Does Load Carrying Influence Sagittal Plane Locomotive Stability?

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ABSTRACT

ARELLANO, C. J., C. S. LAYNE, D. P. O'CONNOR, M. SCOTT-PANDORF, and M. J. KURZ. Does Load Carrying Influence Sagittal Plane Locomotive Stability? *Med. Sci. Sports Exerc.*, Vol. 41, No. 3, pp. 620–627, 2009. **Purpose**: We used methods from dynamical system analysis to investigate the effect of carrying external loads on the stability of the locomotive system and sagittal plane kinematics. We hypothesized that carrying an additional load at the waist would 1) decrease the dynamic stability of the locomotive system and 2) cause changes in the location of the Poincaré map's equilibrium point for the hip, the knee, and the ankle joint kinematics. **Methods**: Lower extremity kinematics were recorded for 23 subjects as they walked on a treadmill at their preferred speed while carrying external loads of 10%, 20%, and 30% of their body weight around their waist. Gait stability was evaluated by computing the eigenvalues of the locomotive system at the instance of heel contact and midswing. Changes in the hip, the knee, and the ankle's equilibrium point of the Poincaré sections were used to determine whether there were changes in the joint kinematics while carrying external loads. **Results**: No significant differences in sagittal plane stability were found between the respective load carrying conditions ($P > 0.05$). Significant changes ($P < 0.05$) in the equilibrium points of the hip and the knee were found at heel contact and midswing. **Conclusions**: The data suggest that humans are capable of maintaining sagittal plane stability while carrying loads up to 30% of their body weight. **Key Words:** POINCARÉ, EIGENVALUE, LOAD CARRIAGE, WALKING, GAIT, LOCOMOTION

Most research on the effect of load carrying on the performance of the locomotive system has focused on the physiological consequences of carrying an external load around the waist or in an external backpack (13,15,30). The human locomotive system involves the integration of neurophysiological and musculoskeletal systems to perform gait. Although there appears to be a linear relationship between the metabolic cost and the amount of load carried (13), it is unknown if load carrying decreases the stability of the gait pattern. Previous computer models of bipedal walking indicate that "carrying a payload at the hip" decreases the stability of the gait pattern in the sagittal plane (28). Load carrying may lead to a less dynamically stable walking pattern in humans due to the effects of the additional gravitational and inertial forces acting on the locomotive system (13). If load carrying does influence the dynamic stability of the gait pattern, it seems reasonable that we would also expect changes in the sagittal plane kinematics. However, there does not appear to be a consensus in the literature as to whether the sagittal plane lower extremity kinematics are altered while carrying an external load. Several investigations have reported no significant differences in sagittal plane joint kinematics while carrying external loads that range from 10% to 64% of an individual's body weight (4,17,35). Alternatively, others have reported an increased amount of knee flexion and ankle dorsiflexion while carrying external loads that range from 20% to 40% of the individual's body weight (23,32). Possibly, the different joint kinematics noted in the literature may represent ways to ensure a stable gait pattern while carrying external loads.

Maintaining dynamic stability is of great importance for the maneuverability and the prevention of a fall that might occur while carrying an external load. Epidemiological data reported by Andersson and Lagerlöf (1) indicate that 53% of workplace musculoskeletal injuries are a result of slipping related falls while carrying a load. Potentially, carrying loads may place the locomotive system in a less stable state where the rate of recovery from an unexpected perturbation may be slower. Because 17% of the disabling injuries in the work environment are due to a fall (18), further studies are necessary to determine whether carrying an external load influences the dynamic stability of the gait pattern. The stability of the gait pattern can be assessed by introducing a perturbation such as a slip or a trip during the gait cycle (6,33). In this case, the gait pattern is more stable if a fall does not occur, and if it takes fewer gait cycles to return

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Submitted for publication October 2007.
Accepted for publication July 2008.

0195-9131/09/4103-0620/0
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DOI: 10.1249/MSS.0b013e31818a0ea4

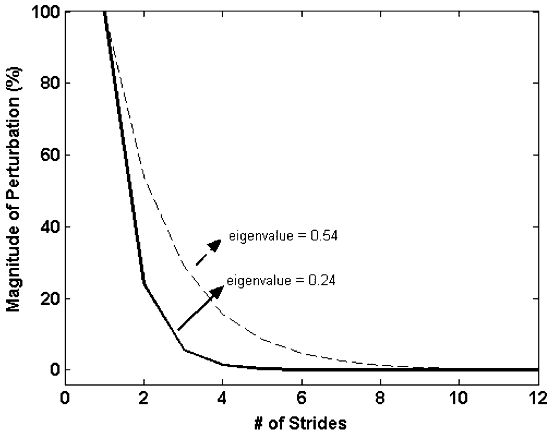

FIGURE 1—Eigenvalues signify the rate of recovery from a perturbation over multiple strides. The smaller eigenvalue (0.24) takes approximately 5 strides to recover whereas the larger eigenvalue (0.54) takes approximately 11 strides to recover. Thus, the faster the rate of recovery, the more stable the gait pattern.

back to the preferred steady-state gait pattern. However, the use of external perturbations does have limitations, in particular because the participant will often switch to a guarded gait after the first perturbation trial (31). Hence, this may limit the ability to quantify the stability of the gait pattern while carrying an external load.

The field of dynamic systems has established complementary mathematical tools for quantifying the stability of the locomotive system that may provide additional insights (11). For example, Floquet analysis is similar to the perturbation analysis in that it evaluates the ability of the locomotive system to return back to its preferred steady-state gait over time. The amount of divergence from the steady-state gait pattern can be quantified based on the eigenvalues of the linearized stride-to-stride Jacobian matrix, which approximates the effect of small perturbations that arise during walking (11,14,19,20,28). The magnitude of the eigenvalues ranges from zero to one and indicates the rate of recovery in response to a perturbation. An eigenvalue that is closer to zero indicates that the locomotive system will rapidly recover from a perturbation, whereas an eigenvalue that is closer to one indicates a slower recovery (Fig. 1). For example, an eigenvalue of 0.24 means that 24% of the perturbation remains after a stride (11) and that this perturbation will be asymptotically reduced over the next consecutive strides (e.g., 5.76% after two strides, 1.38% after three strides, etc.). A system with a larger eigenvalue is considered less stable because it takes longer for the locomotive system to return back to the steady-state gait pattern and has a higher probability of falling if an additional perturbation is encountered during the recovery period (11,28). The use of the eigenvalues to evaluate the stability of the gait pattern has been well supported by experiments with walking robots (28) and has recently been shown to be a valuable tool for distinguishing between fallers and nonfallers in the aging population (14). Potentially, evaluating the eigenvalues of the locomotive system will also provide further insight on how carrying a load influences the dynamic stability of the gait pattern.

The field of dynamical systems has also established the use of a Poincaré map to simplify our understanding of the evolving dynamics of the locomotive system (Fig. 2). The Poincaré map consists of repeatable discrete points in the rhythmical behavior of the locomotive system and is used to calculate the eigenvalues of the system (14,19,20). An exemplary discrete point would be the hip joint angle at heel contact for a series of gait cycles. The average of all points for the hip joint angle at heel contact in the Poincaré map is called the system's equilibrium point (x^*) for that joint and represents the preferred steady-state movement pattern. A significant upward or downward shift of the equilibrium point along the diagonal of the Poincaré map represents changes in the system's preferred joint kinematics (19,20). Potentially, changes in the location of the equilibrium points may provide additional insight on the relationship between the preferred joint kinemics and the stability of the gait pattern while carrying external loads about the waist.

The purpose of this study was to use dynamical system analysis to investigate the effect of carrying a load about the waist on the stability of the locomotive system and the joint kinematics in the sagittal plane. We reasoned that walking with additional loads placed about the waist would presumably challenge the locomotive system and cause greater stride-to-stride instabilities that occur within the joint kinematics during walking. Thus, we hypothesized that stability of the gait pattern in the sagittal plane would decrease as additional loads placed about the waist

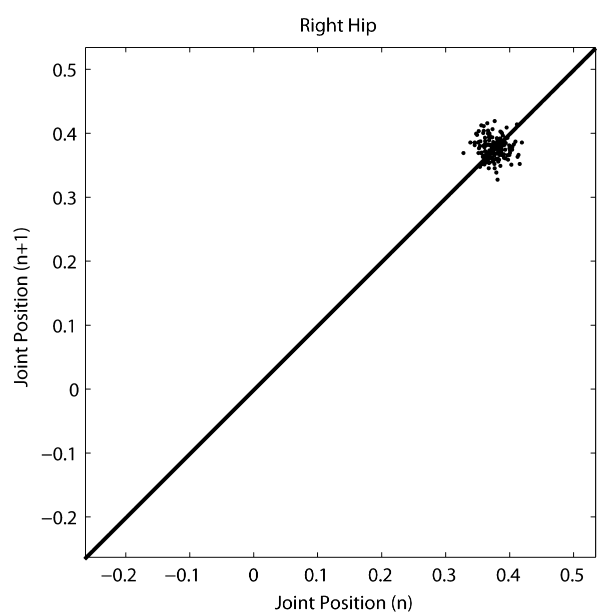

FIGURE 2—Poincarè map for steady-state locomotion. The horizontal axis denotes the value of the joint position (radians) at step n, and the vertical axis denotes the value of the joint position (radians) at the subsequent step $n + 1$. Due to periodicity, the hip angle achieves dynamic equilibrium, which is denoted by a cluster of points along the diagonal line.

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increased. We also hypothesized that additional loads placed about the waist would be accompanied by changes in the location of the equilibrium point of the Poincaré map for the hip, the knee, and the ankle joint kinematics.

METHODS

For a sample size of twenty subjects, an assumed minimal correlation of 0.50 between repeated measures, and a conservative effect size of 0.80, the power analysis before data collection revealed a type II error rate equal or lesser than 0.20 (i.e., power = 80%) to detect differences in stability (the largest eigenvalue, β) between loading conditions. The conservative effect size of 0.8 was based on the data from Hurmuzlu et al. (20), in which values of β were generated from postpolio patients and normal healthy subjects. The effect size in Hurmuzlu et al. (20) was 3.75; however, a value of 0.8 was used because we expected smaller changes in stability between loading conditions because our subjects were healthy young individuals. On the basis of our sample size calculations, 23 subjects were recruited to volunteer in the study (age = 23.8 ± 4.5 yr, mass = 63.9 ± 8.7 kg, height = 1.7 ± 0.1 m). 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 Institutional Review Board.

Subjects walked on a treadmill while carrying external loads of 0%, 10%, 20%, and 30% of their normal body weight. The maximum external load used for this investigation was determined from our pilot investigation where we determined that 30% was the maximum that our subjects could comfortably tolerate for a long-term steady-state walking trial. Before data collection, subjects were instructed to warm-up while walking on the treadmill at a self-selected pace for a total of 6 min. Subjects were instructed to choose a comfortable walking speed that could be maintained for a long duration. The subjects identified their walking speed by manually increasing and decreasing the treadmill speed while the treadmill belt was initially stationary. The self-selected pace (mean ± SD = 0.98 ± 0.24 $m{\cdot}s^{-1}$) was maintained for all the respective external load conditions. The participants completed 154.7 ± 1.3 strides for each of the respective conditions. Each of the external load conditions was presented in a random order. The load was applied using thin lead strips (0.45 kg each) that were firmly attached symmetrically around the waist via a modified hip belt (Fig. 3). We added the weight around the waist because it was close to the center of mass of the subjects and reduced the possibly that the altered lower limb performance was related to the posture of the torso while carrying additional weight. Additionally, if the weight was carried in a backpack, the distance that the weight was from the center of mass may have been different for subjects with different anthropometrics. This could impact the magnitude of the moment that would be created at the hip. Hence, by carrying the load around the waist, we reduced the probability that load placement influenced our results.

For each condition, the subjects walked for a total of 4 min, and we collected biomechanical data from their movement performance from the last 3 min. A high speed (120 Hz) six-camera motion capture system (ViconPeak Inc., Centennial, CO) was used to record the three-dimensional positions of marker triangulations that were placed on the right foot, shank, and thigh segments (Fig. 3). 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 (29). The x, y, and z axes represented the segment's anterior/posterior, medial/lateral, and vertical directions, respectively. 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 (22) with a prescribed limit of 0.01 $m{\cdot}Hz^{-1}{\cdot}Hz^{-1}$. The range for the optimal cutoff frequency in the x, y, and z coordinates ranged from 5 to 7,

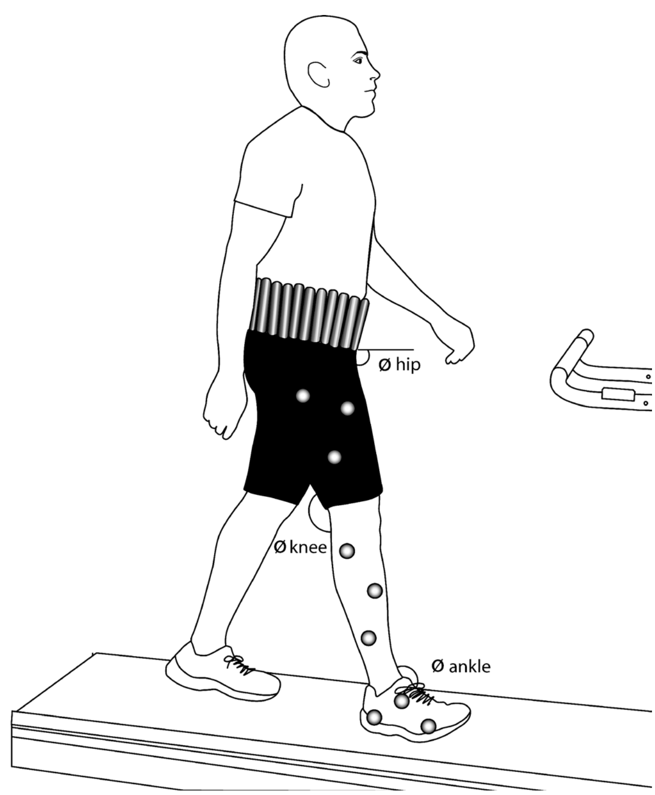

FIGURE 3—Thin lead strips were symmetrically attached around the waist using a modified hip belt while subjects walked on an instrumented treadmill. The relative angles of the hip, knee, and ankle are shown. The hip angle was calculated the with respect to the horizontal axis.

from 3 to 5, and from 4 to 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.

On the basis of 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 the sagittal plane dynamics because they represent the dominant plane of motion during walking (26). The joint angles and velocities from the continuous time series were extracted at each instance of heel contact and midswing that occurred during the gait cycle using customized laboratory software. The instance of heel contact was defined as the maximum position of the heel marker in the forward direction for each step of the gait cycle. The instance of midswing was defined as the maximum knee flexion that occurs during the swing phase of the gait cycle. These discrete points were used to construct the Poincaré maps that were used to determine the stability at these instances of the gait cycle. We selected heel contact because it has been identified as a critical part of the stance phase that is most sensitive to the amount of load carried (18). Midswing was evaluated because recent computer models of gait have indicated that the motion of the swing leg influences the stability of the gait pattern (25). Furthermore, these instances were also selected because they represent repeatable features of the kinematics that can be used to construct Poincaré maps.

The eigenvalues of the system were used to measure the dynamic stability at the instance of heel contact and midswing of the gait cycle (14,19,20,28). We assumed that the dynamics of the locomotive system were captured by the joint positions and velocities of the right ankle, knee, and hip. These variables were used to define the state vector that defined the dynamics of the system (equation 1).

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

The six state variables denote the angular positions (ϕ_1, ϕ_2, ϕ_3) and the angular velocities ($\dot{\phi}_4, \dot{\phi}_5, \dot{\phi}_6$) at the ankle, knee, and hip, respectively. For steady-state human locomotion, the locomotive system achieves dynamic equilibrium. This property was defined by equation 2.

$$x^* = f(x^*) \quad [2]$$

The variable x^* is the equilibrium point in the Poincaré map, and f is the function that describes the change in the location of the equilibrium point from one stride to the next. Ideally, if the gait pattern were completely periodic (i.e., no deviations in preferred joint kinematic trajectory), the function would map to the same point on the diagonal of the Poincaré map from one stride to the next. However, this is not the case because the dynamics of human locomotion slightly fluctuate from stride to stride due to neural errors or disturbances in the coupling of the lower extremity segments. The equilibrium point was estimated by computing the average of all the discrete points in the respective Poincaré maps (19,20).

Perturbations were linearized about the equilibrium point x^* according to equation 3.

$$\delta \mathbf{x}^{n+1} = \mathbf{J}\delta \mathbf{x}^n \quad [3]$$

δ denotes the deviation from the equilibrium point, and **J** is the Jacobian that defined the rate of change of the state variables from one stride (n) to the next (n + 1). $\delta \mathbf{x}^n$ and $\delta \mathbf{x}^{n+1}$ were defined according to equations 4 and 5, respectively.

$$\delta \mathbf{x}^n = [X_n - X^*, X_{n+1} - X^*, X_{n+2} - X^*, X_{n+3} - X^*, \ldots] \quad [4]$$

$$\delta \mathbf{x}^{n+1} = [X_n + 1 - X^*, X_{n+2} - X^*, X_{n+3} - X^*, X_{n+4} - X^*, \ldots] \quad [5]$$

A least-squares algorithm (34) was used to solve for **J** (Equation 6), and the stability of the locomotive pattern was determined by calculating the eigenvalues of **J.**

$$\mathbf{J} = [(\delta \mathbf{x}^{n+1})(\delta \mathbf{x}^n)^{\mathrm{T}}]\,[(\delta \mathbf{x}^n)(\delta \mathbf{x}^n)^{\mathrm{T}}]^{-1} \quad [6]$$

The maximum eigenvalue (β) of the system was used to quantify the stability at the instance of heel contact and at midswing of the gait cycle. A β value that was further away from zero was considered less stable than those that were closer to zero (11,19,20,28). Theoretically, a locomotive pattern that has a β value that is further away from zero requires a longer time to return back to the steady-state gait pattern. The longer the time needed to return back to steady state indicates a less stable gait pattern.

We used a general linear model (GLM) repeated-measures analyses with two within-subjects fixed factors (load and instance of the gait cycle) to determine whether there were any significant differences in the magnitudes of the maximum β between the respective load conditions (0%, 10%, 20%, and 30% of body weight). An additional GLM repeated-measures analysis was used to test for any differences in the equilibrium points at heel contact

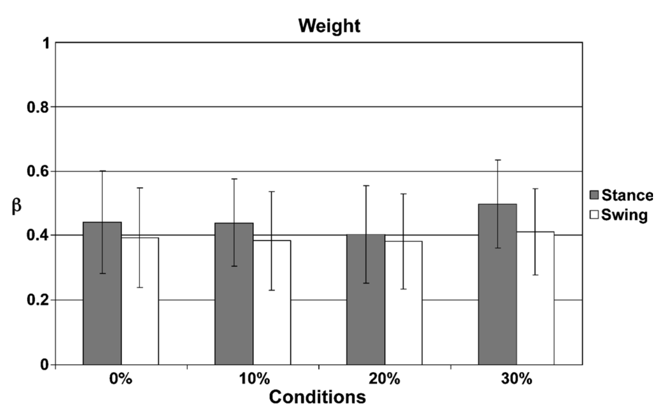

FIGURE 4—β values (mean ± SD) for normal walking and walking with 10%, 20%, and 30% additional body weight. The maximum eigenvalue, β, was computed by sampling the Poincaré section at the instance of heel strike (stance phase) and maximum knee flexion (swing phase).

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and midswing under the respective load conditions. All statistical tests were performed with an alpha level of 0.05.

RESULTS

There was no significant main effect for load (P = 0.270) and no significant interaction between load and instance of the gait cycle (P = 0.581). However, there was a significant main effect for the β values computed at the instance of the gait cycle (P = 0.025; Fig. 4). These results indicate that the participants had the same stability in the sagittal plane at the instance of heel contact and midswing for the respective load carrying conditions. Figures 5 and 6 depict the equilibrium points for each individual joint at heel contact and midswing under the respective load conditions. Graphically, it is apparent that the equilibrium points for the hip and the knee at heel contact increased as the magnitude of the load increased. An increase in the equilibrium point denotes an increase in joint flexion. Compared with unloaded walking, the equilibrium point for the hip and the knee at heel contact significantly shifted upward (i.e., increased joint flexion) as the load increased from 0% to 10% (P < 0.001 and P = 0.012, respectively), 20% (P < 0.001 and P = 0.003, respectively), and 30% (P < 0.001 and P < 0.001, respectively). Complementary changes in the equilibrium points were found for the hip and the knee joint at midswing.

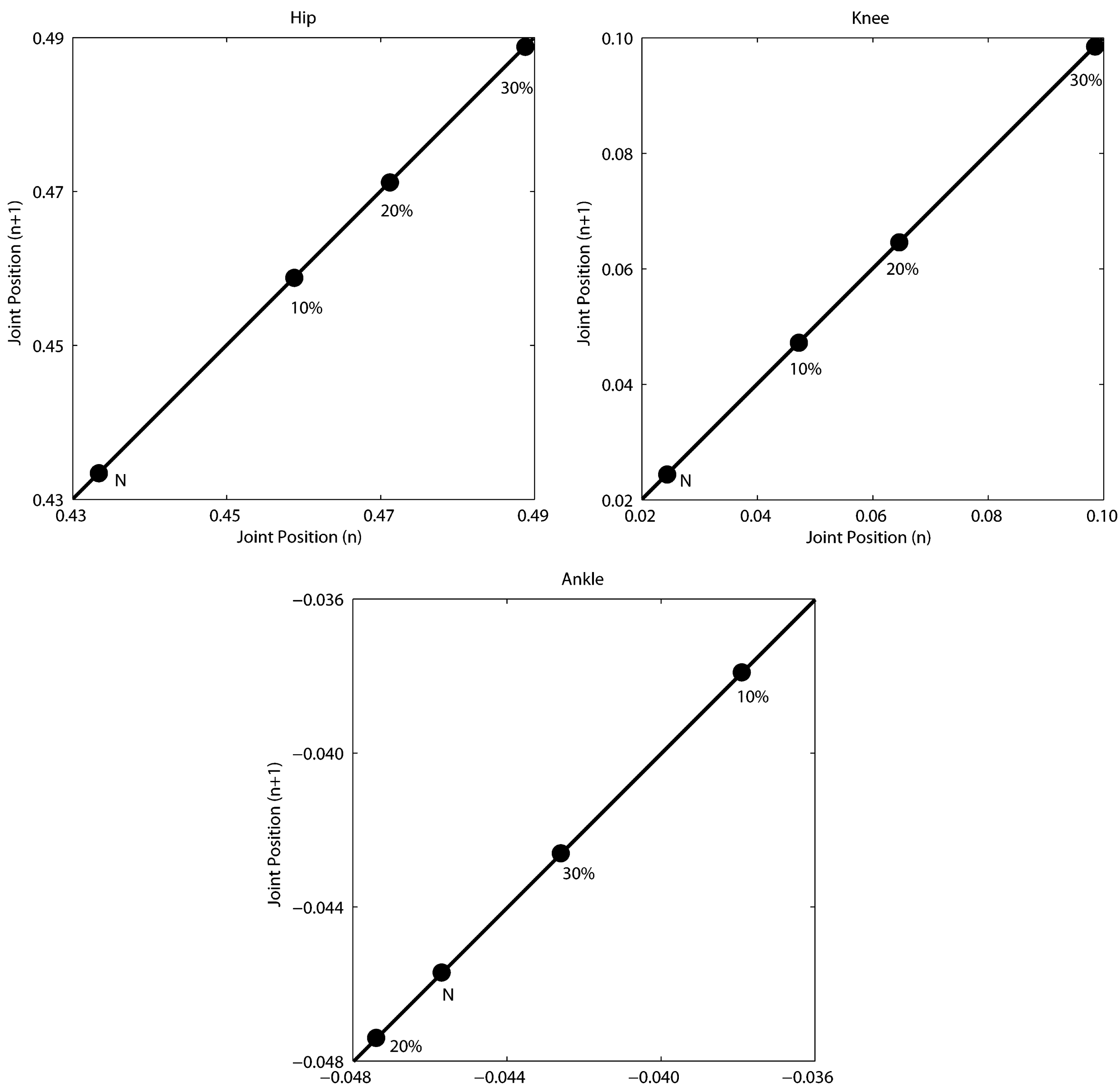

FIGURE 5—Poincaré maps demonstrating the upward and downward shifts in the equilibrium point (χ*) while walking with increased weight. χ* generated at the instance of heel strike and displayed in units of radians. The points represent the mean of all the subjects.

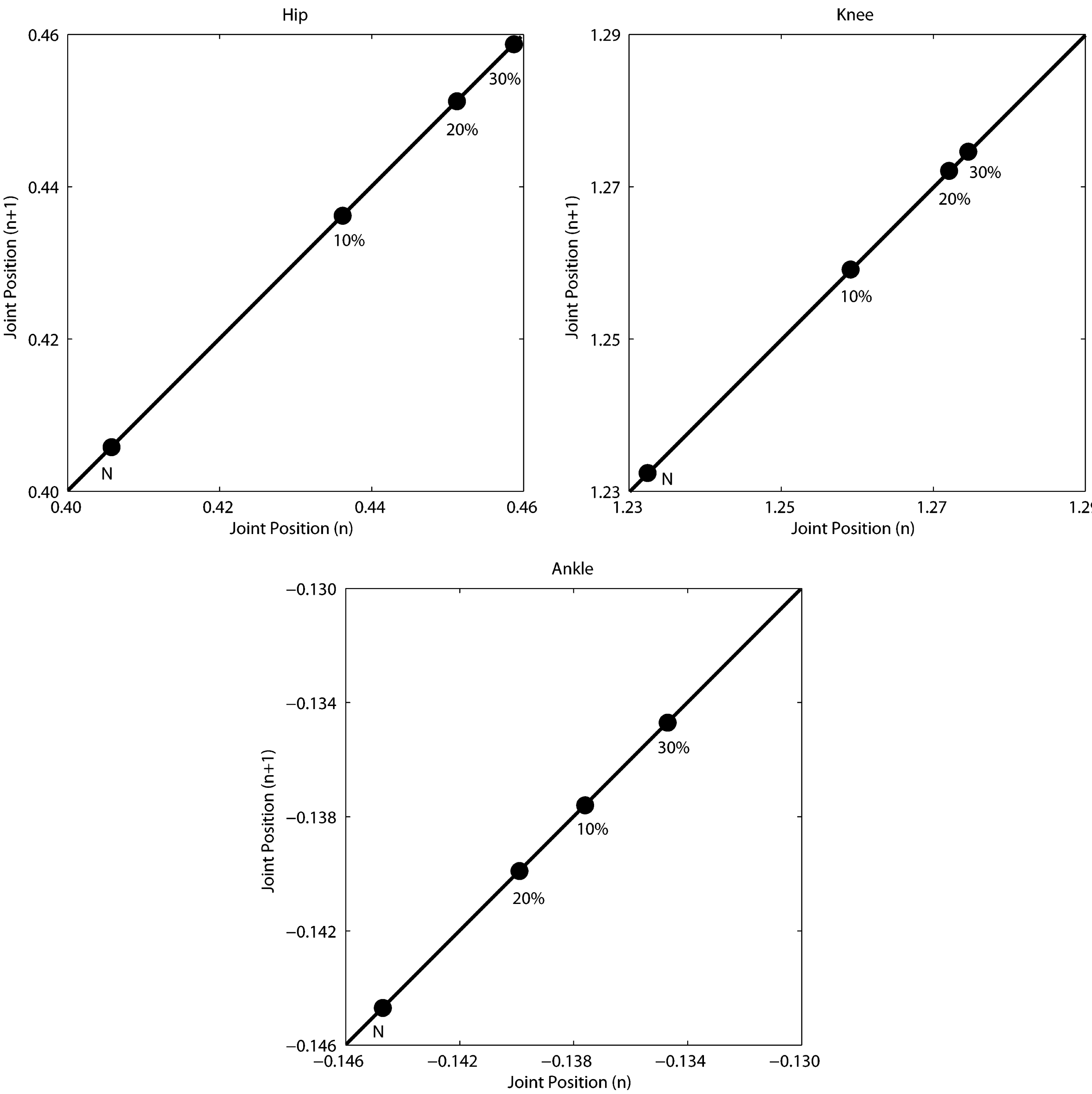

FIGURE 6—Poincaré maps demonstrating the upward and downward shifts in the equilibrium point (χ*) while walking with increased weight. χ* generated at the instance of maximum knee flexion and displayed in units of radians. The points represent the mean of all the subjects.

Compared with unloaded walking, the equilibrium points for the hip and the knee during midswing shifted significantly upward at 10% ($P < 0.001$ and $P < 0.001$, respectively), 20% ($P < 0.001$ and $P < 0.001$, respectively), and 30% ($P < 0.001$ and $P < 0.001$, respectively). The equilibrium points for the ankle at heel contact and midswing did not significantly change with additional loads ($P > 0.05$).

DISCUSSION

The dynamical systems analysis used in this investigation indicated that the participants remained stable in the sagittal plane while carrying loads about the waist of up to 30% body weight. Thus, the hypothesis that load carrying would decrease the stability of the sagittal plane gait dynamics was not supported. However, our results suggest that the participants carried heavier loads about the waist by increasing the flexion of the hip and the knee at heel contact and midswing. We speculate that these changes may have been used to maintain the stability of the gait pattern because these lower extremity variables were used to construct the state vector that defined the dynamics of the locomotive system (equation 1).

Previous experiments on the lower extremity kinematics while carrying loads have been mostly inconclusive. For example, several investigations have reported no significant differences in sagittal plane joint kinematics while carrying external loads (4,17,35), whereas others have reported differences (5,23,32). The results presented here support the notion that humans do have altered hip and knee joint kinematics while carrying loads around the waist. However, it should be noted that the significant changes in the sagittal plane kinematics noted here and elsewhere have been quite small, that is, less than 4° (32). We suggest that the slight changes may indicate that the leg kinematics is tightly controlled by the nervous system while carrying loads about the waist. A similar strategy has recently been reported for the trajectory of the foot under various limb-loading conditions (3,21). Potentially, the tight control of the limb performance allows for a more accurate and stable transfer of the weight from one step to the next. Alternatively, the slight kinematic changes at heel contact may be associated with additional shock absorption, lowering of the center of gravity, and control of the forward momentum of the body (16,23,32). As a result, the kinematic changes demonstrated by our subjects may be indicative of a possible strategy for maintaining dynamic stability while carrying externals loads about the waist of 30% body weight.

It is alternatively plausible that walking speed may have been a contributing factor in maintaining the stability of the gait pattern while carrying loads. It should be noted that the self-selected speeds (0.98 ± 0.24 m·s^{-1}) chosen by our subjects were slower than what has been previously reported (~1.25–1.38 m·s^{-1}) in the load carrying literature (13,23,27,36). Walking slower may have aided the subjects in maintaining dynamic stability while load carrying (7,9). However, we cannot conclude that this was the case because speed was not manipulated in this experiment. Future experiments should explore if walking speed may have influenced the outcomes observed in this study.

The results presented in this investigation indicate that the gait pattern is more stable at midswing than at heel contact while carrying loads in the sagittal plane. These results imply that a perturbation applied midswing would be more rapidly attenuated over several gait cycles than if the same perturbation was applied at heel contact. Potentially, the differences in stability may be related to the fact that heel contact signifies the point in the gait cycle where the body is directly experiencing the effects of the added load. Alternatively, it is possible that the differences may be related to the viewpoint that stance and swing phase dynamics are governed by different neural control mechanisms (10,12). The balance control mechanisms during the swing may be more robust to changes in added loads. However, additional studies are necessary to fully understand the differences in stability between the two phases of the gait pattern and how they may play a role in maintaining the balance of the gait pattern.

The data presented in this study provide further insights on the biomechanics of carrying loads. However, future investigations should consider if the outcomes presented here can be extended to larger loads and different walking speeds. This will help to establish a threshold for stable load carrying. In addition, recent investigations have noted that the frontal plane dynamics may require additional neural control and may be less stable than the sagittal plane dynamics (2,8,24). As a result, our future investigations will consider if the other planes of motion are stable while carrying loads.

CONCLUSIONS

Walking with an external load of 30% body weight about the waist did not influence the stability of the gait pattern in the sagittal plane. Contrary to previous investigations on load carrying (4,17,35), we found a slight amount of increased flexion of the hip and the knee at heel contact and midswing. We speculate that the changes at heel contact may be used to maintain dynamic stability by absorbing shock, lowering the center of gravity, and controlling the forward momentum of the body while carrying loads.

The authors would like to thank Joy Wilson for her expertise in creating the manuscript figures and to the two anonymous reviewers for their feedback that helped to shape the final manuscript. Partial support for this investigation was from the Texas Learning and Computational Center and the NASA NNX07AP91A grants awarded to MJK and the Texas Space Grant Consortium Fellowship awarded to CJA. The results of the present study do not constitute endorsement by ACSM.

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