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RESEARCH ARTICLE

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Lower limb kinematics during treadmill walking after space flight: implications for gaze stabilization

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Abstract We examined the lower limb joint kinematics observed during pre- and postflight treadmill walking performed by seven subjects from three Space Shuttle flights flown between March 1992 and February 1994. Basic temporal characteristics of the gait patterns, such as stride time and duty cycle, showed no significant changes after flight. Evaluation of phaseplane variability across the gait cycle suggests that postflight treadmill walking is more variable than preflight, but the response throughout the course of a cycle is joint dependent and, furthermore, the changes are subject dependent. However, analysis of the phaseplane variability at the specific locomotor events of heel strike and toe off indicated statistically significant postflight increases in knee variability at the moment of heel strike and significantly higher postflight hip joint variability at the moment of toe off. Nevertheless, the observation of component-specific variability was not sufficient to cause a change in the overall lower limb joint system stability, since there was no significant change in an index used to evaluate this at both toe off and heel strike. The implications of the observed lower limb kinematics for head and gaze control during locomotion are discussed in light of a hypothesized change in the energy attenuation capacity of the musculoskeletal system in adapting to weightlessness.

Key words Walking · Kinematics · Stability · Variability · Space flight

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Introduction

Both scientific and anecdotal evidence suggests profound changes in perceptual-motor functioning after space flight (Reschke et al. 1994). These changes pose conern for situations in which movements must be executed reliably and accurately. Locomotion, both on Earth following the completion of a Space Shuttle mission, or on a remote planet surface following a lengthy flight, is subject to compromise by changes in perceptual-motor functioning as a function of inflight adaptation to weightlessness.

Postflight locomotor changes of a biomechanical nature include increased angular amplitude at the knee and ankle, and increased vertical accelerations in the center of mass (Hernandez-Korwo et al. 1983). In addition, Chekirda et al. (1971) noted apparent change in the contact phase of walking, in which the foot appeared to be "thrust" onto the support surface with a greater force than that observed before flight. Chekirda also noted that, in efforts to preserve stability, cosmonauts spread their legs far apart, used their arms more, and used shorter steps after flight. Even with these compensatory changes, both Russian and US investigators have observed disturbances in performance including deviations from a straight trajectory (Chekirda et al. 1971) and a tendency toward loss of balance during walking when turning corners (Bryanov et al. 1977; Homick and Reschke 1977).

Locomoting through a complex and cluttered environment also involves perceptual demands, and a contributing factor to stable and reliable locomotion is the maintenance of stable gaze. Empirical evidence suggests that the head-neck-eye complex operates to minimize angular deviations in gaze during locomotion (Pozzo et al. 1990). Since the head–neck–eye complex is situated on top of the trunk–lower limb complex, the postflight biomechanical changes reviewed above suggest a high potential for negative impact on gaze stabilization strategies. The situation is further compounded by changes in perceptual function. For example, postflight, crew members develop a stronger dependence on visual cues (Anderson et al. 1986); there are documented changes in the ability to detect accelerations; and otolith organ sensitivity has been show to decline throughout the duration of a flight (Watt et al. 1986). In addition, changes in vestibulo-ocular reflex (VOR) gain as a function of space flight have been observed (Berthoz and Grantyn 1986; Viéville et al. 1986), and exposure to weightlessness appears to modify eye-head coordination during target acquisition (Kozlovskaya 1985; Thornton et al. 1988) and ocular saccadic performance (Uri et al. 1989).

When considered together these biomechanical and perceptual changes point toward a highly probable adaptation of head and gaze control during locomotion after space flight. However, strategies used for maintaining gaze stability have not been documented during postflight locomotion. To better understand the functional implications of existing flight-related evidence, especially in terms of the strategies used for coordination among the various perceptuo-motor subsystems, we designed an investigation to examine the role of adaptive modification in head movement control during postflight locomotor performance.

The investigation was designed to address this problem not only in terms of eye–head–trunk coordination, but rather as a problem from the ground up (cf. Jones et al. 1995). We contend an important element of gaze control during locomotion is the management of energy flow through the body, especially during high-energy interactions with the support surface such as those occurring at the moment of heel strike and toe off (McDonald et al., in preparation). The ability to attenuate the transmission of energy through the body is influenced directly by a number of factors. Among these are changes in the characteristics of the musculoskeletal shock absorbers, including the viscoelastic properties of joints (Voloshin et al. 1981). Also important for the management of energy flow through the body is the pattern of joint kinematics seen during locomotion. Of specific relevance is the lower limb joint configuration at the moment of heel strike with the surpport surface. Perry and Lafortune (1993) demonstrated that the absorption capacity could be reduced by excessive foot pronation, suggesting that the joint configuration of the foot-ankle at heel strike contributes directly to the potential transmission of the heel strike shock wave throught the body. Recall that Chekirda et al. (1971) documented postflight changes in foot activity during the contact phase of locomotion.

McMahon and colleagues (1987) suggest that the degree of shock wave transmission during locomotion was extremely sensitive to the degree of knee flexion. They discovered that while the tibial shock was increased with increased knee flexion, the transmission of the shock wave to the head was significantly reduced. It should be noted, however, that Lafortune et al. (1995) after a direct investigation of the role of knee angle on axial stiffness of the lower limb, suggested that increased knee angle at foot impact is less effective than previously thought in attenuating impact shock. Nevertheless, Hernandez-Korwo et al. (1983) noted postflight locomotor changes in both knee and ankle angles.

Grossman et al. (1988) recognized that locomotion induces rhythmic oscillations of the trunk and the head. The predominant frequency of these oscillations is equivalent to the step frequency. Since the head contains both the visual and vestibular systems, any irregularities in these step-dependent oscillations could influence locomotor control. Consequently, we determined it was crucial to examine not only the head–trunk linkage (cf. Assaiante and Amblard 1993) but all the links between the head and the support surface. Appropriate attenuation of the intersegmental energy flow during locomotion will minimize the disturbance of the visual and vestibular systems, and preserve head and gaze stability. However, we suspect that space flight adaptation may compromise this ability and thus lead to impaired head and gaze control. To determine more clearly the role of the lower limb joint complex in this phenomenon we chose to focus attention on two specific locomotor events: heel strike and toe off. These high-energy transitions between the stance and swing phases were considered the most likely events to illustrate changes in locomotor performance since any maladapted effort to manage energy flow would result in inappropriate energy transfer among contiguous body segments and could cause disturbances in both lower limb coordination and head-eye coordination observed during postflight walking.

Subjects and methods

Subjects

Volunteer subjects were recruited from the Astronaut Corps at NASA Johnson Space Center. Data are reported from seven subjects from three Shuttle flights flower between March 1992 and February 1994. Flight duration was either 8 or 9 days. Of the seven subjects, two were first-time fliers and five had flown at least once previously; six were men and one was a woman; and their heights ranged from 1.68 m to 1.85 m. Subject ages ranged from 35 to 49 (mean=41) years.

Apparatus

Before each testing session, passive retro-reflective markers that served as tracking landmarks were affixed at vertex, occipital and temporal positions on the head, and at the acromion process, lateral epicondyle of the humerus, midpoint on the dorsal surface of the distal portion of the radius-ulnar, C7, femoral greater trochanter, lateral femoral epicondyle, lateral malleolus, shoe surface coincident with the posterior surface of the calcaneus (an additional marker was affixed to the same location on the left foot also), and the fifth metatarsophalangeal joint, on the right side of the body. The movement of these markers was recorded simultaneously with four video cameras that sampled video images at 60 Hz. Ambient light was adjusted to allow high contrast between the retro-reflective markers and the surface to which they were attached. The position of each marker in space was determined with the aid of a video-based motion analysis system (Motion Analysis Corp., Santa Rosa, Calif.). In addition to kinematic data acquisition, the subjects were prepared for surface electromyography (EMG) of several muscles on both lower limbs, and for vertical electrooculography (EOG). These EMG and EOG data will not be considered here.

Table 1 Experimental conditions performed

Treadmill speed (km/h)	Visual target at 2 m		Visual target at 30 cm	
	Continuous vision	Periodic occlusion	Continuous vision	Periodic occlusion
6.4	Trials 2 and 4	Trials 3 and 5	Trials 6 and 8	Trials 7 and 9
9.6	Trials 11 and 12	Not performed	Not performed	Not performed

Each subject wore cycling shorts and a sleeveless shirt and the same brand of running shoe pre- and postflight. Foot switches constructed using Interlink Electronics force-sensing resistors were attached to each shoe at the heel and toe and sampled at a rate of 752 Hz through a 12-bit A/D board.

During each test session subjects were required to ambulate on a motorized treadmill (Quinton Series 90 Q55 with a surface area of 51 cm×140 cm) while visually fixating a centrally located Earth-fixed target (light emitting diode, LED) positioned either 2 m or 30 cm from the eyes and at a height coincident with the eyes in the standing position. To prevent injury through falling, each subject wore a full body harness that was attached to an overhead gantry. During nominal performance this harness provided no support and did not interfere with the natural movement of the limbs. Subjects were monitored by a "spotter" at all phases of the experiment to ensure their safety.

Procedures

To evaluate the influence of space flight, data were collected both pre- and postflight. The preflight testing schedule consisted of three testing sessions at 120, 60, and 10 days before launch. The postflight schedule consisted of testing on landing day (2–4 h after landing), then 2 and 4 days after landing (and occasionally 8 days after landing if nominal performance clearly had not been attained by the test session on day 4). In each test session subjects were required to stand upright (for the purposes of calibration), walk at 6.4 km/h (1.77 m/s) and run at 9.6 km/h (2.67 m/s) on the motorized treadmill while visually fixating the LED target. All trials lasted 20 s.

Table 1 illustrates the experimental conditions constituting each data collection session, with the trial number indicating the presentation order within each testing session. Additional walking trials were performed during periodic visual occlusion. Trials 1 and 10 were the standing trials used to calibrate the EOG system, and segmental kinematic data collected during these trials were used to calculate joint configurations during quiet standing.

Only the data obtained from *walking* (6.4 km/h) trials during *near* (30 cm) target visual fixation collected 10 days before flight (referred to as "preflight") and on landing day (referred to as "postflight") will be reported here since we consider this comparison the most likely to reflect any effect of space flight.

Before beginning the experimental trials, subjects were instructed to maintain ocular fixation of the target at all times. During each trial, the spotter monitored the location of the subject on the treadmill and instructed the subject to move forward or backward if necessary. For the walking trials, subjects stood off the treadmill belt while its speed was increased to the criterion. At this point the subject was free to begin walking. A few strides were permitted to allow the subject to become comfortable with the speed and to attain a steady gait. After a verbal "ready" indication from the subject, data collection was begun with the subject continuing to walk and fixate the target for 20 s.

Data analysis

The data presented here comprise a direct evaluation of the patterns of the lower limb joint kinematics observed during treadmill walking after short-duration space flight. The analyses are designed to determine the potential influence of lower limb kinematics on adaptive strategies utilized for head and gaze control during postflight locomotion.[1]

Basic characteristics of the temporal form of the gait pattern are examined since even while locomoting on a treadmill at a fixed speed there is opportunity to trade-off step amplitude and step frequency while maintaining the same forward speed. At the same time the relative duration of the stance and swing components of the step can be adapted. This composition is referred to as the duty factor, a ratio representing the amount of time spent in the stance phase in each step. By analyzing the temporal location of the toe off between two successive foot falls, we can identify the duty factor. The duty factor of bipedal walking is typically reported as approximating 0.6, i.e., the toe off occurs at about 60% of the step. Step-to-step variation of these temporal measures will be presented as a precursor to the joint kinematic analyses. Any changes in these factors could directly influence the frequency and amplitude of the rhythmic oscillations in the trunk and the head.

Several techniques will be presented to evaluate the lower limb locomotion system comprising the hip, knee, and ankle joints. Representing the periodic motion of these joints on the phaseplane, we have documented the *within-cycle variability* over discrete epochs of the cycle, and also at the two discrete events of heel strike and toe off. These analyses are performed on each joint independently, and are intended to document any disturbances in individual joint activity, and where this occurs relative to the phases of the gait cycle. To quantify the *cycle-to-cycle stability* in the gait pattern, a Poincaré map is used to take the continuous dynamics of the joint phase portraits into the discrete regime based on the event-specific iterations at heel strike and toe off. The states of the phase portraits (angular displacement and angular velocity) of the three lower limb joints are used to define a six-dimensional state space. Such a representation allows the exploitation of a specific analysis technique (described in detail below) to evaluate *system stability*. This technique is intended to evaluate the behavior of the three-joint system as a whole so that any changes in a single joint can be assessed at the "system" level. Thus we present independent measures of system component variability, and a measure of system stability as a whole. These measures are intended to determine changes in the nature and source of perturbations to the trunk emanating from the lower limbs during the locomotor cycle.

Marker trajectory data were processed to derive three-dimensional translation information relative to a coordinate frame coincident with the surface of the treadmill. Thus, subjects walked to toward the $+X$ direction and the belt moved in the $-X$ direction, the vertical axis orthogonal to the surface of the treadmill was $+Z$, and the Y axis was orthogonal to the X–Z plane (Fig. 1). The marker trajectories were low-pass filtered at 10 Hz using a finite impulse response filter with a hamming window. The filtered trajectories in X and Z were then used to determine joint angular motions in the sagittal plane for the hip (thigh and knee markers), knee (thigh, knee, and ankle), and the ankle (knee, ankle, toe). Figure 1 illustrates how these joint angles were determined relative to the coordinate frame of reference. The hip (H) angle was measured with respect to the vertical with flexion designated as positive and extension as negative. The knee (K) angle was measured from the projection of the thigh link segment to the tibial link segment, with flexion designated positive and extension as negative. The ankle (A) angle was measured as that between the tibial link segment

[1] Related papers in preparation will address other elements of this project, including neuromuscular activation patterns and head–eye–trunk coordination strategies during treadmill walking.

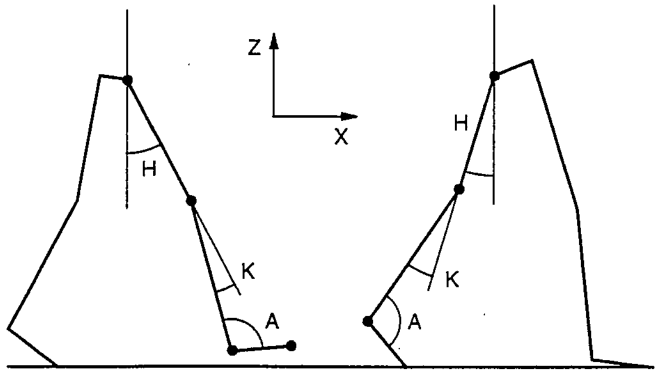

Fig. 1 The convention for joint angle measurements. H hip angle, K knee angle, A ankle angle

and the foot segment, with plantar flexion being greater than 90° and dorsiflexion less than 90°. These three joint angles were considered to be a satisfactory representation of the lower limb dynamic during the task of treadmill walking.

To facilitate the modeling of the oscillatory motion of the lower limb, the equilibrium position was determined for each joint under consideration. This position, determined for each subject, was equivalent to the joint angles measured during quiet standing on the treadmill. Hence, the hip, knee, and ankle joint angles were used to determine the equilibrium point about which the joint motions occurred for each subject. This equilibrium point is represented as the origin (0,0) on the phaseplane and all joint angular displacement data are represented with respect to this origin. Having determined the sagittal plane joint angular displacements, we determined the joint angular velocities with a fourth-order central difference algorithm.

The foot switch signals allowed determination of the moments of heel strike and toe off in the right limb. Foot switch information was not available for all subjects. In such cases reliable kinematic correlates for heel strike and toe off were determined from the toe marker velocity in the Z direction. Determining heel strike and toe off in this manner matched the foot switch information with an error not exceeding ±16.7 ms.

Phaseplane data, using the joint angular displacement and joint angular velocity as the states of the system, were analyzed using three different techniques to evaluate the joint dynamics. The first of these techniques was employed to evaluate the variability of independent joint motion over the course of the full gait cycle. The second was used to evaluate the variability of independent joint configuration at two discrete points in the gait cycle. For both these techniques a measure was constructed to combine the variability in the joint angular displacement and the variability in the joint angular velocity. After normalizing each gait cycle to 60 samples, the variability in the joint angular kinematics observed over the multiple cycles of one trial was quantified using the standard deviation about the mean joint angle and the mean joint angular velocity at the moment of heel strike and at the moment of toe off. The displacement and velocity standard deviation magnitudes were then used to define the diameter of the two orthogonal axes of an ellipse. The area of this ellipse is presented as an index of the variability on the phaseplane. To evaluate the variability over the full gait cycle, the cycle was broken down into five 20% temporal epochs and the variability from each of the 12 samples within each epoch was summated. The phaseplane variability at heel strike and at toe off is presented using those samples at which the named events occurred.

The third technique using the phaseplane data attempted to evaluate system stability. This technique utilized three lower limb joints in combination. The idea of using joint kinematics as state variables and the Poincaré maps to evaluate the stability of human locomotion was first introduced by Hurmuzlu (Hurmuzlu and Basdogan 1994; Hurmuzlu et al. 1994).

First return maps can be represented by the following finite difference equations in an n-dimensional state space:

$$X^k_{i+1}=f_k\,(X_i) \quad k=1,\,...,\,n \tag{1}$$

where x is a vector of state variables ($x=[x^1, x^2, ..., x^n]^T$) and f represents the nonlinear mapping function. The equilibrium values (steady-state) of Eq. (1) are known as fixed points of the map. Assuming that the fixed point of a map is defined as:

$$x^*=x_{i+1}=x_i \tag{2}$$

then one can analyze the stability of a dynamical system via linearizing Eq. (2) in the neighborhood of the fixed point to obtain:

$$\delta x_{i+1}=\mathrm{J}\delta x_i \tag{3}$$

where δx_i and δx_{i+1} represent the perturbations associated with the ith and $(i+1)$th elements of the state vectors and J is a $(n \times n)$ Jacobian matrix. The entries of this matrix are the partial derivatives of the nonlinear mapping functions (f_i, $i=1, ..., n$) with respect to the state variables, given as:

$$a_{kj}=\left.\frac{\partial f_k}{\partial x^j}\right|_{x^*} \quad j=1,\,...,\,n,\;\; k=1,\,...,\,n \tag{4}$$

Such a system is considered stable around equilibrium if all the eigenvalues of the Jacobian matrix lie inside the unit circle (Guckenheimer and Holmes 1983; Parker and Chua 1989). Bifurcations occur if the eigenvalue(s) move outside the unit circle, resulting in structural changes in the system.

Elements of the Jacobian matrix can be obtained easily if the nonlinear mapping functions (f) that return cross-sections of the flow to itself are known. However, the complexity of human locomotion does not permit simple determination of the functions (f) such as in Eqs. (1) or (2). Although mathematical models of locomotion are available in the literature (for a brief summary see Hurmuzlu and Basdogan 1994), we are not aware of any study that identifies an appropriate form of analytical equation or function. Consequently, we experimentally acquired the joint kinematics of human gait and constructed the Jacobian matrix by means of least square regression techniques (cf. Hurmuzlu and Basdogan 1994).

Following the procedures of Hurmuzlu (Hurmuzlu and Basdogan 1994; Hurmuzlu et al. 1994), we first identified the state variables of our system as the hip, knee and ankle motions in the sagittal plane. This resulted in a six-dimensional state space of the form:

$$X_{\text{space}}=\{\Phi_{\mathrm{H}}, \Phi_{\mathrm{K}}, \Phi_{\mathrm{A}}, \dot{\Phi}_{\mathrm{H}}, \dot{\Phi}_{\mathrm{K}}, \dot{\Phi}_{\mathrm{A}}\} \tag{5}$$

where Φ and $\dot{\Phi}$ represent the angular rotations and velocities of the three joints used in defining the conceptual model of the gait dynamics. These state variables were each sampled at the moment of heel strike and the moment of toe off, and the same data were used to construct the Poincaré maps. For each trial a mean value for each state was calculated and designated as the equilibrium value. The steady-state value of each state variable at each event (heel strike or toe off) was assumed to be the statistical average (mean) of all the samples. The deviation from equilibrium was then measured at each iteration for each state by calculating the difference between the mean state value and the state value at that iteration. A multidimensional regression was then performed among the vectors determined relative to the steady-state value in order to approximate the elements of the Jacobian matrix.

The set of equations that formulate this multidimensional fit can be written as

$$\begin{aligned}(Q_{\mathrm{H}})_{i+1}&=a_{11}\,(Q_{\mathrm{H}})_i+a_{16}(\dot{Q}_{\mathrm{A}})_i+p_1\\ &\;\vdots\\ (\dot{Q}_{\mathrm{A}})_{i+1}&=a_{61}\,(Q_{\mathrm{H}})_i+a_{66}(\dot{Q}_{\mathrm{A}})_i+p_6\end{aligned} \tag{6}$$

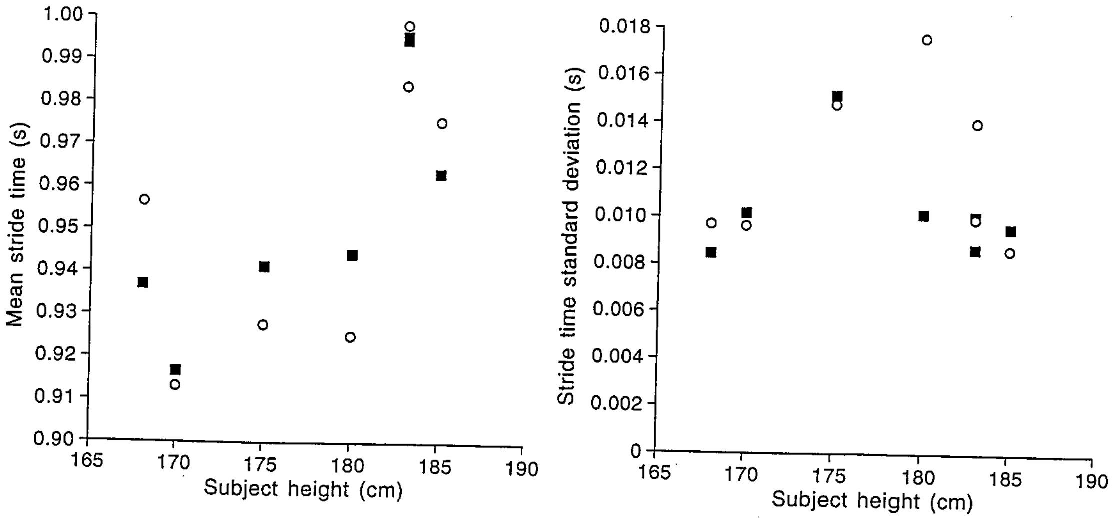

Fig. 2 Mean and standard deviation of the preflight (*filled squares*) and postflight (*open circles*) trial stride time for each of the seven subjects plotted as a function of subject height

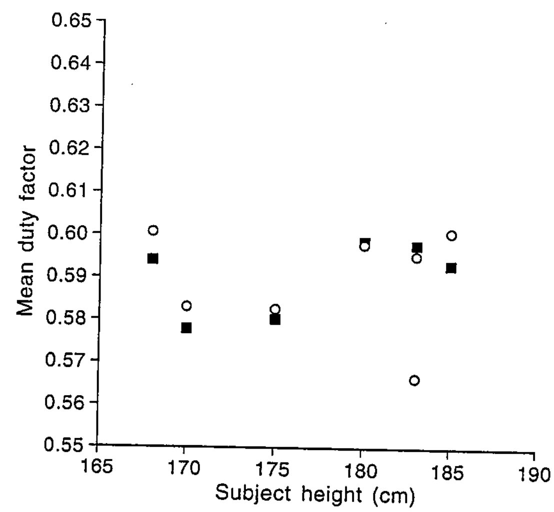

Fig. 3 Preflight (*filled squares*) and postflight (*open circles*) mean duty factor presented for each subject as a function of subject height

where Q_H, ..., $\dot{Q}_A$ represents the column vectors with a number of rows equal to the number of sampled locomotion steps (e.g., Q_H is a column vector indicating the deviation magnitude of the sagittal hip excursion relative to the steady-state hip excursion), a_{ij} values form the elements of the approximated Jacobian matrix, and p_i (i=1, ..., 6) are the constants of the regression.

Finally, we calculated the eigenvalues (λ_i, i=1, ..., 6) of the Jacobian matrix and statistically averaged them for each individual subject to quantify the dynamic stability exhibited by that subject during treadmill walking. According to stability theory, all eigenvalues should lie inside the circle, ($|\lambda_i|<1.0$, i=1, ..., 6) for a stable system (Guckenheimer and Holmes 1983).

Results

Temporal stride characteristics

Temporal stride measures are presented for two reasons. The first is to assess the task-specific performance of the lower limb system. The second is to evaluate a potential confound of the subject population. Subjects were asked to walk at a fixed speed of 6.4 km/h on the treadmill. However, preferred walking speed is closely related to subject height, and given the range in subject height measured in this sample, certain subjects may have had to walk at speeds other than their preferred speed. Evidence exists to suggest performance in a nonpreferred state may not be as stable (Kugler and Turvey et al. 1988). Consequently, we examined several simple temporal characteristics of the gait patterns relative to the height of the subject.

Figure 2 presents the mean stride time, and the standard deviation about this mean, as a function of subject height. The Pearson correlation of mean stride time and subject height was significantly different from zero and remained so after the flight (pre=0.820, post=0.681; $P<0.05$), indicating that mean stride time increased with increasing subject height and was not influenced by flight. The Pearson correlation between the standard deviation of stride time and subject height was neither significantly different from zero, nor did it change after flight (pre=−0.117, post=0.147; $P>0.05$), confirming that no simple linear relationship existed between stride time variability and subject height. Thus differences in subjects' height could not have influenced the postflight results.

Figure 3 illustrates the similarity of the duty factor for each subject pre- and postflight, and the lack of interaction with subject height. The mean duty factor both pre- and postflight was approximately 0.59, so that toe off occurred 59% of the way through the stride after heel strike. Paired t-tests of both the mean duty factor data and the within-trial variability of the duty factor identified no differences in the preflight versus postflight data ($P>0.05$).

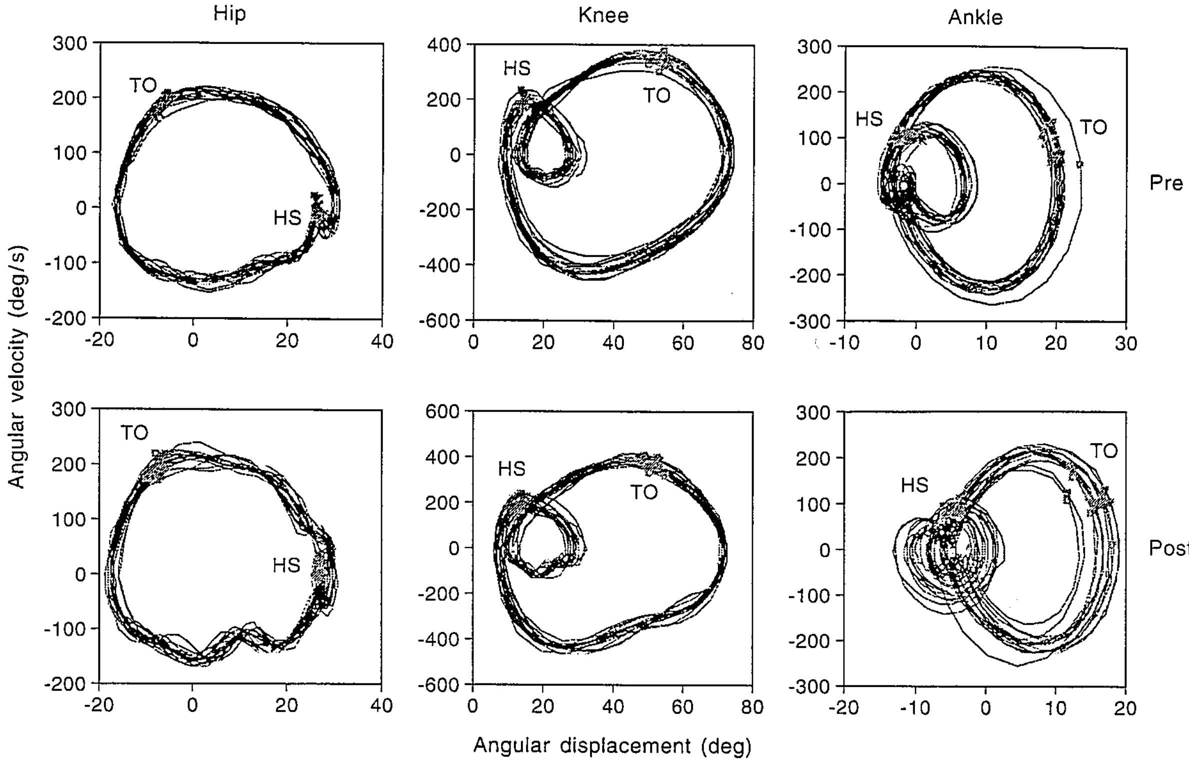

Fig. 4 Exemplar phase portraits of the three lower limb joints. These data are from one subject and illustrate 15 consecutive cycles from one preflight and one postflight trial. The location of heel strike (*HS*) and toe off (*TO*) for each cycle is indicated

Lower limb joint phase portraits

Figure 4 displays exemplar phase portraits, along with the identification of the location of heel strike and toe off, to help illustrate the degree of variability in the joint kinematics within a trial. The quantitative analyses that follow use data in this form to evaluate within-cycle fluctuations, changes in variability at discrete points within each cycle, and system stability.

Within cycle variability

The data presented in this section reflect the within-cycle variability on the phaseplane. Figure 5 presents box plots of the pre- and postflight data for the hip, knee, and ankle joints constructed from the seven subjects. In all three joints, the postflight variability is clearly higher than the preflight variability, at all epochs. Moreover, there are apparent differences in variability magnitude at the different stride epochs. The knee joint appears to have elevated variability around heel contact, and the ankle joint appears to have elevated variability about the swing phase. However, the sizes of the box and whiskers at many epochs in all joints indicate quite substantial individual differences in joint variability. Consequently, repeated measures ANOVAs on each joint revealed no significant flight or epoch effects at the hip and knee joints. Only the ankle joint displayed significantly higher postflight variability at the 0.05 level. Table 2 summarizes these results. In general these data suggest that postflight treadmill walking is more variable than preflight, but the response throughout the course of a gait cycle is joint dependent and, furthermore, the changes are subject dependent.

Discrete event variability

The data in Fig. 6 document the variability on the phaseplane at the moment of heel strike and toe off for each of the three lower limb joints. In most instances, variability is seen to increase after flight. However, paired *t*-tests of these data identified only the postflight increase in knee variability as significant at the moment of heel strike ($P<0.05$), with only the hip joint postflight variability being significantly higher at the moment of toe off ($P<0.05$). While the size of the box and whiskers in postflight measures on all three joints is indicative of substantial individual differences, the significant joint-spe-

Table 2 ANOVA results of phaseplane variability as a function of stride epoch, and flight

	Hip	Knee	Ankle
Epoch	$F(1,33)=3.4$, $P=0.074$	$F(1,33)=1.5$, $P=0.23$	$F<01.0$
Pre vs post	$F(1,33)=2.4$, $P=0.134$	$F(1,33)=2.8$, $P=0.10$	$F(1,33)=7.3$, $P=0.011$

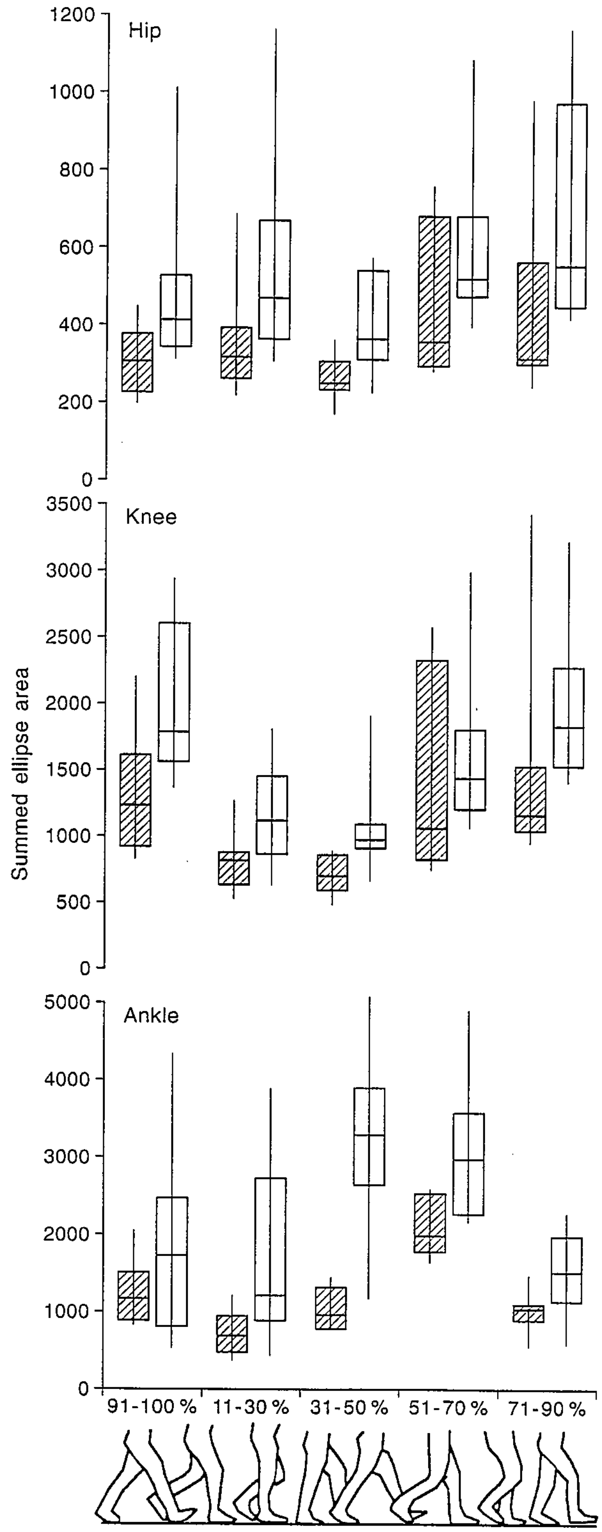

Fig. 5 Box plots of phaseplane variability as a function of stride epoch, from right heel strike to right heel strike. Epochs represent 91%–10%, 11%–30%, 31%–50%, 51%–70%, and 71%–90% of the gait cycle. *Hatched boxes* indicate preflight data, *open boxes* indicate postflight data. The *boxes* represent the 25th, 50th, and 75th percentiles, the *whiskers* represents the 10th and 90th percentiles

cific changes at heel strike and toe off emphasize the importance of these locomotor events.

Dynamic stability assessment

The data presented in Fig. 7 illustrate an index of dynamic stability calculated at the moment of heel strike and toe off during pre- and postflight performance. Paired t-test analyses of the data presented in Fig. 7 identified no significant difference between preflight and postflight at either heel strike or toe off. Furthermore, the stability index magnitude across subjects was quite consistent, as seen in the width of the box and whiskers.

As explained in Subjects and methods, the stability index is based on the eigenvalues of the Jacobian matrix, and a complete loss of stability is identified specifically by the index exceeding unity. Detecting a statistically significant difference in this stability index which never exceeded unity does not denote a qualitative change in the system dynamics from the perspective of nonlinear dynamics. However, such a result could be used to indicate a tendency to less stable behavior. The absence of any notable changes in the stability index is indicative of the preservation of lower limb intersegmental coordination.

Discussion

This investigation was designed to evaluate lower limb joint kinematics during treadmill walking after space flight, with specific reference to head and gaze control. Basic temporal features of the gait cycle, such as stride time and duty cycle, were seen to remain unchanged following flight. However, specific and consistent changes in joint phaseplane dynamics were identified at the moment of heel strike and toe off and, in general, variability was greater in the postflight tests. Nevertheless, the dynamic stability of the lower limb system during the transitions between the stance and swing phases did not seem to change following flight; it is clear, however, that the individual responses to flight need to be investigated further.

At the outset of this project, joint angular dynamics were expected to be significantly perturbed by space flight. The phaseplane variability by stride epoch data indicate an overall (but statistically insignificant) increase in variability at all three joints. The lack of significance can be attributed partly to the substantial individual differences. The expectation was that the susceptibility of the gait cycle to disturbance would be greatest around the heel strike and toe off events. These events represent significant energy exchange with the support surface, either through exaggerated impact at heel strike or through an exaggerated effort to propel the body forward at toe off. Generally, variability was higher in the heel contact epoch of the knee joint, both before and after flight, and this variability was exacerbated in much of the postflight

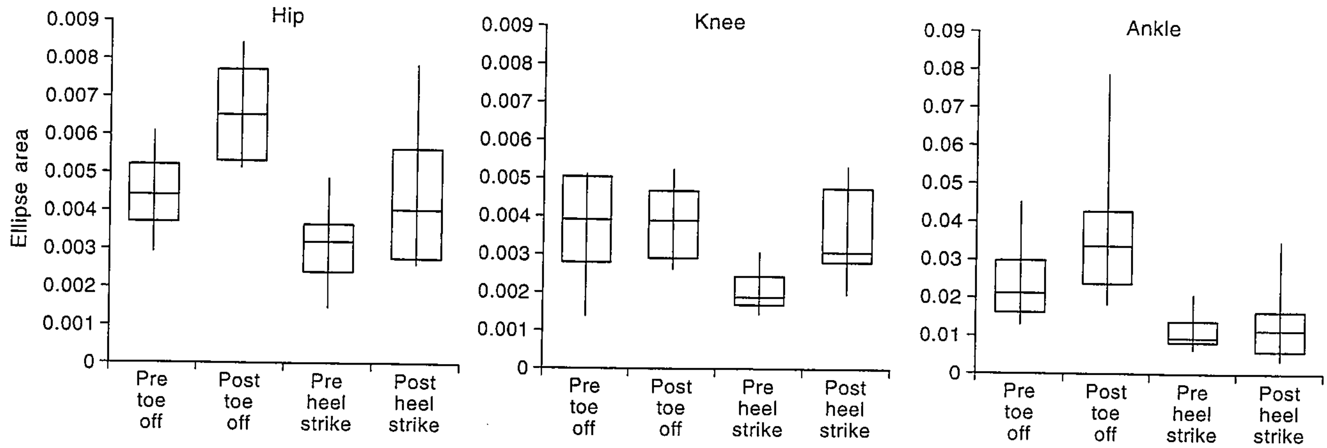

Fig. 6 Box plots of phaseplane variability of the pre- and postflight toe off and heel strike events for the hip, knee and ankle angles

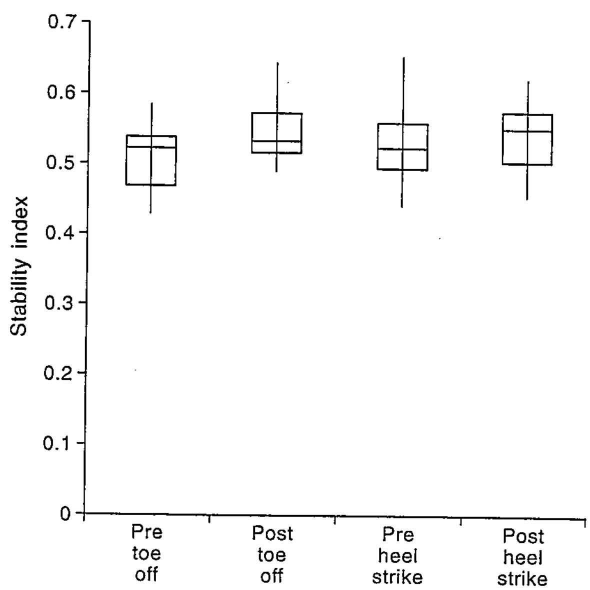

Fig. 7 Box plots of the system stability index for the pre- and postflight toe off and heel strike events

data. Similarly postflight variability in the ankle joint was higher for the epoch containing the toe off, both before and after flight, and this increase was exacerbated postflight. These data lend some indirect support to the possibility that these peak energetic events are the source of postflight disturbances in gait. Since the head and eyes are located atop a multisegmental system, any disturbance can propagate through these segments and consequently disturbances identified in the lower limbs may be related to the reported oscillopsia during walking after flight (Bloomberg et al. 1995).

Confirmation of the significance of these gait events was sought, with analyses focusing on joint variability at the precise moments of toe off and heel strike. At toe off, the initiation of the swing phase, hip joint phaseplane variability was significantly greater after flight than before. At the beginning of the swing phase, the hip is flexing and accelerating to maximum angular velocity, and thus this event is a strong candidate for perturbations of the trunk. At the moment of heel strike, the initiation of the stance phase, the knee joint phase portrait variability was also significantly greater after flight. McMahon and colleagues (1987) have demonstrated that while exaggerated knee flexion during running is energetically inefficient, this strategy does act to change the joint stiffness and consequently reduce the transmission of heel strike energy to the head. The increased variability observed in our data may be a result of attempts by the crew members to adjust the lower limb configuration about the moment of heel strike, indicating both a postflight increase in susceptibility to perturbations at heel strike and explicit attempts to modulate head perturbations resulting from the impact force of heel strike. Although drawing firm conclusions from this single piece of evidence is difficult, we are attempting to elucidate this issue with further analyses of new data.

As well as changes in joint variability, we had anticipated noticeable changes in system stability following flight, indicating at least a decrease of system stability, if not a qualitative change in system dynamics. These changes did not occur, indicating that the increased individual joint variability was insufficient to interfere with the basic pattern of lower limb coordination. The absence of significant changes in the index used to evaluate system stability at both toe off and heel strike is consistent with the subjects successfully walking on the treadmill after flight. However, the relationship between the pattern of joint coordination and the observed joint variability in light of the retained system stability needs to be investigated further. Relatedly, the variability seen in the lower limbs may propagate through the trunk to the head, where the consequences may be more profound for the strategies engaged in maintaining head and gaze stability.

We decided to use a treadmill protocol because it permitted parallel evaluation of full-body segmental kinematics and eye movements during locomotion. Only in this manner was it possible to evaluate head and gaze control strategies during locomotion. However, the use of the treadmill also subjected the locomotor performance to certain constraints. Some evidence suggests that treadmill walking is inherently less variable than overground walking. Nelson and colleagues (1972) observed that treadmill running was characterized by less variable vertical and horizontal velocities than overground running. Similarly, a comparison of the mechanical energies of overground and treadmill walking by Woolley and Winter (1979) found that the stride-to-stride variability of all work measures was significantly greater overground, suggesting that the treadmill constrains walking more rigidly.

We found that the temporal characteristics of the gait patterns were remarkably robust, as demonstrated by the lack of any significant change in either the mean duty factor or the variability of the duty factor. Consequently, subjects seemed to maintain a consistent stance-to-swing ratio, even on landing day. The basic stride data did illustrate a linear correlation between stride time and subject height. This is not surprising given the well-documented allometric relationships found in animal locomotion. However, there was no such relationship between stride time variability and subject height, and an absence of any influence of space flight on this feature. On a treadmill the appropriate locomotory state is well defined, with a specific, unvarying speed and little opportunity for directional error. Since treadmill walking is associated with low tolerance for error, the result of varying beyond the acceptable state is a complete failure in performance. In comparison, overground walking is much more forgiving, with much more opportunity for variance in speed and in directionality.

Indeed, some subjects opted not to attempt the treadmill protocol after flight, which suggests there may be gross changes in locomotor control in some individuals, beyond the relatively subtle changes we observed. The subjects from whom we acquired data performing treadmill walking at the criterion speed had, by definition, attained a relatively high and consistent level of coordination. The possibility of observing qualitative changes in coordination in the lower limb may be extremely slight given this constraint. Moreover, subjects were further constrained by having to fixate and maintain gaze on a visual target, thus further regulating their performance.

Our data are also subject to the unique constraints of space flight related research. Specifically, crew members can begin to readapt to the presence of gravity between the moment of Shuttle landing and the time at which the postflight data are collected (usually 2–4 h after landing). The readaptation rate during this time is particularly high (Paloski et al. 1992), and thus the postflight data must be evaluated with this in mind. In addition, first-time fliers often display more difficulty with postural control after flight than do experienced fliers (Paloski et al. 1992). A larger group of subjects might allow factors such as flight experience and flight duration to be addressed as experimental variables.

Nevertheless, we identified some subtle but consistent changes in postflight lower limb dynamics during treadmill walking. However, the data presented also confirm the heterogeneous nature of human adaptation after space flight. The significant changes at the moment of heel strike and toe off are encouraging for the hypothesized change in the attenuation capacity of the musculoskeletal system. Changes of this nature are intimately linked to changes in the strategies for the maintenance of head and gaze stability. Future analyses will focus on head–trunk coordination patterns in light of these changes, and an evaluation of the energy content in head vibration. Only then will we be able to comment on the functional relevance of these data and determine their operational consequences.

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