0.26 ± 0.05	**.051***
$[t_s - t_2]/t_s$	0.23 ± 0.03	0.25 ± 0.03	.087	0.23 ± 0.03	0.23 ± 0.02	.295	0.24 ± 0.03	0.24 ± 0.04	.902
$[t_s - t_3]/t_s$	0.52 ± 0.05	0.51 ± 0.05	.284	0.52 ± 0.05	0.50 ± 0.04	.092	0.52 ± 0.04	0.49 ± 0.05	**.021***
$[t_3 - t_1]/t_s$	0.22 ± 0.04	0.23 ± 0.04	.344	0.23 ± 0.04	0.24 ± 0.03	.293	0.22 ± 0.04	0.24 ± 0.05	.061
Walking velocity (m/s)	1.36 ± 0.09		-	1.37 ± 0.11		-	1.35 ± 0.10		-
CoP_{AP_mean} (m/ms²)	3.24 ± 0.31	3.18±0.35	.487	3.16 ± 0.53	3.13 ± 0.38	.758	3.23 ± 0.28	3.08 ± 0.32	.061
CoP_{AP_heel} (m/ms²)	1.19 ± 0.14	1.14 ± 0.23	.407	1.19 ± 0.13	1.20 ± 0.15	.803	1.21 ± 0.13	1.17 ± 0.18	.393
$CoP_{AP\_midfoot}$ (m/ms²)	3.00 ± 0.31	3.15 ± 0.37	.094	2.98 ± 0.29	3.14 ± 0.37	.071	2.96 ± 0.27	3.12 ± 0.28	**.023***
$CoP_{AP\_forefoot}$ (m/ms²)	5.01 ± 0.33	5.01 ± 0.39	.959	4.99 ± 0.35	4.99 ± 0.37	.971	4.96 ± 0.33	4.98 ± 0.35	.827
$CoP_{AP\_difference}$ (m/ms²)	3.82 ± 0.30	3.86 ± 0.43	–	3.79 ± 0.31	3.79 ± 0.35	–	3.75 ± 0.31	3.81 ± 0.32	–
CoP_{ML_mean} (m/ms²)	2.29 ± 0.98	2.08 ± 0.97	.415	2.42 ± 0.84	2.12 ± 1.12	.292	2.26 ± 0.93	2.01 ± 0.99	.337
$CoP_{ML\_displacement}$ (m)	0.07 ± 0.02	0.06 ± 0.02	.363	0.07 ± 0.02	0.05 ± 0.01	**.044***	0.06 ± 0.02	0.05 ± 0.02	.428

*P < .01. A, affected side; CL, contralateral side; MFF, medial forefoot; MMF, medial midfoot; LFF, lateral forefoot; LMF, lateral midfoot; N,newton; s: second; m:meter; ms:millisecond; CoP, center of pressure; AP, anteroposterior; ML, mediolateral.

Discussion

The present study investigating the effects of different arm swing conditions on plantar pressure behavior during walking revealed that when the arm swing was held, the onset of terminal stance was earlier and lasted longer, and the mean CoP_{AP} acceleration at the MF was lower on the affected side than on the contralateral side. Besides, the peak force at the MFF was lower on the contralateral side, and the ML displacement of CoP was lower on the affected side when the arm swing was held compared to the free arm swing condition. Finally, the difference in mean CoP_{AP} acceleration between heel and forefoot was lower on the affected side when the arm swing was held compared with the FS.

Alterations in walking dynamics lead to abnormal loading patterns, which can be detected by foot pressure analysis. Understanding the relationship between abnormal loading pattern and proximal biomechanical reason, which is transmitted to the loading pattern, is pivotal in tailoring the treatment for the level of walking. However, a gap exists in the understanding of the linkage between altered walking dynamics, such as arm swing asymmetries, and related loading pattern alterations not only for patients but also for the healthy population. Defining the basic pathomechanics of a healthy population helps understand the biomechanical fundamentals for patients who have variable disabilities such as muscle tonus or coordination abnormalities. Therefore, healthy individuals were examined in the present study, revealing that ipsilateral early onset and prolonged terminal stance duration might be related to the held arm swing during walking.

Terminal stance onset and duration were defined by the difference between the time of the second peak and the deepest peak of the force–time graphics in individual stance duration ($t_2 - t_3/t_s$),[16] which is the transition period from eccentric to isometric contraction of plantar flexor muscles to raise the heel for initiating the propulsion of the body forward.[19] Therefore, early onset and prolonged terminal stance may deal with early heel raise. Early heel rise is primarily because of an enhanced plantar flexor activity usually accompanied by reducing the second force peak (F_2).[19] Predictably, the F_2 on the held side was not significantly different from that on the contralateral side because the healthy individuals in the present study had hypothetically typical plantar flexor activity but held an arm swing on the dominant side.

In late stance, the ipsilateral upper limb swings from the middle line to forward, helping in counterbalancing pelvic external rotation coop with internal trunk rotation.[11,12] As the ipsilateral arm was held in this study, the arm could not go forward. Therefore, the altered counterbalance mechanism might have slightly reduced external pelvic rotation, which might have been compensated by the early onset of late stance phase where the plantar flexors started to increase the relative ipsilateral leg length.

Although the average peak force at MFF on the contralateral side was found to be lower in the held arm condition than in the FS condition, the affected side's MFF was not significantly different (Table 1). During typical walking, the pressure at MFF on the contralateral side coincides with the ipsilateral terminal stance and pre-swing phases when the contralateral arm moves backward to reach the maximum

Table 2. Comparison of the Parameters of Interest Between the Same Sides Under Different Arm Swing Conditions

	Affected Sides		Contralateral Sides	
Parameters of Interest	**Groups**	***P***	**Groups**	***P***
F_{MAX_MFF}	FS-RS	.838	FS-RS	.106
	FS-H	.716	FS-H	**.011***
	RS-H	.442	RS-H	.348
F_{MAX_MMF}	FS-RS	.205	FS-RS	.782
	FS-H	.514	FS-H	.903
	RS-H	.389	RS-H	.874
F_{MAX_LFF}	FS-RS	.655	FS-RS	.857
	FS-H	.643	FS-H	.837
	RS-H	.944	RS-H	.991
F_{MAX_LMF}	FS-RS	.457	FS-RS	.611
	FS-H	.071	FS-H	.303
	RS-H	.371	RS-H	.081
F_{MAX_HEEL}	FS-RS	.302	FS-RS	.436
	FS-H	.072	FS-H	.272
	RS-H	.324	RS-H	.392
Walking velocity (independent from the affected or contralateral side)	FS-RS	.345	FS-RS	.345
	FS-H	.721	FS-H	.723
	RS-H	.220	RS-H	.221
CoP_{AP_mean}	FS-RS	.349	FS-RS	.279
	FS-H	.710	FS-H	.081
	RS-H	.471	RS-H	.241
CoP_{AP_heel}	FS-RS	.889	FS-RS	.228
	FS-H	.163	FS-H	.608
	RS-H	.142	RS-H	.233
$CoP_{AP\_midfoot}$	FS-RS	.602	FS-RS	.747
	FS-H	.341	FS-H	.523
	RS-H	.411	RS-H	.723
$CoP_{AP\_forefoot}$	FS-RS	.355	FS-RS	.573
	FS-H	.118	FS-H	.484
	RS-H	.288	RS-H	.822
$CoP_{AP\_difference}$	FS-RS	.245	FS-RS	.132
	FS-H	**.023***	FS-H	.405
	RS-H	.071	RS-H	.475
CoP_{ML_mean}	FS-RS	.132	FS-RS	.571
	FS-H	.791	FS-H	.592
	RS-H	.068	RS-H	.176
$CoP_{ML\_displacement}$	FS-RS	.961	FS-RS	.644
	FS-H	**.032***	FS-H	.304
	RS-H	**.011***	RS-H	.578

FS, free swing; H, held; RS, restricted swing; MFF, medial forefoot; MMF, medial midfoot; LFF, lateral forefoot; LMF, lateral midfoot; CoP, center of pressure; AP, anteroposterior; ML, mediolateral.

extension range and enforce the externally rotated opposite pelvis through internal rotation by push-off.[20] The contralateral peak force at MFF in held condition may be decreased in late stance because the affected arm cannot move backward. The 3-dimensional kinematic analysis should be analyzed in detail to describe the relationships between the counter trunk and pelvis rotation under different arm swing conditions.

The CoP_{AP} acceleration in this study was investigated by calculating the first derivative of anterior CoP velocity, which decreased at the MF on the affected side than on the contralateral side under the held arm swing condition. Under the free arm swing condition, CoP was at the MF in the midstance and terminal stance phases[21] where the ipsilateral arm moved from back to front to balance the pelvic counter-rotation.[11] Center of pressure being in the MF area during gait also coincides with the duration between contralateral toe-off and initial contact where the body load is carried on the ipsilateral side (single-limb stance).[21,22] The forward progression of CoP in midstance is related to the forward movement control of the whole body's center of mass via majorly single foot rollover function.[21] Therefore, CoP forward velocity and acceleration are important indicators of movement efficiency and stability control.[23] Since angular momentum generated by the arms was reduced ipsilaterally under the held arm condition in the study, the foot rollover function might be altered to slow the center of force in the AP direction, which might be an indicator of demand to increase AP stability control.[8,23–25]

Nolan et al[26] revealed that the CoP velocity in stroke survivors was significantly lower in single-limb stance on the affected side compared to the contralateral sides and healthy controls. The reduced or absence of arm swing may contribute to the slow forward, which is an indicator of AP stance stability, maintaining walking speed (Table 1).[27] Therefore, improving the arm swing range in physiotherapy may contribute to increasing the forward CoP velocity and help improve AP stability in a single stance.

Fuchioka et al[21] revealed that slower CoP might also be related to reduced walking speed. No significant gait velocity difference was found among the participants during any arm swing conditions. Additionally, the mean CoP acceleration in stance was also not different on both sides during any walking condition (Table 2).

The CoP differences in different plantar regions in this study showed that the difference in forward CoP velocity between heel and FF was higher on the affected side than on the contralateral side under held arm condition, although the individual regional CoP velocities were not significantly different. The CoP was in the forefoot region in terminal stance and pre-swing during walking under the free arm condition.[28] The lack of the arm swing caused early onset and extended terminal stance duration and consequently, reductions in CoP velocity and acceleration in the AP direction in a single-limb stance, which coincided with ipsilateral midstance and terminal stance durations.[22] Center of pressure velocity and acceleration reductions might continue in FF and cause the difference in the heel region because the single-limb stance partially coincided with the terminal stance where the CoP was in the forefoot region.

This study also confirmed that the biomechanical alteration of 1 upper extremity affected both sides of the lower extremity during walking. Kim et al[6] showed that the first and the second peak forces were lower and the stance time was longer on the affected and unaffected sides of the hemiplegic patients compared to the healthy controls. In addition, Femery et al[5] found significant differences in specific plantar pressure distribution profile on both sides in hemiplegics. Moreover, it was mentioned that the longer stance time and altered peak forces might deal with stability in stance[29] in hemiplegics and reduced gait velocity in hemiplegics[5] and unilateral transhumeral amputees.[30] Therefore, with increasing stance duration, the pure effect of the altered arm swing might deal with reduced stance stability, which was first defined by Gage et al[29] as the primary necessity for normal walking. However, even though the stance time did not change, the late stance duration in the present study was longer and the onset time of the terminal stance was earlier on the affected side than on the contralateral side when the arm swing was held. On the contrary, the ML displacement of CoP significantly increases with the instability in stance and[31] is reduced on the affected side in this study when the arm swing is lacking (Table 1). Therefore, the terminal stance might increase, although the single-limb stance (MSt and TSt durations) did not change and the stability increased on the affected side by reducing the ML difference in CoP. The lack of arm swing might influence the stance phase stability on the affected and contralateral sides differently in healthy and neurologically affected people.[32]

Altered pedobarographic parameters in the present study, which were the longer *t*3 and *t*3/*t*s of the contralateral side and late stance of the affected side, coincided with the arm in non-neutral position (flexion and extension range) of the affected limb during walking. The arm extension occurred between the contralateral side's initial contact (peak arm flexion) and ipsilateral initial contact (peak arm extension). The early stance on the contralateral side occurred between the peak arm flexion and the mid-line position (crossing the trochanter), and the late stance on the affected side occurred between mid-line and peak extension arm swing. The arm extension in mid-stance on the contralateral lower extremity created counterforce to the rotatory movement of the body during the swing and helped in dynamic stabilization in early stance.[19] Thus, the longer late stance on the affected side might deal with the absence of arm flexion, and the longer early stance (loading response and mid-stance) on the contralateral side might be due to the absence of arm extension, which reduced the dynamic stability on both sides. The lower peak force at the MFF on the contralateral side under the held condition than under the free arm swing condition revealed that the absence of arm extension might result in difficulty in the push-off phase and unstable late stance. Therefore, both the arm flexion and extension range must be increased during gait rehabilitation, particularly hemiplegic gait rehabilitation.

Study Limitations

The present study tried to mimic the reduced and no arm swing conditions in healthy individuals; however, the compensations would differ in neurologically and/or orthopedically affected patients. While attention was drawn to the development of compensatory mechanisms with trunk movements due to restriction of arm swing[3], another limitation is that we could not control trunk movements in our study. The inclusion of 3-dimensional gait analysis in the study may make the results of the study more comprehensive.

The lack of arm swing might interact with early onset and longer terminal stance phase and lower mean CoP_{AP} acceleration at the MF on the affected side in healthy populations. Additionally, the lack of arm swing lowered the peak force at the MFF on the contralateral side and the mean CoP_{AP} acceleration difference between heel and forefoot on the affected side compared with the free arm swing condition. Although compensations would be different in neurologically affected patients than in healthy populations, the results revealed that the unilateral lack of arm swing might be considered in gait rehabilitation for pathologies such as stroke and cerebral palsy, which have a longer duration of the terminal stance phase.

Ethics Committee Approval: Ethical committee approval was received from the Ethics Committee of Istanbul University (Date: May 13, 2016, No: 09).

Informed Consent: Written informed consent was obtained from all participants who participated in this study.

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