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The effect of simulated leg length discrepancy on lower limb biomechanics during gait

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ABSTRACT

Understanding the effects of leg length discrepancy (LLD) on the biomechanics of gait and determining as to what extent of LLD alters gait is essential. A total of 91 biomechanical data were assessed from 14 lower limbs of healthy individuals walking under random conditions: shod only and with a 5, 10, 15, 20, 30 and 40 mm sole lift. Lower limb kinematics and dynamic leg length (DLL) were measured by a motion capture system. Hotelling's T-Square test was used to evaluate the differences in DLLs throughout the gait cycle in conjunction with differences between the sides based on the maximal stance phase and minimal swing phase DLLs. Kinematics were compared using the one-way blocked analysis of variance and Post-hoc analysis by the paired *t*-test. Significant dynamic shortening of the longer limb, mainly during the swing phase, and significant change in maximal stance and minimal swing phase DLL relationship started at a 10 mm lift condition ($p < 0.05$). Thirteen kinematic variables produced a significant angular main effect ($p < 0.05$), with a more flexed position of the longer limb and extended shorter limb beginning at a 5 mm lift. An increase in hip abduction and external foot rotation during the swing phase was also found. This study demonstrates that simulated LLD, as low as 5 mm, causes biomechanical changes in the lower limbs during gait revealed in both kinematics and dynamic leg length, suggesting that LLD, as small as 5–10 mm, should not be ignored.

1. Introduction

Leg length discrepancy (LLD) is a common condition due to either structural deformities originating from true bony leg length differences [1] or a functional deformity originating from abnormal hip, knee, ankle or foot movements [2]. LLD has been associated with several pathological conditions, such as foot pathologies [3], stress fractures and running injuries [4], low back pain [5] and osteoarthritis of the hip and knee joints [6]. Therefore, understanding the effects of LLD on the biomechanics of gait may help illuminate the development of injuries and assist in determining the need for equalization of leg length in preventing injuries and symptom relief.

However, there is still a negligible consensus as to the extent of LLD believed to have an effect on gait [7–11]. Studies have tackled this dispute by evaluating the effects of simulated LLD on lower limb biomechanics during gait. In a recent systematic review [12], the authors concluded that gait deviations might begin to occur at a discrepancy of > 10 mm (hereafter, significant LLD) and rise as LLD increases. Eight studies [1,11,13–18] evaluating the effect of LLD on gait by using lifts to simulate LLD were reviewed. Lift heights ranged between 10 and 50 mm. Only 1 study evaluated the effect of a 10 mm lift on lower limb kinematics and found compensations at the pelvis, knee and ankle, leading to an asymmetrical gait [1]. No studies evaluated the effect on any gait parameter during a lift condition of < 10 mm. The evidence is still limited, therefore, further proof is needed to support and evaluate the effect of LLD ≤10 mm (hereafter, mild LLD) on gait. Moreover, studies evaluated the variations in gait biomechanics by measuring the changes in lower limb kinematics [1,11,13–18]. This could be further substantiated by measuring the functional changes in leg length during gait by measuring the sum effect of gait kinematics on the dynamic leg length (DLL) [19,20]. At present, this is the only functional measurement for LLD suggested in the literature. Measuring methods such as radiography [21] and current clinical accepted methods [22,23] are not proficient in measuring the dynamic changes in leg length, since they are performed in a single static position.

The goal of this study was to investigate the effects of simulated mild to significant LLD on lower limb kinematics and on the functional leg length by implementing a DLL measurement. We hypothesized that participants would alter their gait pattern to dynamically lengthen the shorter limb and conversely, shorten the longer limb which would be

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reflected in both kinematics and DLL.

2. Methods

2.1. Participants

Seven healthy participants, average age 23 (19–27), weight 62.5 kg and 168 cm height were recruited via a convenience sampling. Inclusion criteria included no history of musculoskeletal injury or pain during the last year and a European shoe size of 39–45 (only these lift sizes were available for data collection). In addition, a musculoskeletal examination was performed by the main investigator which included an assessment of the range of motion, muscle-tendon length and skeletal alignment in order to exclude any lower limb malalignment (i.e., asymmetrical hyper pronated feet [24], limited range of motion or structural LLD > 5 mm). The study was approved by the Medical Center's Ethics Committee.

2.2. Procedures

Participants underwent a gait laboratory evaluation using a three-dimensional motion analysis system (Vicon®, Oxford Metrics, UK), according to the PlugInGait model (PGM) [25], with a sampling rate of 120 Hz. Thirteen reflective passive skin markers were placed on the subject's pelvis and lower limbs according to the PGM protocol.

All 7 participants were evaluated while walking during the following 13 random conditions: shod only with their own sneakers, and another 6 conditions for each leg separately with an extra sole lift of 5, 10, 15, 20, 30 and 40 mm (Image 1). By measuring 7 participants and intervening on both lower limbs, a total of 14 lower limbs and 91 measurements were randomly assessed for all participants. The long limb was defined as the leg with a lift; the contra lateral limb was defined as the short limb. Simulated LLD was achieved by placing a shoe lift along the entire length of the shoe (from heel to toe) to simulate an effect throughout the gait cycle. The lift was made of a high-density ethylene vinyl acetate material and attached with velcro straps to the bottom of the subject's own shoes, hence, no marker positions were altered between conditions. Participants were given 2 min to acclimate to each lift and only after concurring that they were comfortable with the lift, data capture began along a 14 m gait lab walkway. Participants walked at their self-selected speed and rested for 2 min between data collections of each condition. The order of data collection was randomized between sides and heights of the lift. Six gait cycles for the left and right side were randomly sampled for analysis.

2.3. Data reduction and analysis

Dynamic leg length (DLL) was measured as the absolute distance from the hip joint center to the heel marker (HDLL), to the ankle joint center (ADLL) and the forefoot marker (FDLL) [19,20]. Virtual trajectories of the hip and ankle joint centers were determined by the PGM and heel and forefoot markers were placed according to the PGM protocol [26]. Three trajectories facilitated the measurement of leg length throughout the gait cycle, where each measurement interacted in a different manner with lower limb kinematics and the gait cycle (Image 2) [19,20].

DLLs were measured throughout the gait cycle. In addition, maximal stance and minimal swing phase DLLs were also used. Mid to terminal stance was defined as the stance phase; mid-swing was defined as the swing phase. The maximal stance and minimal swing phase DLLs values were chosen due to the mechanical requirements of gait. Those values would be most affected where maximal functional leg length during stance is required to clear the contralateral side and the minimal swing phase length is required to clear the foot off the ground. Thus, three DLL components were analyzed:

1. Differences in DLLs between sides throughout the gait cycle (longer minus shorter limb) to detect time of significant differences with respect to lift heights.
2. Ratio between maximal stance and minimal ipsilateral swing phase DLLs measuring changes in the functional length of each lower limb.
3. Differences between maximal stance and minimal contralateral swing phase DLLs to evaluate the interaction between sides.

Gait kinematic data included hip flexion-extension, adduction-abduction, internal-external rotation, knee flexion-extension, internal-external rotation, ankle dorsiflexion-plantarflexion, foot internal-external rotation with respect to the lab. Maximal values were analyzed at defined events of the gait cycle: initial contact, loading response, mid-stance, terminal-stance, foot off and swing phase. Initial contact and foot off were determined using the vertical ground reaction force. Loading response was defined as 0%–10% of the gait cycle, mid-stance defined as 10%–30% and terminal stance as 30%–50% [27]. Gait variables were normalized to 51 data points.

2.4. Statistical analysis

The Shapiro-Wilk test verified the normality assumption. Differences in DLLs throughout the gait cycle were assessed using the one-sample Hotelling's T-Square test at 51 sample points during the gait cycle to find the difference between the longer and shorter limb when lifts were placed on the left side (classified as group A) and when the lifts were placed on the right side (classified as group B). The Hotelling's T-Square test was also used to evaluate the significance of the difference between the shorter and longer limb by obtaining the ratios between the maximal DLLs achieved during the stance phase compared to the minimal ipsilateral DLLs during the swing phase (MaxStance/MinSwing Ratio) and finding the differences between the maximal DLLs during the stance phase and the contra lateral minimal DLLs during the swing phase (MaxLong-CONTRALATMinShort Diff.). P-values were corrected by the Benjamin-Hochberg (BH) procedure, thus guaranteeing a false discovery rate (FDR) control of 0.05 per measure. Differences were also correlated with the heights of the lifts by the Spearman's rank correlation.

One-way blocked analysis of variance (ANOVA) was used to screen the kinematic changes that are affected from the lift heights while adjusting to the participants' random effect, lift side and their interaction. The lift height p-values obtained from these models were BH adjusted. Post-hoc analysis by a paired t-test evaluated the angular changes between each mode and shod condition. Significance level was set at 0.05.

3. Results

3.1. DLLs differences between sides throughout the gait cycle

No statistically significant difference was found in the DLL throughout the gait cycle when walking shod only and with a 5 mm lift condition (Fig. 1). A negative significant difference indicated a significant shortening of the longer limb, elongation of the shorter limb or both, whereas a positive significant difference indicated an opposite change in the DLL. A negative significant difference started at a 10 mm lift condition in FDLL during terminal stance to mid-swing phase (Fig. 1). A rise in the DLL difference occurred as the lift height increased, mainly during the swing phase but also during the mid-stance phase in the ADLL (30 mm condition) and HDLL (20–40 mm conditions). However, a positive significant difference occurred in conditions of 20–40 mm height lifts in FDLL during terminal swing to early mid-stance phase and in HDLL during the pre-swing-initial swing phase.

3.2. Maximal stance to ipsilateral minimal swing phase DLLs ratios

All subjects when walking shod presented a similar maximal stance