Gait & Posture 69 (2019) 60–65

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Full length article

The effect of unstable shoe designs on the variability of gait measures

Mona Khoury-Mireb[a,*], Deborah Solomonow-Avnon[a], Nimrod Rozen[b], Alon Wolf[a]

[a] Biorobotics and Biomechanics Lab, Faculty of Mechanical Engineering, Technion-Israel Institute of Technology, Haifa, Israel
[b] Department of Orthopaedic Surgery, Ha'Emek Medical Center, Afula, Israel

ARTICLE INFO

Keywords:
Gait variability
Gait stability
Unstable footwear

ABSTRACT

Background: Unstable footwear designs are popular as training devices to strengthen human neuromuscular control, and many studies have evaluated their effect on gait parameters in comparison to conventional footwear designs. However, there is minimal research on variability of gait measures during walking with unstable shoes. Therefore, the study objective was to compare variability of gait measures between stable and unstable shoe configurations, in conjunction with kinematic and kinetic changes.

Methods: Fifteen healthy male subjects walked in both a stable and unstable footwear device configuration while full-body gait kinematic and kinetic data was collected. Averages and standard deviations of gait trials were compared between the two configurations at different stages of each step.

Results: Comparison of gait variability between both footwear configurations revealed that variability of frontal-plane foot center of pressure offset, transverse-plane ankle moment, and frontal-plane shoulder angle decreased significantly while walking in the unstable configuration, while transverse-plane spine angle variability increased. No changes in variability of gait measures at the knee, hip, or pelvis were observed. Kinematic and kinetic changes were observed throughout the whole body with the unstable shoe.

Conclusion: Our findings suggest that the unstable device used in the study may reduce gait variability at the two extremes of the kinematic chain (i.e., foot, ankle, and shoulders), but increase variability of spine rotation angle. This may suggest a compensatory mechanism to maintain both stability and adaptability, and may have potential clinical implications for gait retraining and enhancing dynamic gait stability and joint stability, pending further investigation.

1. Introduction

Natural gait is a repeatable consistent function in healthy populations [1], yet complex stride-to-stride variability exists [2]. Although "variability" and "stability" are not interchangeable terms, unstable walkers (i.e., young children, the elderly, Parkinson's disease patients) may have greater stride-to-stride variability [2], and increased variability may be a risk factor for falls in certain populations [3]. However, when discussing association between variability and stability, one must differentiate between global and local stability. In general, increased movement variability on the global (macroscopic) level may have negative impact on gait performance and should be minimized [4]. On the other hand, increased variability on an individual joint (microscopic) level may be a favorable indication of increased adaptability and flexibility, in allowing exploration of different movement patterns to find the optimal strategy, or the ability to overcome external perturbations [2,5]. However, the relationship between joint-level variability and dynamic stability may be specific to an individual's circumstances and health status. For example, reduced joint-level variability may be an indication of pathology, but may reduce risk of falling [2,6]. Yet, excessive joint variability may provide evidence for development of degenerative joint diseases [7]. Given the complex relationship between variability and dynamic stability, investigation of variability while walking with unstable shoes is of great interest.

Unstable footwear designs have been widely studied and implemented over the last decade as devices to strengthen muscles and train neuromuscular control [8–11], including in treatment protocols for pathologies including degenerative diseases of lower-limb joints [8–10]. It is well known that unstable footwear alters gait pattern and neuromuscular activity [12–19], with several of these alterations identified as contributors, or potential contributors, to its clinical benefit [9,10,20,21]. As such, studies have largely focused on differences in these parameters between unstable and stable (control) shoes, but there is minimal literature regarding gait variability in unstable shoes.

* Corresponding author at: Biorobotics and Biomechanics Lab (BRML), Faculty of Mechanical Engineering, Technion-Israel Institute of Technology, Haifa, 32000, Israel.

E-mail address: monakh@technion.ac.il (M. Khoury-Mireb).

https://doi.org/10.1016/j.gaitpost.2019.01.017
Received 9 April 2018; Received in revised form 19 December 2018; Accepted 11 January 2019
0966-6362/

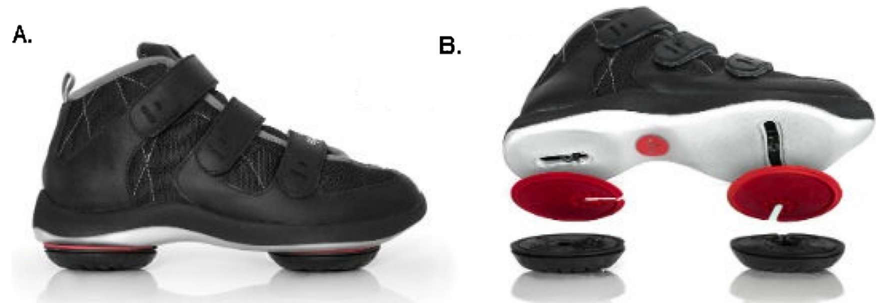

Fig. 1. Biomechanical device with: (A) adjustable elements in Neutral (unstable) configuration, (B) no elements attached (stable configuration).

Several previous studies exist using the Masai Barefoot Technology (MBT) rocker-bottom shoes (rounded sole in the anterior-posterior direction, and a soft heel), two of which report variability of full-body gait parameters and electromyographic parameters [22,23], one which reports variability of temporal gait characteristics only [24], and all of which were conducted on a treadmill at a pre-defined speed. Another study on a Biodyn unstable sandal construction (flexible heel element and freely moving pivot point under the foot arch) examined variability of spatio-temporal gait parameters only, on a treadmill at habitual speed [25]. An additional study on a shoe with an irregular randomly-deforming midsole, constructed for the purpose of the study, examined lower-limb kinematics and electromyography during treadmill-walking at a pre-defined speed [26]. Previously, our research group, using a unique unstable shoe device with two convex elements attached to a shoe sole, reported variability of foot center of pressure (COP) parameters with different levels of instability (convexity) during overground walking at habitual speed [17], but did not compare results with a conventional shoe. To the best of our knowledge, there are no studies that compare variability of full-body gait measures between an unstable and conventional (stable) shoe during overground walking at habitual speed. Treadmill-walking reduces natural variability of gait, potentially masking changes in variability, which may be disadvantageous in a study primarily interested in variability [27]. Further, measuring gait variability at a pre-determined speed differing from habitual speed may mask natural gait variability, as it has been shown to exhibit a U-shape over a range of walking speeds, in which it is increased when speed is either slower or faster than subject-preferred speed [28].

One might assume that a shoe that provokes dynamic instability, or a greater risk of falling, might lead to greater stride-to-stride variability, as seen in individuals prone to falling [2]. However, this is not necessarily the case, as footwear-generated instability may cause enhanced activation of stabilizing muscles in order to limit degrees of freedom in the joint, leading to more stability [29]. Indeed, both increased [22–26] and decreased [17] variability in gait parameters have been observed for short-term use of different unstable shoes. Thus, the conflicting results warrant investigation of gait variability for different unstable shoe structures to characterize their unique mechanisms and benefits or disadvantages.

The present study was devised to: (1) quantify and compare variability (standard deviation; STD [2]) of full-body gait measures between a stable shoe construction (SS) versus unstable shoe construction (US), and (2) assess kinematics and kinetics during gait in SS and US, using a unique footworn device. Objective 2 was undertaken to aid in interpretation of variability results. We hypothesized that there would be: (1) less variability (reduced STD) in gait measures, and (2) significant differences in mean gait measures in US compared to SS. To the best of our knowledge, this study is the first to report variability of full-body gait parameters while overground walking in US at a comfortable self-selected speed, and compare gait variability measures between SS and US of the specific study device.

2. Methods

2.1. Participants

The cohort included 15 healthy anthropometrically-similar male subjects (Shoe size = French 43, Age = 26.3 ± 2.6years, Height = 174.93 ± 3.65 cm, Mass = 71.5 ± 9.34 kg). Exclusion criteria were any orthopedic, musculoskeletal, or neuromuscular pathology. The study was approved by the Ethics Sub-Committee and participants received full explanation before giving written informed consent.

2.2. The biomechanical system

The APOS biomechanical device (APOS System, APOS—Medical and Sports Technologies Ltd. Herzliya, Israel) was used [18]. It includes two adjustable convex-shaped biomechanical elements attached to the feet by means of a shoe sole with two mounting rails (Fig. 1). A shift of the elements in the transverse plane of the foot causes a corresponding directional shift of COP [30]. The device was used in two conditions: (1) without biomechanical elements, representing SS, and (2) with the elements attached and configured to a neutral condition [18] (see Supplement), representing US.

2.3. Experimental protocol

Subjects were given a several-minute period (< 3 min, varying across subjects) prior to data acquisition to become accustomed to each shoe configuration. Subsequently, gait analyses were performed in both US and SS, in random order, at subject-selected walking speed. Preliminary testing revealed that gait speed did not vary significantly across device configurations or walking trials, thus speed was not controlled and all successful gait trials were included in the analysis, such that the natural walking pattern, and associated variability, was measured. Walking speed was reported in the analysis.

Three-dimensional motion analysis was performed using an 8-camera Vicon motion analysis system (Oxford Metrics Ltd., Oxford, UK) for kinematic data capture, at a sampling frequency of 100 Hz. Ground reaction forces (GRFs) were recorded by two 3-dimensional AMTI OR6–7-1000 force plates placed in tandem in the center of a 10-m walkway, at a sampling frequency of 1000 Hz. Full-body kinematic data was collected simultaneously while subjects walked over the walkway. Subjects performed 6–8 successful walking trials in each shoe configuration. A successful trial was defined as one in which the dominant foot landed exclusively and entirely within the confines of one of the force plates. Methodology used to avoid intentional force plate targeting can be found in the Supplement. A standard marker set was used [31]. Joint angles were calculated based on marker locations using 'PlugInGait' (Oxford Metrics Ltd., Oxford, UK), and joint moments (normalized by body mass) were calculated via 'PlugInGait' using inverse dynamic analyses. A Woltring filter (MSE 10) was used to smooth marker trajectories. All analyses were performed for the subject-reported dominant limb (that used to kick a ball).

Upper and lower-body gait parameters were recorded with respect to both shoe configurations. Medio-lateral COP offset was defined as the perpendicular distance between the GRF origin on the force plate and the toe-heel line formed by the reflective markers [18]. Antero-posterior COP offset was defined as the perpendicular distance between the GRF origin on the force plate and a perpendicular line bisecting the toe-heel line. Additionally, ankle, knee, and hip angles and moments, and pelvis, spine, and shoulder angles were calculated. Mean and STD values were calculated for the set of 6–8 gait trials in each shoe condition during two substages of stance phase, Midstance (MS; 16.7–50 % stance phase) and Terminal Stance (TS; 50–83.3 % stance phase).

2.4. Statistical analysis

Data was analyzed using MATLAB. Wilcoxon signed-rank tests were used to compare parameters between shoe configurations. A *p*-value below 0.05 is considered statistically significant, and all *p*-values are two-sided.

3. Results

Difference in average and STD of walking speed between US and SS was not statistically significant (US speed = 1.354 ± 0.097 m/s, SS speed = 1.339 ± 0.098 m/s, $p_{speed} = 0.5614$; US speed STD = 0.035 m/s, SS speed STD = 0.045 m/s; $p_{speed\ STD} = 0.1688$).

3.1. Foot center of pressure

Mean and STD of COP offset are shown in Table 1B and A, respectively. Mean values of antero-posterior axis COP offset at MS and TS revealed a significantly more anterior offset in US, compared to SS (10 mm and 7 mm more anterior on average, respectively). There were no significant changes in mean medio-lateral COP offset, or STD of antero-posterior COP offset. Along the medio-lateral axis, STD values were significantly decreased in US compared to SS (45% and 53% for MS and TS, respectively) (Fig. 2A).

3.2. Kinetic and kinematic variables

Average gait parameters and STD values for MS and TS are shown in Table 2B and A, respectively. Significant differences in hip, ankle, spine, and shoulder kinematics were observed. Specifically, there was a significant decrease in mean hip flexion (10% at MS), increase in mean ankle dorsiflexion (56% and 8% at MS and TS, respectively) and inversion (21% at TS), increase in mean spine extension (11% at MS) and contralateral flexion (43% at TS), and decrease in mean internal spine rotation (67% at MS) in US. Additionally, there was a significant increase in variability of spine internal rotation angle (39% at MS) (Fig. 2B), and decrease in variability of shoulder adduction/abduction angle (38% and 49% at MS and TS, respectively) (Fig. 2C) in US.

Significant differences in ankle and knee kinetics were observed. Specifically, there was a significant increase in mean ankle dorsiflexion moment (13% and 7% at MS and TS, respectively) and knee extension moment (202% and 14% in MS and TS, respectively) in US. There was a significant decrease in variability of ankle internal rotation moment (24% and 22% at MS and TS, respectively) in US (Fig. 2D).

Table 1
Variability (A) and mean values (B) of medio-lateral and antero-posterior COP offset at Midstance (MS) and Terminal stance (TS).

	A. Variability (STD between steps) [mm]					
	MS			**TS**		
	UNST	**ST**	***p*-val**	**UNST**	**ST**	***p*-val**
***M-L axis**	2.252	4.142	***0.000**	2.380	5.148	***0.000**
A-P axis	10.840	9.460	0.303	7.711	8.800	0.135
	B. Mean Values [mm]					
	MS			**TS**		
	UNST	**ST**	***p*-val**	**UNST**	**ST**	***p*-val**
M-L axis	1.688	3.614	0.151	3.368	5.466	0.083
***A-P axis**	0.945	-9.089	***0.008**	77.645	70.353	***0.003**

*and bold indicates significant value ($p < 0.05$) or parameters for which significance was found for either variability or mean values. UNST = unstable configuration (shaded in gray), ST = stable configuration, M-L axis = medio-lateral foot axis, A-P axis = antero-posterior foot axis. Anterior and lateral directions are positive.

4. Discussion

The current study was conducted to quantitatively compare gait variability between stable and unstable footwear configurations. In accordance with our hypothesis, results showed less stride-to-stride variability during gait in US, as seen by reduced stride-to-stride STD in COP location, transverse-plane ankle moment, and frontal-plane shoulder angle. Contrary to our hypothesis, spine rotation angle exhibited greater variability in US. These results may reflect a compensatory mechanism for maintaining stability, specific to US in the investigation device.

A previous study examining variability of COP trajectory using the same device with different levels of instability (convexity of biomechanical elements), found less variation of stride-to-stride COP trajectory with increasing instability [17]. This is in agreement with the decrease in COP trajectory STD with US found in the current study. Decrease in variability of ankle and shoulder parameters with US may suggest that ankle and shoulder joints play a key role in maintaining stability during gait in the study shoe. It has been shown that unstable shoes increase ankle muscle activity during walking [32,33]. A stiffer joint may resist displacement from external perturbation, preventing excessive joint motion and injury. This may support the notion that the decreased ankle joint variability is related to increased muscle activation, resulting in a stiffer, less variable motion. This is corroborated by the increased magnitudes of ankle moment observed in US. Changes in ankle moment may have also been used to adjust COP in order to maintain stability [34]. With respect to the shoulder, further studies on upper-body movement during unstable gait are required for comparison, as they are currently lacking in the literature. However, variability was decreased in US, possibly indicating stiffening of the shoulder joint, and abduction was increased. These phenomena are consistent with a well-known mechanism of maintaining balance, in which one stiffly holds the arms further away from the body, such as when walking on a balance beam. The hip joint functions as a major load-bearing joint and is stable in nature, while the knee joint, although a less stable joint, is spanned by strong protecting muscles that maintain joint stability. We therefore suggest that there is no further need for compensation in these joints when presented with the investigated level of instability.

Although decrease in variability was found for COP location and ankle and shoulder parameters with US, increase in variability was found for transverse-plane spine angle. One might speculate that it is compensatory for the decreased variability at the ankle and shoulder, in order to maintain flexibility and adaptability, and hence global dynamic stability. Since the ankle and shoulders are relatively "locked" in a position to limit degrees of freedom for maintaining stability, as we assume may be the case here, then increased variability of spine rotation angle may be a necessary tradeoff for maintaining balance. This, however, remains to be investigated.

Our gait variability results were not in agreement with Stoggl et al. [22]. They compared STD due to short-term usage between unstable (MBT) and conventional shoes, and found greater variability with MBT. This dissimilarity is likely a factor of the difference in structure of the devices investigated. The present study device may be more stable along the antero-posterior axis in MS and TS, during which both elements are in contact with the ground, supporting the forefoot and hindfoot, while in the MBT shoes only one curved aspect of the shoe sole is in contact with the ground, providing the sagittal-plane rocker

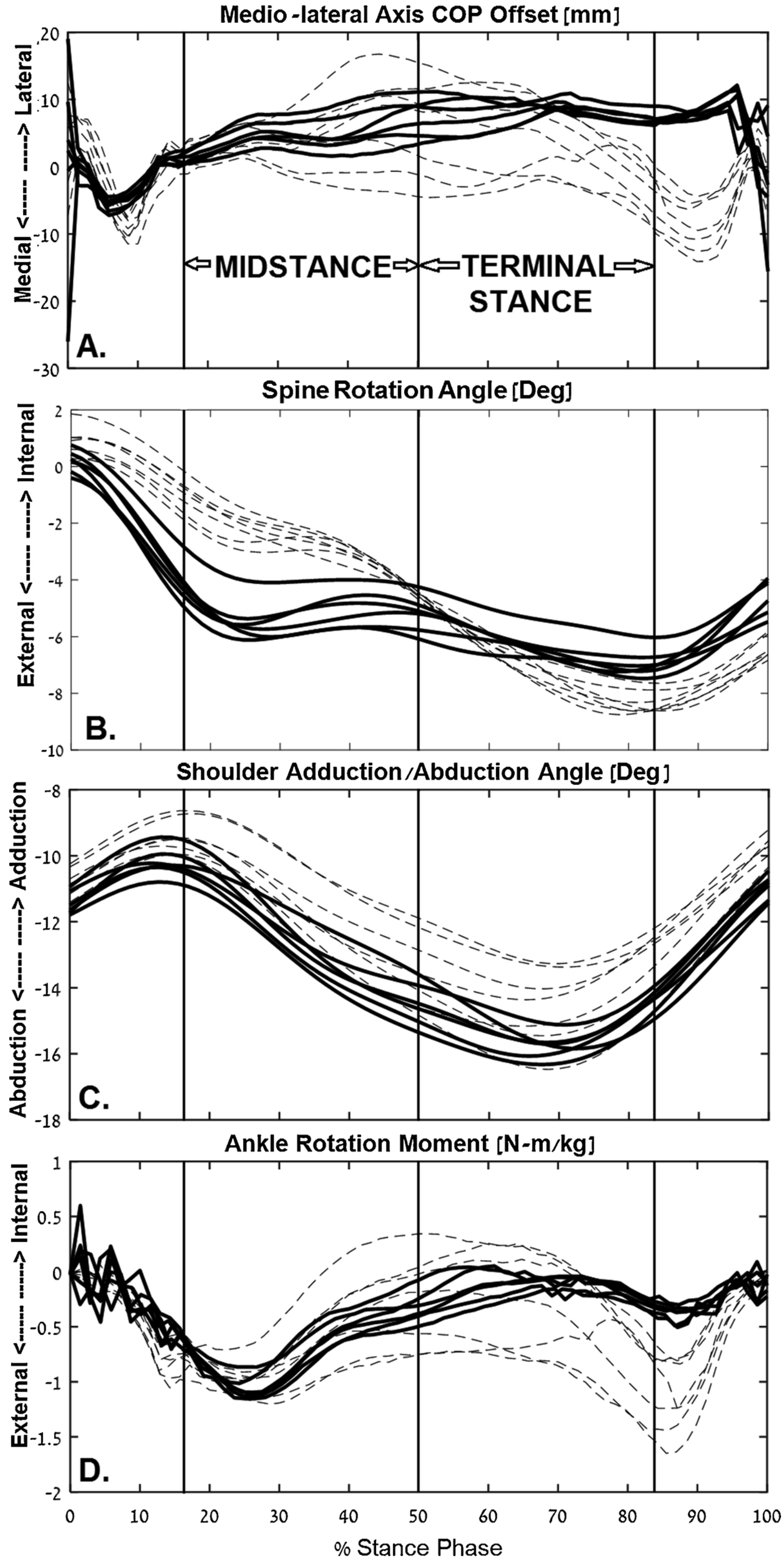

Fig. 2. Representative subject's graphs for (A) medio-lateral axis COP offset, (B) spine rotation angle, (C) shoulder adduction/abduction angle, and (D) ankle rotation moment for all gait trials in both the unstable shoe configuration (bold lines) and stable shoe configuration (dashed lines). The midstance and terminal stance substages, for which study parameters were reported, are indicated.

effect. Indeed, Stoggl et al. [22] found greater variability in multiple sagittal-plane gait measures, while we did not find any. In the medio-lateral direction, the convexity of the study device elements provides a rocker effect; however, variability in this plane could lead to falling. The adaptations in the ankle joint may occur prior to MS in order to reduce variability in MS, and this should be investigated further. Contrary to this, the MBT shoes have a soft heel, which may still be in contact with the ground during at least some portion of MS. This may

Table 2
Variability (A) and mean values (B) of gait kinetic and kinematic variables at Midstance (MS) and Terminal stance (TS).

	A. Variability (STD between steps)						B. Mean Values					
	MS			TS			MS			TS		
	UNST	ST	*p*-val	UNST	ST	*p*-val	UNST	ST	*p*-val	UNST	ST	*p*-val
						ANGLES [°]						
***Hip Flex(+)/Ext(-)**	1.448	1.510	0.679	0.921	0.873	0.762	7.540	8.418	***0.026**	-2.864	-2.687	0.454
Hip Add(+)/Abd(-)	0.935	0.732	0.083	0.645	0.604	0.679	1.423	1.532	0.389	0.883	1.193	0.073
Hip Int(+)/Ext(-) Rot	1.135	1.028	0.389	0.880	0.905	0.804	10.301	10.106	0.639	9.787	9.984	0.330
Knee Flex(+)/Ext(-)	1.650	1.751	1.000	1.108	1.210	0.679	6.396	6.459	0.978	1.438	1.259	0.599
Knee Add(+)/Abd(-)	0.696	0.724	0.847	0.426	0.413	0.639	3.727	3.990	0.454	1.939	1.978	0.890
Knee Int(+)/Ext(-) Rot	1.502	1.656	0.978	1.605	1.824	0.277	-6.288	-6.514	0.561	-8.893	-9.312	0.277
***Ankle D(+)/P(-) Flex**	0.831	0.901	0.330	0.813	0.877	0.561	2.431	1.559	***0.004**	10.221	9.421	***0.018**
***Ankle Inv(+)/Ev(-)**	0.361	0.364	0.303	0.504	0.546	0.978	0.975	0.759	0.135	2.551	2.113	***0.030**
Ankle Int(+)/Ext(-) Rot	1.718	1.844	1.000	2.399	2.089	0.229	-21.125	-20.620	0.639	-13.797	-12.671	0.135
Pelvis Ant(+)/Post(-) Tilt	0.815	0.732	0.489	0.766	0.724	0.421	10.159	9.800	0.169	10.970	10.623	0.135
Pelvis Obliq	0.531	0.422	0.277	0.425	0.390	0.804	1.909	1.849	0.454	-1.467	-1.370	0.847
Pelvis Int(+)/Ext(-) Rot	1.349	1.125	0.252	1.237	1.385	0.454	2.073	2.356	0.389	-1.253	-1.138	0.762
***Spine Flex(+)/Ext(-)**	0.934	1.049	0.303	0.788	0.955	0.229	-6.268	-5.634	***0.018**	-4.867	-4.437	0.359
***Spine Lat Flex**	0.816	0.866	0.890	0.878	0.816	0.489	3.316	3.215	0.762	-2.195	-1.540	***0.026**
***Spine Int(+)/Ext(-) Rot**	0.890	0.640	***0.004**	0.799	0.803	0.389	0.332	1.019	***0.012**	-4.222	-4.348	0.847
***Shoulder Flex(+)/Ext(-)**	1.384	1.666	0.121	2.263	2.816	***0.048**	-12.944	-12.851	0.890	3.326	2.042	0.083
***Shoulder Add(+)/Abd(-)**	0.932	1.226	***0.008**	1.037	1.333	***0.041**	-13.173	-12.550	***0.048**	-17.002	-16.632	0.359
Shoulder Int(+)/Ext(-) Rot	3.360	2.625	0.229	2.707	2.594	0.229	2.393	1.797	0.720	-6.910	-6.720	0.720
						MOMENTS [N-m/kg]						
Hip Flex(+)/Ext(-)	5.470	4.772	0.135	3.753	3.528	0.561	26.781	26.486	0.679	-14.374	-16.108	0.121
***Hip Add(+)/Abd(-)**	3.738	3.656	0.720	4.022	4.153	1.000	37.613	35.528	***0.005**	50.560	50.210	0.847
Hip Int(+)/Ext(-) Rot	0.958	0.960	0.720	1.003	1.080	0.762	-3.429	-3.263	0.454	9.538	10.031	0.121
***Knee Flex(+)/Ext(-)**	4.833	4.834	0.934	3.457	3.396	0.934	-1.329	1.298	***0.015**	-31.827	-27.956	***0.002**
Knee Add(+)/Abd(-)	2.846	2.696	0.421	3.059	4.028	0.083	33.269	32.352	0.229	17.824	16.923	0.847
Knee Int(+)/Ext(-) Rot	0.861	0.807	0.599	1.088	1.175	0.524	4.461	4.145	0.151	12.329	12.364	0.978
***Ankle D(+)/P(-) Flex**	5.428	5.068	0.561	4.742	5.500	0.208	39.639	35.139	***0.018**	103.137	96.162	***0.000**
Ankle Inv(+)/Ev(-)	0.862	0.767	0.489	1.072	1.201	0.330	4.402	4.178	0.277	12.707	12.809	0.890
***Ankle Int(+)/Ext(-) Rot**	1.407	2.266	***0.030**	1.557	3.071	***0.000**	-5.798	-3.878	***0.018**	0.927	1.730	0.252

*and bold indicates significant value ($p < 0.05$) or parameters for which significance was found for either variability or mean values. UNST = unstable configuration (shaded in gray), ST = stable configuration, Flex/Ext = flexion/extension, Add/Abd = adduction/abduction, Int/Ext Rot = internal/external rotation, D/P Flex = dorsi/plantar flexion, Inv/Ev = inversion/eversion, Ant/Post = anterior/posterior, Obliq = obliquity, Lat Flex = lateral flexion.

further explain why variability in gait parameters was increased with the MBT shoe, as the material properties of the shoe sole at the heel, as well as the structure of the sole, may give the user a feeling of uncertainty that provokes exploration of gait strategies to achieve optimal performance, while still providing structural stability. Thus, reduction in variability due to US in the present study may be unique to unstable shoes similar in structure. Additionally, Stoggl et al. [22] calculated variability over the whole stance phase, rather than specific substages. Calculations around initial contact and toe-off do not reflect the purpose of this study because they are influenced by the device structure. Specifically, around initial contact and toe-off, the curved aspect of the edges of the posterior or anterior hemispherical elements, respectively, come in contact with the floor, and present a different instability condition than the one we wished to examine. Therefore, we presented data at MS and TS only, during which both hemispheres are relatively parallel to the ground.

Results for gait parameters (Table 2B) showed significantly greater ankle dorsiflexion, ankle dorsiflexion moment, and knee extension moment in US (MS and TS). This is in agreement with a previous study from our research group [35] that found that an anteriorly-translated COP resulted in the same changes in these parameters that we observed in US, in which COP was significantly more anterior. However, in the present study, since COP was not intentionally displaced, as was in Haim et al. (2010), we cannot discern whether changes in sagittal-plane parameters were a result of an anteriorly-displaced COP, due to configuration of the device elements, or vice versa.

Significant effect of MBT footwear on lower-limb biomechanics has been previously shown in a systematic review undertaken by Tan et al. [12]. Their meta-analysis exhibited notable differences in effects. They found some significant effects on sagittal-plane hip and knee kinematics that indicated a trend towards reduced motion, whereas we did not. Additionally, they found decreased dorsiflexion in TS, while we found increased dorsiflexion in MS and TS. Tan et al. noted, however, that ankle results were contradictory, and some studies found increased dorsiflexion throughout stance phase, similar to our results. It must be acknowledged that parameters analyzed in Tan et al. were slightly different than those in our study in some cases (i.e., peak values, values pooled over the entire stance phase, etc.), and they did not have available data for all of parameters that were analyzed in our study. Farzadi et al. [13] undertook a systematic review of unstable shoes in general, revealing varying impact on gait between studies, and suggested that it may be due to different adaptation strategies in the presence of instability, as well as certain strategies adopted by different test populations, among other factors such as gender, age, and time allowed for adaptation to the shoes. They also suggested that knee and hip sagittal-plane kinematics tend to change in accordance with changes in ankle dorsi/plantarflexion. We did not observe a similar phenomenon, and thus the device and unstable configuration of our study may minimize variation of adaptation strategies among subjects, and preserve a more homogenous gait. This remains to be analyzed in different study cohorts.

In studies of this type, one might question whether an increase or decrease in variability is the favorable outcome with unstable shoes. The answer is beyond the scope of this study; however, it might depend on the treatment objectives or type of unstable footwear. For example, the makers of the MBT shoe claim that it provides an unstable surface for walking, simulating barefoot walking on soft ground, which was suggested to have advantages for the locomotor system [19]. The MBT shoe is thus marketed to the general public as well as pathological populations, having exhibited clinical benefit, or potential clinical benefit, such as reduction in joint pain in osteoarthritis and potentially improved posture control [8,12,36]. The shoe investigated in the present study was designed for treatment purposes and shifts COP [30] in such a way to alter biomechanics in a manner that provides clinical benefit, or potential clinical benefit, such as joint load reduction and decreased pain [9,10,20]. Whereas the MBT shoe cannot be modified, the convex elements of the study device can be custom-positioned for optimal clinical benefit. The element of instability provoked by the study shoe "forces" the user to adopt a specific COP location that optimizes biomechanics associated with clinical benefit. The simultaneous instability is meant to induce motor learning to retrain gait pattern. Indeed, training in the study shoe causes positive changes in gait pattern over a one-year period when measured during barefoot walking [9,10]. Training in MBT shoes over a ten-week period caused reduction in variability, measured while walking in the MBT shoe, compared to a conventional shoe, also suggesting motor learning and gait retraining [22,23]. Yet, to date there have not been any studies that explicitly correlate the clinical benefits with change in gait variability. Further study is required to fulfill this objective.

It should be noted that the instability level in the present study was within a range that is not overly challenging to the neuro-muscular system. According to our results, healthy adults may cope with the challenge by enhancing muscle activity and stiffening the joints. Higher levels of instability could require different strategies, and therefore should be separately investigated. Additionally, it should be noted that prolonged use of the study device in the unstable configuration might alter its impact. Stoggl et al. reported that long-term usage caused the variability to decrease to that of a conventional shoe [22], later supported by Stoggl and Müller [23]. Finally, the results are applicable to our distinct cohort of healthy young males only.

5. Conclusion

The present study exhibits significantly decreased variability of gait parameters at the two extremes of the kinematic chain (i.e., foot, ankle, and shoulders) while walking in a unique unstable shoe. The induced instability may be met with an attempt to limit degrees of freedom at the ankle, via control of ankle moment, in order to maintain dynamic stability. Decreased variability of frontal-plane shoulder angle in the unstable shoe, along with increased shoulder abduction, may signify a well-known compensatory mechanism to maintain stability by holding the arms further away from the body, as when walking on a balance beam. Results also exhibited a concurrent increase in variability of transverse-plane spine angle, which may be employed in order to maintain adaptability and flexibility, thus maintaining global dynamic stability. The study results are likely a product of the specific footwear design, and could have substantial implications in design and use of unstable training devices. Further study is required to confirm the proposed mechanism and determine potential clinical benefits of the changes in variability with the unstable shoe design, and is crucial for understanding dynamic stability and its relationship to variability of gait measures.

Conflict of interest

None.

Acknowledgement

The authors thank APOS–Medical and Sports Technologies Ltd. for their generosity in contributing the devices used in the study.

References

[1] M.P. Kadaba, H.K. Ramakrishnan, M.E. Wooten, J. Gainey, G. Gorton, G.V.B. Cochran, Repeatability of kinematic, kinetic, and EMG data in normal adult gait, J. Orthop. Res. 7 (1989) 849–860.
[2] J.M. Hausdorff, Gait dynamics, fractals and falls: finding meaning in the stride-to-stride fluctuations of human walking, Hum. Mov. Sci. 26 (2007) 555–589.
[3] M.J.P. Toebes, M.J.M. Hoozemans, R. Furrer, J. Dekker, J.H. Van Dieën, Local dynamic stability and variability of gait are associated with fall history in elderly subjects, Gait Posture 36 (2012) 527–531.
[4] K.M. Newell, D.M. Corcos, Variability and Motor Control, Human Kinetics, Champaign, IL, 1993.
[5] R. Bartlett, J. Wheat, M. Robins, Is movement variability important for sports biomechanists? Sport Biomech. 6 (2007) 224–243.
[6] H.R. Fallah Yakhdani, H.A. Bafghi, O.G. Meijer, S.M. Bruijn, N. van den Dikkenberg, A.B. Stibbe, B.J. van Royen, J.H. van Dieën, Stability and variability of knee kinematics during gait in knee osteoarthritis before and after replacement surgery, Clin. Biomech. 25 (2010) 230–236.
[7] M.D. Lewek, J. Scholz, K.S. Rudolph, L. Snyder-Mackler, Stride-to-stride variability of knee motion in patients with knee osteoarthritis, Gait Posture 23 (2006) 505–511.
[8] B.M. Nigg, C. Emery, L.A. Hiemstra, Unstable shoe construction and reduction of pain in osteoarthritis patients, Med. Sci. Sports Exerc. 38 (2006) 1701–1708.
[9] D. Solomonow-Avnon, A. Herman, D. Levin, N. Rozen, E. Peled, A. Wolf, Positive outcomes following gait therapy intervention for hip osteoarthritis: a longitudinal study, J. Orthop. Res. 35 (2017) 2222–2232.
[10] A. Haim, G. Rubin, N. Rozen, Y. Goryachev, A. Wolf, Reduction in knee adduction moment via non-invasive biomechanical training: a longitudinal gait analysis study, J. Biomech. 45 (2012) 41–45.
[11] P. Federolf, B. Nigg, Unstable shoes, Footwear Sci. 4 (2012) 71–72.
[12] J.M. Tan, M. Auhl, H.B. Menz, P. Levinger, S.E. Munteanu, The effect of Masai Barefoot Technology (MBT) footwear on lower limb biomechanics: a systematic review, Gait Posture 43 (2016) 76–86.
[13] M. Farzadi, Z. Nemati, M. Jalali, R.S. Doulagh, M. Kamali, Effects of unstable footwear on gait characteristic: a systematic review, Foot. 31 (2017) 72–76.
[14] D. Solomonow-Avnon, D. Levin, M. Elboim-Gabyzon, N. Rozen, E. Peled, A. Wolf, Neuromuscular response of hip-spanning and low back muscles to medio-lateral foot center of pressure manipulation during gait, J. Electromyogr. Kinesiol. 28 (2016) 53–60.
[15] D. Solomonow-Avnon, A. Wolf, A. Herman, N. Rozen, A. Haim, Reduction of frontal-plane hip joint reaction force via medio-lateral foot center of pressure manipulation: a pilot study, J. Orthop. Res. 33 (2015) 261–269.
[16] Y. Goryachev, E.M. Debbi, A. Haim, A. Wolf, The effect of manipulation of the center of pressure of the foot during gait on the activation patterns of the lower limb musculature, J. Electromyogr. Kinesiol. 21 (2011) 333–339.
[17] M. Khoury, A. Haim, A. Herman, N. Rozen, A. Wolf, Alteration of the foot center of pressure trajectory by an unstable shoe design, J. Foot Ankle Res. 8 (2015) 67.
[18] A. Haim, N. Rozen, S. Dekel, N. Halperin, A. Wolf, Control of knee coronal plane moment via modulation of center of pressure: a prospective gait analysis study, J. Biomech. 41 (2008) 3010–3016.
[19] B. Nigg, S. Hintzen, R. Ferber, Effect of an unstable shoe construction on lower extremity gait characteristics, Clin. Biomech. 21 (2006) 82–88.
[20] D. Solomonow-Avnon, A. Haim, D. Levin, M. Elboim-Gabyzon, N. Rozen, E. Peled, A. Wolf, Reduction of hip joint reaction force via medio-lateral foot center of pressure manipulation in bilateral hip osteoarthritis patients, J. Orthop. Res. 34 (2016) 1762–1771.
[21] H. Tateuchi, M. Taniguchi, Y. Takagi, Y. Goto, N. Otsuka, Y. Koyama, M. Kobayashi, N. Ichihashi, Immediate effect of Masai Barefoot Technology shoes on knee joint moments in women with knee osteoarthritis, Gait Posture 40 (2014) 204–208.
[22] T. Stöggl, A. Haudum, J. Birklbauer, M. Murrer, E. Müller, Short and long term adaptation of variability during walking using unstable (Mbt) shoes, Clin. Biomech. 25 (2010) 816–822.
[23] T. Stöggl, E. Müller, Magnitude and variation in muscle activity and kinematics during walking before and after a 10-week adaptation period using unstable (MBT) shoes, Footwear Sci. 4 (2012) 131–143.
[24] P. Federolf, K. Tecante, B. Nigg, A holistic approach to study the temporal variability in gait, J. Biomech. 45 (2012) 1127–1132.
[25] U. Granacher, R. Roth, T. Muehlbauer, R.W. Kressig, T. Laser, K. Steinbrueck, Effects of a new unstable sandal construction on measures of postural control and muscle activity in women, Swiss Med. Wkly. 141 (2011).
[26] C. Apps, T. Sterzing, T. O'Brien, R. Ding, M. Lake, Biomechanical locomotion adaptations on uneven surfaces can be simulated with a randomly deforming shoe midsole, Footwear Sci. 9 (2017) 65–77.
[27] J.B. Dingwell, J.P. Cusumano, P.R. Cavanagh, D. Sternad, Local dynamic stability versus kinematic variability of continuous overground and treadmill walking, J. Biomech. Eng. 123 (2001) 27.
[28] J.B. Dingwell, L.C. Marin, Kinematic variability and local dynamic stability of upper body motions when walking at different speeds, J. Biomech. 39 (2006) 444–452.
[29] A.S.P. Sousa, A. Silva, R. Macedo, R. Santos, J.M.R.S. Tavares, Influence of long-term wearing of unstable shoes on compensatory control of posture: an electromyography-based analysis, Gait Posture 39 (2014) 98–104.
[30] M. Khoury, A. Wolf, E.M. Debbi, A. Herman, A. Haim, Foot center of pressure trajectory alteration by biomechanical manipulation of shoe design, Foot Ankle Int. 34 (2013) 593–598.
[31] M.P. Kadaba, H.K. Ramakrishnan, M.E. Wootten, Measurement of lower extremity kinematics during level walking, J. Orthop. Res. 8 (1990) 383–392.
[32] C. Price, L. Smith, P. Graham-Smith, R. Jones, The effect of unstable sandals on instability in gait in healthy female subjects, Gait Posture 38 (2013) 410–415.
[33] J. Romkes, C. Rudmann, R. Brunner, Changes in gait and EMG when walking with the Masai Barefoot Technique, Clin. Biomech. 21 (2006) 75–81.
[34] M. Vlutters, E.H.F. van Asseldonk, H. van der Kooij, Center of mass velocity-based predictions in balance recovery following pelvis perturbations during human walking, J. Exp. Biol. 219 (2016) 1514–1523.
[35] A. Haim, N. Rozen, A. Wolf, The influence of sagittal center of pressure offset on gait kinematics and kinetics, J. Biomech. 43 (2010) 969–977.
[36] P.R. New, J.M. Pearce, The effects of Masai Barefoot Technology footwear on posture: an experimental designed study, Physiother. Res. Int. 12 (2007) 202.