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The independent effect of added mass on the stability of the sagittal plane leg kinematics during steady-state human walking

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SUMMARY

This study investigated the independent effect of added mass on the stability of the leg kinematics during human walking. We reasoned that adding mass would influence the body's inertial state and thus challenge the ability of the leg to redirect and accelerate the total mass of the body while walking. We hypothesized that walking with added mass would reduce the stability of the leg kinematics. Lower extremity sagittal plane joint kinematics were recorded for 23 subjects as they walked on a treadmill at their preferred speed with and without added mass. The total mass of each subject was manipulated with combinations of simulated reduced gravity and added load. The stability of the leg kinematics was evaluated by computing the eigenvalues of the Poincaré map (i.e. Floquet analysis) that defined the position and velocity of the right hip, knee and ankle at heel-contact and mid-swing. Significant differences in stability were found between the various added mass conditions (*P*=0.040) and instant in the gait cycle (*P*=0.001). *Post-hoc* analysis revealed that walking with 30% added mass compromised the stability of the leg kinematics compared with walking without additional mass (*P*=0.031). In addition, greater instability was detected at the instance of heel-contact compared with mid-swing (*P*=0.001). Our results reveal that walking with added mass gives rise to greater disturbances in the leg kinematics, and may be related to the redirection and acceleration of the body throughout the gait cycle. Walking with added mass reduces the stability of the leg kinematics and possibly the overall balance of the walking pattern.

Key words: inertia, gait, Floquet multipliers, nonlinear.

INTRODUCTION

Maintaining a stable walking pattern is a complex motor task since the body's acceleration is constantly changing throughout the gait cycle. For example, during the first half of stance, the center of mass decelerates until reaching mid-stance and then accelerates during the second half of stance (Cavagna and Margaria, 1966). Furthermore, the velocity of the center of mass is transitioned from a downward to an upward direction by the legs during double support (Donelan et al., 2002). Hence, humans must perform external work by the legs to control the acceleration of the body's mass during walking (Cavagna and Margaria, 1966). The ability to change the acceleration of the body is governed by the body's inertial state or resistance to change in its motion. If the body's inertia grows too large, then it is possible that the leg kinematics become less stable because the legs may not be able to accelerate properly and redirect the body's mass throughout the gait cycle. Although this seems plausible, the influence of added mass on the stability of the leg kinematics during walking remains unknown.

The influence of mass alone on the biomechanics of walking has been experimentally explored by first adding weight to the subject and then counteracting the forces due to gravity with a body weight suspension system (De Witt et al., 2008; Grabowski et al., 2005). The upward lifting force provided by the suspension system compensates for the gravitational forces such that the subject's weight is maintained but the subject's overall mass is increased. Thus, adding mass without adding weight increases the amount of mass that the legs must redirect and accelerate throughout the gait cycle. Using the combination of body weight support and added weight, it has been demonstrated that walking under the influence of added mass alone increases metabolic cost and affects the velocity of the center of mass (De Witt et al., 2008; Grabowski et al., 2005). These results suggest that the acceleration and redirection of the body's mass throughout the gait cycle is important for the maintenance of the steady-state walking pattern. Potentially, these factors may also play a role in the stability of the leg kinematics.

Human walking is constantly subjected to local disturbances that arise in the natural couplings of the lower extremity dynamics and neural noise (Dingwell and Kang, 2007; Faisal et al., 2008; Hasan, 2005). These local disturbances must be dissipated within or across several strides to maintain a stable walking pattern. If the locomotor system fails to dissipate these disturbances, they will grow uncontrollably and may result in a loss of stability (Dingwell and Kang, 2007). The amount of variation in the walking pattern is often used as a metric for assessing stability (Dingwell et al., 2001; Dingwell and Marin, 2006; Gabell and Nayak, 1984; Hausdorff, 2007). However, there is less experimental evidence to support the notion that changes in the amount of variability are equated with the ability of the legs to dissipate the local instabilities that arise in the walking pattern (Dingwell et al., 2001; Dingwell and Marin, 2006). The difficulty in making this connection most probably lies in the fact that variability measures (i.e. standard deviation and coefficient of variation) do not provide insight into how the motor system responds to local disturbances over continuous strides.

Floquet analysis is a well-established technique from theoretical mechanics that has been used to quantify the rate at which local disturbances, which are naturally present in the walking kinematics, are dissipated (Dingwell and Kang, 2007; Full et al., 2002; Granata and Lockhart, 2008; Hurmuzlu and Basdogan, 1994; Hurmuzlu et al., 1996; McGeer, 1990). The walking pattern is considered to have greater stability if it is capable of dissipating these small disturbances at a faster rate. The rate at which disturbances are dissipated is quantified by the magnitude of the largest eigenvalue of the Jacobian which defines the evolution of the walking pattern from one stride (or section of the Poincaré map) to the next (Guckenheimer and Holmes, 1983). An eigenvalue that is close to zero indicates that the walking pattern will rapidly dissipate small disturbances. Alternatively an eigenvalue that is further away from zero indicates a slower dissipation of the disturbances that occurred during the walking pattern. For example, an eigenvalue of 0.24 means that 24% of a disturbance remains after a stride and that the disturbance will be asymptotically reduced over several strides (Fig. 1). A walking pattern with a larger eigenvalue (e.g. 0.54 *versus* 0.24) is considered less stable because it takes longer to dissipate the local disturbances.

The purpose of this study was to investigate the effect of added mass on the stability of the sagittal plane leg kinematics during walking. Stability was defined here to be the rate at which small disturbances in the leg kinematics are dissipated from one stride to the next. We added mass to each subject to directly increase the body's inertial state. We hypothesized that increasing body mass alone would lead to a decrease in the stability of the leg kinematics during steady-state walking.

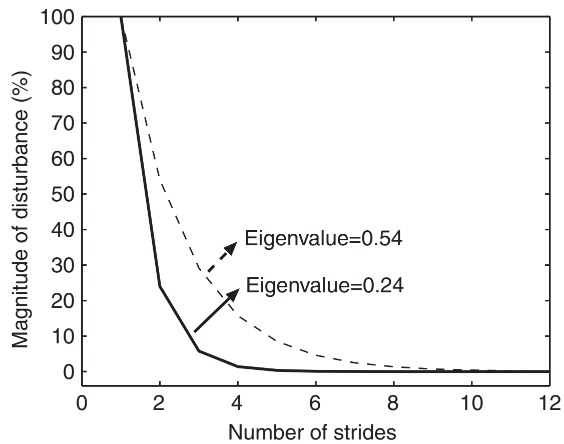

Fig. 1. Eigenvalues signify the rate at which disturbances in the leg kinematics are dissipated over multiple strides. The smaller eigenvalue (0.24) indicates that disturbances dissipate at a faster rate whereas the larger eigenvalue (0.54) dissipates disturbances at relatively slower rate. Thus, the faster rate at which disturbances are dissipated, the more stable the walking pattern.

MATERIALS AND METHODS

We collected data from 23 volunteer subjects (age 23.8±4.5 years, mass 63.9±8.7 kg, height 1.7±0.1 m; mean ± s.d.). All subjects were healthy young adults with no injuries or known pathological problems. All subjects underwent a verbal interview, and read and signed an informed consent document that was approved by the University's Committee for the Protection of Human Subjects prior to experimental data collection.

To begin the experiment, each subject put on neoprene wetsuit bottoms followed by a support vest (Biodex Medical Systems, NY, USA) that was secured around the torso and upper thighs. The neoprene wetsuit bottoms helped reduce chafing during walking and improved the comfort and fit of the support vest. The subjects initially selected a comfortable walking speed that could be maintained for a long duration by manually increasing and decreasing the treadmill speed. Once their self-selected speed was chosen, the subjects warmed up while walking on the treadmill for a total of 6 min. A self-selected speed was used in this investigation because it best represented the subject's natural walking dynamics. This notion is based on previous experimental data that has shown that changes in walking speed above or below a preferred gait influences the dynamic stability of the walking pattern (Dingwell and Marin, 2006; England and Granata, 2007). The self-selected speed (0.98±0.24 $\mathrm{m\,s^{-1}}$, mean ± s.d.) was maintained for all the respective inertia conditions.

The subjects completed walking trials with 0%, 10%, 20% and 30% of added body mass. Each of the mass conditions was presented in a random order. First, thin lead strips (0.45 kg each) were firmly attached symmetrically around the hips using a modified hip belt (Fig. 2). By adding the lead strips to the subject, we increased both weight and mass. To compensate for the increased forces due to weight, an upward vertical force was applied (i.e. equal to the amount of added weight) by a customized body weight suspension system that easily attached to the support vest worn by each subject. The subject stood quietly on the treadmill while in line with the overhead pulley as the upward lifting force was applied in the direction of gravity. A strain gauge load cell (Omega Engineering, Stamford, CT, USA) was placed in parallel with a cable-spring-pulley and supplied a voltage proportional to the applied lifting force. The lifting force recorded by the strain gauge load cell was monitored in real-time with the aid of a desktop computer. To apply the lifting force to the subject, rubber springs were stretched using a hand winch. Once the desired lifting force was achieved, the hand-winch was locked in place to allow the lifting force to be constantly applied throughout the entire duration of each walking trial. The average lifting force (mean ± s.d.) applied to each subject was 14.27±2.06 lbs, 27.26±3.83 lbs and 40.40±6.22 lbs while walking with 10%, 20% and 30% added mass, respectively.

For each condition, the subjects walked for a total of 4 min and we collected lower extremity kinematics during the last 3 min. A high speed (120 Hz) six-camera motion capture system (ViconPeak, Centennial, CO, USA) was used to record the three-dimensional positions of marker triangulations that were placed on the right foot, shank and thigh segments. A 5 s standing calibration was collected to determine the anatomical reference system for each segment. Subjects were instructed to stand upright, distribute their weight evenly on both feet, and keep both knees in a locked position. The location of the markers during the standing calibration trial was used to correct any misalignment of the local reference vectors that defined each of the respective lower extremity segments (Nigg et al., 1993). The position data for all markers were filtered using a zero-lag Butterworth filter, and the selected cutoff frequencies were determined using the Jackson Knee Method with a prescribed limit of 0.01 $\mathrm{m\,Hz^{-1}\,Hz^{-1}}$ (Jackson, 1979). The range for the optimal cutoff frequency in the x, y and z coordinates ranged from 5–7, 3–5 and 4–7 Hz, respectively. The Jackson Knee Method was used because it allowed for the maximum attenuation of noise in each respective coordinate while still preserving the relevant content of the signal.

Based on the filtered marker positions, we calculated the sagittal plane joint angular positions and velocities of the ankle, knee and hip. We evaluated the stability of only the sagittal plane leg kinematics because they represent the dominant plane of motion during walking (Mah et al., 1994). The joint angles and velocities from the continuous time series were extracted for each instance of heel-contact and mid-swing that occurred during the gait cycle using customized laboratory software. The instant of heel-contact was

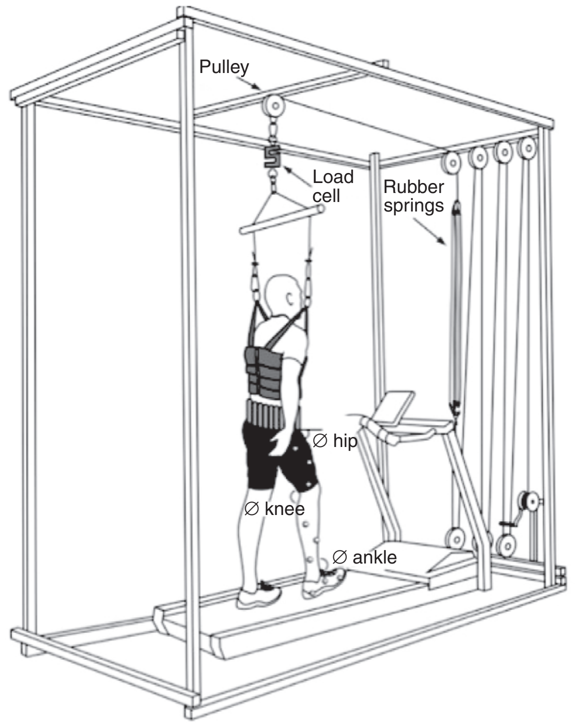

Fig. 2. Body weight suspension system. Thin lead strips were symmetrically attached around the waist using a modified hip belt while subject walked on an instrumented treadmill. To compensate for the added weight, an upward vertical force was applied to allow subject to experience only the effects of inertial forces.

defined as the maximum position of the heel marker in the forward direction for each stride. The instant of mid-swing was defined as the maximum knee flexion that occurs during the swing phase of the gait cycle. These discrete points were used to construct Poincaré maps that were used to determine the stability of the leg kinematics at these instances of the gait cycle. Several technical articles on the use of Floquet analysis have indicated that the eigenvalues of the Jacobian will not depend on the choice of the Poincaré section for steady-state walking (Dingwell and Kang, 2007; Hurmuzlu and Basdogan, 1994). We chose to partition the stability of the leg kinematics into the stance and swing because previous biomechanical studies have indicated that these phases of gait may be under different balance control mechanisms (Frenkel-Toledo et al., 2005; Gabell and Nayak, 1984). Additionally, the instances of heel-contact and mid-swing were selected because they represent repeatable features of the leg kinematics that can be reliably extracted from the kinematic data for the construction of the Poincaré maps.

The eigenvalues of the Poincaré map were used to measure the stability of the leg kinematics at the instant of heel-contact and mid-swing during the gait cycle (Granata and Lockhart, 2008; Hurmuzlu and Basdogan, 1994; Hurmuzlu et al., 1996; McGeer, 1990; Tedrake et al., 2004). The sagittal plane joint positions and velocities of the right ankle, knee and hip were used to define the a state vector that defined the dynamics of the leg as:

$$x = [\phi_1 \phi_2 \phi_3 \dot{\phi}_4 \dot{\phi}_5 \dot{\phi}_6]^T . \quad (1)$$

The six state variables denote the angular positions ($\phi_1\phi_2\phi_3$) and angular velocities ($\dot{\phi}_4\dot{\phi}_5\dot{\phi}_6$) at the ankle, knee and hip respectively. For steady-state human locomotion, the walking pattern achieves dynamic equilibrium. This property was defined by the following relationship:

$$\boldsymbol{x}^* = f(\boldsymbol{x}^*) . \quad (2)$$

The variable $\boldsymbol{x}^*$ represents the state vector in the Poincaré map, and f is the function that describes the change in the location of the state vector from one stride to the next. Ideally, if the walking pattern was completely periodic (i.e. no deviation from the preferred joint kinematics), the function would map to the same point on the diagonal of the Poincaré map. However, this is not the case because the dynamics of legs slightly fluctuate from stride-to-stride. The equilibrium point was estimated by computing the average of all the discrete points in the respective Poincaré maps (Hurmuzlu and Basdogan, 1994; Hurmuzlu et al., 1996).

Perturbations were linearized about the equilibrium state vector $\boldsymbol{x}^*$ according to the following equation:

$$\delta_{x_{n+1}} = \mathbf{J}\delta_{x_n} , \quad (3)$$

where δ denotes the deviation from the equilibrium state vector, and $\mathbf{J}$ is the Jacobian which defined the rate of change of the state variables from one stride (n) to the next (n+1). δ_{x_n} and $\delta_{x_{n+1}}$ were defined as:

$$\delta_{x_n} = [x_n - \boldsymbol{x}^*, x_{n+1} - \boldsymbol{x}^*, x_{n+2} - \boldsymbol{x}^*, x_{n+3} - \boldsymbol{x}^*, \ldots] , \quad (4)$$

$$\delta_{x_{n+1}} = [x_{n+1} - \boldsymbol{x}^*, x_{n+2} - \boldsymbol{x}^*, x_{n+3} - \boldsymbol{x}^*, x_{n+4} - \boldsymbol{x}^*, \ldots] . \quad (5)$$

A least squares algorithm (Tedrake et al., 2004) was used to solve for $\mathbf{J}$ (Eqn 6), and the stability of the leg's kinematics were determined by calculating the eigenvalues of $\mathbf{J}$:

$$\mathbf{J} = [(\delta_{x_{n+1}})(\delta_{x_n})^T][(\delta_{x_n})(\delta_{x_n})^T]^{-1} . \quad (6)$$

The maximum eigenvalue (β) of $\mathbf{J}$ was used to quantify the stability of the leg kinematics at the instant of heel-contact and at mid-swing of the gait cycle. A β value that was further away from zero was considered less stable than those that were closer to zero. Theoretically, a walking pattern that has a β value that is closer to one requires a longer time to return back to the steady-state walking pattern. The longer the time needed to return back to steady state presumably indicates a less stable gait pattern.

To be cautious in our Floquet analysis, we tested the possibility that the calculated eigenvalues were a result of correlated noise in the walking pattern rather than instabilities within the leg kinematics. We used an algorithm developed by Theiler et al. (Theiler et al., 1992) to generate surrogate data sets from the original discrete points that were used to construct the Poincaré maps. The surrogation algorithm is based on a phase randomization process that destroys any deterministic features of the data but maintains the mean, variance and power spectra (Miller et al., 2006; Theiler et al., 1992). The maximum eigenvalue (β_{SUR}) was computed for each surrogate, and a paired t-test was used to test if the original (β) and surrogate (β_{SUR}) values were significantly different. A significant difference would provide evidence that the β values from the original data were not due to correlated noise in the walking pattern.

The number of subjects used in this investigation was based on an *a priori* power analysis. Based on a sample size of twenty subjects and a conservative effect size of 0.8, the power analysis revealed a type II error rate equal or lesser than 0.20 (i.e. power $\geq$80%) to detect differences in stability (the largest eigenvalue, β) between inertia conditions. The conservative effect size of 0.8 was based on the data from Hurmuzlu et al. (Hurmuzlu et al., 1996), in which values of β were reported for post-polio patients and normal healthy subjects. An effect size of 3.75 was computed by taking the

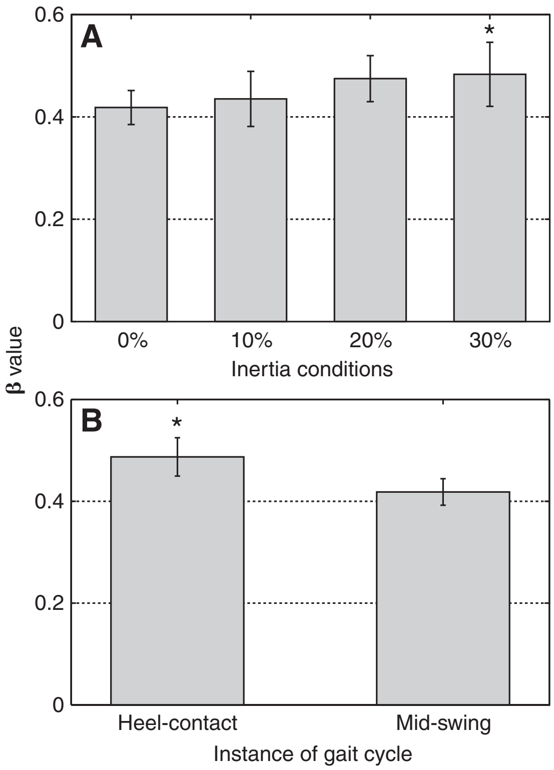

Fig. 3. (A) The β values (means ± s.d.) for the inertia main effect for normal walking and walking with 10%, 20% and 30% inertia. The asterisk indicates that subjects were significantly less stable while walking with 30% inertia when compared with walking without added inertia. (B) The β values (means ± s.d.) for the instance of the gait cycle main effect. The asterisk indicates less stability at heel-contact compared with mid-swing.

difference in the mean β values between both groups and dividing by the standard deviation of the healthy group. With an effect size of 3.75, the power analysis yielded an extremely low sample size of only four subjects. Since our subject population was healthy and because we expected smaller changes in stability between mass conditions, we decided to utilize a conservative effect size of 0.8. With an effect size of 0.8, the power analysis yielded an objective sample size of twenty or more subjects. We used a general linear model (GLM) analysis with two within-subjects fixed factors (mass and instant of the gait cycle) to compare the values of β among the respective mass conditions (0%, 10%, 20% and 30% of body mass). An additional GLM analysis was used to test for any possible differences in the equilibrium points at heel-contact and mid-swing under the respective inertia conditions. All statistical tests were performed with an α-level of 0.05.

RESULTS

Overall, added mass decreased the stability of the leg kinematics during walking. Statistical analyses revealed significant main effects for mass (P=0.040) and instant of the gait cycle (P=0.0001). Compared with walking without added mass, the average β for the mass main effect increased by 15.5% at the highest added mass condition of 30% (Fig. 3A). The main effect for the instant of the gait cycle revealed that the average β value was greater at heel-contact compared to mid-swing (Fig. 3B), indicating greater instability in the leg kinematics at heel-contact. However, no significant mass-by-instant of the gait cycle interaction was found (P=0.751). *Post-hoc* analysis indicated that the leg kinematics were significantly less stable while walking with 30% added mass when compared to walking without added mass (P=0.031). Furthermore, our trend analysis indicated a significant increasing linear trend for β as mass increased (P=0.007), providing further indication that additional mass influenced the stability of the leg kinematics while walking.

Our surrogate analysis revealed a significant difference between the mean values computed for the original data series (β) and the surrogate data series (β_{SUR}, refer to Table 1). The results confirm that the calculated β values were probably not the result of correlated noise in the walking data.

The equilibrium values computed at heel-contact and mid-swing are given in Table 2. Compared with normal walking, the equilibrium point for the hip at heel-contact significantly shifted upward (i.e. increased joint flexion) as mass increased from 0% to 10% (P=0.033), 20% (P=0.011) and 30% (P=0.029). The equilibrium point for the knee at heel-contact did not significantly change across mass conditions (P=0.472). However, the equilibrium point for the ankle joint at heel-contact significantly shifted upward at 10% additional mass (P=0.006) compared with walking without added mass, but was not significantly different at 20% (P=0.185) and 30% (P=0.323) added mass.

Changes in the equilibrium points were found for the hip and knee joint at mid-swing. Compared with normal walking, the equilibrium points for the hip and knee at mid-swing shifted significantly upward at 10%, 20% and 30% (all values for P<0.001, respectively). The equilibrium points for the ankle at mid-swing did not significantly change with increased mass (P>0.05).

DISCUSSION

The outcomes of this investigation supported our hypothesis that increasing body mass alone would lead to a decrease in the stability of the sagittal plane leg kinematics during steady-state walking. Based on our experimental design, we infer that the leg kinematics were less stable because the added mass influenced the inertial state of the body throughout the walking gait cycle (De Witt et al., 2008). Changes in the body's inertial state appear to create small disturbances in the leg kinematics while walking. Based on our results, we suggest that controlling the acceleration of the body mass throughout the gait cycle may be essential for maintaining walking stability.

The results presented in this investigation also indicate that the leg kinematics were more stable at mid-swing than at heel-contact while walking with additional mass. Most probably the differences in stability of the leg kinematics may be related to the fact that heel-contact signifies the point in the gait cycle where the body is undergoing larger changes in velocity as the body's mass is transferred from the trailing leg to the leading leg (Donelan et al., 2002). Alternatively, it is possible that the differences in the stability of the leg kinematics at heel-contact and mid-swing may be related to the viewpoint that stance and swing phase dynamics

Table 1. Group means and standard deviations for the β values generated from the original and surrogate data series

	Heel-contact		Mid-swing	
	Original	Surrogate	Original	Surrogate
β	0.49±0.04*	0.43±0.03	0.42±0.03*	0.38±0.01

*Significant difference between the original and the surrogate data (P=0.001).

Table 2. Equilibrium values of the sagittal plane joint rotations for each inertia condition while walking at steady state

	Heel-contact (deg.)			Mid-swing (deg.)		
Inertia	Hip	Knee	Ankle	Hip	Knee	Ankle
0%	24.83±3.70	1.40±3.51	–2.62±3.37	23.25±3.47	70.61±5.37	–8.29±5.23
10%	25.48±3.49*	1.68±4.16	–1.51±3.36*	24.65±3.64*	72.15±5.19*	–7.89±5.16
20%	25.88±3.60†	1.84±4.57	–2.09±3.22	25.47±3.68†	72.82±5.60†	–8.27±5.50
30%	25.80±3.56‡	1.93±4.01	–2.18±3.33	25.22±3.87‡	72.43±5.39‡	–8.61±5.37

Significance was set at P=0.05; *significant difference between 0% and 10%; †significant difference between 0% and 20%; and ‡significant difference between 0% and 30%. Values are means ± s.d.

are governed by different balance control mechanisms (Frenkel-Toledo et al., 2005; Gabell and Nayak, 1984). This notion is partly supported by experimental evidence that has shown that humans respond differently to disturbances experienced in the different phases of the gait cycle (Dietz et al., 2004). However, additional studies are necessary to better understand the differences in stability of the leg kinematics during the stance and swing phase.

Our results indicated that adding an additional 30% of body mass influences the equilibrium points of the Poincaré sections at heel-contact and mid-swing. However, it should be noted that the significant changes in the sagittal plane joint kinematics were quite small (i.e. less than 2 deg.). We suggest that these slight changes are important given the fact that previous research has indicated that limb posture plays a crucial role in the stabilization of gait (Biewener and Daley, 2007). Alternatively, it is possible that the noted changes in the joint kinematics were small since the largest eigenvalue did not approach one in any of our experimental conditions. This may indicate that the local disturbances due to added mass were not sufficiently large enough to require considerable changes in the joint kinematics.

There are several limitations to our experimental methods that need to be considered. Our method of reducing the gravitational forces was accomplished by applying a vertical lifting force with a fixed pulley system and support vest that was worn around the torso. The vest may have assisted with the stability of the leg dynamics by providing additional torso control. It is also possible that small horizontal forces were introduced if the subject did not stay directly below the fixed pulley. Potentially, these horizontal forces may have influenced our measures of stability. However, we are skeptical that these horizontal forces would have influenced the outcomes of our study because they would be considerably smaller than the vertical lifting forces applied to the body during our experiments. We additionally noted that the vertical lifting forces were more variable as additional mass was added to the subject. It is difficult to determine if these variations were due to the design of the body weight support system or disturbances present in the walking patterns of our subjects. Since we found differences in the largest eigenvalue and the equilibrium points, we suspect that these changes were probably a reflection of the disturbances present in the walking pattern rather than a flaw in the design of our apparatus.

It should also be noted that the participants selected a walking speed that was slower than that typically reported in the literature (~1.5 $\mathrm{m\,s^{-1}}$). This is probably a result of our experimental protocol. Subjects in this study manually increased and decreased the treadmill speed while starting from a standing position as opposed to first starting at a slow walking speed and then at a high walking speed as described by Martin et al. (Martin et al., 1992). A faster walking speed would increase the need to accelerate the body mass. As such, we may have detected differences in the stability of the leg kinematics for all the experimental conditions by having the subjects walk faster.

Although Floquet analysis has demonstrated that the elderly individuals and individuals with Polio have a less stable walking pattern, the possible mechanisms behind these losses of stability has not been identified (Dingwell et al., 2008; Hurmuzlu et al., 1996). Our results might provide initial insights into the understanding how stability in the leg kinematics may be lost. For example, it is possible that the less stable walking pattern seen in the elderly may be related to an inability of the legs to effectively accelerate and redirect the body mass throughout the gait cycle. Furthermore, our results may be useful for estimating the stability of the limb kinematics during lunar and Martian extra vehicular activities where the astronaut may be carrying additional mass while in a reduced gravity environment. In this case, the additional mass may play a role in how stable the leg kinematics are while walking across the terrain of various celestial bodies.

In summary, we found that added mass reduces the stability of the leg kinematics during steady-state walking. These results indicate that the inertial state of the body plays a role in the stability of the leg kinematics and may be related to how the body is redirected and accelerated during walking. Effective acceleration of the body mass throughout the gait cycle appears to be an important factor for stable limb kinematics and possibly walking balance.

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