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# The Effects of Unilateral Ankle Loading on the Long-Range Correlation of Spatiotemporal Gait Parameters during Treadmill Walking

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**The Effects of Unilateral Ankle Loading on the Long-Range Correlation of Spatiotemporal Gait Parameters during Treadmill Walking**

A THESIS

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By

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## Abstract

Long-range correlation has been observed in the time series of the human neuromuscular system and indicative of a healthy system. However, this analysis has not been used for an scenario of asymmetrical loading during walking. This study aimed to understand the effect of unilateral ankle loading on long-range correlation of spatiotemporal gait parameters in healthy young adults during treadmill walking. We used four unilateral ankle loads (A0, A25, A50, and A75, representing the increase of the moment of inertia of the leg about the knee joint by 0%, 25%, 50% and 75%, respectively) and attached it on the non-dominant leg. We used a modified lower-extremity marker set to collect kinematic data. Subjects walked on a treadmill at their self-selected speed for five minutes under each load condition. For data analysis, we used the toe and heel markers to identify the gait events of heel strike and toe off. We divided each five minute trial into three 100-second segments to investigate potential time effect. We calculated the mean and standard deviation of step length and step time, and conducted a detrended fluctuation analysis (DFA) and calculated the scaling exponent for every 100 seconds. For statistical analysis, we conducted three-way (2 side x 3 time x 4 load) ANOVA with repeated measures on step length and time. Our results demonstrate that mean step time and step length showed three-way interaction, in which step length increased with the load for both sides whereas step time increased on the loaded side but decreased on the unloaded side. Asymmetry between two legs decreased in step length while asymmetry in step time remained over time and across load conditions. DFA results revealed long-range correlation in step length and step time; however, step length showed only a time effect whereas step time showed neither a time nor a load effect. Our results suggest that there might be different control mechanisms on regulaitng spatiotemporal varialbes and its long-range correlation while walking with unilateral ankle load.

**Keywords:**

Walking, treadmill, ankle load, long-range correlation, detrended fluctuation analysis, unilateral loading.

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## Introduction

Long-range correlation has been observed in the time series of the human neuromuscular system and indicative of a healthy system. Walking is an essential physical activity for daily life and is a complex activity that requires coordinated motions at the lower limbs to move the body forward while maintaining balance in an upright position. While a healthy population often displays a relatively symmetrical pattern between the left and right sides, this gait symmetry can be compromised due to pathology or injury (Sadeghi, 2003; Sadeghi, Allard, & Duhaime, 1997; Sadeghi, Allard, Prince, & Labelle, 2000). One method of augmenting gait asymmetry is by loading one leg but keeping the other leg intact. This unilateral loading paradigm has been utilized to investigate the effects of inertial manipulations on gait patterns in both clinical populations (Mattes, Martin, & Royer, 2000; Selles et al., 2004; J. D. Smith & Martin, 2013; J. D. P. M. Smith, Philip E. PhD, 2011) and healthy individuals (Haddad, van Emmerik, Whittlesey, & Hamill, 2006; Kodesh, Kafri, Dar, & Dickstein, 2012; Noble & Prentice, 2006; J. D. Smith & Martin, 2007; J. D. Smith, Royer, & Martin, 2013; J. D. Smith, Villa, & Heise, 2013). In contrast to many studies on the effect of asymmetrical loading on spatiotemporal gait parameters, little is known to what degree asymmetrical loading affects the long-range correlation among the hundreds of steps during treadmill walking.

## The Statement of the Question

This study aimed to understand the effect of unilateral ankle loading on the long-range correlation of spatiotemporal gait parameters in healthy young adults during treadmill walking.

**The Rationale**

Typical gait of healthy adults has often been characterized as a symmetrical pattern (J. Hamill 1984; Sadeghi et al., 2000). However, patterns of subtle asymmetry exist in both kinematic and kinetic measurements (Hirokawa, 1989; Sadeghi, 2003; Sadeghi et al., 1997; Sadeghi et al., 2000). Hirokawa (1989) suggested that during walking the dominant limb primarily provides a propulsive force during push-off while the non-dominant limb primarily supports the posture. Sadeghi et al. (1997) reported that propulsion was related to the limb with predominantly muscle power generation whereas support and control functions were associated with the limb having predominantly power absorption behavior. A further study by Sadeghi (2003) showed the lower limbs generally behave symmetrically when the total behavior of the limbs is considered and the existence of local asymmetry due to the functional discrepancy between the lower limbs is recognized as locomotor compensation.

Changes in gait symmetry at the interlimb level may have clinical significance for patients who suffer from movement disorders such as hemiplegia (MP Griffin, 1995) and amputation (David J. Sanderson 1997; Sadeghi, Allard, & Duhaime, 2001; D. A. Winter & Sienko, 1988) and typically display a more asymmetrical gait. Clinical efforts have been commonly exerted to improve gait symmetry and overall motor function in these patients (MP Griffin, 1995; D. A. Winter & Sienko, 1988). Griffin et al. (1995) examined the symmetry properties for 34 gait variables at a self-selected walking speed in a group of 31 hemiplegic subjects and found symmetry played a weak role in promoting walking speed in these subjects. Several other studies investigated gait asymmetry between the residual limb and the intact limb in people with unilateral amputation by loading the residual limb with the same inertial property as that of the intact limb, and found greater gait asymmetry and higher energy cost for the matched inertia loading (Mattes et al., 2000; J. D. Smith & Martin, 2013; J. D. P. M. Smith, Philip E. PhD, 2011). Winter et al. (1988) suggested that the terms of motor adaptations and compensations are more relevant for gait rehabilitation as people with lower limb amputation may adopt a new asymmetrical pattern within the constraints of the residual limb and the prosthesis. In below-knee amputee gait, for example, hip extensors on the residual limb compensate for the loss of ankle function during push-off, and the intact limb also displays compensatory muscle activities to control the collapse of the trunk promptly (David J. Sanderson 1997; Sadeghi et al., 2001; D. A. Winter & Sienko, 1988).

External loading at lower extremities has been widely used for gait perturbation study since it increases the moment of inertia of the loaded leg, which affects both kinematic and kinetic patterns during the initial and end of the stance phase and the whole swing phase (Royer & Martin, 2005; Selles, Bussmann, Wagenaar, & Stam, 1999). Hence, it lengthens swing time and decreases stance time of the loaded leg, and the unloaded leg is conversely affected. Kinematic and kinetic changes of asymmetrical loading are well documented in previous research (Haddad, van Emmerik, Whittlesey, & Hamill, 2006; Noble & Prentice, 2006; J. D. Smith & Martin, 2007; J. D. Smith, Royer, & Martin, 2013; J. D. Smith, Villa, & Heise, 2013). It has been reported that walking patterns adopt a new inertial property in a quite short time (40~50 strides) and reach a new steady state (Noble & Prentice, 2006; J. D. Smith & Martin, 2007; J. D. Smith, Villa, et al., 2013). However, previous gait asymmetry adaptation studies mainly focused on the effect of on and off of one load hence it is still not clear to what degree different loads affect the adaptation pattern.

Despite many studies investigating the influence of lower limb inertia on walking dynamics in both healthy (Haddad et al., 2006; Kodesh et al., 2012; Noble & Prentice, 2006; J.

D. Smith & Martin, 2007; J. D. Smith, Royer, et al., 2013; J. D. Smith, Villa, et al., 2013) and clinical populations (Mattes et al., 2000; Selles et al., 2004; J. D. Smith & Martin, 2013; J. D. P. M. Smith, Philip E. PhD, 2011)  the efforts were mainly concentrated on the mean and step-to-step variability (such as standard deviation or coefficient of variation) of kinematic and kinetic variables. From a dynamical systems perspective, fluctuation at one-time point (one walking step) may have a profound effect on the performance at another time point (another walking step which is hundreds of steps apart) (Dingwell & Cusumano, 2010). This long-range correlation may shed light on the neuromuscular mechanisms regulating gait dynamics. However, to our knowledge, the effect of unilateral loading on the long-range correlation of spatiotemporal varaibles is largely unknown, particularly under different loading conditions.

Long-range correlation has been observed in the time series of various human neuromuscular system. This correlation is a fractal-like, scale-free fluctuation correlation and has been widely applied to diverse fields such as DNA sequence (Peng et al., 1992; Peng et al., 1994), heart rate dynamics (Peng, Havlin, Hausdorff, et al., 1995; Peng, Havlin, Stanley, & Goldberger, 1995), brain activities (R. J. Smith et al., 2017), and stride interval of human walking (Dingwell & Cusumano, 2010; Hausdorff, Peng, Ladin, Wei, & Goldberger, 1995; Hausdorff et al., 1996). When the system deteriorates due to disease (Hausdorff et al., 1997) or external constraints (Bohnsack-McLagan, Cusumano, & Dingwell, 2016; Hausdorff et al., 1996), time series fluctuations become more uncorrelated or random. Based on these findings, it has been argued that long-range correlation is regulated by the central nervous system, and this theory was further supported by remained long-range correlations in patients with severe diabetic peripheral neuropathy (Gates & Dingwell, 2007).

Detrended fluctuation analysis (DFA) is a common approach to calculating fluctuation correlations, and it has been widely used to quantify stride interval variability in human walking (Dingwell & Cusumano, 2010; Hausdorff et al., 1995; Hausdorff et al., 1996). DFA yields a scaling exponent, α, that quantifies not only the randomness of the system but also the statistical persistence or anti-persistence in a scalar time series. The scaling exponent provides the information on how deviations in time series are statistically correlated within small or large time windows (Dingwell & Cusumano, 2010). While persistent fluctuations have been argued as a critical marker of the healthy physiological system, Dingwell and Cusumano (2010) suggested that uncorrelated or anti-persistent may reflect the tightness of control against the constraints of the system.

Taken together, we propose that understanding the effect of unilateral ankle loading on the long-range correlation of spatiotemporal gait parameters will provide further insight into neuromotor control strategies for walking. Specifically, we expected that different spatiotemporal gait patterns will be observed in their mean values and reflected in the long-range correlation. In this study, we manipulated the amount of unilateral ankle load during 5-minute treadmill walking and investigated the effect of unilateral ankle loading on the long-range correlation of spatiotemporal gait parameters using DFA.

**The Hypotheses**

1. As the unilateral ankle loading increases, step time and length will increase on the loaded leg and decrease on the unloaded leg. Step time and length asymmetry will decrease over the 5-minute walking trial.

2. As the unilateral ankle loading increases, the scaling exponent of DFA will increase on the loaded leg and decrease on the unloaded leg. The scaling exponent asymmetry will decrease over the 5-minute walking trial.

**Delimitations and Limitations**

Delimitations

This study included college-age male healthy subjects. The subjects were not diagnosed with any mental or physical condition that would prevent them from walking on a treadmill. Exclusion criteria for potential participants :

- Inability to walk on the treadmill for 20 minutes.
- Preexisting injuries or conditions that would prevent them from walking on a treadmill.
- Preexisting injuries or conditions that could be adversely affected by physical activity.
- Current medications that would affect movement, mental capacity or concentration.

Our data provided insight into healthy young male adult population's motor control strategies for asymmetrical inertia manipulation. Our results may not be generalizable to people with different sexes and/or at different ages or people with physical or intellectual disabilities or other medical conditions.

Limitations

We used each subject's preferred treadmill speed in this study. Therefore, our reesuls may not be generalizable to different speeds of treadmill walking. Further, we acknowledge that there may be differences in motor strategy and gait pattern between overground and treadmill walking in our subjects. Our results may not be generalizable to overground walking at the same or different speeds.

## Literature Review

Walking is a complex movement pattern that requires coordinating of the lower limbs to propel the body forward while maintaining balance in the upright position. An effective coordination strategy depends on the central nervous system's ability to use appropriate muscle activation patterns to achieve the desired movement trajectory. Unilateral leg loading has been utilized to understand the effect of altered inertial properties from one side on human walking motor adaptation.

Several studies on healthy populations suggested that approximately 40 - 50 stride of adaptation period is required to adjust to the altered leg inertia properties and to achieve the steady-state in both kinematics and kinetics (Noble & Prentice, 2006; J. D. Smith & Martin, 2007; J. D. Smith, Villa, et al., 2013). The period of adaptation was the same in both load addition and removal (Noble & Prentice, 2006; J. D. Smith & Martin, 2007; J. D. Smith, Villa, et al., 2013). These studies used treadmill walking at speed (1.5 ~ 1.57 m/s) which was slightly higher than reported preferred walking speeds for healthy adults. And the ankle loading (~ 2kg) was approximated the difference between the intact and prosthetic lower limbs previously reported for unilateral, transtibial amputees (Mattes et al., 2000). They also observed small changes or similar patterns in joint kinematic across load conditions while joint kinetics exhibits larger changes during steady-state (Noble & Prentice, 2006; J. D. Smith, Villa, et al., 2013). Selles et al. (2004) suggested that amputees alter their muscle coordination to maintain ideal kinematic walking patterns as a means of adjusting to new inertial properties of their prosthetic limb. The results from healthy population studies (Noble & Prentice, 2006; J. D. Smith, Villa, et al., 2013) also support kinematic invariance strategy for adaptation to leg inertial discrepancy as the joint moments were refined over time to achieve a similar kinematic pattern that was observed in the unloaded baseline condition.

To gain insight of joint kinetic control strategy of asymmetrical ankle loading, J. D. Smith, Villa, et al. (2013) and J. D. Smith, Royer, et al. (2013) investigated joint kinetics during swing phase by using intersegmental dynamics. During the swing phase of both loaded and unloaded legs, the interaction moments were interpreted as opposing the motions of the joints as they were out of phase with the net moment. On the other hand, the gravitational and muscle moments were interpreted as either counterbalancing the effects of the interaction moments or producing the motions at the joints as they were in phase with the net moment for the majority of swing (J. D. Smith, Royer, et al., 2013). With increasing unilateral load, while the unloaded side remained unchanged, the absolute magnitudes of all three moments on the loaded side increased systematically, but the relative contributions were different (J. D. Smith, Royer, et al., 2013). As foot mass increased, relative contributions of interaction moments about the knee (21%) and hip (8%) and gravitational moment about the ankle (44%) increased, whereas the relative contributions of muscle moments about all three joints declined (J. D. Smith, Royer, et al., 2013). This result suggests that the neuromuscular system is relying more on the interaction and gravitational moments for producing the swing to economize muscular effort (J. D. Smith, Royer, et al., 2013).

Haddad et al. (2006) suggested that gait adaptations to unilateral ankle loading mainly occur at the interlimb level in the healthy population. Their study examined changes in coordination patterns between limb couplings at intralimb and interlimb level on a stride by stride basis. Bilateral kinematic data from 5-min treadmill walking at preferred walking speed were assessed with randomized six load conditions (0.90, 1.8, 2.7, 3.6 or 4.5 kg and a no-load ) on the non-dominant leg. Continuous relative phase (CRP) was applied over three interlimb (thigh–thigh, leg–leg and foot–foot) and four intralimb (thigh–leg and leg–foot in both the loaded and unloaded limbs) couplings. Changes in coordinative patterns were quantified utilizing both cross-correlation and root-mean-square difference (RMS) measures. The cross-correlation findings suggested that the spatiotemporal pattern of coordination across the stride changes at the interlimb but not at the intralimb. However, significant RMS changes were observed at all interlimb level and at the unloaded side intralimb level in response to the load increase over both stance and swing phase. In the unloaded side, leg-foot intralimb coupling showed significant RMS increase only for the stance phase. Although both interlimb and intralimb showed changes in response to unilateral changes in limb inertial properties, changes in interlimb coordination were greater than in intralimb coordination, in that both magnitude (RMS) and temporal evolution (cross-correlation) changed to a large extent.

Kodesh et al. (2012) showed bilateral changes in step time and length symmetry of the combined effect of unilateral leg loading and walking speed. The study examined spatiotemporal gait data from overground walking under four randomly sequenced test conditions on self-selected speed (SS), fast speed (F), the self-selected speed with leg loaded (LSS), and the fastest attainable speed with leg loaded (LF). For the loading condition, a 3 kg mass was fixed to the ankle of dominant leg. Symmetry index (SI) for step time and length was calculated using the formula:

$$SI = \frac{R - L}{0.5 \times (R + L)} \times 100$$

While the unilateral leg loading increased step time asymmetry with longer step time for the loaded leg, fast speed with unilateral leg loading particularly increased step length asymmetry, with shortening in the unloaded leg and lengthening in the loaded leg. The major change between the LSS and the LF conditions was the compensatory decrease in step length of the unloaded leg. It appears that the step shortening of the unloaded leg counterbalanced the step lengthening of the loaded leg to maintain optimal stride length and generate the fastest possible walking speed. However, the result of this study only showed the short-term adaptations as they examined several overground walking trials.

DFA is a modification of root mean square analysis which takes advantage of the process that removes local trends, therefore, is relatively unaffected by the nonstationarity of the time series. DFA has been widely used to determine the fractal scaling properties and long-range correlations in diverse fields from DNA sequences (Peng et al., 1992; Peng et al., 1994), heart rate dynamics (Peng, Havlin, Hausdorff, et al., 1995; Peng, Havlin, Stanley, et al., 1995), and stride interval (Bohnsack-McLagan et al., 2016; Dingwell & Cusumano, 2010; Gates & Dingwell, 2007; Hausdorff et al., 1995; Hausdorff et al., 1996).

Hausdorff et al. (1995) first employed DFA on stride interval of 9-min overground walking at self-selected speed in healthy human walking and showed that the fluctuations exhibit long-range correlations ($\alpha = 0.83$). And they showed that scale-free correlation in stride interval was not exhibited in the surrogate data set of randomly shuffled time series. In the further study, Hausdorff et al. (1996) tested 1-hour of overground walking at three different speed (preferred, slow, and fast) and 0.5-hour of metronomic walking at three different paces. The metronomic paces were computed from each subject's mean stride interval of each three previous 1-hour walking trials. Regardless of speed, 1-hour walking trials showed long-range correlations ( $\alpha =$ 0.9, 0.84, 1.00 at slow, preferred, and fast walking, respectively). However, all metronomic paced walking trials yielded $\alpha \approx 0.5$ and, in many cases, the relationship between log (F(n)) and log (n) was not linear.

While some authors have argued that long-range correlation and persistent reflects an inherently "healthy" system, and uncorrelated and anti-persistent fluctuations indicate pathology (Gates & Dingwell, 2007; Hausdorff et al., 1997), Dingwell and Cusumano (2010) proposed an alternate explanation. Dingwell and Cusumano (2010) suggested that anti-persistent in a time series of the system may due to required tight control as deviations were followed by deviations in the opposite direction for rapid correction. In their study, they conducted 5-minute treadmill walking trials at each of 5 speeds (80% to 120% of preferred walking speed). Their resuls showed that only stride speed (SS) exhibited anti-persistent whereas stride time (ST) and stride length (SL) exhibited statistical persistent. Dingwell and Cusumano (2010) argued that it is equivocal to interpret that the system in this experiment is simultaneously both "healthy" and "unhealthy". Instead, the loss of statistical persistence can be interpreted as the increased control effort to achieve the desired task goal as the statistical persistence became anti-persistent by constraints such as metronome pace (Hausdorff et al., 1996) and treadmill speed (Dingwell & Cusumano, 2010).

The adaptation strategy to increased leg inertia is achieved by refining joint moment interactions (J. D. Smith, Royer, et al., 2013; J. D. Smith, Villa, et al., 2013) to keep the kinematic invariant (Noble & Prentice, 2006; Selles et al., 2004; J. D. Smith, Royer, et al., 2013; J. D. Smith, Villa, et al., 2013). While those joint kinematic and kinetic mainly occur in the loaded side (Noble & Prentice, 2006; Selles et al., 2004; J. D. Smith, Royer, et al., 2013; J. D. Smith, Villa, et al., 2013), spatiotemporal changes occur at the interlimb level to counterbalance each other for maintaining optimal stride (Haddad et al., 2006; Kodesh et al., 2012). Since DFA yields the scaling exponent α as an indicator of a control process (Dingwell & Cusumano, 2010), utilizing DFA on unilateral ankle loading will provide further insight to understand motor adaptation and compensation strategy to asymmetrical gait dynamics.

## Methods

Subject Recruitment and Data Collection at the Lab

Ten male subjects participated in this study. Mean (standard deviation) of age was 24.5 (3.7) years, height 1.78 (0.1) m, and weight 79.0 (7.9) kg. This study was approved by Georgia State University Internal Review Board. Every subject signed an informed consent form.

The subjects came to the Biomechanics lab at Georgia State University for the data collection for one session which took about 75 minutes of time. The subjects were asked to wear compression shorts for proper marker placement on the body.

Upon arrival we described the study to the subjects and asked them to sign an Informed Consent form before their participation. Then, we measured the subject's weight and height, and attached reflective markers at toe, heel, ankle, tibia, knee, and thigh and hip on both sides. We used a 7-camera Vicon motion capture system to record these markers. A T-pose was captured for five seconds which required the subject to stand and hold the arms out 90 degrees to the body.

The subjects walked straight along a 10-meter walkway three times at their comfortable speed. We recorded the time with a stopwatch and calculated the average of their overground walking speed. This average speed was considered as the subject's comfortable walking speed, and it was used for treadmill walking. We asked the subject to kick a ball on the floor, and the leg that the subject used to kick the ball was considered as the dominant leg.

Then, we tested four load conditions: no ankle weights (A0), 25% ankle load (A25), 50% ankle load (A50), and 75% ankle load (A75), the last three conditions representing the increase of moment of inertia of the leg about the knee joint by 25%, 50% and 75%, respectively (David A. Winter, 2009). We tightly secured the load above the ankle and did not provide the subject with much time for practice. The subjects walked on a treadmill for 5 minutes under each load condition and the four load conditions (A0, A25, A50 and A75) were randomized across the subjects. We started to collect each 5-minute walking trial when the treadmill speed reached the desired speed but the subject might have not been totally adapted to the load. Enough rest was provided between trials to remove the potential residual load effect and muscle fatigue.

### Data Analysis

Marker data were processed using the Vicon Nexus software and exported for further analyis. We used customized MatLab programs to calculate spatiotemporal gait variables and conduction the DFA analysis.

Step length and step time were calculated with heel markers. Step length was defined as the anterior-posterior distance along the axis of progression between heel strike of the contralateral leg to heel strike of the ipsilateral leg. Step time was defined as the time between heel strike of the contralateral leg to heel strike of the ipsilateral leg. Heel strike events were found based on the heel marker velocity and then the step time was calculated by the difference between consecutive heel strikes of loaded and unloaded legs. Step length was calculated based on the trajectory of the heel marker at heel strike events, and the length of the passed treadmill obtained by multiplying step time to treadmill speed. Mean and standard deviation of step length and step time was calculated for every 100 seconds of a trial on each leg.

Five minutes of walking trial was divided into three 100-second segments to investigate the time effect of unilateral loading on walking adaptation. This 100-second interval was determined by considering both the adaptation time to inertia properties (40~50 strides) (Noble & Prentice, 2006; J. D. Smith & Martin, 2007; J. D. Smith, Villa, et al., 2013) and the reported minimum series length for DFA (64 observations) (Didier Delignieresa, 2006). Hence, mean and standard deviation of step length and step time was calculated in the first 100-second, the second 100-second, and the third 100-second. And to verify if the time factor affected long-range correlation, DFA was conducted on each of three 100-second segments of step length and step time data.

DFA was utilized to quantify serial correlations on step length and step time on each side of the leg. Each time series of N steps was integrated as below:

$$y(k) = \sum_{i=1}^{k}[I(i) - I_{avg}]$$

Where $I(i)$ was the ith step data, and $I_{avg}$ was the average step data. Next, the integrated time series was divided into equal, non-overlapping segments of length *n*. Within each segment, the series was linearly detrended. The squares of locally detrended fluctuation was averaged over the entire data set and the square root of the mean residual, *F(n)*, was calculated as below:

$$F(n) = \sqrt{\frac{1}{N}\sum_{k=1}^{N}[y(k) - y_n(k)]^2}$$

This process was repeated over all time scales to provide a relationship between *F(n)* and *n*. Slopes of log[*F(n)*] vs. log(*n*) plots was obtained by linear regression to compute the scaling exponent, α. If the value at one step is completely uncorrelated with any previous values, α = 0.5. If there are short-term correlations, the initial slope might be different from 0.5, but α would approach 0.5 for large window sizes. An 0.5 < α =< 1 indicates a persistent long-range correlation. This indicates that deviations were more likely to be followed by deviations in the same direction. An α < 0.5 indicates that deviations in one direction were more likely to be followed by deviations in the opposite direction (anti-persistence). This suggestes a more tightly controlled process which generates more frequent corrections of deviations (Dingwell & Cusumano, 2010).

Statistical Analysis

Dependent variables included step length and time as well as the scaling exponent of step length and time. Three-way (2 side x 3 time x 4 load) analysis of variance (ANOVA) with repeated measures was conducted on mean of step length and time as well as the scaling exponent of step length and time. Pairwise comparisons with Bonferroni adjustments was conducted when necessary. For all statistical process, SPSS Statistics was used, and the significance level was set to alpha = 0.05.

## Results

Subjects

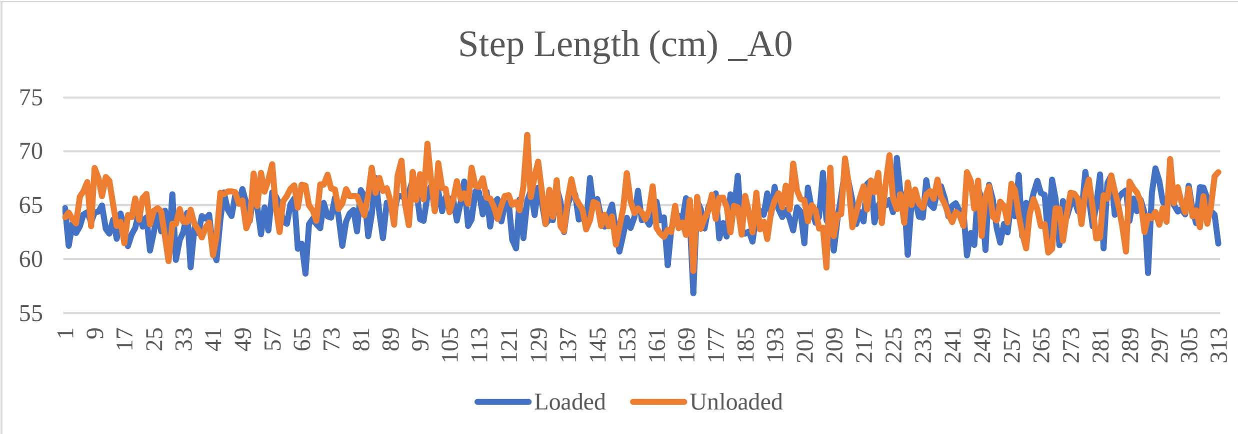
The participants included ten male college students. Mean +/- standard deviation was age: 24.5 ± 3.67 years, height: 177.5 ± 5.10 cm, weight: 87.3 ± 23.5 kg, preferred walking speed: 1.34 ± 0.10 m/s). All participants had the right leg as the dominant leg.

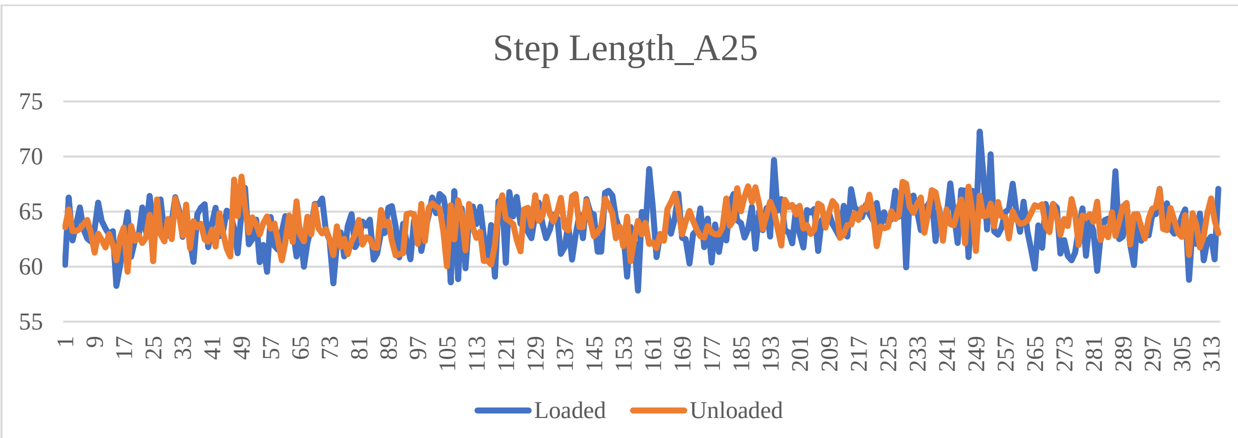
Tabel 1: Participants Characteristics.

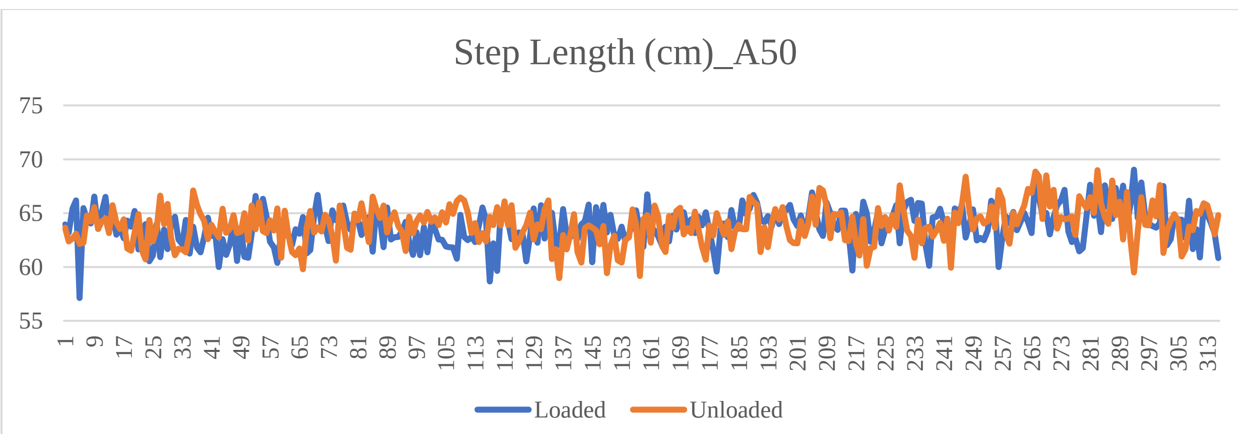
| N | Gender | Age | Height (cm) | Weight (kg) | Treadmill Speed (m/s) |
|---|---|---|---|---|---|
| 10 | Male | 24.5 (3.67) | 177.5 (5.10) | 87.3 (23.52) | 1.34 (0.10) |

Note tha tall values except N and Gender are given Mean (SD)

Step Length


Step Length (cm) _A0
75
70
65
60
55
1 9 17 25 33 41 49 57 65 73 81 89 97 105 113 121 129 137 145 153 161 169 177 185 193 201 209 217 225 233 241 249 257 265 273 281 289 297 305 313
Loaded
Unloaded


Step Length_A25
75
70
65
60
55
1 9 17 25 33 41 49 57 65 73 81 89 97 105 113 121 129 137 145 153 161 169 177 185 193 201 209 217 225 233 241 249 257 265 273 281 289 297 305 313
Loaded
Unloaded


Step Length (cm)_A50
75
70
65
60
55
1 9 17 25 33 41 49 57 65 73 81 89 97 105 113 121 129 137 145 153 161 169 177 185 193 201 209 217 225 233 241 249 257 265 273 281 289 297 305 313
Loaded
Unloaded

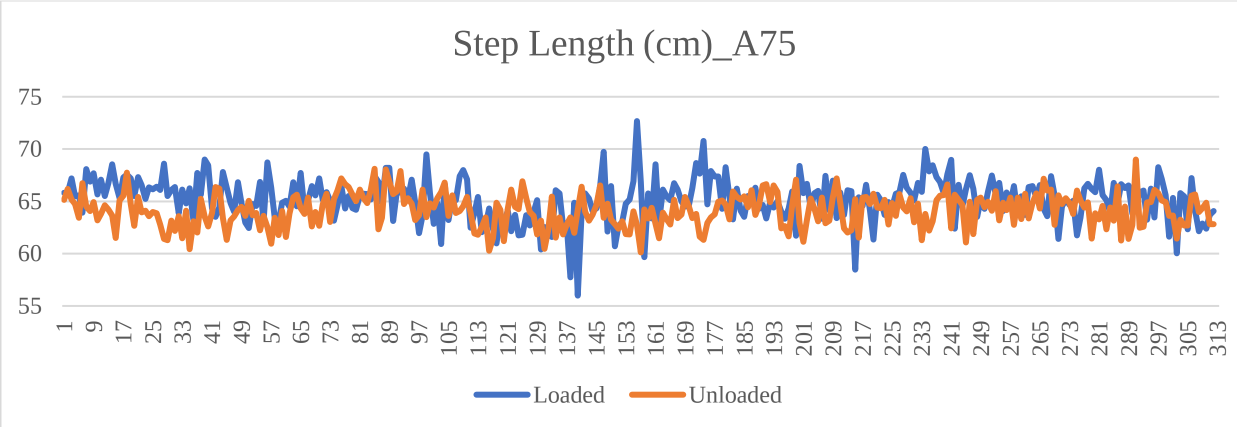



Figure 1: Time history of step length (cm) from a representative subject at each loading conditions (x-axis denotes step count)

Table 2 : The mean and standard deviation of step length (cm) : Mean (SD)

| | Loaded Side | | | | Unloaded Side | | | |
|---|---|---|---|---|---|---|---|---|
| | A0 | A25 | A50 | A75 | A0 | A25 | A50 | A75 |
| 100 sec | 62.7 (3.59) | 63.6 (3.30) | 64.1 (3.59) | 64.0 (4.21) | 63.6 (3.42) | 63.6 (3.30) | 64.1 (3.60) | 63.6 (3.55) |
| 200 sec | 63.0 (3.82) | 63.1 (3.61) | 64.4 (3.50) | 64.3 (3.74) | 63.0 (3.82) | 63.4 (3.06) | 64.4 (3.50) | 64.3 (3.74) |
| 300 sec | 63.4 (3.61) | 63.4 (3.33) | 63.6 (3.84) | 64.7 (3.77) | 63.4 (3.61) | 63.4 (3.33) | 63.6 (3.40) | 64.7 (3.77) |

Unilateral ankle load increased step length in both loaded and unloaded sides, but with a different extent across different time scales and between two sides (Table 2). Statistical analysis revealed that there was a three-way interaction on step length ($F(6,48) = 8.88$, $p = 0.001$). Pairwise comparison by the load factor revealed an increase of ankle load resulted in a significant increase on both loaded and unloaded side. On the loaded side, there were significant increases from A0 to A75 in the first 100 sec ( $p = 0.03$), A0 to A50 ( $p = 0.02$) and A75 ( $p < 0.001$), and A25 to A75 ( $p = 0.03$) in the second 100 sec, A0 to A75 ( $p < 0.001$), and A25 to A50 ( $p = 0.03$) and A75 ( $p = 0.01$) in the third 100 sec (Fig. 2). On the unloaded side, significant differences were found from A0 to A50 ( $p$ = 0.02) and A75 ( $p$ < 0.001), and A25 to A75 ( $p$ = 0.03) in the second 100 sec, and A0 to A75 ( $p$ < 0.001), and A25 to A50 ( $p$ = 0.03) and A75 ( $p$ = 0.01) in the third 100 sec (Fig. 3).

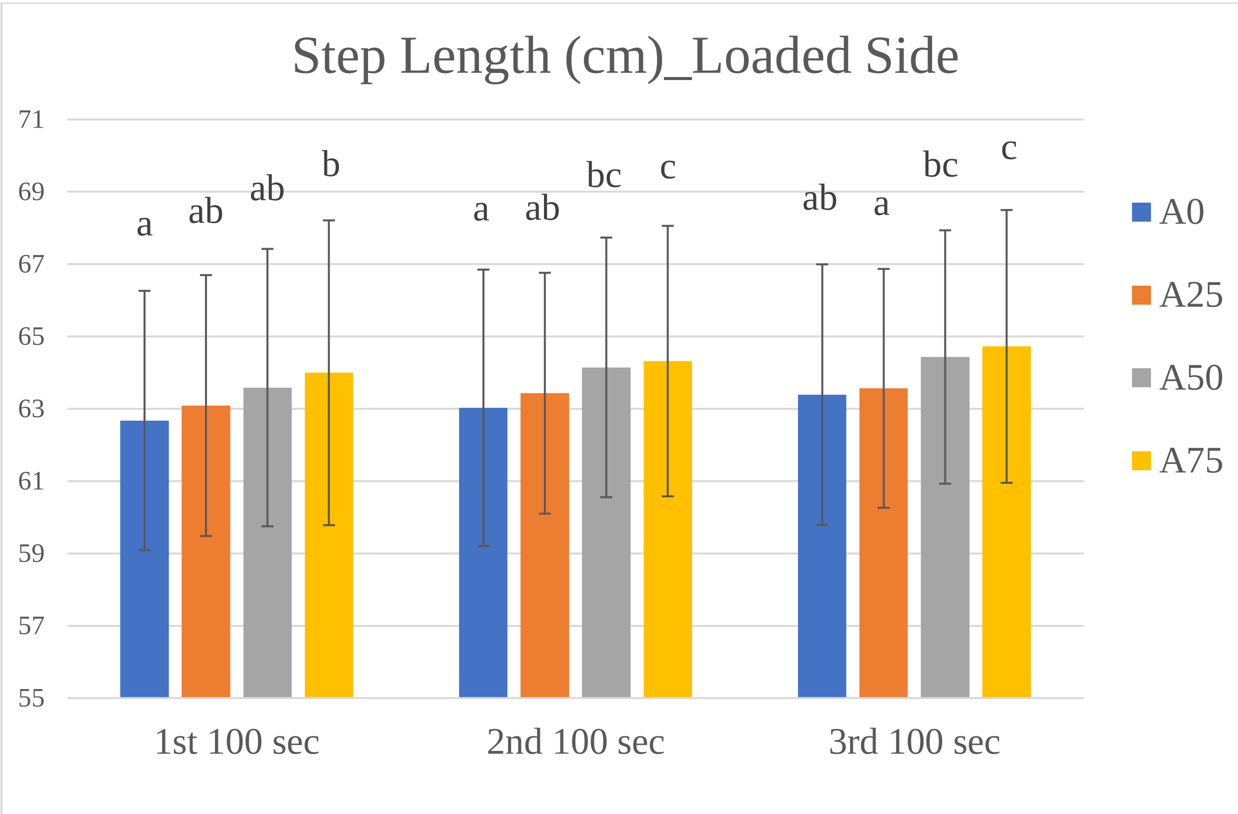



Figure 2: The mean and standard deviation of step length (cm) on the loaded side at every 100 sec by loading conditions (different letters above the bar graph denote statistical differences at $p$ <0.05, i.e., a is different from b, and b is different from c)

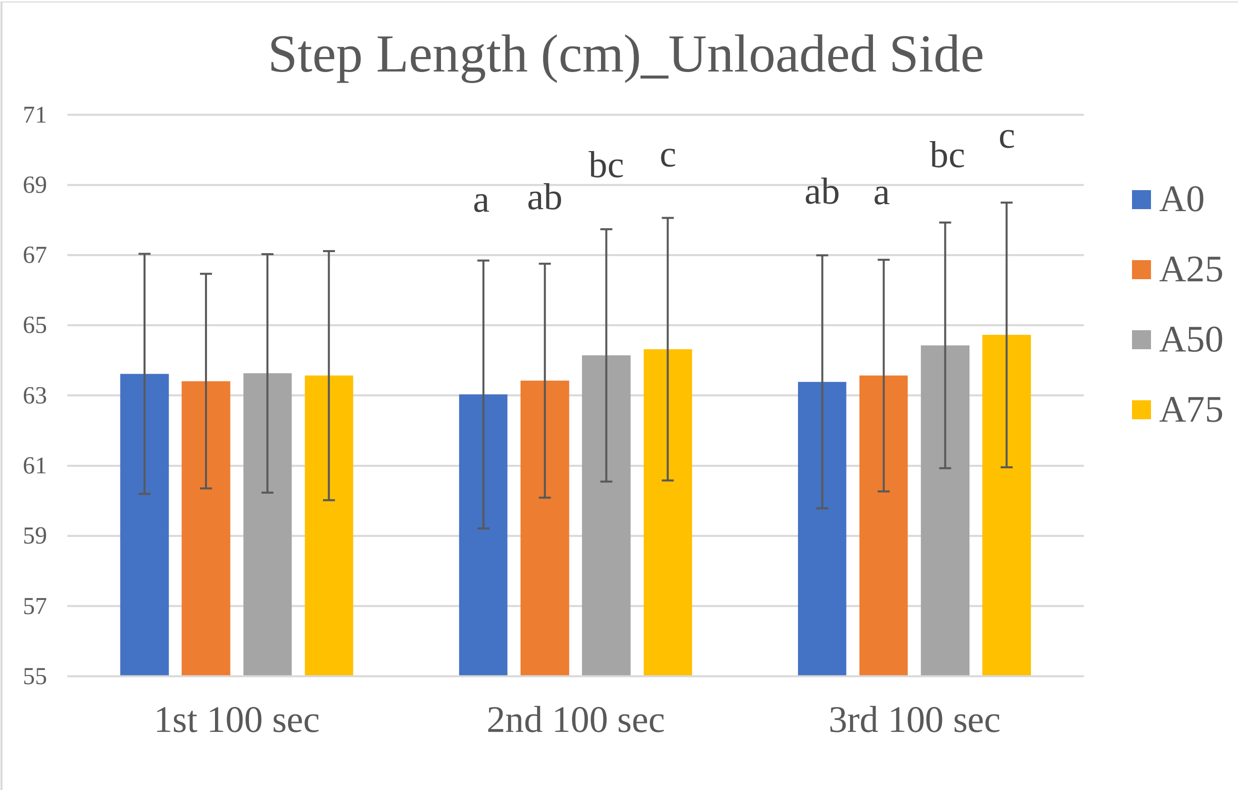
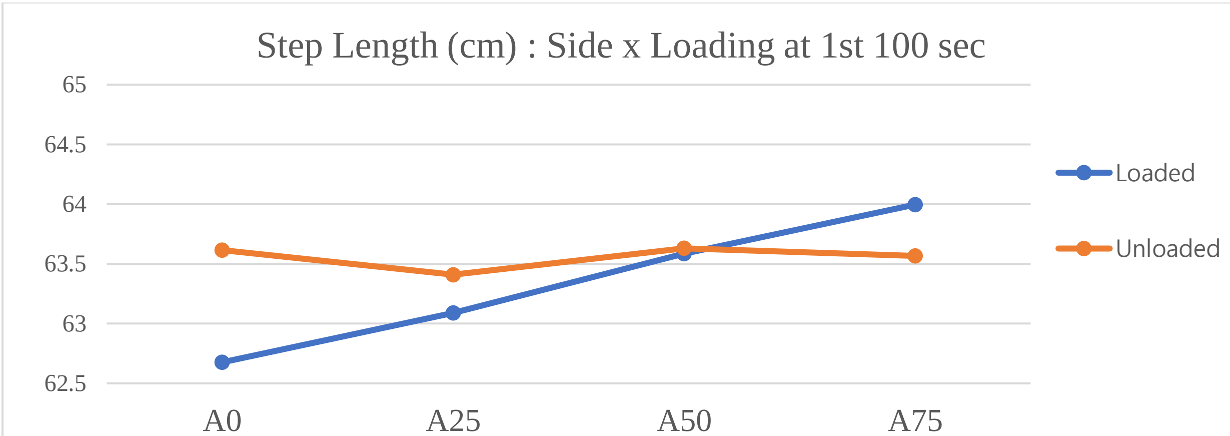


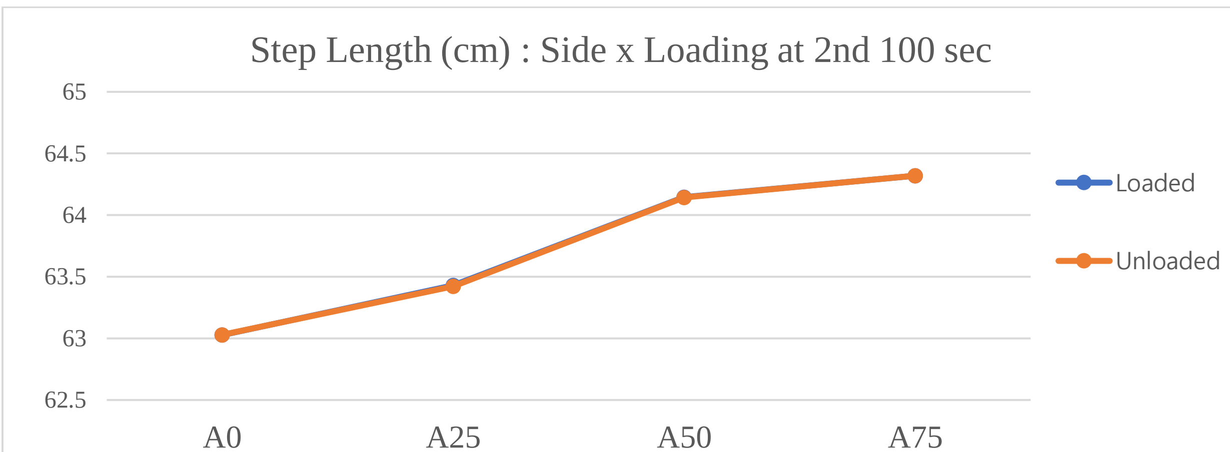

Figure 3: The mean and standard deviation of step length (cm) on the unloaded side at every 100 sec by loading conditions (different letters above the bar graph denote statistical differences at $p$ <0.05, i.e., a is different from b, and b is different from c)

Simple main effect analysis revealed two way (side x loading) interaction at 1st 100 sec ($F(3,27) = 8.90$, $p < 0.001$) (Fig. 4A), and loading effect at 2nd 100 sec ($F(3,27) = 12.93$, $p < 0.001$) (Fig. 4B) and 3rd 100 sec ($F(3,27) = 13.57$, $p < 0.001$) (Fig. 4C)

A.




B.




C.

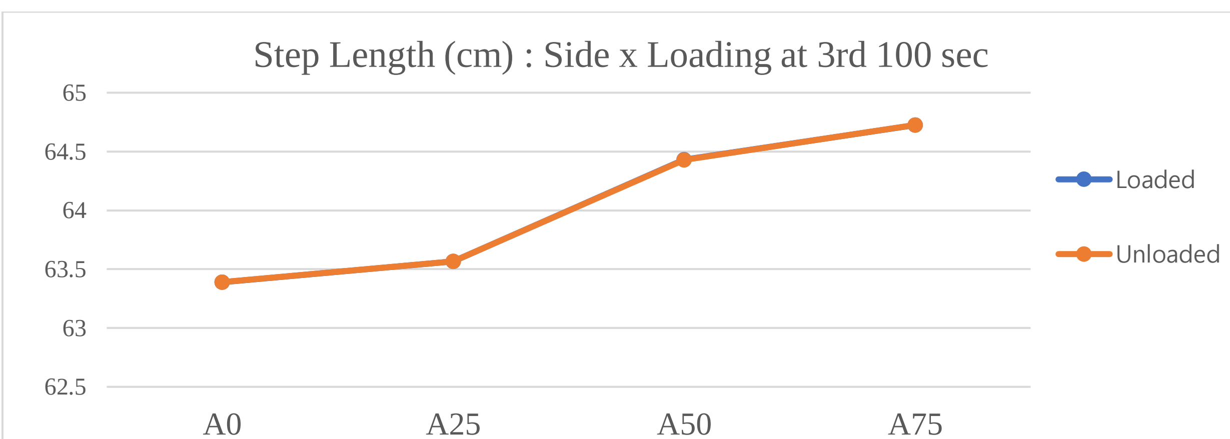



Figure 4A, B, C: Side x loading interaction of step length (cm) at (A) 1st 100 sec, (B) 2nd 100 sec and (C) 3rd 100 sec.

DFA revealed α > 0.5 of scaling exponent across all side, time and loading conditions (Table. 3). Statistical analysis revealed that there was only a time effect ($F(2,18) = 14.6$, $p$ <0.001). Post-hoc analysis reported that the DFA scaling exponent was different between the first 100 sec and the second 100 sec (Fig. 5).

Table 3 : The mean and standard deviation of the scailing exponent α of step length : Mean (SD)

| | Loaded Side | | | | Unloaded Side | | | |
|---|---|---|---|---|---|---|---|---|
| | A0 | A25 | A50 | A75 | A0 | A25 | A50 | A75 |
| 100 sec | 0.78 (0.18) | 0.66 (0.10) | 0.64 (0.07) | 0.62 (0.10) | 0.75 (0.22) | 0.67 (0.11) | 0.64 (0.07) | 0.72 (0.18) |
| 200 sec | 0.64 (0.16) | 0.65 (0.12) | 0.62 (0.13) | 0.62 (0.14) | 0.63 (0.16) | 0.70 (0.18) | 0.62 (0.13) | 0.61 (0.15) |
| 300 sec | 0.69 (0.18) | 0.65 (0.13) | 0.76 (0.16) | 0.72 (0.22) | 0.69 (0.18) | 0.65 (0.14) | 0.75 (0.14) | 0.73 (0.23) |

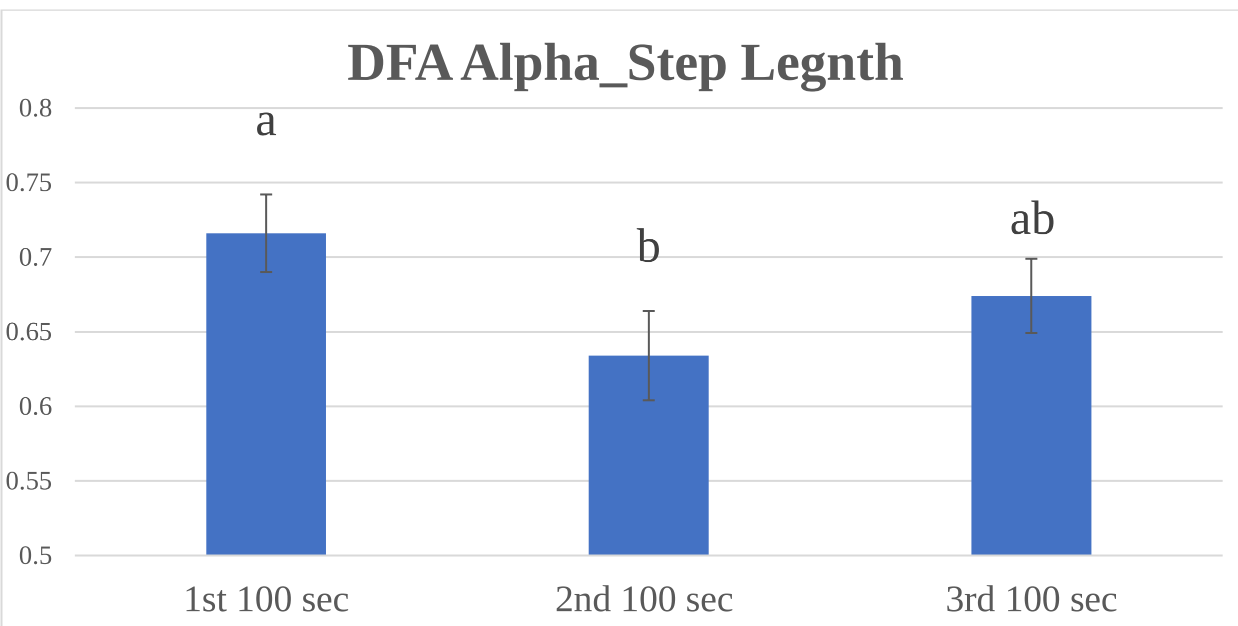
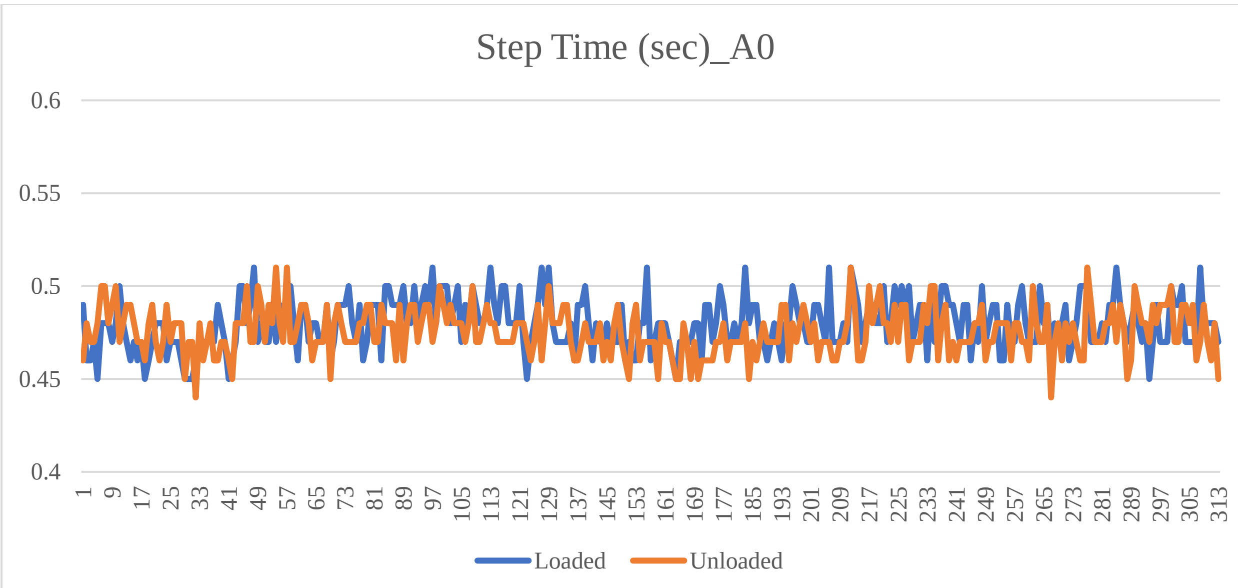
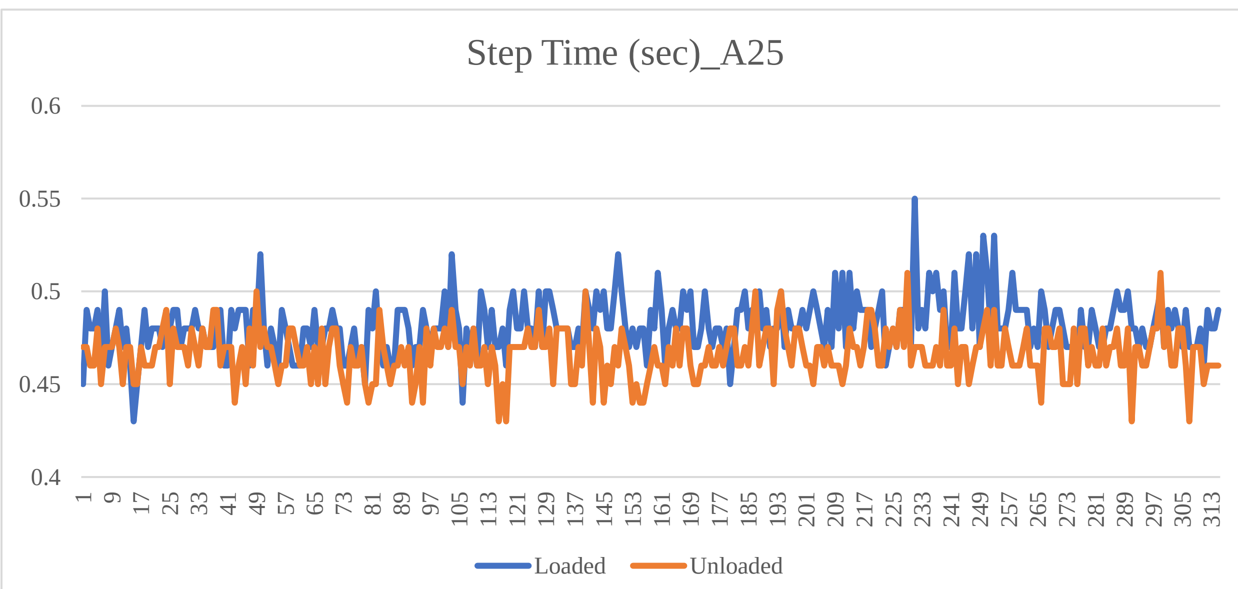


Figure 5: The mean and standard deviation of the scaling exponent α of step length by the time (different letters above the bar graph denote statistical differences at $p$ <0.05, i.e., a is different from b, and b is different from c)

Step Time



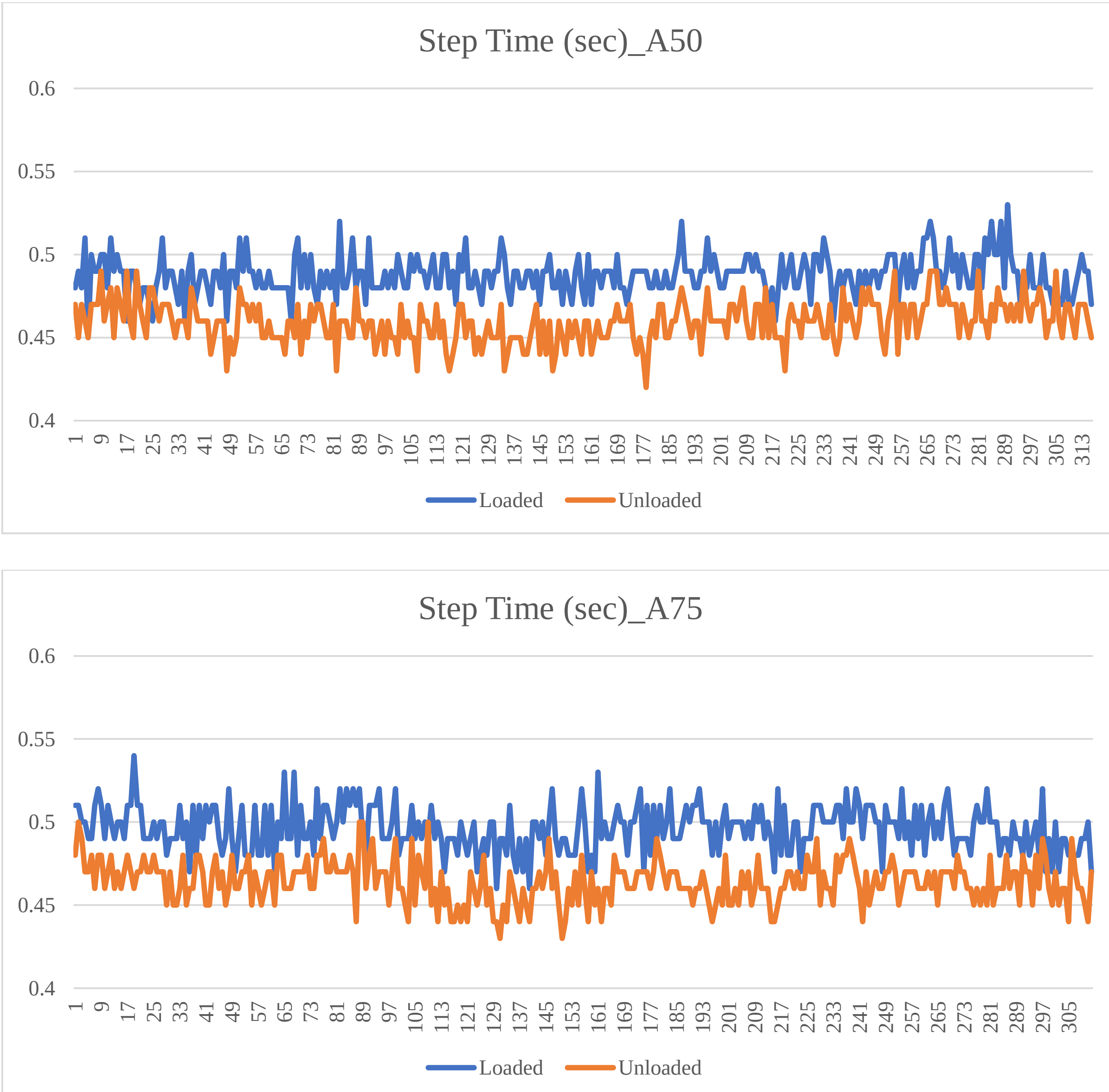



Figure 6 : Time history of step time (sec) from a representative subject at each loading conditions (x-axis denotes step count)

Unilateral ankle load increased step time on the loaded side and decreased step time on the unloaded side, and this effect varied across different time scales and between the two sides (Table 4). A three-way interaction was found on step time ($F(6,48) = 3.58$, $p < 0.05$). Pairwise comparison by the load factor revealed significant differences on both loaded and unloaded side. On the loaded side, there were significant increases as the ankle load increased from A0 to A25 ( $p = 0.01$), A50 ( $p < 0.001$) and A75( $p < 0.001$), and from A25 to A75 ( $p < 0.001$) at every 100 sec (Fig. 7) whereas unloaded side resulted in significant decrease from A0 to A50 ( $p = 0.01$) and A75 ( $p < 0.001$) at the first 100 sec, and A0 to A75 ( $p = 0.04$) at the second 100 sec, and A0 to A50 ( $p = 0.02$) and A75 ( $p = 0.01$) at the third 100 sec (Fig. 8).

Table 4 : The mean and standard deviation of step time (sec): Mean (SD)

| | Loaded Side | | | | Unloaded Side | | | |
|---|---|---|---|---|---|---|---|---|
| | A0 | A25 | A50 | A75 | A0 | A25 | A50 | A75 |
| 100 sec | 0.51 (0.03) | 0.53 (0.03) | 0.54 (0.03) | 0.54 (0.03) | 0.52 (0.03) | 0.52 (0.03) | 0.51 (0.03) | 0.5 (0.03) |
| 200 sec | 0.51 (0.03) | 0.52 (0.03) | 0.54 (0.03) | 0.54 (0.03) | 0.52 (0.03) | 0.51 (0.03) | 0.51 (0.03) | 0.51 (0.03) |
| 300 sec | 0.521 (0.03) | 0.53 (0.03) | 0.53 (0.03) | 0.55 (0.03) | 0.52 (0.03) | 0.52 (0.03) | 0.51 (0.03) | 0.51 (0.03) |

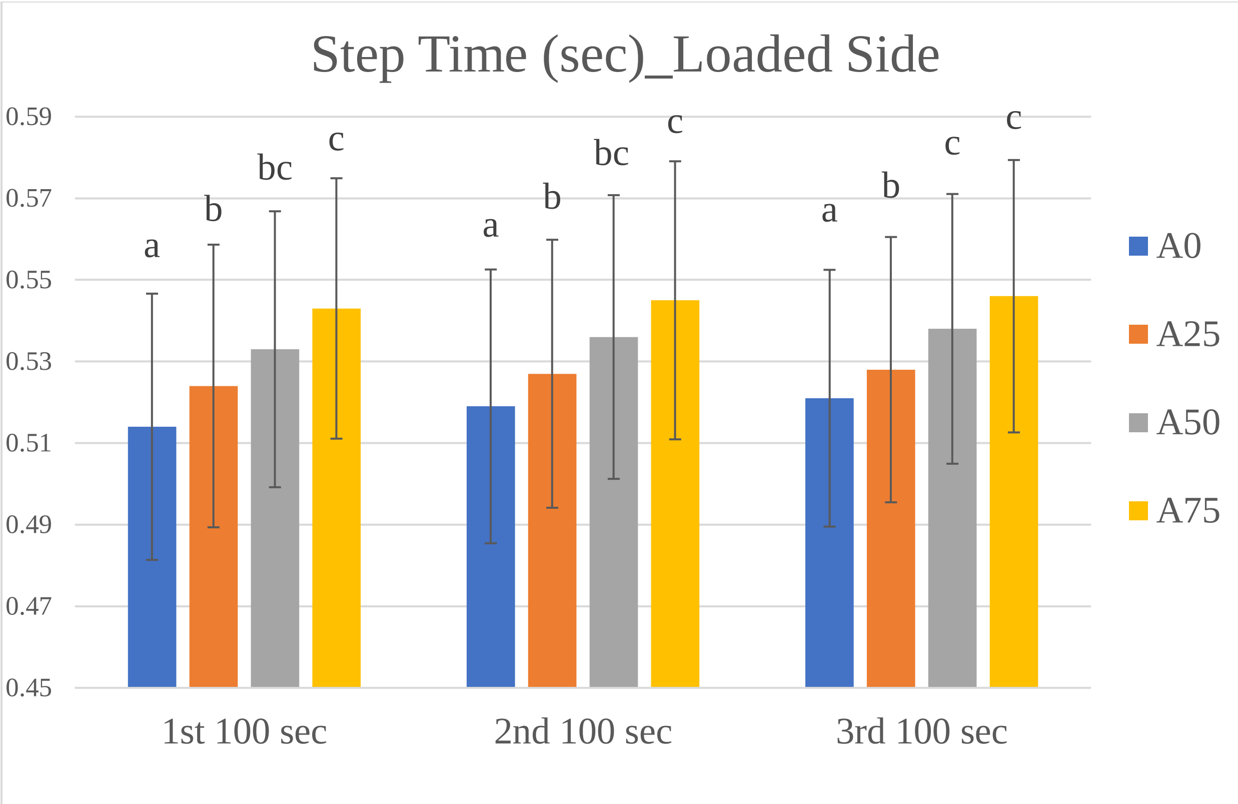



Figure 7: The mean and standard deviation of step time (sec) on the loaded side at every 100 sec by loading conditions (different letters above the bar graph denote statistical differences at *p* <0.05, i.e., a is different from b, and b is different from c)

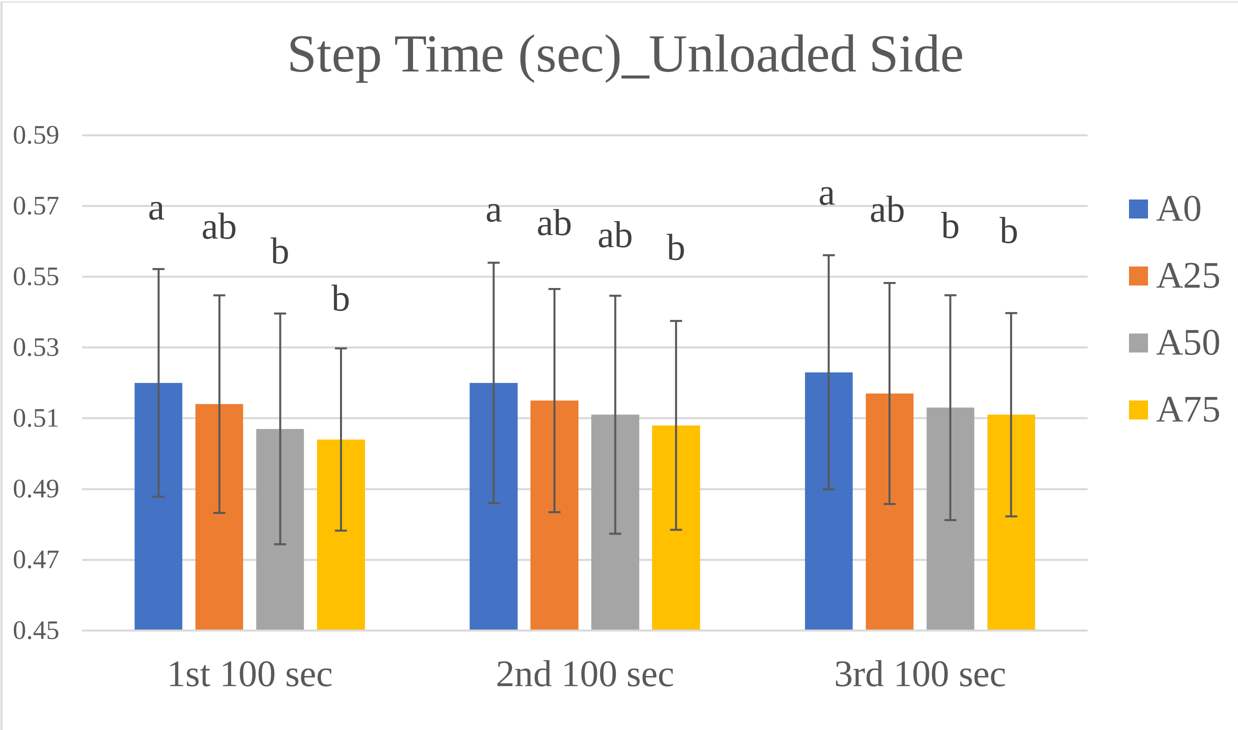
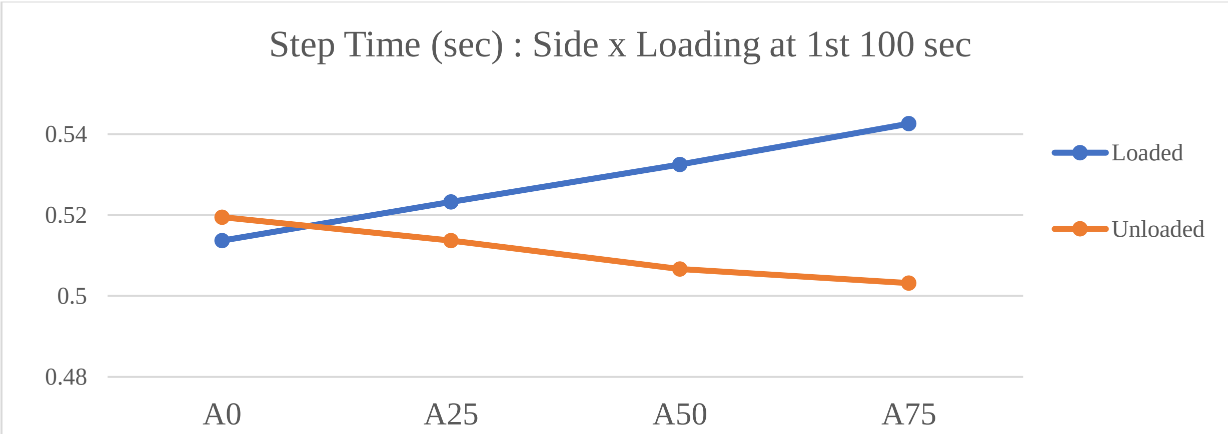


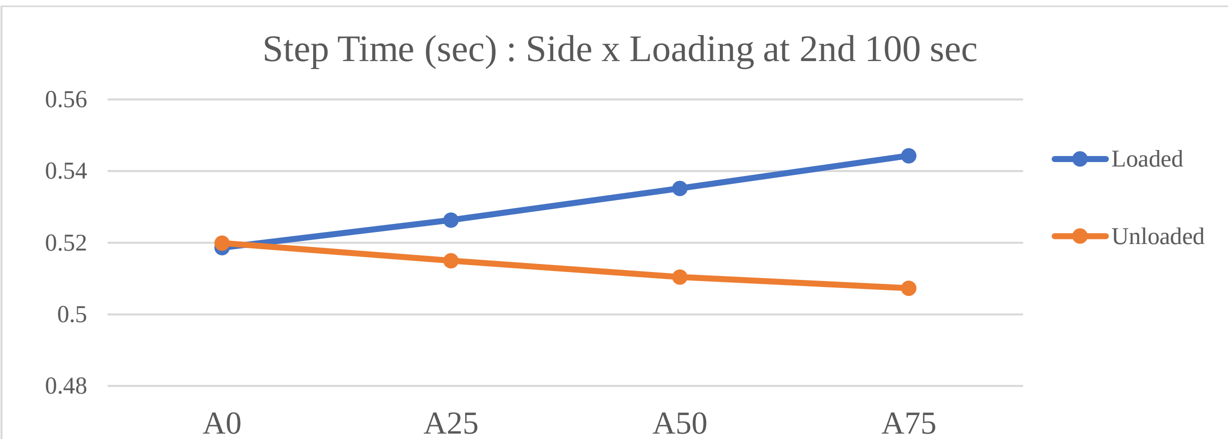

Figure 8: The mean and standard deviation of step time (sec) on the unloaded side at every 100 sec by loading conditions (different letters above the bar graph denote statistical differences at $p$ <0.05, i.e., a is different from b, and b is different from c)

Simple main effect analysis revealed two-way interactions at every 100 sec (side x loading, $p < 0.001$) (Fig. 9 A, B, C) and, at A0 condition ($F(2,18) = 7.02$, $p < 0.001$) (Fig. 10A) and at A75 condition ($F(2,18) = 4.69$, $p < 0.001$ ) (Fig. 10D). Side and time effect was found at A25 condition (side : $F(1,9) = 7.91$, $p = 0.02$, time : $F(1.3, 11.7) = 5.09$, $p = 0.04$) (Fig. 10B) and A50 condition (side : $F(1,9) = 48.4$, $p < 0.001$, time : $F(2,18) = 22.2$, $p < 0.001$) (Fig. 10C).

A.




B.




C.

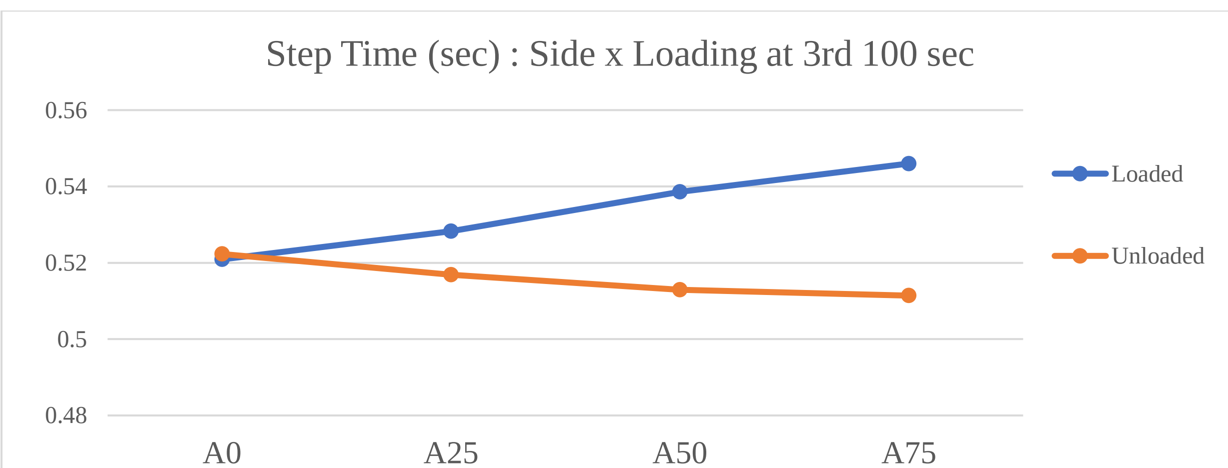
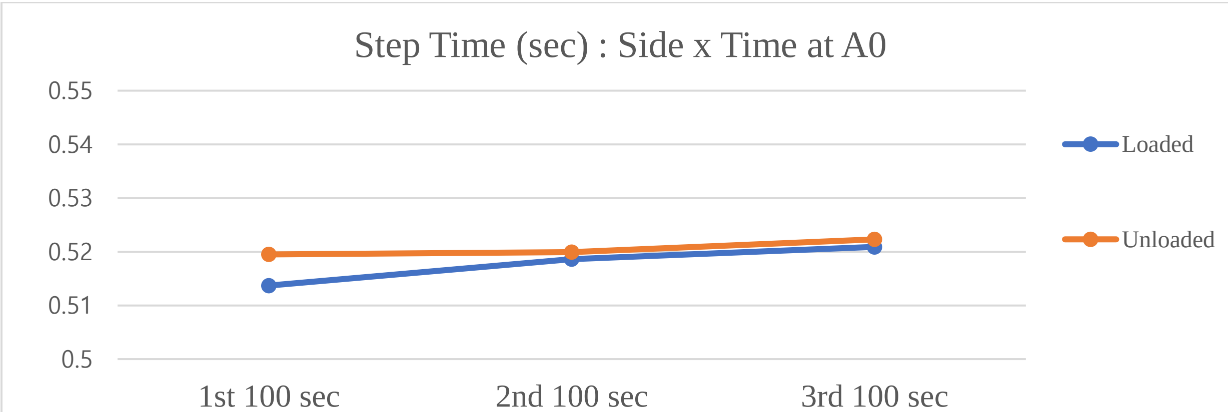
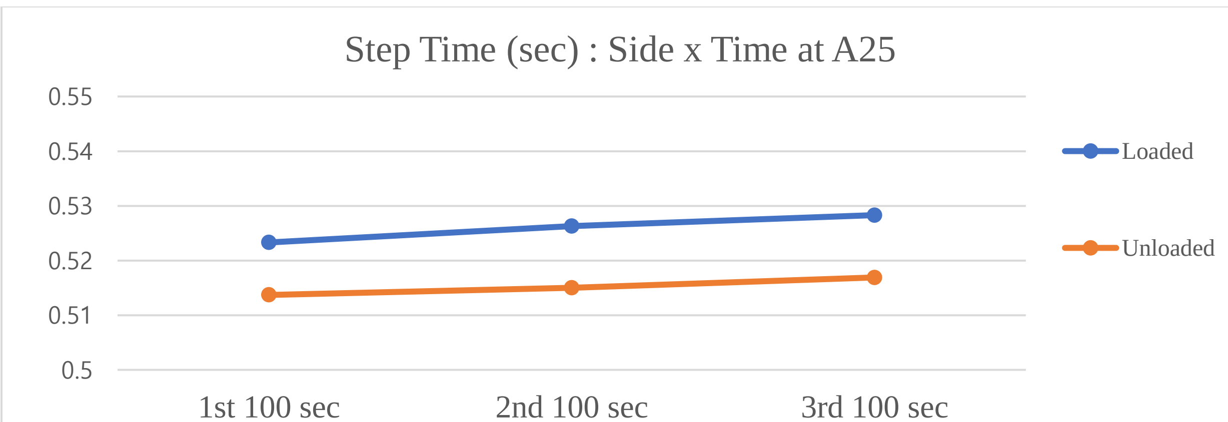



Figure 9A, B, C: Side x loading interaction of step time (sec) at (A) 1st 100 sec, (B) 2nd 100 sec and (C) 3rd 100 sec.

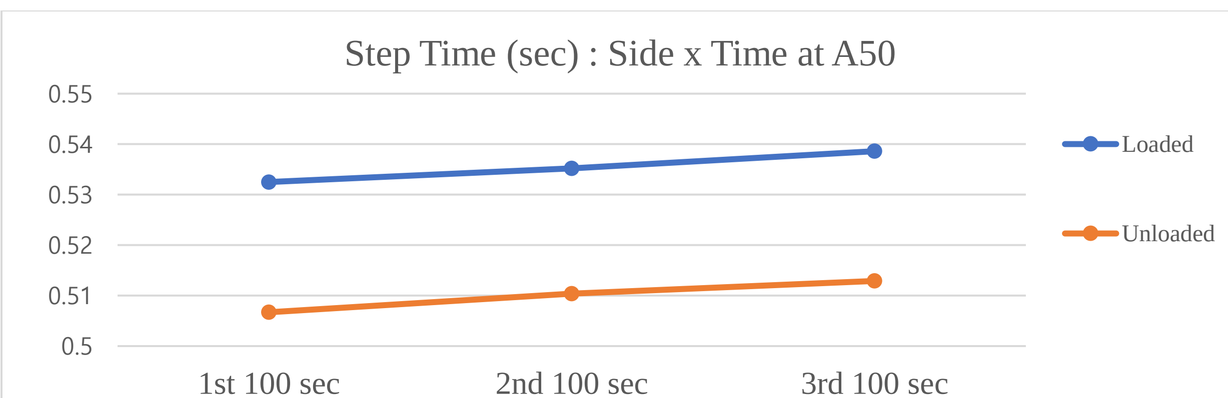
A.


Step Time (sec) : Side x Time at A0
0.55
0.54
0.53
0.52
0.51
0.5
1st 100 sec
2nd 100 sec
3rd 100 sec
Loaded
Unloaded

B.


Step Time (sec) : Side x Time at A25
0.55
0.54
0.53
0.52
0.51
0.5
1st 100 sec
2nd 100 sec
3rd 100 sec
Loaded
Unloaded

C.


Step Time (sec) : Side x Time at A50
0.55
0.54
0.53
0.52
0.51
0.5
1st 100 sec
2nd 100 sec
3rd 100 sec
Loaded
Unloaded

D.

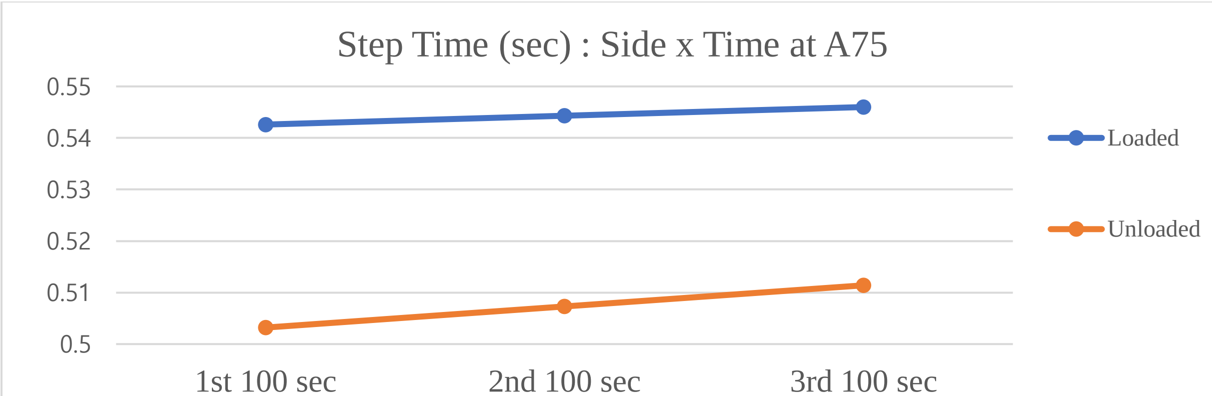



Figure 10A, B, C, D: Side x time interaction of step time (sec) at (A) A0, (B) A25, (C) A50 and (D) A75 conditions

DFA revealed α > 0.5 of scaling exponent across all side, time and loading conditions and the scaling exponent appeared to be consistent at different conditions (Table. 5). Statistical anlaysis revealed no significant main effects or interactions.

Table 5 : The mean and standard deviation of the scailing exponent α of step time : Mean (SD)

| | Loaded Side | | | | Unloaded Side | | | |
|---|---|---|---|---|---|---|---|---|
| | A0 | A25 | A50 | A75 | A0 | A25 | A50 | A75 |
| 100 sec | 0.72 (0.05) | 0.69 (0.04) | 0.68 (0.05) | 0.66 (0.05) | 0.68 (0.06) | 0.61 (0.03) | 0.67 (0.05) | 0.71 (0.04) |
| 200 sec | 0.70 (0.04) | 0.64 (0.04) | 0.54 (0.04) | 0.59 (0.04) | 0.65 (0.04) | 0.63 (0.04) | 0.66 (0.05) | 0.65 (0.03) |
| 300 sec | 0.62 (0.06) | 0.67 (0.04) | 0.64 (0.05) | 0.68 (0.04) | 0.61 (0.05) | 0.60 (0.05) | 0.66 (0.05) | 0.60 (0.03) |

**Discussion**

As unilateral ankle loading increased, step time of the loaded side increased, and the unloaded side decreased whereas both sides of step length increased. Asymmetry between legs induced by unilateral loading was preserved on step time over time, however no asymmetry was exhibited on step length. These findings support our first hypothesis on the adaptation of step time but only partially on the adaptation of step length. Further, our results did not support the hypotheses on the long-range correlation of step length and time during treadmill walking in young adults.

A three-way (side x time x load) interaction was found in both step length and step time. Step length of both sides increased as the unilateral loading increased, however the unloaded side started to increase after the first 100 sec while the loaded side increased at the beginning of the walking. However, step time of the loaded side increased with the load while the unloaded side decreased with the load. Our hypothesis expected a trend of opposite direction between the loaded and the unloaded side, but only step time showed this compensatory trend.

Two-way interactions were found in both step length and step time. Side x loading interaction was observed in step length at the 1st 100 sec, and this interaction indicates a significant difference between two sides in A0 loading condition at the 1st 100 sec due to sustained step length of the unloaded side. Step time showed two-way interactions at every 100 sec (side x loading), and A0 and A75 loading conditions (side x time). Step time difference between the two sides showed a significant increase by loading conditions at every 100 sec, however, two sides showed different time trend at A0 and A75 conditions. At A0 condition, loaded side significantly increased from 1st 100 sec to 2nd and 3rd 100 sec while unloaded side was sustained, however, at the A75 condition, the loaded side was not markedly changed while unloaded side significantly increased by time.

Varraine, Bonnard, and Pailhous (2000) suggested that the intentional control of stride length is a fundamental basis for environmental navigation. The lengthening of stride length was carried out by an increase in the propulsive force and an increase in swing duration on the ipsilateral leg while the shortening of it mainly came from a decrease of swing duration (Varraine, Bonnard, & Pailhous, 2000). Our findings of step length increase on the unloaded side may be due to the intentional control of stride length on this side as our premilinary analysis found that the vertical ground reaction force increased with ankle load on both sides.

Step length showed a significant difference between the loaded and the unloaded sides only at the first 100 sec of the A0 condition whereas step time showed significant differences over time and across all loading conditions except for the A0 condition. Our hypothesis expected both step length and step time asymmetry between two legs would decrease during 5 minutes walking, but only step length reached symmetries while step time kept asymmetry over time. Our results suggest that there might be separate neuromotor control mechanisms on regulating step length and time when adapting to unilateral ankle load over time.

DFA was conducted on step length and step time to verify the effect of unilateral loading on long-range correlation. Both step length and step time yielded scaling exponent of long-range correlation ($1 > \alpha > 0.5$). While scaling exponent of step time was not affected by side, time or loading, there was a time effect on scaling exponent of step length. Unlike the mean values of step time and step length, scaling exponent of these two variables did not exhibit any significant increase, decrease or asymmetry imposed by unilateral ankle loading. These findings reject our second hypothesis in that scaling exponent of DFA would increase on the loaded side as the loading weight increased and would decrease on the unloaded side, and the asymmetry between the legs would decrease over time.

Scaling exponent of step length was significantly different from the first 100 sec to the second 100 sec, and it may indicate adaptation period of step length. This time effect on step length DFA is in line with the findings of step length increase after 100 sec of walking on the unloaded side by the loading weight increase. Previous studies reported that approximately 40 - 50 stride of adaptation period is required for the modified inertia properties of the legs to reach the steady-state in both kinematics and kinetics (Noble & Prentice, 2006; J. D. Smith & Martin, 2007; J. D. Smith, Villa, et al., 2013). As 100 sec was approximately 90 strides in our experiment, our observed 100-sec delayed response of step length on the unloaded side is consistent with those previous studies.

Long-range correlation has been suggested to be indicative of central nervous system control, and uncorrelated fluctuations indicate its deterioration (Gates & Dingwell, 2007; Hausdorff et al., 1997). As scaling exponent of both step length and step time in our data exhibited this statistical persistence over time and across all loading conditions, these findings seem to agree with that the fluctuation of gait cycles of a healthy population may be regulated by the central nervous system. The insignificance of the DFA values suggests that our loading conditions may not be challenging to interrupt the long-range correlation of spatiotemporal gait parameters in healthy young male adults, or the regulation of long-range correlation of spatiotemporal gait parameters might be insensitive to external load conditions. This suggests that the control of spatiotemporal gait parameters may be separate from or priotized over the regulation of long-range correlation at the different central nervous system. This motor control policy may change in populaions at different ages or with different medical conditions.

**Limitations and Future Studies**

Our study was conducted on the treadmill which has many advantages in gait studies; however, treadmill walking might not adequately reflect natural gait dynamics which overground walking might have. Previous study of overground walking with unilateral loadig suggested that the lengthening of step time on the loaded leg was more prevalence than the shortening of step time of the unloaded leg (Kodesh et al., 2012). However, our findings indicated significant difference in both the loaded side and the unloaded side.

Kodesh et al. (2012) tested different speeds in addition to unilateral loading in the study and found the change of step length is sensitive when faster speed is combined with unilateral loading such that the unloaded side step length decreased while the loaded side step length increased. Since our data of the unloaded side step length is contrary to the study of Kodesh et al. (2012) and it is not clear why both sides increased together, it would be reasonable to include different speed in unilateral loading study to reveal the control mechanism of step length in the future. Taken together, our study helped better understand the effect of unilateral inertia perturbation on the spatiotemporal control and long-range correlation of lower limb dynamics and serve as a reference for the assessment of gait fractal dynamics and rehabilitation for people with disabilities.

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Appendix 1: Approved IRB consent form

Georgia State University
Department of Kinesiology and Health
Informed Consent

Title: Long range correlation of gait variables during treadmill walking in young adults
Principal Investigator: Jianhua (Jerry) Wu, Ph.D.

I. Purpose:

We invite you to take part in a research study because you are a college age student. We will study the long range correlation of your walking pattern when you walk on a treadmill.

II. Procedures:

There will be 20 college age subjects in this study. You will come to the Biomechanics lab at Georgia State University for three sessions on three separate days. Each session will take about 75 minutes of your time. We will first confirm with you that you do not have any health problems that would prevent you from walking on a treadmill. Then, we will measure your weight and height. Next, you will walk barefoot on a treadmill. The treadmill measures the force between your feet and the treadmill. We will also attach reflective markers at your toe, heel, ankle, knee, hip, shoulder, elbow, wrist, and temple joints at both sides. We will use a motion capture system to record these markers. We will study the long range correlation of step time and peak impact force during treadmill walking.

At session 1, we will ask you to walk straight along a 10-meter walkway three times. We will use a stop watch to estimate your average walking speed. We consider this speed as your comfortable walking speed. We will use this speed for your treadmill walking. We will test 4 conditions: walking without ankle weights, and walking with ankle weights of about 1.4, 2.8, and 4.2 pounds at both sides. You will walk on a treadmill twice for each condition. Each time you will walk for 5 minutes. We will provide enough rest between trials.

At session 2, we will measure your comfortable walking speed as in session 1. We will ask you to kick a ball on the floor. The leg that you use to kick the ball will be your dominant leg. Then, we will test 4 conditions: walking without ankle weights, and walking with ankle weights of about 1.4, 2.8, and 4.2 pounds at your non-dominant leg. You will walk on a treadmill twice for each condition. Each time you will walk for 5 minutes. We will provide enough rest between trials.

At session 3, we will first measure your comfortable walking speed and stride frequency as in session 1. Then, we will use a metronome (clicking clock) to test 4 conditions of stride frequency during walking. These conditions will be walking without the metronome, walking with each step matching the metronome, walking with each right stride matching the metronome, and walking with every two right stride matching the metronome. You will walk on a treadmill twice for each condition. Each time you will walk for 5 minutes. We will provide enough rest between trials.

III. Risks:

You will experience no more risk from this study than in your daily life. All of our equipment is commonly used in lab studies. We will take every precaution to insure the safety of the equipment. We will place soft landing mats around the treadmill. You are free to slow the pace, take a rest, or withdraw from this study at any time. There are no social or psychological risks.

IV. Benefits:

You will learn about your walking patterns. We will gain the knowledge of long range correlation of walking pattern in young adults. We hope to use this knowledge for future study with persons with disabilities. Also, you will receive $10 for each session that you participate.

V. Voluntary Participation and Withdrawal:

Participation in research is voluntary. You do not have to be in this study. You can drop out at any time if you change your mind. You will not lose benefits if you drop out.

VI. Confidentiality:

We will keep your records private to the extent allowed by law. Only the principal investigator and research assistants will have access to your information. We may share your information with the GSU Institutional Review Board and the Office for Human Research Protection.

Your information will be confidential. We will give you a study identification number (i.e. participant 01) instead of using your name and initials. We will store your information on a password and firewall protected computer in a locked office. Only the principal investigator and research assistants will have access to the data. We will not use your personal information for presentation and publication of the research results. We will summarize study findings in group form. We will not identify you personally.

VII. Contact Persons:

If you have any questions, contact Dr. Jerry Wu at 404-413-8476 (jwu11@gsu.edu) or Mr. Toyin Ajisafe at 404-413-8056 (tajisafe1@student.gsu.edu). If you have any questions or concerns about your rights as a participant in this research study, you may contact Susan Vogtner in the Office of Research Integrity at 404-413-3513 or svogtner1@gsu.edu.

VIII. Copy of Consent Form to Subject:

We will give you a copy of this consent form to keep.
If you are willing to volunteer for this research, please sign below.

Participant:

| ____________________ | ____________________ | ____________ |
|---|---|---|
| Printed Name | Signature | Date |

Principal Investigator:

| ____________________ | ____________________ | ____________ |
|---|---|---|
| Printed Name | Signature | Date |